Force Feedback In Surgery: Physical Constraints and Haptic Information

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Abstract

Force feedback is widely assumed to enhance performance in robotic surgery, but its benefits have not yet been systematically assessed. Further, the implementation cost of force sensors that allow force feedback is prohibitive, due to the stringent design requirements imposed by the surgical environment. In this dissertation, we address both sides of this cost/benefit analysis of force feedback. We proposed a novel force sensor design targeting force feedback in surgery, as well as investigate the specific benefits allowed by force feedback.

We demonstrate the benefit of force feedback in surgery through a series of psychophysical experiments. By investigating performance on tasks with and without force feedback, we find that the primary benefit of force feedback is that interaction forces are reduced. This results in an increase in patient safety, because high forces correlate directly with tissue trauma. Two mechanisms enable force reduction and other benefits: 1) force feedback transforms environmental interaction forces into mechanical constraints and 2) forces act as a source of information to the surgeon. Mechanical constraints passively reduce intrusions into environmental structures (and, thereby, forces) due to the interactions of the compliance of the hand and the stiffness in the environment. Because this benefit is completely passive, it happens without cognitive response by the user. Accordingly, these benefits occur instantaneously, on the time scale of mechanical interactions. Force feedback also allows additional manipulation strategies that take advantage of these physical interactions, potentially reducing mental workload of the surgeon. We also find that force feedback provides information to the surgeon. While the number of ways that forces can potentially inform surgeons is large, the interaction between training and other sources of information is complex. We find that training is necessary, in some contexts, to take full advantage of the informational benefits of force feedback in surgery.

We propose a three axis force sensor design using strain gages and the Shape Deposition Manufacturing (SDM) technique, where components are embedded inside a pourable epoxy. The performance of the SDM based sensor (0-2 N range, 0.15 RMS calibration error, 0.15 N drift over five minutes) is comparable to a similarly sized metal element sensor built using standard strain gage design techniques. The resulting three axis sensor is small enough for minimally invasive surgical tasks, waterproof, and insensitive to temperature changes. Adapting the SDM design for mass production is straightforward because no machining is required of the SDM sensors (sensors are cast in reusable molds), resulting in a high performance, low cost, disposable sensor.

Contents

Acknowledgments

Sports metaphors are the commonly accepted comparisons used when describing a difficult process, such as getting a Ph.D. Getting a Ph.D. is like running a marathon, or getting a Ph.D. is like a baseball game that goes on for too long. Unfortunately, I have no major sports experience (does marching band count?) to compare to my time in graduate school. So, I'll fake it.

Graduate school is like Mike Tyson's Punch Out on the original Nintendo Entertainment System. For those of you who didn't waste your childhood playing video games, Mike Tyson's Punch Out was a boxing game where you played Little Mac, a young, smiling, up-and-comer trying to fight his way through a series of cartoonish opponents twice his size for a chance to go against Mike Tyson. I feel this is very close to the graduate school experience. In your first few bouts as a graduate student, you are easily able to handle Glass Joe's scientific computing class, or Von Kaiser's stochastic systems. At some point, the classes get more difficult, and you're surprised at how much effort you are putting into Piston Honda's nonlinear control theory. But you persevere. You learn the patterns of your opponents, and once you have those figured out, winning is straightforward. Go Little Mac!

Research turns out to be the same way. In the beginning, King Hippo is hard to defeat. But once you figure out his secret, you are able to KO a conference paper with little effort. There are ups and downs in research (boy, was Bald Bull tough), but every challenge is, in the end, fun.

Until you get to Mike Tyson.

Mike Tyson (a.k.a. Kid Dynamite) is the dissertation. You have your skills, you've practiced a lot to this point, but Tyson is something else. You still have to figure out his pattern and come up with a counterattack, but do so as he pounds you in the face over and over again. Same with the dissertation. It's the same game, but now it's harder, with all sorts of new challenges. Complicated equipment. Time deadlines. Post graduation plans. Career. All pounding you in the face while you are trying to figure out your thesis. And, there may be something else. There may be an element of luck there, confusing your efforts. You make gambles that may or may not pay off. Iron Mike is the toughest one to beat, by far.

In the end, you're not going to get the KO. You're not going to get the TKO, either. Realistically, you're just trying to go all fifteen rounds and win by decision. I think that represents the graduate experience pretty well. Fun for the most part, until the end, when you're just trying to leave the ring on two feet.

Little Mac had a coach named Doc Louis that encouraged Mac when he was down and gave him tips for the upcoming fights. I don't know if Doc actually had a doctorate degree, but his advice was always sound, and his support never wavered. I think Doc Louis is the real reason that Little Mac is able to beat Mike Tyson. Without that support and encouragement, there is no way that Little Mac would have kept stepping into that pixelated ring, day after day.

My family, friends, and coworkers are the Doc Louis that helped me beat Iron Mike. I am grateful and lucky to have such a supportive group in my corner. They are the ones that helped me through the tough times while also making graduate school fun.

My family has been a never ending source of encouragement and love. Thank you, Mom and Dad, for your support these six years. Being so far away from family and home was tough, but knowing a loving home was there gave me strength. I also owe my parents thanks for their words of wisdom. My Mom would always assert the importance of a good dinner, something that I have taken to heart, especially when putting in long hours working. As for Dad, I'll get a job soon, I promise. I also want to thank my brother David for his support during my time in graduate school. David is also pursuing a Ph.D., so he knows what the struggle is like. More importantly, he is able to relate to the parents why graduate school is hard and takes a while to finish.

My friends and roommates have made the time that I don't spend in lab fun and stress free. When I moved to Boston from California in 2000, I knew no one here. I moved into the dorms to try to meet people and enjoy cinder block walls, cramped rooms, and shared bathrooms. Luckily, the cave-like living conditions of Child Hall forced everyone into the common areas, where I met many friends and future roommates. Many thanks to Scott Schnee, Aly Shamji, Becca Norwick, and Derek Greenfield. We made an overpriced apartment with an overbearing landlord a home. I also want to thank the Survivor crew, Dan Matross and Becca Norwick, for keeping alive a ritual that provided a center to my life. Kathy Aidala has been a wonderful friend for the past five years, keeping me interested and active in the Harvard Wind Ensemble, the Harvard Juggling Club, and highly competitive squash matches. It's been fun. Many thanks to Jason Gallicchio for Dragonboating and China, for the Helmet, and continuing to show me how fun engineering can be. A final thanks to all of my friends who have participated in my numerous studies. You have truly made this dissertation possible.

The person who has done the most to enable my enjoyable graduate school experience is my adviser, Rob Howe. The best decision I have ever made was to come work for Rob. He has helped make my time here fun, challenging, and enjoyable in so many ways. While I do want to thank Rob for his role as an adviser (I have learned much about engineering, problem solving, communication), what I respect him for the most is how he has set up a work environment that provides intellectual freedom with little external stress. I have been able to explore interesting tangents and decide what I want to work on during my time here. There have always been interesting problems to investigate and resources available to do the work. Whenever there has been a deadline, Rob has always been willing to work toward it in parallel with the graduate students. Thank you, Rob, for this wonderful experience.

I can't imagine a better group to work with than the Biorobotics Lab. Every single member I consider my friend as well as coworker. Everybody is smart and fun to be around. I will miss the lab the most after leaving here. Thank you to Anna Galea, Heather Gunter, Solomon Diamond, Marius Linguraru, Paul Novotny, Amy Kerdok, Aaron Dollar, Shelten Yuen, Petr Jordan, Masashi Nakatani, and Yuri Ishihara. It has been an honor and a pleasure working with you. Special thanks to Yuri Ishihara whose help and knowledge of surgery has been invaluable for my final experiment. I also want to thank all the undergraduates who I have worked with during my time in graduate school - Jaime Lee, Cami Lau, Ross Feller, and Diana Lee. Your enthusiasm and competence have been amazing. I have had so much fun working with you. Playing volleyball on the sand court with you guys is one of my fondest memories of graduate school. Another thank you to Amy Kerdok for being my compatriot during these final months. We did it!

Many thanks to Doug Perrin, who has provided support in many ways throughout his time here. Support for my research, emotional support when enthusiasm was waning, support across the board. If life were a big game of Battlefield 2, Doug (he likes to be known as Dr. Perrin) would be the Support class, and he'd be great at it (yes, Paul, you can be the Sniper). I will never forget the nights that Doug and I stayed up all night, trying to finish a paper or a project. Sometimes we had a deadline, and we were working because we had to. Other times, we worked all night just so we could say we finished a project in just one night. If that's not fun, I don't know what is. Thank you, Doug, for being a wonderful friend.

The final member of the Biorobotics lab that I want to acknowledge is Ryan Beasley. I couldn't ask for a better friend to experience graduate school with. Ryan and I started in the Biorobotics lab at the same time and took all the same classes. We spent so much time together, working on coursework, and I have nothing but fond memories. We would work until midnight, every weekday, when Ryan would need to catch the T to go home. Saturday mornings, I would go over to Ryan's place and we would work some more, with Ryan's wife Joanna being so supportive and making us lunch, even though she had her own graduate studies to focus on. We would take Sunday morning off, then start the whole process over again. Much of my happiness and success in graduate school is due to Ryan. The summers that we spent juggling, throwing the frisbee, learning to play squash are treasured memories. Ryan is also a fantastic person to talk to about research. Whenever I'm confused about my work or unsure of the next step (which happens daily), I know I can talk to Ryan and he can make sense of it all. I'm terrified of the future when Ryan's not going to be right there to talk to about work, or movies, or video games, or anything. Ryan Beasley, you're my hero.

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I also would like to acknowledge the wonderful collaborators that I have worked with. Research is easy when collaborating with bright, enthusiastic people. Many thanks to Professor Susan Lederman, Dr. Nick Stylopoulos, Dr. Patrick Jackson, Dr. Ben Lee, Dr. Nikolay Vasilyev, and Dr. Pedro del Nido. This dissertation would not have been possible without their effort and generosity. I also would like to thank my funding sources, the National Institutes of Health (R01 HL073647-01) and the National Science Foundation (EEC-9731748 and Graduate Research Fellowship).

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For Monica

Chapter 1

Introduction

Robotic telesurgery attempts to supplement the performance of the surgeon with a robot, overcoming some of the difficulties encountered with minimally invasive surgery while reducing trauma for the patient. In typical robotic telesurgery, a robotic manipulator passes through a small port in the body to interact with internal tissue. The surgeon controls the motion of the robot by moving an interface mechanism whose motion is mapped to the robot. This is in contrast to semi-autonomous robotic surgeries [50], in which the robot executes precsie motions without direct surgeon control. A complete robotic telesurgical system would consist of two interface mechanisms (one for each hand) and their corresponding instruments, as well as an endoscopic camera and monitor for visual feedback. The system can provide precise positioning, motion scaling [47], dexterity enhancement through added degrees of freedom [33], and hand tremor reduction [118, 27]. Even with these benefits, however, many surgical procedures are still challenging and time consuming [132].

One obvious feature missing from current robotic surgical systems is the ability to transmit the sense of touch. This ability, also known as force feedback, would involve sensing interaction forces at the instrument tip and recreating those forces against the surgeon's hands. The lack of force feedback in current robotic surgical systems is a notable one because surgeons are accustomed to having the sense of touch in open surgeries. Further, surgeons often note that force feedback would be a desirable feature in robotic surgical systems. Why, then, don't current systems incorporate force feedback? One reason is that high-fidelity force information is not essential for all surgical tasks [10], as surgeons regularly execute a wide variety of minimally invasive procedures using hand-held instruments that provide little haptic information. Similarly, even though current commercial robotic surgery systems provide no force feedback from the instruments, surgeons have demonstrated the ability to use these systems to perform delicate procedures such as coronary artery bypass grafting [106, 111]. Despite this demonstrated ability to work without force information, dexterity with current minimally invasive instruments, manual or robotic, is clearly less than optimal. However, we do not yet understand the role of force sensation in surgical tasks, so it is unclear how force feedback could benefit minimally invasive systems. Principled investigations into the precise role of force feedback in different surgical tasks could allow better technology design that improves surgical performance and patient outcome. The goal of this dissertation is to elucidate these principles and provide a foundation for incorporating force feedback into robotic surgery.

Force feedback is an important potential feature of surgical robots. While a number of different approaches exist for improving surgical performance with a robot, force feedback offers the unique possibility of providing a benefit to the surgeon with little increase in mental workload. Because surgeons are accustomed to manipulating tissues in the presence of forces, restoring this sense of touch through force feedback can draw upon that previous experience. Furthermore, force feedback requires no prior knowledge (such as preoperative imaging or registration) to provide a benefit. The goal of the present study is to examine the objective benefits of force feedback in surgery to allow a structured cost/benefit analysis. Although force feedback is likely to be useful, a benefit must be clearly demonstrated because the technological cost associated with force feedback is known to be high.

1.1 Forces In Surgery

Force perception is important in surgery because the lack of force feedback fidelity in laparoscopic procedures has been shown as a cause of errors [100], see [132] for a review. Although forces are transmitted in minimally invasive surgery [6, 95], they are not the same as in open procedures [41]. Studies have investigated the magnitude of forces both on instruments during surgery [59, 56, 29] and on the surgeons hands [99, 97]. These studies, however, do not make the link between perceived or applied forces, and task performance.

Forces feedback could assist surgery through a number of mechanisms. Vibratory forces have been shown to aid performance in a puncturing task [67]. Spatially distributed forces have been shown to be important in palpation tasks such as lump detection [75]. Here we focus on the forces that arise from tool based interaction with soft tissues, which encompass many surgical tasks.

1.2 Force Feedback and Task Performance

A complete telerobotic, force feedback enabled surgical robot incorporates a remote manipulator, force sensing at the remote end, and force recreation to the user. Previous work on force feedback in surgery has focused on involving individual components or subsets of the complete system. No work has investigated a complete force feedback enabled telerobotic surgical system.

With the rise of commercial haptic interface technology and increases in computing power, surgical simulation has become a viable technique for mimicking surgical interactions. Surgical simulators have the interface (both visual and haptic feedback) of an actual surgery, but the feedback is produced by a computer simulation instead of a real physical environment. Simulators are useful for training [85, 110], planning [86, 108], and skill assessment [91]. These tasks can even be tailored to mimic a specific patient (e.g. [83]). Most previous work in surgical simulation is simulating the surgical environment realistically enough so that training done on the simulator applies to the real surgery. This is a tradeoff between accurately computing the deformations and tool/tissue interactions [3, 25, 54], using models that accurately characterize the loading of real tissue [101], and speed of computation [2]. Some work has looked at the necessary haptic fidelity needed [12, 71]. There is also a large body of work generating simulators tailored to a specific surgical operation, from needle insertion [28, 37] to suturing [49] to cutting with scissors [129]. A number of general simulators exist as well [88, 119]. While all of these are based on the hypothesis that forces are necessary for an accurate simulation, none examine how these forces translate into a performance benefit in actual surgery.

Although a number of surgical teleoperators exist, and some provide force feedback, little is known about the mechanisms by which force feedback improves surgical performance. Even without force feedback, teleoperators can increase dexterity [33] and allow manipulation on a finer scale, such as microsurgery [47]. While some teleoperators do provide force feedback, research has focused on device design [53, 69, 1], neglecting to assess the mechanisms by which force feedback benefits surgery. Some work has examined time delays that exist for long distance [92, 14], but we focus on cases where the surgeon interface is close to the surgical robot where time delays are negligible. Some investigators have focused on optimizing teleoperator dynamics for surgery [60, 17], making the assumption that force feedback is useful when discriminating compliances. Finally, some work has shown that teleoperators missing some axes of force feedback may provide similar benefits as those with full force feedback capabilities [104].

Virtual fixtures are a feature of teleoperation systems that are meant to restrict motions to lie along a path by providing an opposing force, or prevent users from entering restricted area [21, 48]. Virtual fixures can also be used to restrict users to a certain velocity or force. While this is a benefit that robots can provide in surgery [27], it is not true force feedback, thus is limited in application due to the need to integrate additional information (such as preoperative imaging) to form the boundaries.

Sensory substitution is a way of providing force information back to the user by using senses other than the natural force propagation pathway through the hand. An example would be a visual representation of applied force [66, 116] or a scaled vibration applied to the hand. While this is not true force feedback, as all physical dynamics between the hand and the interface are removed, the relative benefits between sensory substitution and force feedback should reveal insights to the mechanisms of force feedback benefit. The benefits that sensory substitution provide may be subject to an increased amount of cognitive processing, increasing mental workload and possibly increasing the time required to respond to that information.

A considerable body of work has also appeared on the development of force feedback technology, including the design of force-sensing surgical instruments [5, 84, 80, 7, 117], and force feedback instruments [98, 93, 133]. It is generally recognized that force sensing would benefit surgery, consequently researchers have developed a number of sensing technologies, including optical [46, 73, 94], strain gage based [5], and piezoelectric [24]. Force sensor costs tends to be high, however, even for low performance devices because the surgical environment requires that they be small, sterilizable, water resistant, and robust to changes in temperature. Knowing the mechanisms of force feedback benefit may lead to insights simplifying force sensor design.

Some work has investigated the benefit of force feedback outside the realm of surgery. The addition of force feedback may reduce musculoskeletal loading [26]. Other investigations point to reductions in mental workload [90, 19, 76] or increases in user satisfaction [64].

Task based performance evaluation using force feedback outside the realm of surgery has centered on interaction with stiff objects [107], however, which does not necessarily mimic the interaction with soft tissue that dominates surgical manipulation.

Several studies have attempted to directly answer the question of how force feedback helps in surgery. Few of these studies have focused on the role of force feedback in manipulation of soft tissue, which is the central aim of most surgical procedures. The bulk of the work is on distinguishing material properties when grasping [123, 13, 8, 45] or when cutting [39]. Kitigawa et al. demonstrated the usefulness of force feedback during a knot tying task with fine suture, but did not use a surgical environment [66]. Other task based studies include suturing [116], and Nissen fundoplication [96]. Additional benefits to the patient may include increased safety, due to knowledge of trauma causing forces, or increased speed of certain manipluations, leading to a decreased operation time. Clearly, a number of potential benefits of force feedback to surgery exist, and a principled framework is necessary for a broad analysis.

1.3 Overview of Experiments

The presented work establishes hypotheses on the mechanisms of benefit of force feedback in surgery and establishes a framework through which further analysis can be conducted. A motivational experiment is presented in Chapter 2, demonstrating various benefits of force feedback in surgery and showing that some benefits exist independent of surgical experience. Chapter 3 examines the passive mechanical benefit of force feedback, which can be modeled and quantified. This approach can be used to evaluate the relative role of force feedback as a passive constraint versus an information source. The role of force feedback as an information source is further developed in Chapter 4, where an experiment is presented that examines the tradeoff in performance in a task with two information sources, vision and force feedback. Two designs for inexpensive forces sensors are presented in Chapter 5, where the designs are motivated by the constraints of robotic surgery and the mechanisms by which force feedback can provide a benefit. Chapter 6 presents a final experiment examining the benefits of force feedback in a cannulation task, using a two instrumented, force feedback enabled surgical robot. Conclusions integrating the role of force feedback in surgery, passive and informational benefits, and the role of training are presented in a final chapter.

Chapter 2

Analysis Of Blunt Dissection

2.1 Introduction

Ask a surgeon if force feedback is needed for robotic surgery, and the answer is predictably yes. The basis for this intuitive answer is perhaps less immediate. High-fidelity force information is certainly not essential for all surgical tasks, as surgeons regularly execute a wide variety of minimally invasive procedures using hand-held instruments that provide little haptic information. Similarly, current commercial robotic surgery systems provide no force feedback from the instruments, yet surgeons have demonstrated the ability to use these systems to perform delicate procedures such as coronary artery bypass grafting [111, 106]. Despite this demonstrated ability to work without force information, dexterity with current minimally invasive instruments, manual or robotic, is clearly less than optimal. What is lacking is an understanding of the role of force sensation in surgical tasks that would allow a principled assessment of the benefits of force feedback systems.

In this study, we experimentally evaluate the role of force feedback in blunt dissection, a surgical manipulation task frequently employed in minimally invasive surgery. Our hypothesis is that force feedback is useful in this context when there is a large contrast in mechanical properties along the dissection plane between adjacent regions of tissue. Subjects in the experiments used a laboratory telesurgical system with high fidelity force feedback to dissect a relatively stiff lumen from a softer substrate. We compare their abilities to perform this task with varying degrees of force feedback. Further, we evaluate performance variation with amount of previous surgical experience. The results presented here indicate that force feedback allows more precise dissection with lower applied forces and fewer errors, independent of surgical training.

2.2 Methods and Materials

We selected dissection as the focus of these experiments because it is an important surgical task, accounting for 25-35% of the time spent on most surgical procedures [103]. Additionally, dissection ranks second in terms of the estimated effort required for performing a surgical task. Dissection is most often performed using scissors or specialized dissectors such as hooks and coagulators. Regardless of the instrument used, dissection is composed of three distinct phases: (1) tissue recognition, (2) accurate instrument positioning, and (3) tissue cutting/spreading. While carrying out the dissection, the surgeon tries to minimize tissue trauma and preserve surrounding structures. We have chosen to use a hook dissector because of its simplicity and popularity in general laparoscopic surgical procedures.

2.2.1 Telemanipulation System

The experiments use a laboratory teleoperation testbed consisting of two Phantom haptic interface devices (Model 1.5, SensAble Technologies, Inc., Woburn, Mass.)[16]. One Phantom acts as the surgeon master controller and the other acts as the surgical robot. The master is an unmodified Phantom with the stylus attachment. Subjects control the motion of the surgical robot by moving the stylus, held in a pen grasp, where the tip of the stylus maps to the proximal end of the instrument shaft.

The instrument used for the blunt dissection task is a right angle hook with a depth of 1 cm, a diameter of 0.9 mm, and a rounded tip. The hook is attached to a 50 cm rigid shaft that passes through a fixed pivot, simulating the incision into the patients abdomen. The surgical robot is attached to the proximal end of the shaft with a two degree-of-freedom joint that prevents rotation of the instrument (Fig. 2.1).

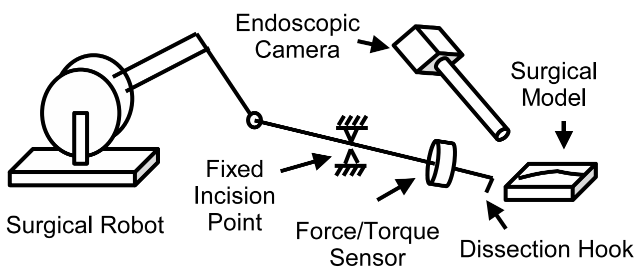


Figure 2.1: Surgical setup

Forces are sensed at the tip of the instrument by a six-axis force/torque sensor (Nano43 transducer, ATI Industrial Automation, Apex, North Carolina) built into the instrument shaft. The master and surgical robot are controlled with the bilateral force feedback controller (i.e. position feedforward and force feedback) traditionally used in teleoperated systems[107]. The Phantom control computer samples the forces at 1 kHz and transforms the forces to the proximal end of the shaft by assuming that the instrument shaft acts a perfect lever. That force is scaled for the appropriate experimental condition and then reproduced by the surgeon master controller; ideally, this results in the user feeling the forces that would be experienced if the stylus tip was attached directly to the proximal end of the instrument shaft. The force on the master is thus given by

$$f_{master} = g_{ff} A(x_{robot}) f_{sensor} \tag{2.1}$$

where g_{ff} is the force feedback gain and A is the position dependant matrix that transforms the sensor force fsensor so as if acting on the proximal end of the instrument shaft.

The teleoperation system, including the master, the surgical robot, and the force sensor, are controlled by a 333 MHz Pentium computer running Windows NT. The surgical robots position is controlled using proportional position/velocity control, independent of force feedback, defined by

$$f_{robot} = k_p(x_{master} - x_{robot}) + k_d(\dot{x}_{master} - \dot{x}_{robot}) \tag{2.2}$$

where x_{master} is the position of the tip of the interaction stylus, xrobot is the position of the connection between the surgical robot and the proximal end of the instrument shaft, and $k_p = \begin{bmatrix} 0.5 & 1.0 & 0.5 \end{bmatrix}^T$ N/mm and $k_d = \begin{bmatrix} 0.0001 & 0.0010 & 0.0005 \end{bmatrix}^T$ Ns/mm. These values were empirically derived to provide uniform stiffness in the portion of the workspace used for these trials while maintaining stability of the teleoperation system [16]. The control algorithm is implemented in Visual C++ along with the force/torque sensor interface.

2.2.2 Visual Feedback

The subjects received visual feedback from a fixed surgical endoscope, camera, and light source (Telecam SL NTSC/Xenon 175, Karl Storz Endoscopy-America, Inc., Culver City, Cal.), to provide the same visual feedback encountered in minimally invasive procedures. The relative orientation between the master controller and the monitor is approximately the same as the orientation between the endoscope camera and the instrument, to minimize the mental effort of relating visual and instrument frames [121]. However, lack of depth perception and the laparoscopic movement constraint at the incision point remain sources of difficulty.

2.2.3 Surgical Models

The surgical models used are intended to simulate a vital structure such as an artery embedded in its surrounding tissue. Two types of model were constructed: in one the artery was visible through the tissue and in the other the tissue completely obscured the artery. These models contain materials of different stiffness on the order of the pertinent biological tissues to provide realistic stiffness contrast. Further, the models are straightforward to dissect with a fixed endoscopic view and an instrument with fixed orientation.

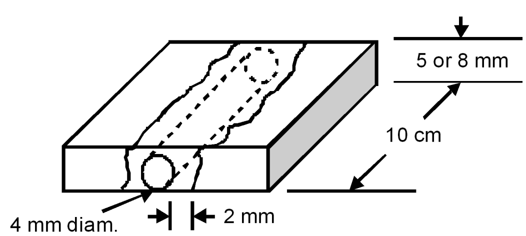


Figure 2.2: Surgical model

The material chosen to simulate the tissue bed is a laboratory-made clay. The artery is represented by a stiffer clay material (Weatherstrip and Caulking Cord, Mortite, Inc., Kankakee, Ill.) in cylindrical strips 4 mm in diameter. The tissue bed clay is colored pink to provide visual contrast to the gray artery material. The key feature of the clay tissue model is that it is a reproducible material that captures the plastic failure that is the goal of blunt dissection procedures. This provides the advantage of repeatability over biological tissue, removing the effects of model variation between trials. To quantify the material properties, we measured the steady dragging force of the blunt dissection hook embedded 5 mm into the model tissue material as 0.5 N, and embedded into the model artery material as 3.5 N.

A uniform and easily replicated process was used in the construction of these models. To fabricate each model, we placed a straight 10 cm length of artery on a mass of clay, then compressed the model with a flat plate to a uniform height. For the model where the artery was visible, the tissue was compressed to a height of 5 mm. For the obscured artery case, the model was compressed to 8 mm and the model was then flipped and squared off to regular dimensions, so that the artery was at the bottom of the resulting model (Fig. 2.2).

2.2.4 Protocol

Subjects carried out several dissection tasks with varying levels of force feedback provided by the teleoperation system. Subjects were instructed to expose the artery, clearing away tissue 2 mm on each side of the artery as well as removing any tissue on top of the artery. The subjects were also told to minimize the number of errors, defined as any scratch or puncture of the artery that produced visible damage to the artery, corresponding to 1.0 N. Aside from the primary goal of minimizing errors, subjects were instructed to minimize the area of tissue disturbed outside of the region to be exposed. Finally, after meeting the above two requirements, subjects were to expose as much of the artery as possible in the allotted time.

In every case, the subject was to start at the same point and progress down the artery, working to clear both the sides and the top of the artery at the same time. In the trials where the artery was not initially visible, the subjects were to find and then expose the artery. Subjects were informed that the artery was always generally straight and centrally located within the model. Lastly, the subjects were to always use the same motion when clearing away tissue, that of a small scrape or dig with the hook instrument.

Subjects trained approximately 15 minutes in order to become familiar with the system and the task. By the end of the training, subjects were required to 1) reliably execute the correct scraping motion and 2) reliably remove excess clay from the hook. Longer times were allowed for some participants to ensure a similar level of proficiency; no subject required longer than 20 minutes.

Each subject participated in several trials of 5 minutes each, where each trial involved a force feedback scaling of 0% (no force feedback), 37%, or 75%. The 75% scaling level was the highest gain available that maintained high fidelity and stability. To examine the effect of previous surgical experience on performance, subjects from 5 different levels of training were chosen (Table 2.1). A higher group number corresponds to a higher level of previous experience. Subjects in group 1 participated in 6 trials, where each force feedback level was repeated for one model with a visible artery and one with an obscured artery, as a pilot study. After finding little difference in performance between the visible artery and obscured artery trials, and in the interest of surgeons' time, subjects in groups 2-5 used only the visible artery model (3 trials each). 20 subjects, 12 male and 8 female, participated in the study.

Table 2.1: Subject groups

Group #	Subject group	n (M/F)	Mean Age	Trials
1	Graduate students	8 (3/5)	25	6
2	3rd-4th year Med. students	3 (2/1)	25	3
3	1st-3rd year residents	3 (2/1)	28	3
4	Senior residents	3 (2/1)	33	3
5	Attending surgeons	3 (3/0)	45	3

The graduate student group had little to no formal surgical background, but were familiar with the robotic equipment used in the experiment. The 3rd and 4th year medical students were knowledgeable about various surgical procedures, yet had little hands-on surgical training. The 1st-3rd year surgical residents had some hands-on experience with both general and laparoscopic procedures. The senior residents had extensive experience with general and laparoscopic procedures and were well versed in surgical technique. The attending (permanent staff) surgeons had the greatest surgical training and experience, with expert knowledge of laparoscopic procedures.

A final set of trials was conducted using a porcine liver and gall bladder to validate the clay surgical model. The liver/gall bladder interface is a reasonable comparison to the contrasting stiffnesses of the artery/tissue model [35]. Further, removing the gall bladder from the liver is a common surgical task for blunt dissection. Three trials of one minute each at the three force feedback levels were performed by a medical student, a senior resident, and an attending surgeon for a total of nine trials. The liver and gall bladders used were harvested and frozen, then defrosted prior to the trials.

All forces encountered by the instrument tip were recorded by the software. To avoid recording force data when cleaning the tip of excess tissue, a button on the stylus was used to pause the logging of data. Peak and root-mean-square (RMS) force values were then calculated from the complete force record. Errors are defined as exceeding a force threshold when contacting the artery. An observer noted the times of possible error occurrences during each trial. The force log was later examined at the times of the possible errors to determine if the subject did exceed the force threshold of 1.0 N. Area affected was calculated using a digital image of the completed models. The area affected was segmented from the image by hand and then measured using software. Finally, length dissected was extracted by a similar method, using the digital image to measure the length of artery exposed.

2.2.5 Measures

Four different outcome measures were examined for each trial. The applied forces, the number of errors, the length of dissection, and the area of tissue affected were chosen to best characterize the performance of a subject. The applied forces, number of errors, and total area affected correlate directly with tissue trauma. The length dissected, given a fixed trial duration, provided a measure of productivity.

2.2.6 Statistical Analysis

To determine statistical significance of our experimental conditions (force feedback scaling, training, artery visibility) we used the nonparametric Friedman test of k related samples in place of a conventional multi-factor ANOVA. The nonparametric test was chosen because of the relatively small sample and the lack of information concerning the distribution of the variables under study. The statistical analysis included examining the RMS forces, the peak forces, the length dissected, the area affected per cm dissected, and the number of errors per trial. The SPSS statistical analysis software package (Version 10.1, SPSS Inc., Chicago, Ill.) was used to carry out the analysis. A p value of less than 0.05 was considered statistically significant.

2.3 Results

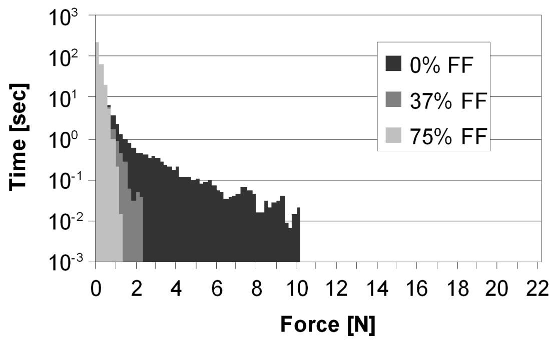
Force feedback significantly reduced the magnitude of the forces applied at the instrument tip during dissection, independent of previous surgical knowledge. Fig. 2.3a-e show histograms of the force samples for all subjects and all trials with the visible artery. Subjects applied high force levels for longer durations when force feedback was not available. Conversely, during trials with force feedback, less time was spent applying higher forces; forces above 3 N were of negligible duration (less than 5 ms on average over 5 minute trial) for 75% force feedback scaling, and above 4 N were negligible for 37% scaling. Further, the greater the force feedback gain, the less time was spent applying larger levels of force. For the graduate student group, these results also apply whether or not the subject can initially see the artery (Fig. 2.3a and f). In addition, neither the average RMS force nor the average peak force for the occluded artery trials are statistically different from the visible artery trial results ($F(1{,}7) = 0.175$, $p = 0.688$; $F(1{,}7) = 0.526$, $p = 0.492$). We therefore consider the visible artery case for the rest of the analysis.

Task results for the porcine tissue trials yield a similar force profile (Fig. 2.4). Again, as force feedback scaling increased, peak force (8.59 N, 4.88 N, 3.08 N) and RMS force (1.65 N, 1.40 N, 0.54 N) decreased.

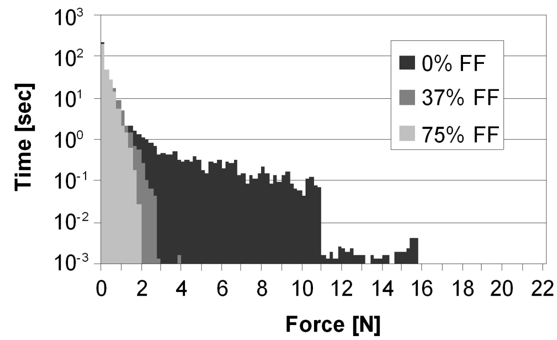
Figs. 2.5 and 2.6 show the RMS and peak forces of each of the subject groups. The addition of force feedback significantly reduced the RMS force by 30% to 65% ($F(2{,}30) = 49.13$, $p < 0.001$) and the peak force by a factor of 3 to 6 ($F(2{,}30) = 58.69$, $p < 0.001$). Again, higher force feedback gain resulted in a reduction of forces applied across all subject groups. The factor of training was also significant for both RMS and peak force applied ($F(4{,}15) = 6.33$, $p = 0.003$; $F(4{,}15) = 7.69$, $p = 0.001$). A polynomial contrast shows both a significant linear and cubic trend that an increase in surgical experience results in higher RMS and peak applied forces ($F(4{,}15) = 6.331$, $p = 0.003$; $F(4{,}14) = 7.685$, $p = 0.001$).

The average number of errors during a trial was also affected by the addition of force feedback (Fig. 2.7). Increased force feedback led to a significant reduction in the average errors ($F(2{,}30) = 12.54$, $p < 0.001$). Training level of the subjects was also a significant factor, where an increase in previous surgical training led to a maximum of a 7 fold increase in the number of errors committed of the untrained group ($F(4{,}15) = 5.39$, $p = 0.007$).

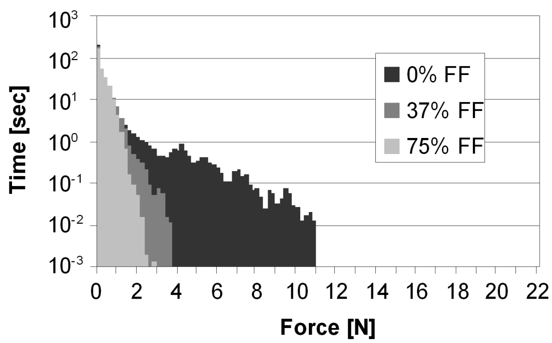
Two measures that were not significantly affected by the addition of force feedback were the productivity measures (Figs. 2.8, 2.9). The length of artery dissected did not change



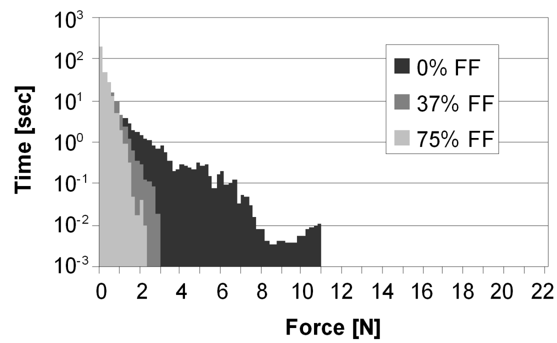
(a) Graduate students (n = 8)



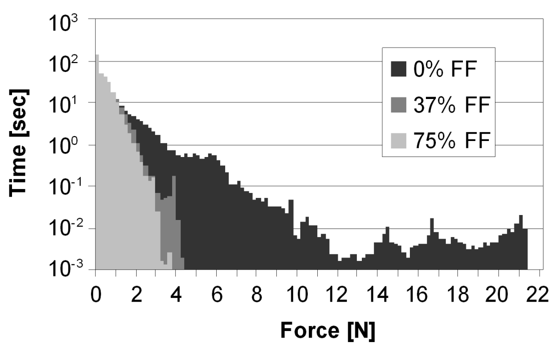
(b) 3rd-4th year medical students (n = 3)



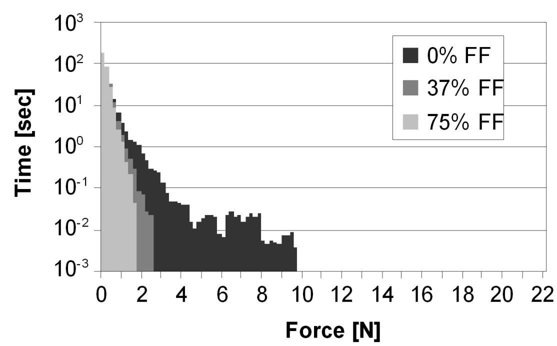
(c) 1st-3rd year surgical residents (n = 3)



(d) Senior surgical residents (n = 3)



(e) Attending surgeons (n = 3)



(f) Grad. students, occluded artery (n = 8)

Figure 2.3: Average time spent applying different levels of force, for each level of force feedback. Bins (0.2 N spacing) represent the average time spent applying the corresponding force. Every figure represents a total of five minutes at each force feedback level. (a)-(e) Visible artery trials for all subject groups. (f) Occluded artery trials for subject group 1.

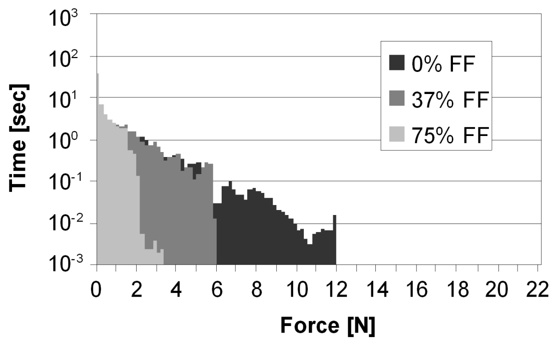


Figure 2.4: Histogram of forces applied for blunt dissection of gall bladder from liver surface

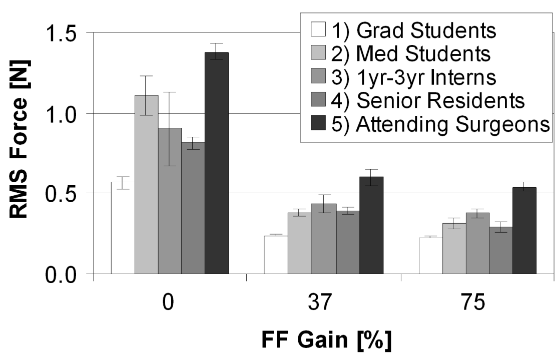


Figure 2.5: Average RMS force applied versus force feedback gain. Error bars show standard error.

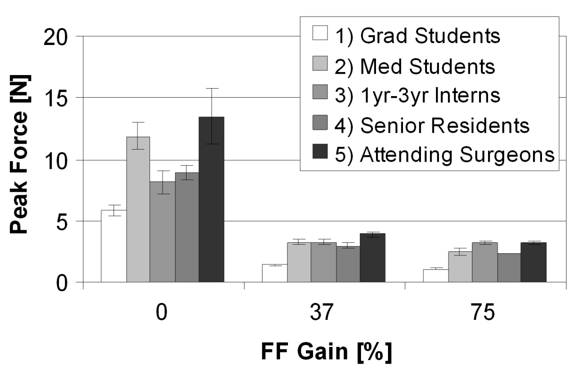


Figure 2.6: Average peak force applied versus force feedback gain. Error bars show standard error.

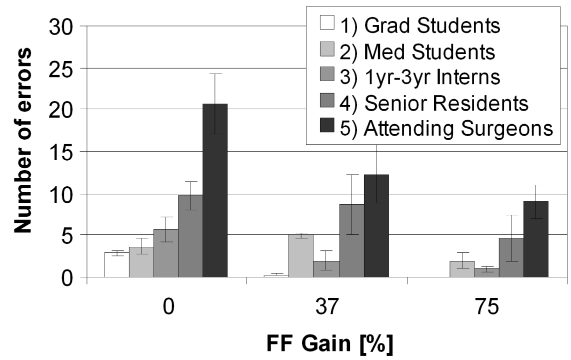


Figure 2.7: Average number of errors vs. force feedback gain. Error bars show standard error.

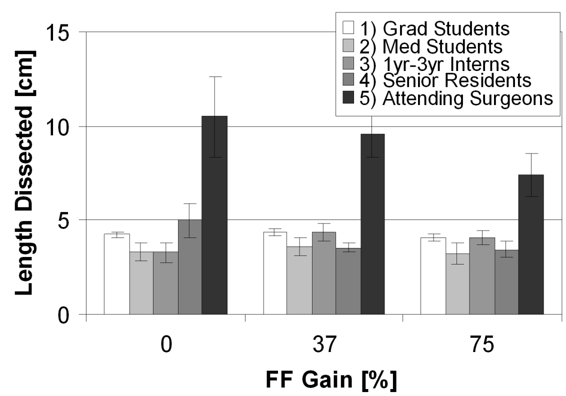


Figure 2.8: Length dissected vs. force feedback gain. Error bars show standard error.

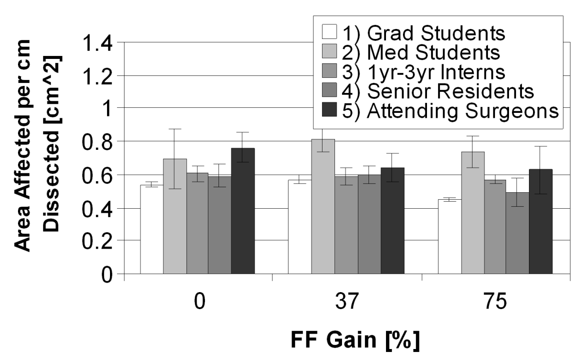


Figure 2.9: Area affected per cm dissected vs. force feedback gain. Error bars show standard error.

significantly over different levels of force ($F(2,30) = 2.673$, $p = 0.09$). Similarly, the amount of area affected was not influenced by the addition of force feedback ($F(2,30) = 1.371$, $p = 0.269$). The factor of training did not significantly alter the trends or levels of length dissected or area affected except for the length dissected by the attending surgeons ($F(3,13) = 0.321$, $p = 0.810$; $F(4,15) = 0.869$, $p = 0.505$). The surgeons were able to dissect more than twice the amount, on average, as any other group. Also, for only the surgeon group, there was a trend that increased force feedback levels resulted in decreased length dissected; however, this trend did not reach significance.

2.4 Discussion

In this study we examine the effects of force feedback on a blunt dissection task, where we hypothesized that the addition of force feedback improves surgical performance. Our results show that force feedback improves performance by reducing the overall forces applied, thus reducing tissue trauma. Force feedback also aids surgical performance by reducing the number of accidental incursions into sensitive structures. These results hold across all levels of previous surgical experience. Therefore, in this task surgeons do not appear to acquire skills through extensive experience in minimally invasive surgery that decrease the benefits of force feedback. The addition of force feedback did not seem to affect the productivity of subjects, as measured by the length of the artery dissected and the amount of surrounding tissue accidentally affected. We also observed trends with respect to surgical experience; as level of experience increases, applied forces and accidental incursions increase. No significant trend was observed with respect to surgical experience and the productivity measures except that the attending surgeon group dissected significantly more artery than all other groups.

Conditions in this study simulated the essential aspects of laparoscopic hook dissection in minimally invasive surgical procedures. Visual feedback was provided by a standard surgical laparoscope, and the instrument control mode included the fixed pivot at the incision point. While the mechanical properties of the synthetic clay models were not identical to actual tissue, the key behavior in this task is plastic deformation under traction loading. In this respect, the clay material replicates the behavior under blunt dissection with electrocautery of friable tissue such as liver parenchyma and thin layers of connective tissue such as the liver-gallbladder junction [38]. This is verified by the trials executed using excised animal tissue as the experimental environment, where we observe a similar force profile as the clay artery trials, along with comparable force ranges. It is important to note that these results pertain to tissues where plastic deformation forces dominate in blunt dissection. This is the case in many surgical procedures because blunt dissection is useful for separating tissue planes joined by relatively weak connective tissue [103]. These results may also apply to blunt dissection of tissues with significant elastic and viscous forces, but further experimental investigation is required.

2.4.1 How Force Feedback Benefits Surgery

This study leads to the hypothesis that there are two mechanisms that produce the benefits of force feedback in surgery. At high levels of force feedback, we speculate that the intrinsic mechanical properties of the tissues being manipulated are transformed into physical constraints on the surgeons motions. Subjectively speaking, it is difficult to move the instrument into a damaging configuration because a large force on the hand will oppose any motion that involves contact between the instrument and the tissue. Further, this constraint not only acts as a safety barrier, reducing the forces applied and the number of errors, but the constraint can also act as a guide to the surgical instrument. For instance, when the instrument is positioned between two structures of different stiffness, accurate dissection can simply be achieved by first applying a minimal force to press the instrument against the stiffer tissue. Then, the instrument can be dragged along the surface of the stiffer tissue while relying on the force feedback to maintain a uniform and safe contact force between tissue and instrument.

As the level of the provided force feedback decreases, the benefit of force feedback is hypothesized to arise less as a physical constraint and more as a supplemental source of information. Because the forces are now harder to perceive, the surgeon must devote increased mental processing capacity to recognize and interpret this additional information. Thus, at low levels of force feedback, a conscious response is required to take advantage of the available forces. However, because we observed similar force profiles with no change in productivity at the gains of 75% and 37%, forces at these gains likely act as physical constraints, not solely as a supplemental information source. Further experimental investigation is required to understand the role of force magnitudes in this lower range.

Based on the experiments with real tissues, we conjecture that the relationships among force feedback gain and performance measures will persist over a range of mechanical properties; in particular, the same constraint mechanism functions in both cases. The variation in performance measured in this study as a function of force feedback gain was large and repeatable, shown to a high degree of statistical significance.

2.4.2 Surgical Experience

An interesting result is the variation in performance with respect to previous surgical experience. While all subject groups benefited from the addition of force feedback, the attending surgeon group had entirely different performance levels then the other four groups. Specifically, attending surgeons consistently applied higher forces and committed more errors across all levels of force feedback scaling. On the other hand, the surgeons were able to expose twice the average length of artery as any other group, without a higher level of unwanted area affected. The attending surgeons were clearly using a different performance tradeoff then other groups.

Several hypotheses can be made about why attending surgeons applied higher average forces. One possibility is that surgeons had a different prior expectation of their performance. Because they expect to expose a certain amount of artery in a trial, they disregard the possibly excessive forces applied to achieve their goal. One reason the surgeons may disregard these forces is that the feedback, both haptic and visual, does not accurately represent the range of subtle cues encountered in actual surgery. For instance, when an instrument contacts tissue, the surgeon not only receives haptic feedback, but observes deformation, color change, and functional change (e.g. blood vessels rupturing). Without these coordinated cues, an experienced surgeon may not register the haptic signal alone as a damaging force.

Another reason the surgeons may have disregarded the haptic force feedback is that the interface mechanism is different than a laparoscopic dissector handle. Thus, the surgeons may not have regarded (consciously or subconsciously) the task as a surgical one, so were not drawing upon their expertise.

Another possible explanation (as observed by [13, 99]) is that the surgeons expected to apply a certain amount of force because dissection procedures require controlled damage to tissue (e.g. the liver capsule is severed when separating the gall bladder from the liver). Surgeons are trained to apply appropriate forces that cause this controlled damage. Subjects who do not have extensive surgical experience may limit force application to prevent irreversible damage to tissue, slowing the dissection process.

The effect of force feedback on the length of artery dissected by surgeons is also notable. An increase in the force feedback scaling level served to decrease the length dissected by the surgeons. An explanation offered by the surgeons is that they are accustomed to receiving some (however slight) force feedback during laparoscopic surgery. When that feedback is removed, as in the 0% scaling case, the surgeons were unaware of the magnitude of forces applied and attempted to dissect as much artery as possible. With the addition of force feedback, the force cues coincided with the visual cues and the surgeons sacrificed speed to lower forces.

The graduate students consistently applied lower peak and average forces while maintaining the same level of length dissected and area affected. They were also the only subject group familiar with the Phantom haptic interface device used as the basis for the telemanipulation system. One hypothesis is that the familiarity with the equipment allowed a more rapid understanding of the force information being provided. This may speak to the relative importance of task training versus equipment training in teleoperated environments. Regardless, the graduate student subject group still derived the force magnitude and error reduction benefit of force feedback.

Observing a similar force trend across all levels of previous surgical experience strongly points to the physical constraint benefit of force feedback. The advanced training and experience of expert surgeons do not diminish the value of this enhancement mechanism. An additional benefit of force feedback to experienced surgeons may be to reduce mental workload. Experienced laparoscopic surgeons have developed perceptual and motor skills to deal with the constraints of minimally invasive surgical techniques, and are able to use visual information to guide fine motions to avoid generating large forces. This visual approach would probably require significant cognitive processing and attention, however, so the aid provided by the physical constraint benefit of force feedback may serve to reduce that cognitive workload.

From this study, the benefit of force feedback is clear when accurate instrument positioning is required and/or when the involved structures are sensitive and surrounding tissue trauma has severe implications. Microsurgical procedures meet all these conditions and may be considered the likely candidates for dexterity enhancement by instruments with force feedback capabilities. Presently the visual acuity, dexterity and tactile sensitivity of the surgeon define the limits of microsurgical procedures. The use of force feedback would allow scaling of forces up to perceivable levels, providing the aforementioned advantages to the microsurgical realm [40].

Chapter 3

Physical Constraint Hypothesis

3.1 Introduction

The benefits of force feedback have been demonstrated in numerous environments and systems. Teleoperated tasks such as hazardous material handling and remote surgery have both shown an increase in performance with the addition of force feedback [107, 60]. Virtual environments have also benefited when the user is presented with force information [120]. Force information may also serve to establish a more powerful feeling of presence in a remote environment [102]. However, the mechanisms by which force feedback improves performance have not been fully investigated. Knowledge of these mechanisms would allow optimal interface design and highlight tasks where force feedback would be most beneficial.

A common approach to analysis of force feedback is to create a model of the interacting limb or the interface mechanism. Finger and wrist impedance models have been developed to examine keyboard design [26] and haptic controller design [42, 70]. Limb impedances have also been found to change over time due to leaning effects and perceived task difficulty [78]. These impedance models, however, were derived at a fixed limb position and do not take into account the desired motion that is present when executing a task. Models of interface mechanisms have also been developed for use in conjunction with limb impedance to determine the optimal way to transmit forces. For example, models for bilateral telemanipulators are used to analyze stability while also minimizing the difference between remote and local forces [44]. Again, these models do not provide insight into the mechanisms by which forces improve performance. Some research has been presented on combining a teleoperator model and a hand impedance model to examine how forces help in teleoperation [23]; Daniel and McAree concluded that forces below 30 Hz act to transfer energy and forces above 30 Hz act as an information source. While these results are an important foundation, they do not relate the effects of force feedback to task characteristics and performance metrics.

We propose that a model of the operator's hand and the haptic interface mechanism combined with knowledge of the operators desired motions can lead to insight into how force feedback improves operator performance. Our primary hypothesis is that force feedback exists as a physical constraint, passively restricting the motion of the operator. Secondly, we conjecture that a model of hand impedance can be used to derive the operator's desired motion trajectory in a task, allowing us to separate the passive benefit of force feedback that acts as a constraint from the informational benefit of force feedback that leads to voluntary motion changes. We present an experiment where users interact with a stylus attached to a robotic interface. Users move the stylus at a constant speed until encountering a tactile stimulus, either a vibration or a force resisting the direction of motion. Based upon their reaction time, we demonstrate that the force stimulus passively restricts the users motion before they can voluntarily react. Further, we show that a second order model of the hand/stylus system can be used to quantify the constraint and informational contributions of force feedback.

3.2 Methods and Materials

In this experiment, subjects execute a motion, and then reverse their motion upon feeling a haptic stimulus. In the 150 ms before the users can voluntarily respond to the stimulus [9], they will continue in the direction of the original motion. During this pre-voluntary time, when force feedback that resists the direction of motion is present, the hand will not travel as far as compared to the case where the stimulus is a vibration that provides no net force to the user. Because subjects cannot voluntarily respond during the pre-voluntary phase, we can also model the motion of the hand during that time using passive mechanical components. This model can be used to analyze the magnitude of the physical constraint and determine the users voluntary motion after responding to the stimulus.

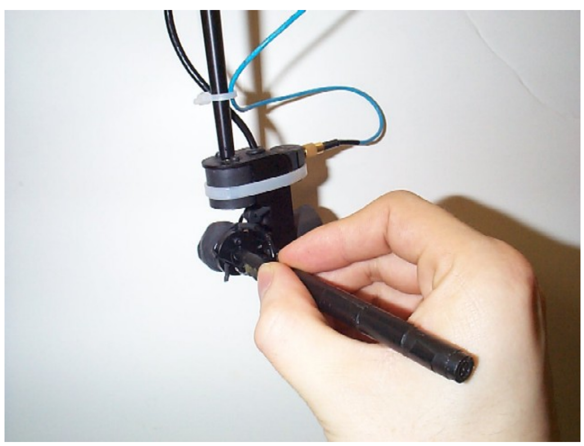
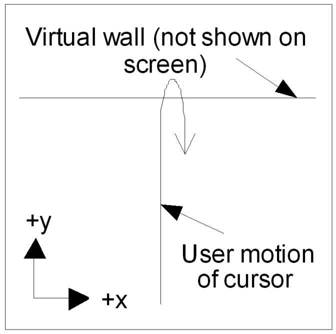


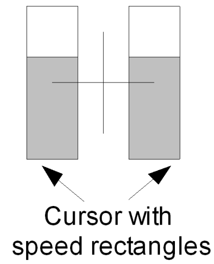
Figure 3.1: Stylus grasp configuration

3.2.1 Experimental Design

A haptic interface device (Phantom 1.5, Sensable, Woburn, MA) was used to record the participants trajectory data as well as provide haptic stimuli during the experiment [16]. We define the positive x direction to be to the right and positive y to be upward with respect to the user. Subjects grasped the stylus like a pen to control the motion of the tip of the stylus in the x, y plane (Fig. 3.1). A one-dimensional accelerometer (PiezoBeam 8630B50, Kistler, Amherst, NY) with a resolution of 0.005m/s^{2}was rigidly attached to the stylus gimbal to



(a) Diagram of workspace showing user motion



(b) Diagram of cursor with speed rectangles displaying a current speed above the desired speed target

Figure 3.2: Visual feedback

record accelerations in the y-dimension. Position and acceleration data were recorded at 1000 Hz for each trial.

Subjects were instructed to move the tip of an interface stylus upward (positive y direction) in a straight-line trajectory at a constant speed until encountering a wall. The wall would be signified by the presence of either a force resisting the motion of the stylus or a vibration. Note that the only information on wall position presented to the user was through the haptic pathways. After contacting the wall, subjects were instructed to reverse their direction of motion as soon as possible and exit the wall.

A computer monitor provided feedback of the participants current position in the x,y plane and velocity in the positive y direction. The position of the tip of the interface stylus was mapped to the position of a cursor with a scale factor of 1 (Figure 3.2a). A one to one mapping between stylus tip and cursor (i.e. a 1 cm motion of the stylus tip would move the cursor 1 cm on the screen) was used to remove any scaling effects. Velocity in the positive y-dimension was displayed using rectangles on either side of the cursor (Figure 2b). The height of the rectangles increased with increasing vertical speed, up to a maximum height. The desired constant velocity that participants were asked to move at corresponded to half of the maximum rectangle height. The workspace of the experiment in the x, y plane was 16 cm by 12 cm.

The path that the subject was asked to follow always began at the bottom center of the workspace. The wall stimulus would appear when the subject maintained a speed +/- 5% of the target speed of 40 mm/s for 100 ms. These numbers were determined through pilot studies to allow a reasonable success rate yet still restrict the participant's initial velocity to a narrow range. If the participant did not meet the speed criterion within the workspace, the trial was repeated. Subjects were given a training period where they practiced moving the stylus at a constant speed. All subjects were able to trigger the wall stimulus regularly (on average, one out of three attempts) within five minutes of training.

The motion of the participants forearm was restricted through the use of a brace rigidly attached to the armrest of the participants chair. The brace was used so that the wrist was the principal joint used to move the stylus. To avoid any audio stimulus, particularly during the vibration trials, participants wore headphones playing noise in the frequency of the vibration.

3.2.2 Stimuli

Two types of stimuli were used in the experiment to represent a wall. The first type was a force vibration along the x dimension at 250 Hz with commanded amplitude of +/-1.0 N. The vibration frequency was chosen to maximally stimulate the rapidly adapting receptors in the fingertip. This stimulus was simply on or off depending on whether the subject was above or below the boundary of the wall. The second type of stimulus was a force in the negative y direction proportional to the distance traveled into the wall. This force is effectively a spring force with stiffness k_{wall}. Three force feedback gains of 33%, 67%, and 100% were used to scale the spring force during the experiment. Since the spring force was the only force encountered, the three levels of force feedback gains were equivalent to three different wall spring stiffnesses. The wall force can thus be expressed as

$$F_{wall} = -G_{FF} k_{wall}(y_{cursor} - y_{wall}) \tag{3.1}$$

where y_{wall} is the y-position of the start of the compliant wall, G_{FF} is the force feedback gain, k_{wall} is the wall stiffness, and y_{cursor} is the y-position of the cursor. A wall stiffness of $k_{wall} = 0.54$ N/mm was used in the experiment to give a perceptually relevant range of wall compliances.

The subjects motion was restricted to the x/y plane to simplify the workspace of the experiment to two dimensions and maintain the orientation of the accelerometer. A spring model (proportional error) was again used to generate the forces necessary to restrict motion in the z dimension

$$F_z = -k_z(z_{cursor}), \tag{3.2}$$

with $k_z = 0.54$ N/mm.

3.2.3 Hand model

A linear, second order system was used to model the impedance of the hand/stylus system. The model consists of a spring, mass, and damper as the link between a desired position input and an actual position output (Figure 3.3). This type of model was chosen due to its success in characterizing limb impedances [42, 61] and as an attempt to find a low order model that still encapsulates physical features relevant to force feedback. The model is also similar to one earlier proposed by Kuchenbecker *et al.* [70] who modeled the impedance of the wrist in a similar stylus grasp configuration. While the hand stylus system does not behave as a linear second order system over all ranges of input, a second order model has been shown to encapsulate the essential dynamics for the small excursions used here.

Writing a force balance for the hand/stylus model in contact with a compliant environment yields

$$k_{hand}(x_d(t) - x_a(t)) + b_{hand}(\dot{x}_d(t) - \dot{x}_a(t)) - m_{hand}\ddot{x}_a(t) = F_{wall}(t) \tag{3.3}$$

where k_{hand}, b_{hand}, m_{hand} are the parameters of the second order hand/stylus model, $x_d(t)$ is the desired hand motion from the central nervous system, $x_a(t)$ is the observed trajectory, and $F_{wall}(t)$ is the wall force.

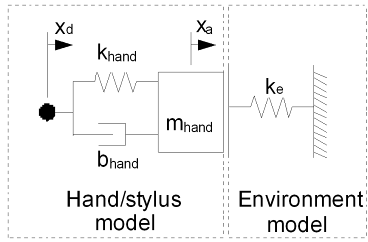


Figure 3.3: Hand and environment models

In order to differentiate between the passive benefits of force feedback and the changes in voluntary motion that result from force feedback, we estimate the desired trajectory of the hand (the commanded hand trajectory from the central nervous system) from the observed trajectories. We can do so in a four-step process using the model described above:

1. Find an estimate of the desired trajectory $x_d(t)$ for the first 150 ms after the user has encountered a stimulus by averaging all of the trajectories after a vibration stimulus occurred. Because the vibration applied no net force, the observed motion should closely match the desired motion.

2. Fit the parameters of a second order model (k_{hand}, b_{hand}, m_{hand}) to the first 150 ms of observed position and force data for the force feedback cases using the above average as the desired trajectory. The fit parameters should be similar across all inputs since the user has not yet voluntarily responded.

3. Construct a mathematical description of the relationship between the input (desired trajectory) and the output (observed trajectory) using the average of all fit model parameters for that subject.

4. Using the above relationship and assuming that the hand parameters remain relatively constant after the user responds to the stimulus, estimate the desired trajectory after 150 ms by applying the inverse of the relationship to the observed motion.

We now describe the steps of the above process in detail.

First, the parameters for the proposed hand model were estimated for each trial. The parameters can be fit using least squares by first expressing the force balance equation (3.3) at all sample times to be used for the fit as

$$A p_{hand} = F_{wall} \tag{3.4}$$

where A is a concatenation of the systems states over n samples

$$A = \begin{bmatrix} x_d(t_0) - x_a(t_0) & \dot{x}_d(t_0) - \dot{x}_a(t_0) & \ddot{x}_a(t_0) \\ x_d(t_1) - x_a(t_1) & \dot{x}_d(t_1) - \dot{x}_a(t_1) & \ddot{x}_a(t_1) \\ \vdots & \vdots & \vdots \\ x_d(t_n) - x_a(t_n) & \dot{x}_d(t_n) - \dot{x}_a(t_n) & \ddot{x}_a(t_n) \end{bmatrix}, \tag{3.5}$$

p_{hand} is the column vector of hand parameters

$$p_{hand} = \begin{bmatrix} k_{hand} & b_{hand} & m_{hand} \end{bmatrix}^T \tag{3.6}$$

and F_{wall} is the column of wall forces at all sample times

$$F_{wall} = \begin{bmatrix} F_{wall}(t_0) \\ F_{wall}(t_1) \\ \vdots \\ F_{wall}(t_n) \end{bmatrix}. \tag{3.7}$$

In these expressions, t_0 represents the time at which the user first contacts the wall and t_1 through t_n are the subsequent sample times (taken at 1 kHz). The time t_n= 150 ms was chosen so that the parameter estimates would only incorporate the passive hand dynamics and not any cognitive response to the force stimuli [9]. The stretch reflex is an active (although not cognitive) response that occurs in response to limb flexion at 30 ms [79]; we will consider the stretch reflex as part of the passive hand model as is commonly assumed [61]. Note that F_w is the commanded force to the Phantom; we assume that F_{wall} matches closely with the force actually applied to the hand because the frequency content of F_{wall} is within the bandwidth of the Phantom [16].

We will assume $\overline{x}_d(t)$ to be the average of all actual trajectories from t_0 to t_n for the vibration case. Because only vibrations were applied in that case and no net forces, the actual trajectory should closely follow the desired trajectory, assuming steady state prior to the stimulus, so $\overline{x}_a(t) = \overline{x}_d(t)$. This desired trajectory should be the same for all cases up to t_n, since users did not have a chance to voluntarily respond. The velocities $\dot{x}_d(t)$and $\dot{x}_a(t)$ are found by differentiating high order (18 terms) polynomial fits to $\overline{x}_d(t)$ and $x_a(t)$, respectively. A polynomial solution is used to minimize high frequency noise in the derivative. The acceleration $\ddot{x}_a(t)$ is measured using the accelerometer. Once the data is expressed as (3.4), a least squares minimization is used to find the values of k_{hand}, b_{hand}, m_{hand} that minimize the error $Ap_{hand} - F_{wall}$. For our estimation, all hand parameter values were constrained to be non-negative.

An estimate of the desired trajectory for each trial can be constructed once the average hand parameters across all trials for a given subject have been found, assuming that k_{hand}, b_{hand}, m_{hand} remain constant. The desired trajectory is found by solving the force balance (3.3) for $x_d(t)$ using Laplace transforms (see App. A)

$$x_d(t) = \text{deconv}(x_a(t), h(t)) - g(t) \tag{3.8}$$

where $h(t)$ is the impulse response of the system model and $g(t)$ is the response due to initial conditions.

3.2.4 Experimental Design

Six people, aged 19 - 26 volunteered for the study. Participants described themselves as right handed with no known abnormalities in either hand.

The experiment conducted was a one factor, four level repeated measures design with an independent variable of stimulus type. The four levels of stimulus were vibratory, 33%

force feedback gain, 67% force feedback gain, and 100% force feedback gain. Each subject completed 96 trials and received the same stimulus presentation order, in which the levels of stimulus were counterbalanced against order. The placement of the wall stimulus in the y-dimension, given that the subject had met the speed constraint, was also counterbalanced across trials.

3.3 Results

Figure 3.4 shows the average intrusion trajectory for each wall stimulus type for a typical subject. Each line is the average of 24 trajectories. We observe that as the force feedback gain is increased, average intrusion into the wall decreases. Note that the intrusion distance is reduced even before 150 ms when voluntary response can begin. This same trend is observed for all subjects up to 150 ms (Fig. 3.5) ($F(3{,}429) = 1562$, $p < 0.001$). The average maximum incursion reached in 150 ms decreases by an average of 80% across subjects. Because a reduction in incursion distance is occurring before voluntary response, for all subjects, force feedback is acting as a physical constraint to the motion of the hand.

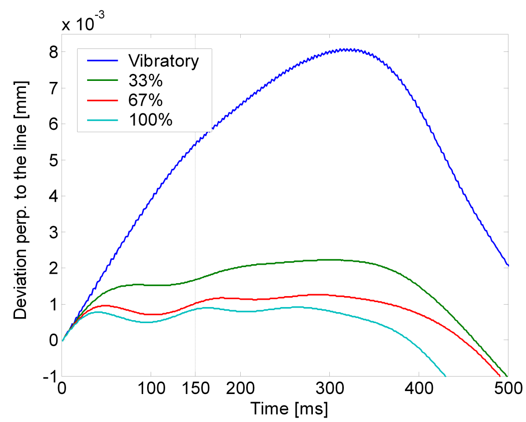


Figure 3.4: Average trajectories for a typical subject

A sample estimate of hand force over time is shown in Fig. 3.6, along with the forces that result from the individual elements. The small oscillations in the damping force are due to the velocity estimation process. The results of the hand parameter fits are shown in Table 3.1, with normalized values graphed in Fig. 3.7. Although trends in each of the normalized parameters were significant with respect to force feedback level (stiffness $F(2{,}286) = 4.77$, $p < 0.01$; damping $F(2{,}286) = 28.12$, $p < 0.001$; mass $F(2{,}286) = 7.17$, $p < 0.005$), the means for each parameter were within one standard deviation of one another. The average variance in force accounted for by the model (VAF), given by

$$\text{VAF} = 1 - \frac{\text{mean}\left[(F_{commanded} - F_{calculated})^2\right]}{\text{var}\,(F_{commanded})}$$

is shown in Table 3.2, with an average VAF of 96% across all trials.

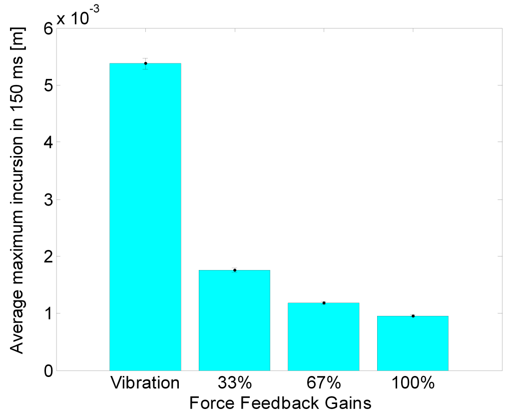


Figure 3.5: Average maximum incursion in 150 ms for all subjects

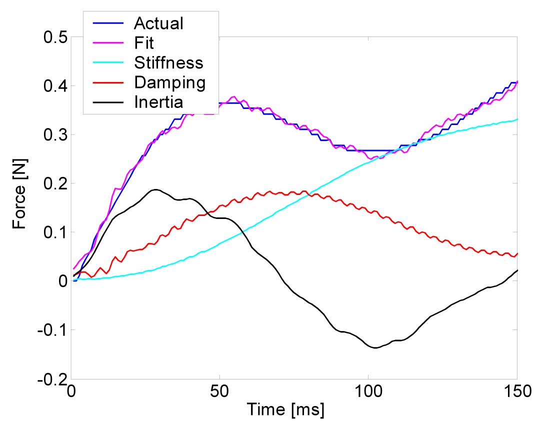


Figure 3.6: Typical force fit up to 150 ms

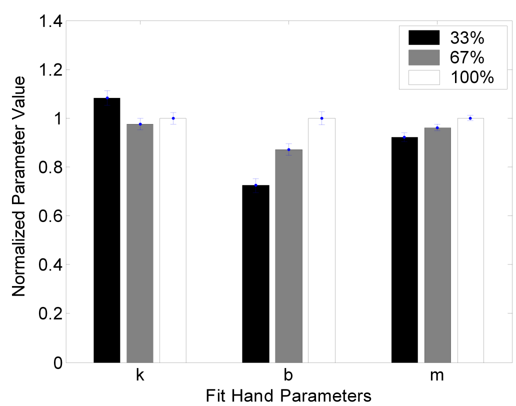


Figure 3.7: Hand parameters estimated across subject and force feedback gain. Error bars show standard error.

Table 3.1: Average fit hand parameters per subject

Subject	k_{hand}[N/m]	b_{hand}[Ns/m]	m_{hand}[kg]
1	136.2	1.62	0.183
2	113.7	2.66	0.206
3	69.1	3.97	0.183
4	60.0	3.36	0.164
5	37.0	3.19	0.153
6	85.6	2.44	0.168
Mean	83.6	2.90	0.200

Table 3.2: Average VAF in force across FF conditions

Subject	33%	67%	100%	Mean
1	95.60%	94.68%	96.36%	95.55%
2	97.14%	97.29%	96.49%	96.97%
3	97.92%	97.09%	95.90%	96.97%
4	97.24%	97.12%	97.35%	97.24%
5	95.76%	96.08%	96.59%	96.14%
6	95.64%	95.93%	95.95%	95.84%
Mean	96.55%	96.37%	96.44%	96.45%

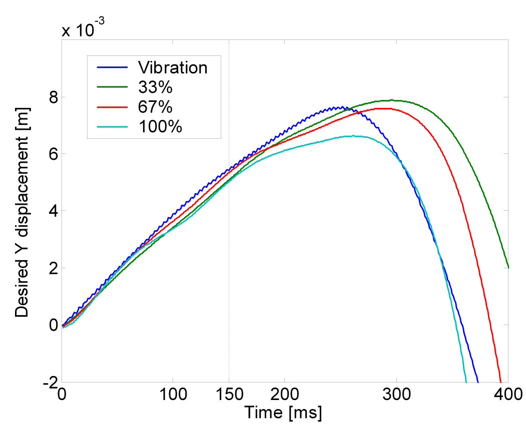


Figure 3.8: Average inferred commanded trajectories for a typical subject

Using the assumption that hand parameters are constant across all force feedback levels, the average of all fit hand parameters, per subject, were used to extract the desired trajectory $x_d(t)$ from each observed trajectory $x_a(t)$ including $t > 150$ ms. Figure 3.8 shows the average extracted desired motion for different levels of stimulus for a typical subject. Each line is again the average of 24 trials. Note that all trajectories are nearly collinear up until 150 ms, at which point they diverge. The average turnaround time (time when the desired trajectory reaches a maximum) for each subject and wall stimulus decreases by up to 70 ms for increasing force feedback level, not considering the vibration stimulus ($F(2{,}286) = 48.101$, $p < 0.001$) (Figure 3.9). The vibration stimulus condition was significantly different from the 100% case and the 33% case, shown using multiple t-test comparisons ($p < 0.005$, $p < 0.001$). However, all average turnaround times were found to be larger than the 100% condition average turnaround time, per subject.

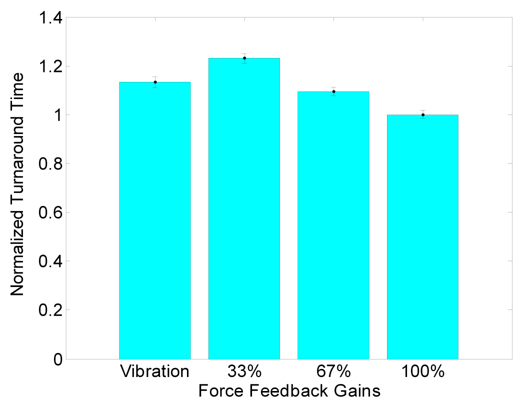


Figure 3.9: Turnaround times of desired trajectories. Error bars show standard error.

3.4 Discussion

Our hypotheses were that force feedback causes a physical, passive error reduction before users can voluntarily respond to a contact stimulus, and that a physical model of the hand/interface system would allow an examination of the effects of force feedback magnitude on desired motion. Two different types of stimuli were used to examine the first hypothesis; one, a vibration that provided information of contact but no net force, the other a virtual spring with different levels of stiffness. In order to investigate the second hypothesis, a second order model was chosen to represent the hand/interface system. Our experiment was designed so that users were at a consistent, steady state condition before the stimulus in order to reduce variation between trials and increase the accuracy of fit for a low-order mechanical model.

We observe that force feedback does indeed act as a physical constraint to the motion of the hand/stylus system, where an increase in force feedback gain can lead to a dramatic reduction in incursion distance before 150 ms. We also observe that a second order model works well at encapsulating the passive motion of the hand/stylus system, consistently accounting for the observed behavior across varying levels of input and force feedback gain.

Using the second order model and assuming that hand parameters do not greatly change after 150 ms, we observe desired trajectories that match closely with the actual motion when there is no net force on the system. However, as force feedback gain increases, the turnaround time of the desired motion decreases for all subjects. Thus, the benefit of force feedback is two-fold: forces passively constrain the motion of the hand and provide information used to alter desired motion.

Several assumptions are made that lead to the above conclusion. The primary one is that the hand/stylus system behaves as a second order system, with desired position as input and actual position as output. Previous studies have attempted to fit low order models to various limb impedances and have met with varying success, with models breaking down when unable to encapsulate higher order effects, specifically those due to additional degrees of freedom [42, 70]. In our case, we have taken precautions to reduce variation between trials and restrict the dominant motion to the wrist. Even so, we observe a slight trend in the estimation of the damping parameter that varies with force feedback level. The main point, however, is that we are using the lowest order model that captures essential system dynamics. Using higher order models may achieve better fits, but at a cost of a large variation in the values of the fit parameters given the short time frame ($<$150 ms) used to fit. An additional assumption was that the hand parameters remained relatively constant after the 150 ms cutoff. This assumption was reasonable given that the estimates of $x_d(t)$ and the hand parameters m_{hand}, b_{hand}, and k_{hand} were relatively consistent across trials. The main emphasis is that a mechanical model can be used to separate the two benefit mechanisms of force feedback; if other methods are used to fit the model, the assumption that the hand parameters stay constant may not be necessary.

Several factors contributed to inaccuracies of the fit model, independent of the low model order. Primarily, the fitting technique requires the knowledge of both the desired trajectory and the output trajectory to extract the parameters. The desired trajectory varied between trials while the desired trajectory used to find mhand, bhand, and khand was an average of all the subjects trajectories in the vibration case. The discrepancy between the two could lead to inaccurate fits. Another source of error in the fitting technique is that, over the short fitting time, the magnitude of the forces due to the individual mechanical components are not always of the same order. Therefore, a large change in one parameter will not result in a large change in force relative to the force due to the other components, causing a range of acceptable fit values.

3.4.1 Benefits of Force Feedback

Based upon the results of the described experiment, we can make some observations concerning the nature of force feedback and its benefits. A primary observation is that there is a fundamental difference between force feedback and other forms of tactile feedback, such as vibration. Force feedback has the capability to reduce errors without requiring cognitive attention. The implications are that the benefits of force feedback can occur before 150 ms, and that taking advantage of force feedback does not necessarily increase mental workload. Vibration feedback, however, is strictly an information source, so the user has to devote attention to derive benefit and the error reduction benefit can only occur after 150 ms. It follows, then, that force feedback might be more useful than vibratory stimuli in situations where inaccurate motions can cause serious errors in a short time frame. Another possibility is in complex environments or difficult tasks where the users mental workload is already high. A situation that fits both of these criteria is robotic surgery [50]. During a surgical procedure, surgeons often execute complex and physically demanding tasks with delicate tissues. If soft tissues generate significant constraint forces as hard surfaces do (a question for further investigation), force feedback would serve to reduce mental workload while passively restraining offending motions into sensitive tissues.

Another requirement for force feedback to provide this passive error reduction benefit is that forces must be generated in a direction opposite to the motion causing the error. Therefore, the passive benefit of force feedback is dependent on how error is defined. An example of where force feedbacks passive benefit would not help is when the user needs to exceed a force threshold to achieve a goal. Increasing the force feedback gain will only serve to make it more difficult to achieve the threshold. Severing tough connective tissue in a surgical task might be one example of this case. Further, the forces generated by force feedback need to be high enough to affect the motion of the hand/interface system for force feedback to achieve a passive benefit.

Using a model-based approach to analyzing the benefits of force feedback allows us to examine another possible benefit of force feedback, that of increased positioning resolution. A common example motivating the use of force feedback deals with attempting to move ones hand or a tool in a straight line. A free motion using only information-based forms of feedback, such as visual signals, is difficult and results in an imperfect straight line. However, moving a tool, such as a pencil, in a straight line is trivial when using a ruler as a guide. The ruler constrains the motion of the pencil to lie exactly alongside the ruler. Using a ruler transforms a task requiring precise position control and mental effort into a simple task requiring the user only to push the pencil against the ruler. In a similar manner, force feedback can reduce the mental workload and positioning control accuracy needed when attempting to position a tool alongside an environmental structure. Returning to the robotic surgery example, if a surgeon needs to position a dissector along the edge of an organ, he or she can take advantage of the intrinsic stiffness of the organ to balance the force of contact, resulting in the dissector being positioned exactly next to the organ without exceeding a force threshold and damaging the organ.

A final benefit of a model of both the hand impedance and the desired motion in response to stimuli is that one can establish a design rule that relates force feedback gain, environmental stiffness, exploration speed, and maximum error. For instance, if error is defined as maximum incursion into a structure (as it was with our experiment) and environment stiffness and force feedback gain are fixed, then the hand model and desired trajectory can be used to find the maximum exploration speed that will not result in exceeding a specified maximum incursion. Or, in the robotic surgery example, given typical exploration speeds of the surgeon, the model permits determination of the minimum force feedback gain required to guarantee that incursions never exceed a certain incursion threshold, for a given environment stiffness.

We have described an analysis of a constrained force feedback experiment using a low-order model. To extend these results to more general force feedback environments and still retain a quantitative predictive ability, the models of the hand/interface system and the environment should be augmented. The hand system, for instance, will behave differently along different axes of motion [122] and at different points in the robot workspace [17]. Also, not all environments can be modeled as a simple spring. An example of a more complex environment would be surgery, where tissues are viscoelastic and highly nonlinear [35]. Finally, desired trajectories may be different for different levels of force on the hand. Choosing different force feedback gains, however, can bring a range of environment forces and stiffnesses into the force levels on the hand encountered in our described experiment.

Chapter 4

Force Feedback Under 3D Ultrasound

4.1 Introduction

The use of 3D echocardiography allows surgeons to carry out a number of minimally invasive procedures on a beating heart, without using bypass and its associated side effects. These procedures can be performed because ultrasound imaging is not blocked by opaque blood. One such procedure is the repair of an atrial septal defect (ASD), a hole in the atrial septum allowing blood flow between the right and left atrium. We are developing a procedure that involves anchoring a Dacron patch over the hole using expanding wire anchors [112]. The patch and anchors are placed using instruments that operate through small ports in the heart wall. This approach has been successfully demonstrated in *in-vivo* animal experiments. Successful deployment of the anchor is difficult; both proper positioning (localizing the anchor deployment tube over the patch and tissue surrounding the ASD) and proper force application (a force over a minimum threshold) are required for successful deployment (Fig. 4.1). We focus on the force application/regulation part of this task because the role of the sense of touch in surgery is still not well understood. Further complicating the situation is the imperfect visual information from the 3D US. Understanding how the tradeoff between visual information and haptic information relates to performance will allow better tools for surgeons, both in the form of visualization techniques and force feedback for surgical robotics.

Previous studies have investigated performance on tasks with combinations of visual and haptic feedback [18, 131, 115, 74, 57, 65, 15, 124, 36]. An important difference here is the nature of the visual feedback. In most previous studies, the visual feedback takes more of the form of sensory substitution, with a graphical representation of a signal from a force sensor. Our study uses 3D US, whose rendering is a direct viewing of deformation (albeit degraded). An increased degree of processing is therefore necessary for subjects to extract force information. Furthermore, the amount of visual feedback of deformation and therefore force is different depending on the stiffness of the material deforming. Finally, the visual feedback in most of the previous studies returned absolute force information. Direct view only provides feedback on the relative force being applied.

We investigate the role of force feedback in the anchor deployment phase of ASD repair, a force control task under imperfect visual feedback (3D US). Force feedback is hypothesized

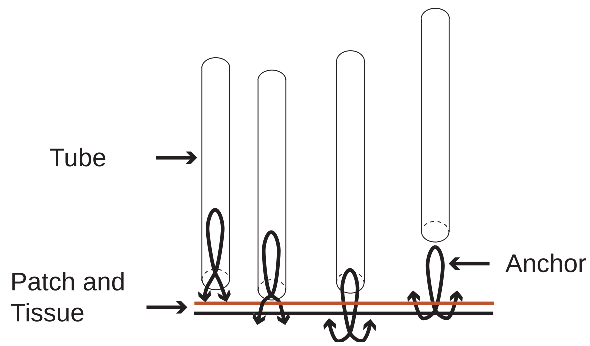


Figure 4.1: Anchor deployment in ASD repair

to improve performance over only visual feedback based on the results of the above prior work. Understanding of the interaction between vision and the sense of touch is important in surgical robotics, where dexterity is enhanced but force feedback is lost[33]. Also, understanding how force feedback can improve performance when visual feedback is limited, as suggested by [36], would help determine the necessary feedback quality for a given surgical task. We carry out a force control task mimicking the anchoring step in ASD repair to investigate these issues.

4.2 Methods

This experiment investigates the ability of surgeons to use force feedback to regulate the interaction force between a surgical instrument and tissue. The experiment mimics the task of anchor deployment, where the surgeon presses an anchor deployment tube against a patch covering tissue (Fig. 4.1). The key factor to success in this task is proper regulation of the force between the tube and the patch/tissue. Forces that are too high cause the patch to laterally slide into the hole, resulting in the anchor grabbing the patch and not the tissue. Forces that are too low will result in failure of the anchor to puncture both patch and tissue, again resulting in an unsuccessful deployment. Consequently, we attempt to encapsulate the force threshold part of the anchoring task.

Tasks are executed using 3D US as visual feedback to examine the effect of a novel and imperfect visualization modality on performance. Subjects carry out the same force regulation task with different force information feedback modalities to determine the ways in which force feedback affect performance.

4.2.1 Telemanipulation System

We used two Phantom haptic interface devices (Model 1.5, SensAble Technologies, Inc., Woburn, Mass.) as a laboratory teleoperation system [16]. Teleoperation is used to investigate different forms of force feedback while maintaining the same interface. One Phantom acts as the surgeon master controller and the other acts as the surgical robot. The master is an unmodified Phantom with the stylus attachment. Subjects control the motion of the surgical robot by moving the stylus, held in a pen grasp, where the tip of the stylus maps to the proximal end of the instrument shaft. The port was placed at the middle of the length of the instrument shaft so that motions of the hand were the same scale as instrument motions. The surgical instrument was a tube (14 gauge blunt needle, diameter 2.1 mm) identical to the one on the anchor deployment device (Fig. 4.2).

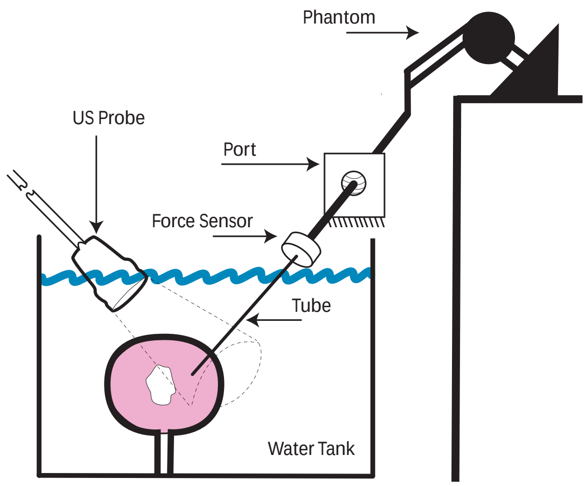


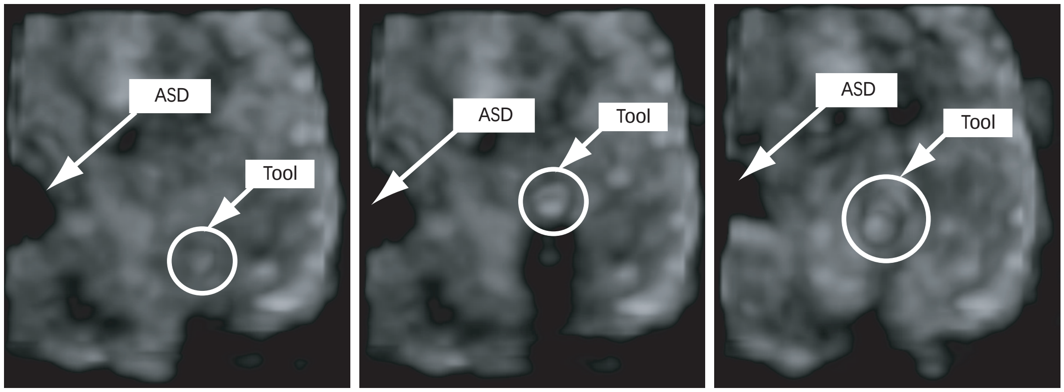
Figure 4.2: Surgical robot and surgical environment

Axial forces (along the shaft of the instrument) are sensed by a one-axis force sensor with an RMS noise level of 0.01 N (LCFD-1KG, Omega Engineering, Inc., Stamford, CT) built into the instrument shaft. The surgical robot is controlled with a standard position feedforward scheme traditionally used in teleoperated systems [107]. When in force feedback mode, the Phantom control computer samples the axial forces at 1 kHz and transforms the forces to the proximal end of the shaft, removing the transmission of friction forces at the port.

The teleoperation system, including the master, the surgical robot, and the force sensor, are controlled by a 2.0 GHz Athlon computer running Windows XP. The surgical robots position is controlled using proportional position/velocity control, independent of force feedback, with gains $k_p = 0.2$ N/mm and $k_d = 0.00035$ Ns/mm. These values were empirically derived to provide uniform stiffness in the portion of the workspace used for these trials while maintaining stability of the teleoperation system [16]. The control algorithm is implemented in Visual C++ along with the force sensor interface. All forces and positions were logged at 1 kHz.

4.2.2 Visual Feedback

The subjects received visual feedback from a 3D US system (Sonos 7500, Philips Medical Systems, Andover, MA) (Fig. 4.3). The system records three dimensional volume information using a phased array ultrasound transducer, then renders a 2D view for display. Objects of different mechanical impedance are rendered with different intensities. Thresholding allows segmentation of these objects from the rendered view, thus opaque blood can appear transparent while tissue is still visible. Image quality is not perfect, however, as the segmentation introduces irregularities at surface boundaries. Also, rigid instruments



(a) Frame 1 (b) Frame 1 (c) Frame 1

Figure 4.3: US images for the start, middle, and end of a trial. Tool starts above the tissue, subject moves the tool to the target, then the tool is pressed into the tissue.

introduce scattering and shadowing artifacts to the visual display because of their high mechanical impedance. The anchor deployment tube was coated with a low impedance plastic to reduce these effects.

To minimize the mental effort of relating visual and instrument frames, the rendering view was chosen to match closely with the relative positioning of the master and display [121]. The view remained constant across all trials and subjects. The rendered view displayed a volume of 3 cm x 3 cm x 2 cm with a voxel resolution of approximately 0.5 mm.

The experiment was performed in a water tank to allow US imaging. The probe was mounted at a 30-degree angle to the tissue to allow full visualization without interference with the motion of the instrument.

4.2.3 Force and Vibration Feedback

We provided three modalities of force information feedback during the experiment. The first was only the rendered view from the ultrasound (US); subjects had to determine the level of force application based on the observed deformation. The second modality was the US view combined with force feedback (US+FF). The surgical master would push back on the subject's hand with a force proportional to the one sensed by the force sensor. That force is scaled by a 75% gain, chosen to provide an intuitive level of force to surgeons for this specific experiment.

The third modality was the US view combined with a tactile vibration (US+V) that was chosen to provide near optimal feedback on the surgeon's applied force without applying a net force to the hand or requiring the surgeon to shift gaze. The vibration force commanded to the master was

$$F_v(t) = \begin{cases} 0 & F_a(t) < 1.5 \\ 0.2 + 0.4(F_a(t) - 1.5)\sin(500\pi t) & F_a(t) \geq 1.5 \end{cases} \tag{4.1}$$

where $F_a(t)$ is the measured contact force.

This caused the user to feel a vibration as soon as the proper threshold was reached and give a scaling cue if the user continued to apply a force over the threshold. The vibration in the motors also manifested as an easily perceptible auditory cue. From pilot studies, users were easily able to apply 1.5 N of force with high accuracy and precision with this feedback modality.

4.2.4 Tissue Target

An excised porcine atrial septum was used to match the tissue mechanics during an actual surgery while maintaining repeatability. The heart tissue was mounted to a rigid wire frame 4 cm in diameter for ease of positioning (Fig. 4.4). An artificial defect (8 mm diameter) was created by excising the central part of the septum to mimic the true environment and provide a recognizable landmark in the US view. The tissue was harvested 2 months prior to the experiment, drained of blood, and kept viable in a 10% Formalin solution. The same tissue was used for all trials.

Two positions on the tissue were chosen as targets for the force application. More than one target position was used so subjects did not anticipate and remember the exact hand motion necessary to execute the task. Also, as stiffness of the tissue may impact both the force and visual feedback [36], positions of different stiffnesses were chosen by choosing positions with different distances from the ASD (Fig. 4.4). The measured stiffnesses of the lower stiffness target position (closer to the ASD) and the higher stiffness target position (farther from the ASD) were 160 N/m and 240 N/m, respectively.

The patch and the anchor deployment system are not included in this experiment to reduce the number of variables. Only the deployment tube and tissue are used to investigate the key factor of accurate force application. Without the patch, the optimal force needed to ensure proper anchor deployment while avoiding puncture (as determined by pilot studies) is 1.5 N.

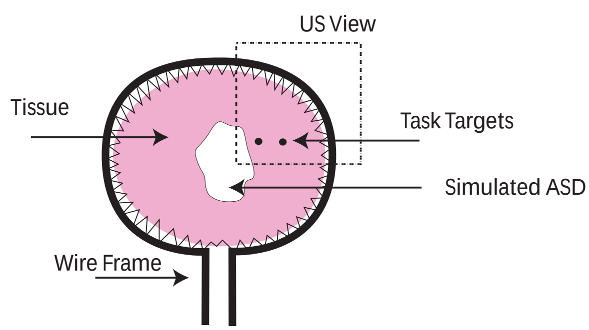


Figure 4.4: Tissue with target positions for force application

4.2.5 Protocol

Subjects were instructed to move the deployment tube to the correct position, then apply the tube against the tissue with the correct amount of force (1.5 N). Subjects then held down a button on the stylus interface for one second to signal when they felt the correct force amount was being applied. These trials were executed under three different forms of force information and at two different locations on the tissue. Before each trial, subjects were informed of the force information condition (US, US+FF, or US+V) and the desired position (closer to or farther from the ASD). The accuracy of the force application was the only error criterion given to subjects (subjects were not asked to trade off speed for accuracy).

Subjects trained for approximately 10 minutes to familiarize themselves with the teleoperation system and to learn the feedback when applying the correct force under different feedback conditions. During training for the US and the US+FF cases, subjects were verbally informed when the 1.5 N threshold was reached. For the US+V case, subjects learned to apply the lowest amount of force such that the vibration occurred. Under all cases (feedback and position), subjects trained until the threshold could reliably be applied three times in a row.

Eight surgeons were subjects, all with backgrounds in minimally invasive surgery (more than 3 years) but having little experience with manipulation under 3D US. The range of backgrounds varied from surgical residents to attending surgeons. Each subject performed 5 trials of each combination of feedback type and target position, for a total of 30 trials per subject.

4.2.6 Measures

Four different outcome measures were examined for each trial to characterize the performance of a subject: the mean force during the final second; the coefficient of variation of each subject's final second mean force (standard deviation divided by the mean); the rate of a successful anchor placement (whether the final second mean force was within 0.5 N of the target force); and the trial time. The coefficient of variation was included to examine subjects' repeatability in force application (precision) without the scaling effect of mean. Even though time was not told to the subjects as a specific error measure, time is included to examine whether subjects were trading off time with force.

4.2.7 Statistical Analysis

To determine statistical significance of our experimental conditions we used a repeated measures ANOVA with within subject variables of force feedback condition and stiffness condition. The statistical analysis included success rate, mean force, coefficient of variation, and time. The SPSS statistical analysis software package (Version 13.0, SPSS Inc., Chicago, Ill.) was used to carry out the analysis. A p value of less than 0.05 was considered statistically significant.

4.3 Results

Success on a trial was evaluated by determining if the mean force in the last second of each trial (during which time the subject was pressing the stylus button) was within +/- 0.5 N of the target application force of 1.5 N (Fig. 4.5). Feedback condition significantly influenced success rate ($F(2,14)=9.07$, $p<0.02$), with the US+V feedback resulting in the highest average success rate of 96%. The main effect of stiffness did not significantly affect success rate ($F(1,7)=0.004$, $p=0.9$) because the effect was opposite for the US and US+FF cases. Thus, the significant effect of stiffness manifested in the interaction term between

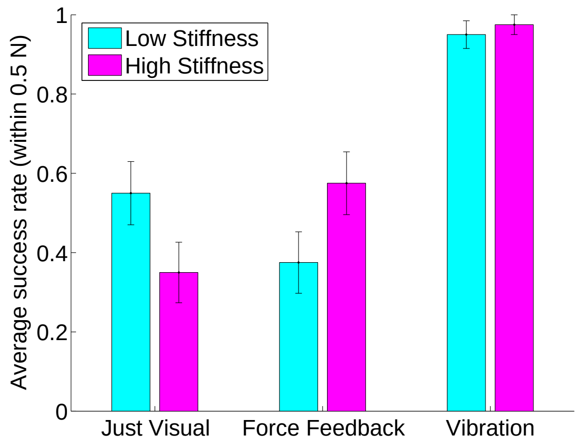


Figure 4.5: Success rate (within 0.5 N of target). Error bars show standard error.

stiffness and feedback type (F(2,14)=8.95, p<0.02). Average success rates of approximately 56% were achieved under the low stiffness, US and the high stiffness, US+FF cases. For the other cases of high stiffness, US and low stiffness, US+FF the average success rate was below 38%.

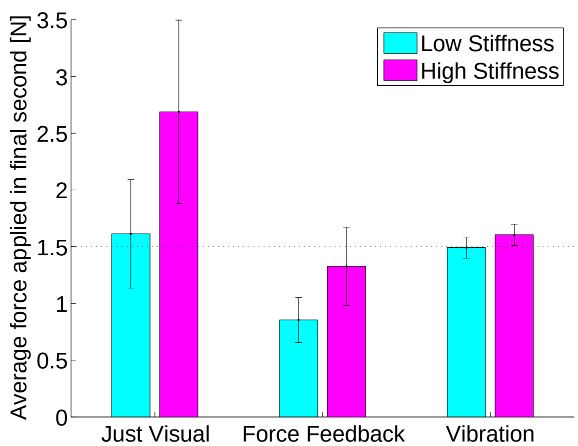


Figure 4.6: Mean force for different conditions. Dotted line shows target force and error bars show 95% confidence interval of mean estimation.

Mean force applied during the last second of each trial (Fig. 4.6) was significantly lower for low stiffness targets (F(1,7)=22.8, p<0.002). Feedback condition also significantly affected mean force (F(2,14)=15.68, p<0.005). The highest average forces were applied under the US feedback, where mean forces for both the low and high stiffness cases were above the target force of 1.5 N. Forces were consistently below the target force when subjects received US+FF feedback, applying an average of 0.85 N and 1.32 N for the low and high stiffness cases, respectively. Subjects were most accurate with the US+V feedback case, with both means being within 0.1 N of the target force. Intersubject variation is highest with only ultrasound feedback (Fig. 4.7) and minimal with the vibration feedback.

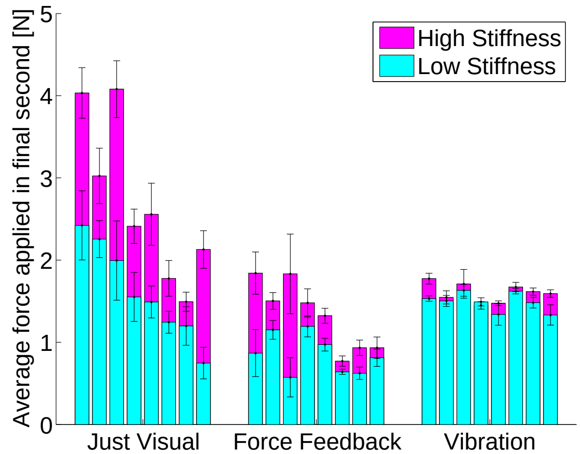


Figure 4.7: Mean force for each subject. Subjects are in same order for all three conditions. Error bars show standard error.

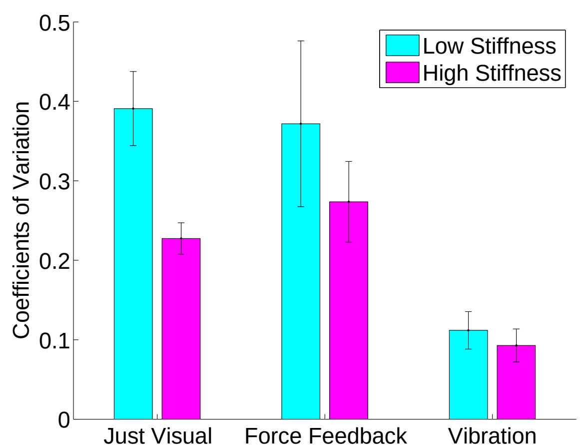


Figure 4.8: Average coefficient of variation. Error bars show standard error.

Coefficient of variation of the *mean* force applied during the last second was analyzed to determine the effect of feedback and stiffness on precision of force application (Fig. 4.8). A lower stiffness target position significantly increased the coefficient of variation ($F(1{,}7){=}9.25$, $p{<}0.02$). Feedback condition also significantly affected the coefficient of variation ($F(2{,}14){=}9.12$, $p{<}0.02$), with the vibration feedback case having coefficients at least twice as small as the other two conditions. A pairwise comparison, however, reveals that the only two significantly different feedback conditions were US and US+V ($p{<}0.001$). The difference between US+FF and US+V almost reached significance ($p{=}0.051$).

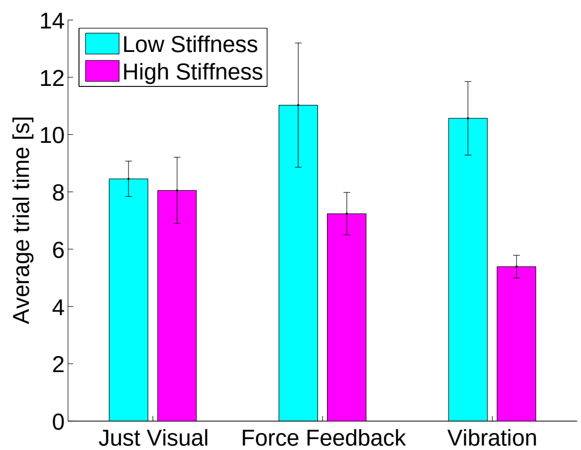


Figure 4.9: Average trial time. Error bars show standard error.

Average time needed to complete a trial was analyzed to assess any performance tradeoffs (Fig. 4.9). Having a low stiffness target position significantly reduced the average time spent per trial4.4 ($F(1{,}7){=}12.9$, $p{<}0.01$), reducing time by up to 5.3 seconds in the US+V feedback condition. Feedback condition did not significantly affect trial time ($F(2{,}14){=}0.94$, $p{=}0.365$), with all feedback conditions averaging from 7 to 10 seconds per trial.

4.4 Discussion

In this experiment we tested the hypothesis that the addition of force feedback to a force application task when visual information is suboptimal would result in improved performance. Our results suggest that performance is dependent on the stiffness of the object being pushed and that the addition of force feedback does not improve performance nearly as much as an indication of the specific target force level.

A goal of this study is to understand the tradeoff between vision and haptics as they relate to performance in a force control task. A number of previous studies have established the degree of force control precision (using visual feedback of force) of the finger[18, 131], elbow and wrist [115], or with a probe [74]. Jones demonstrated that the feedback of a visual representation of force improves accuracy and precision on a force control task over force feedback alone [57]. These studies, however, used a visual representation of force more akin to sensory substitution. While sensory substitution has been shown to aid performance in a force control knot tying task [65], little work has been done examining the tradeoffs between force feedback and direct vision in a force control task (the situation most often encountered in surgery). Cao showed that direct view of deformations can provide force information and associated performance benefits on a simulated task [15]. Desai investigates direct view of deformations in conjunction with force feedback, but only for the identification of stiffnesses [124]. Finally, Gerovich *et al.* examine visual and haptic feedback in a position control task (where the visual feedback is simulated deformation) and find that force feedback may not be necessary unless visual feedback is limited [36].

A recent hypothesis about the relationship between vision and touch is that the central nervous system uses maximum likelihood estimation (MLE) to determine dominance [31]. Therefore, the sense with the lowest variance in the estimate of the salient parameter is weighted more than the other sense. Because there is a clear dominance issue in our results (in some cases, subjects do worse when force feedback information is added), we will discuss our results in the context of maximum likelihood estimation and information.

This framework leads to a hypothesis of why subjects performed better with just US in the low stiffness case than the high stiffness case. With low stiffness, for a small change in force, there is a large visual deformation. Conversely, with high stiffness, a small force of the deployment tube against the tissue will only result in a small visual deformation. Thus, when interacting with objects of differing stiffnesses, objects with lower stiffness will have more information in the visual channel. A reason that the forces were higher in the high stiffness case is that subjects had to apply a certain deformation to get a visual signal out of the noise; that same deformation will result in higher forces for a high stiffness tissue.

The MLE theory can also account for why the US case performed well in the low stiffness case and the force feedback helped in the high stiffness case. This could be explained by dominance—when force feedback is added, force dominates because of the perceived low variance in the force signal. In the high stiffness case, there is a large amount of signal in the force feedback channel while there is relatively little in the visual, so the dominance is correct which results in an increase in performance. An alternative explanation is that mechanical work is the salient parameter governing this force control task, taking into account the displacement over which a force is applied, as investigated by [114].

An explanation of the low overall forces in the force feedback condition can be given by extending this statistical framework to include expectation and prior knowledge. Previous work has shown that the central nervous system behaves according to a Bayesian framework during force control tasks, integrating a prior expectation of force with current sensation for control [68]. Work in softness discrimination has also shown this anticipatory behavior [72]. The low forces observed with the US+FF condition may be a result of surgeons' prior expectations of the forces encountered in surgical tasks. Another explanation, also proposed by [68], is that subjects trade off force control correctness with force control effort. At high forces, the effort needed becomes significant enough to affect subjects' idea of the target force.

4.4.1 Application To Surgery

Several differences exist between this study and an actual 3D US-guided ASD repair. A primary difference is that we are using excised, static tissue when the real surgery is carried out on living, dynamic tissue. The use of the formalin solution to preserve the tissue also stiffens the tissue somewhat, changing the mechanical interaction. Surgeons may also be using different expectations and training for force interaction with tissue due to the interface of the surgical robot because different muscle groups are utilized [58]. The scaling of hand motions to tip motions due to port positioning is a specific example of this interface difference. In our task, the port was placed to provide uniform scaling, while in actual surgery, the port is often placed to scale down the motion of the hand.

Another difference between this study and actual surgery is the quality of the visual feedback. During surgery, the relative position of the heart and the ultrasound probe can change with the beating of the heart. Often, the ASD moves out of the scope of the probe and needs to be relocated. This effect, combined with the complexity of patch deployment, temporary occlusions by tools, and noise introduced by fluid flow all serve to degrade the information in the visual channel [109]. Returning to the statistical framework, the increased variance in the visual channel information may cause other more informative modes of feedback to dominate.

The benefit of force or any type of informational feedback clearly depends on the task and the consequences of error. For instance, in the anchor deployment task, over application of force could result in a puncture. Therefore, force feedback (which minimizes peak forces[128]) can improve overall performance. Another example of correct feedback is in applying optimal suture tying forces using a robot; feedback of forces through sensory substitution improves force accuracy even over direct contact [65].

That being said, a final point to address is the feasibility and ease of use of any feedback in augmenting surgical performance. For instance, even though the vibration feedback allowed subjects to accurately and precisely control tool tip force, most surgeons preferred the force feedback. One reason given is that the vibration was distracting; it was difficult to pay attention to anything else. In surgery, when the surgeon needs to coordinate a number of complex motions based on a range of subtle cues, the presence of one powerful cue may wash out other subtler cues. Potentially, there is less cognitive workload imposed by a natural feedback such as force feedback. Another hypothesis is that surgeons are used to having forces and use of the robot with just visual feedback removed all force feedback. In either case, force feedback has been shown to be useful across a number of surgical tasks [124, 128, 60], while the vibration feedback given here is useful in just this one task.

The results from this experiment lead to a number of interesting questions about the benefit of different forms of feedback in surgery. An obvious hypothesis just mentioned is mental workload of different forms of feedback in surgery. Force feedback is costly and not necessarily a perfect information source for all tasks, but it can potentially provide benefit (both passive and informational [127]) with little increase in cognitive demand because of its intuitive nature.

Chapter 5

Force Sensors For Surgical Robots

5.1 Introduction

An obvious feature missing from current robotic surgical systems is force feedback. In spite of many surgeons desire for this feature and force feedback's demonstrated benefit [128, 107], force feedback remains unimplemented in surgical systems. The difficulty is with the ability to accurately sense the interaction forces between the instrument and the environment. The leading surgical robot, the DaVinci (Intuitive Surgical, Sunnyvale, CA) [33], has the capability of recreating forces against the hand, yet does not implement force feedback because of the lack of force sensors at the instrument tips. Implementing a force sensor is difficult, however, due to the stringent design requirements imposed by the surgical environment. The force sensor needs to fit through a small port (5 mm to 12 mm) and transmit the sensed force information back outside the body. Though the sensor is small, previous studies indicate that the force sensor should sense up to 5 N of force [94] for general procedures. Because of the friction forces between the instrument and the port, the sensor should be located on the distal end of the instrument. An added difficulty is introduced with the use of cable drives for articulated instrument wrists; in order to avoid sensing the internal forces of the cables, the force sensors need to be located on the tip of the instrument. Additional design requirements include being robust to temperature differences (going from room temperature to internal body temperature), waterproof, and sterilizable.

A number of previous designs have been proposed for force sensors in surgery. A three axis force sensor that was independent of moments was developed for laparoscopic instruments [5]. The design was limited, however, in that it could not sense grip forces. A three axis force sensing gripper has been proposed using a force sensing resistor for the third axis [125]. The device was limited by its low resolution due to the force sensing resistor. Optical force sensor design have also been investigated [94]. In addition to the specific problems mentioned, the previous designs are also intricate, requiring many man hours to construct. All of the devices based on strain gages suffer from the arduous process of mounting and bonding the gages effectively. Therefore, sterilization becomes necessary to be able to reuse the expensive sensor.

We propose a novel three axis force sensor design for use with surgical robots. The force sensor takes advantage of the high sensitivity and resolution of strain gages while removing the need for a complicated bonding ritual by embedding the gages inside a pourable epoxy. This approach is based on Shape Deposition Manufacturing (SDM) [20, 81], a technique for constructing structures with embedded components using a cycle of machining and casting. Our SDM based force sensor approach allows for a small sensor that can fit through laparoscopic ports that can be mass produced and thus can be disposable. The sterilization issue is thus avoided. No complex machining of the sensor is required, as the sensor is casted with a reusable mold. A key feature of our design is the incorporation of a heat shield into the sensor, removing the effects of temperature on strain gages embedded inside an epoxy. Our specific design can potentially be attached to the grippers of any surgical robot, making the approach also ideal for research. The same techniques proposed here, though, can also be used to make force sensors integrated into any instrument. Our current design can fit two grippers through a 12 mm port, and the same technology can be straightforwardly applied to generate a 10 mm or 5 mm pair of sensors. We also present a metal element based three axis force sensor design for comparison. This design also uses six strain gages and is similarly sized, built using conventional strain gage based force sensor design techniques.

5.2 Sensor Design And Construction

We focus our design description on the SDM based force sensors. While the metal sensors use the same number of strain gages and are similarly sized, the techniques used in their construction are well established [82].

5.2.1 Components

Our force sensor design uses six silicon strain gages as the transducer elements. We chose silicon strain gages because of their small size (1 mm x 0.25 mm), high sensitivity (high ratio of applied strain to resistance change) and relatively low cost (SS-037-022-500P, Micron Instruments, Simi Valley, California, USA). A full sensor worth of gages (6 gages) is approximately $30 in gage cost.

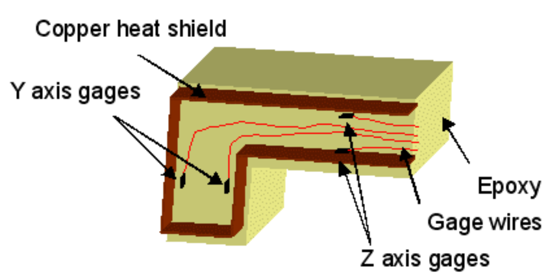


Figure 5.1: Cross-section of SDM force sensor. X axis gages are located at the same distance from the base as the Z axis gages, but are in and out of the plane of the diagram.

To avoid the bonding step normally associated with strain gages, we embed the gages *inside* the element of our force sensor instead of bonding to the surface. To do so, we use a pourable two part epoxy that hardens upon curing. We evaluated a number of materials and epoxies for our element material. Several properties were desired in an element material, including: high stiffness, low creep (so that the strain would not change over time with a given load), low viscosity when mixed (so the epoxy can fill all parts of the mold), and low curing temperature (so the epoxy can cure inside a wax mold). The epoxy that was chosen (Resin 105 Fast Cure, West System, Bay City, Michigan, US) demonstrated all of these qualities.

A key component of our sensor design is the heat shield used to equalize temperature variation between gages. Because epoxy is an insulator, heat takes time to propagate from one side of the sensor to another. This negates the standard temperature rejection scheme of examining the difference in strains on opposite sides of a bending beam. Now that one strain gage is at a different temperature than its opposite, a difference in the two strain gage readings due to strain is indistinguishable from a difference in temperature. With the heat shield in place, the temperature equilibrates quickly, and differences between the two gages once again depends only on strain. We used standard copper braid normally used for wicking solder as the heat shield (Ungar-Wick #4, Ungar Products, Apex, North Carolina, USA).

An aluminum anchor is used as the connector that anchors the epoxy to the surgical grasper (of a surgical robot, for example). The anchor attaches to the grasper using set screws. The epoxy rigidly bonds to the metal because the anchor is roughed before pouring the epoxy and holes are drilled into the anchor to let the epoxy fill (Fig. 5.2).

5.2.2 Construction Process

Our force sensor design process utilizes a two pour casting process. The first pour embeds the strain gages while the second embeds the copper heat shield. There are two phases to the first pour; placement of the individual gages and forming the element. Wax molds were used in all pours to form the element structure. The molds were machined with standoffs to aid in gage placement and wire soldering. Machine wax was chosen as a mold material because of its ability to be machined with high precision and because it did not permanently bond with the epoxy used when curing. Each wax mold component was formed through precise CNC milling.

The first phase of the primary casting involves positioning the strain gages on the wax standoffs and soldering the gages to the lead wires (Fig. 5.3). These standoffs mechanically held the wires in place by slightly embedding them in the surface after the wires were heated. The strain gage was placed, then soldered by hand.

The second phase of the primary casting was forming the element. After the individual strain gages were attached to each mold piece, the mold was brought together along with the aluminum anchor. The lead wires were routed out of the channel through which the epoxy was poured. After pouring and curing the epoxy, each mold piece was removed, leaving the wires and strain gages embedded in the epoxy.

A second epoxy pour was used to embed the copper heat shield around the initial element. The copper braid was wrapped around the element, and held in place using solder. A second mold was then used to form the final outline of the sensor.

When constructing a force sensor by hand, an experienced builder can construct a sensor with 3 man hours of manual manipulation. The design approach, however, easily lends itself to a mass production approach, since the pouring is a two step process. Given a frame that holds the gages in place every time, a mass produced, disposable force sensor becomes reasonable.

5.2.3 Element Description

The force sensor is a dual beam configuration. One strain gage is locate on one of each of the four sides of the first bending beam, allowing the sensation of the two bending moments of the proximal beam. Two final strain gages are located in the final axis of the second bending beam (Fig. 5.4). The face of the second bending beam becomes the new grasper face.

The sensor can be mounted on a number of different graspers, and the design can easily be adapted to accommodate a number of different instruments. We have attached the sensor to the end of an instrument used by the Laprotek system (Fig. 5.5). Future designs will incorporate different metal grasper faces to accomodate different surgical tasks (such as needle driving).

5.2.4 Metal Design

The metal element design uses an aluminum element. The aluminum is anodized to prevent electrical conduct through the strain gages and any stray wire contact with the element (Fig. 5.7). To reject moments, we designed the element as a serial chain of moment rejecting flextures (Fig. 5.6) [82]. Two strain gages are mounted to the back of each flexture. A similar set screw based attachment design as the SDM sensor is used for affixing the sensor to a grasper. Note that the metal design requires precise machining for every sensor.

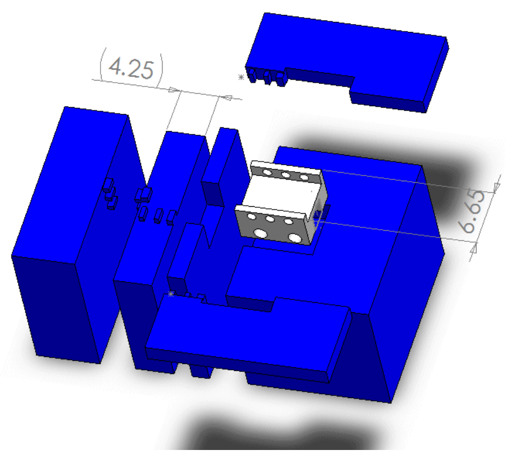
5.3 Sensor Characterization

5.3.1 Calibration

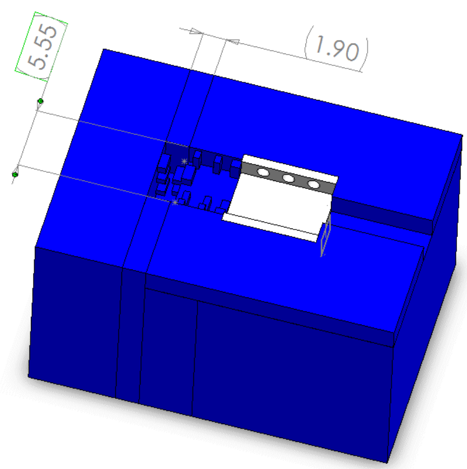
Due to the differences in design of the SDM sensor and the metal element sensor, we were able to take advantage of different calibration schemes for each. For both sensors, individual strain gages resistances were sensed through a Wheatstone bridge in a quarter bridge configuration. The voltage was then amplified by an instrumentation amp circuit with a gain of 100.

To calibrate the measured voltages from the strain gages with actual force for the SDM sensor, known loads were applied along each cardinal axis. All the gage voltages and known loads were then taken together, and the calibration matrix was found using a linear least squares method. Temperature effects were compensated for by allowing the temperature of the sensor to vary during calibration. Estimated forces versus known forces are shown in Figure 5.9a-c to demonstrate the calibration as well as the linearity. An RMS error of 0.15 N in calibration was achieved for the SDM based sensor.

The metal force sensor was rigidly mounted to a commercial, previously calibrated, high resolution force sensor (Mini 40, ATI) to quickly calibrate all axes simultaneously. A mounting bar epoxied to the face of the metal force sensor was used to interface with the commercial force sensor. Forces applied by hand were measured by both sensors, and a linear



(a)



(b)

Figure 5.2: Molds apart and together, showing aluminum attachment anchor. Top mold piece not shown. Epoxy enters from the channel on the right to fill the mold. Dimensions are in millimeters.

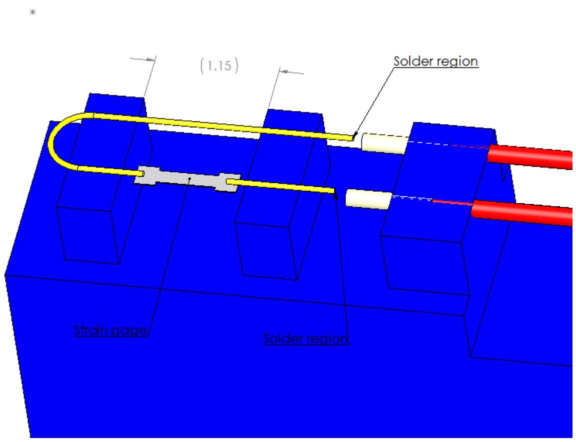


Figure 5.3: Closeup of wax standoffs used to position strain gages inside sensor. Standoffs also held wires in place while soldering. Dimensions are in millimeters.

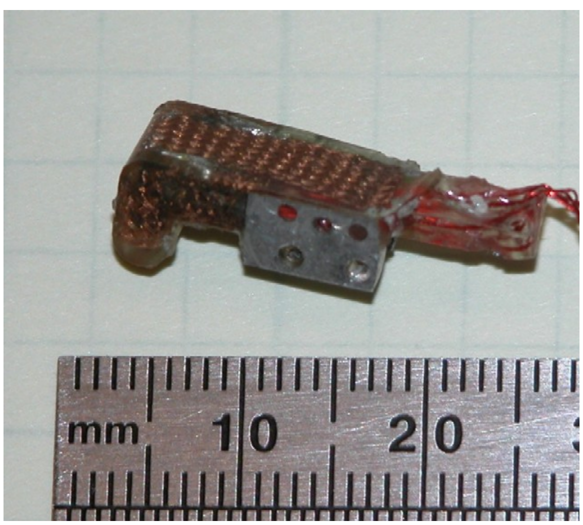


Figure 5.4: Completed three axis SDM based force sensor. Note heat shield and uniform wire exit.

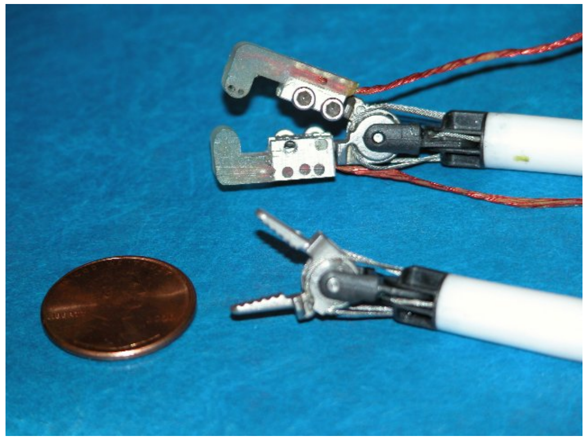


Figure 5.5: Force sensors attached to graspers of a surgical robot

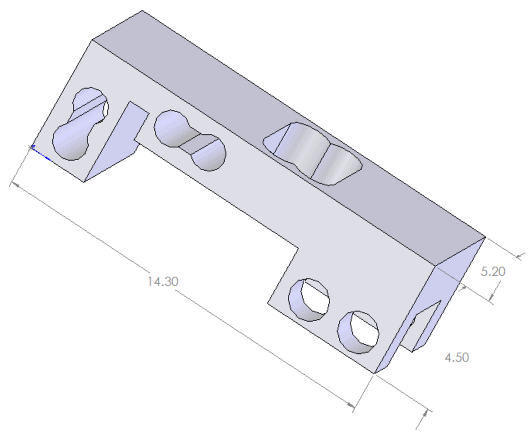


Figure 5.6: Metal force sensor element design. Dimensions are in millimeters.

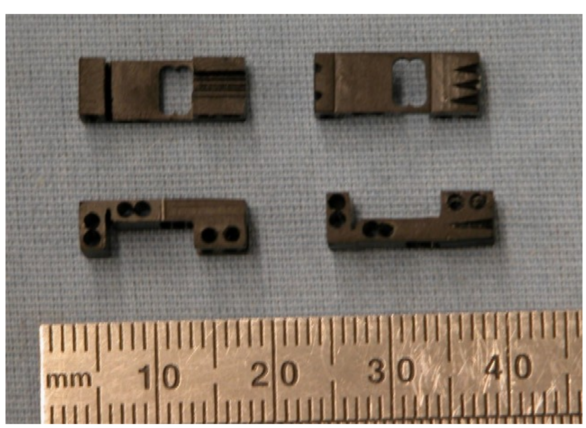


Figure 5.7: Anodized aluminum force sensor elements before strain gages have been attached. Notches reduce stress concentration in wires routed around corners.

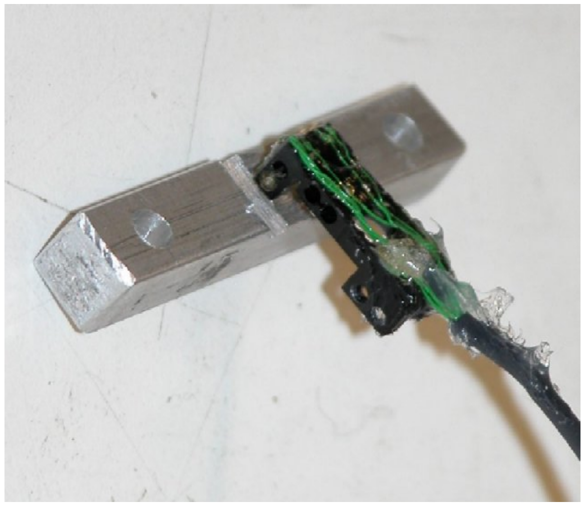
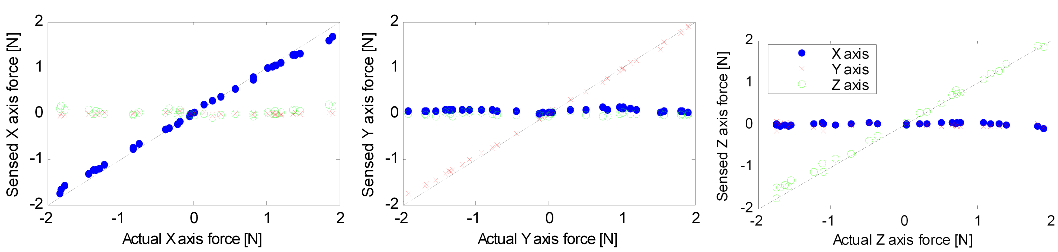


Figure 5.8: Metal element force sensor with strain gages and wiring. Green wires are lead wires to strain gages. Silicone is used at junction of wires and metal element as flexible strain relief. The force sensor is shown attached to a mounting bar used for calibration.



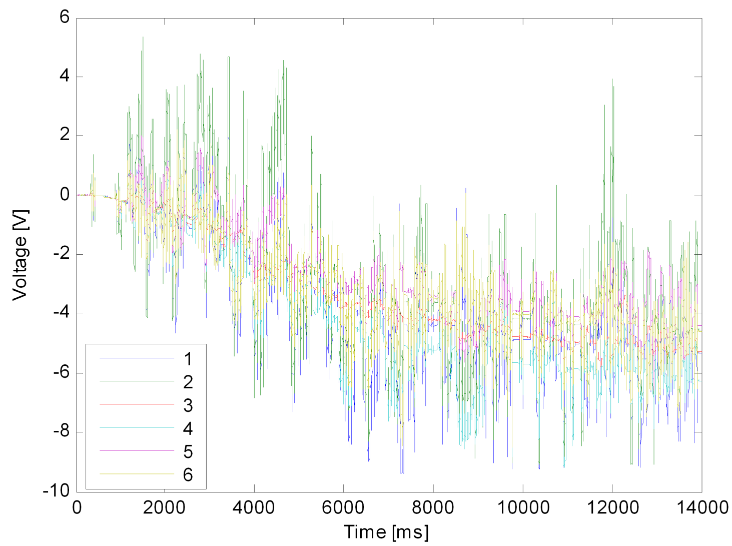
(a) X axis (b) Y axis (c) Z axis

Figure 5.9: Linearity of the SDM force sensor in the X, Y, Z axes

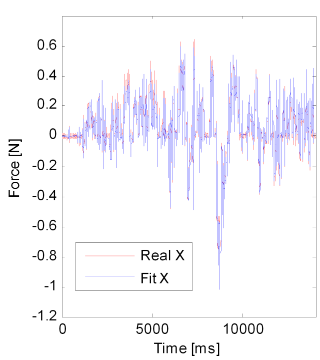
least squares fit was used to find the calibration matrix. This arrangement also allowed the simultaneous application of torques during calibration, increasing the moment insensitivity of the force sensors. As with the SDM sensors, the temperature of the sensor was allowed to vary during calibration to increase temperature insensitivity. Raw strain gage voltages and the corresponding fits are shown in Figure 5.10. The resulting calibration and linearity is shown in Figure 5.11, with an average RMS error in calibration of 0.07 N 5.1.

5.3.2 Noise Characterization

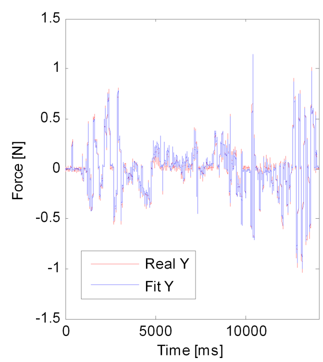
To characterize the noise and resolution of the sensors, the unloaded sensors were placed in an enclosed container and allowed to equilibriate. Data was then taken for 20 seconds. Because the major source of noise is high frequency noise, resolution is coupled with sampling frequency. At 1000 Hz sampling, the RMS noise of both sensors was approximately 0.1 N.



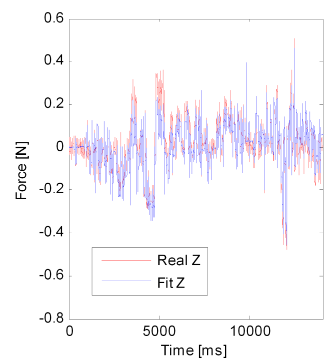
(a) Raw voltages



(b) X axis



(c) Y axis

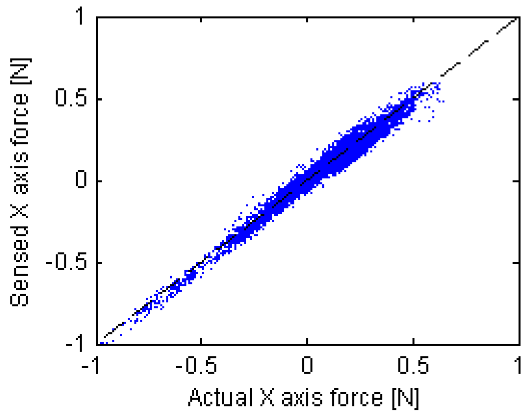


(d) Z axis

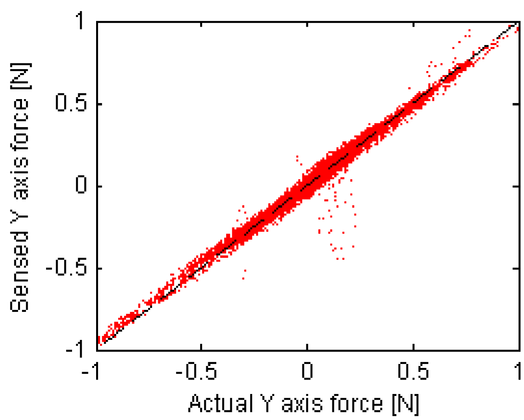
Figure 5.10: Raw strain gage voltages and fit forces metal force sensor in the X, Y, Z axes. Note the large offset drift due to temperature in the raw voltages that is not present in the fit forces, demonstrating temperature compensation.

Table 5.1: RMS errors in calibration for four metal element force sensors in Newtons

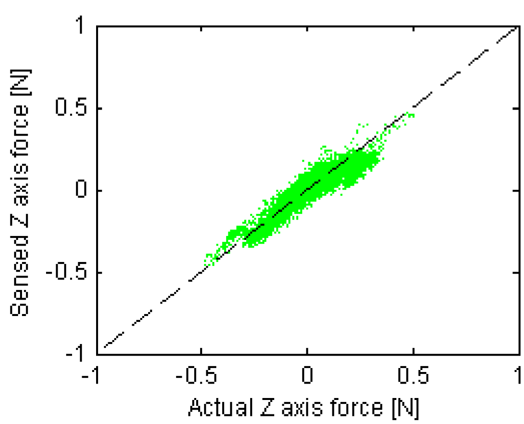
Sensor #	X axis	Y axis	Z axis
1	0.0456	0.0326	0.0691
2	0.0433	0.1145	0.0609
3	0.0702	0.0944	0.0926
4	0.0668	0.0692	0.0547
Avg	0.0564	0.0776	0.0693



(a) X axis



(b) Y axis



(c) Z axis

Figure 5.11: Linearity of the metal force sensor in the X, Y, Z axes

5.3.3 Temperature Drift

To examine the benefit of the heat shield, a sensor was casted without the inclusion of a heat shield. After calibration, sensors were allowed to equilibriate in open air. With the non-heat shielded sensor, we measured a drift of 1.2 N over 5 minutes. The sensor with the heat shield drifts only 0.15 N in 5 minutes, almost an order of magnitude improvement.

5.4 Discussion

While the benefits of force feedback are potentially large (limits interaction forces which directly correlates to trauma), to this point the benefit has been outweighed by the cost of force sensors. The restrictions of the surgical environment make the design of a robust force sensor costly. These restrictions include small size, difficult to access environments, and the addition of complexity to the system.

We present a proof of concept force senor design based on the SDM approach of embedding components into a castable substrate. Our three axis design has performance characteristics of a similarly sized metal element force sensor based on standard design techniques. The benefits of the SDM approach are many. First, a small enough force sensor can be constructed so that two grippers can fit through a 12 mm port. The force sensitivity of strain gages is retained without the complex bonding process normally associated with strain gages. No complex machining of the sensor is required (because it is casted). This design feature can lead to a straightforward mass production scheme, making the sensors low cost, disposable, and removing the issue of sterilization. The sensor is robust due to its monolithic construction. The force ranges can be easily adjusted by depth of gage placement within the element. Wire management is straightforward, with all wires exiting the sensor at the same point, and intrinsic strain relief provided by the epoxy. The sensors are also ideal for research, since the sensors can attach to a number of different graspers.

Material selection is a key limiting factor in the SDM design. Pourable materials often suffer from creep and hysteresis. While we chose the epoxy to have low creep properties, we can still see viscoelastic effects in the time response of the sensor. A number of ways exist to improve this, including implanting various fibers inside the epoxy, or choosing a different material type altogether.

One of the features of this approach is that it lends it self to mass production by automating the casting process. The major hurdle preventing that is the difficulty associated with handling the small strain gages. Specifically, bonding lead wires to the strain gage lead wires is difficult. Potentially, a closer interaction with strain gage manufacturers in addition to a frame for holding lead wires and heat shield would lead to a straightforward automated casting process.

Differences in calibration lead to varying degrees of temperature and moment insensitivity. The metal based sensor is relatively moment insensitive, due to its design and calibration scheme, as compared to the SDM design. This is a potential drawback, especially in surgical tasks which are dominated by large interaction torques, as is the case with suturing. Depending on what inputs are emphasized during calibration, both sensors can be more or less moment or temperature insensitive, with a trade off in force accuracy. An advantage of the metal approach is that moments and temperatures cause similar variations in strain gage output, both causing an overall increase in output of paired gages. Further, the axis are somewhat independent of each other. A simple differencing scheme will remove both of these error sources. Further, the axis are somewhat independent of each other. In the SDM design, because the sensor is simply a beam, the effect of moments and temperatures are coupled across axes and are difficult to carry out. A further progression of the SDM approach would be to randomly place the gages within the element and to calibrate. The previous observations reveal that the metal approach can always get more moment insensitivity or temperature insensitivity for fewer gages, because it takes advantage of the mechanics and precise positioning to cause error sources to cause similar variation.

A number of directions exist for future sensor improvements. Examining different grasp faces to suit different surgical tasks is an area of future research. Reducing the size of the force sensors would increase applicability to more surgical arenas. Finally, resolution can potentially be increased by embedding some of the amplification electronics within the sensor. If some of the instrumentation circuitry could be reduced to that size, noise would be reduced because of the shorter wires between gage and amplifier. Further, multiplexing circuitry could reduce the number of wires needed to connect to the sensor.

Chapter 6

Force Feedback For Cannulation

6.1 Introduction

Surgery requires executing complex motions in a three dimensional environment. Many tasks, including suturing, dissection, and anastamosis require precise positioning and orientation along a path to lead to a successful outcome [77]. For instance, placing a suture involves guiding a needle through a precise path so as not to tear tissue, while positioning the entry and exit point of the needle to successfully join two tissues [29]. A primary challenge surgeons face with minimally invasive, endoscopically guided procedures is that the visual feedback is two dimensional, while the task environment is three dimensional [132]. Through training, surgeons can use other depth cues to carry out tasks in three dimensions [105]. However, interpreting three dimesnional space with such limited sensory information may tax cognitive abilities [130, 4]. Consequently, we hypothesize that additional information in the form of force feedback would aid performance in minimally invasive tasks.

Previous studies of force feedback in surgery have focused on tasks with limited degrees of freedom. For example, previous work on force feedback in blunt dissection investigated a single-handed, three-axis positioning task that required no changes in orientation (Chapter 2). Kazi's work investigated force feedback in a number of tasks including cannulation and palpation, also using a one-handed, 3 axis positioning system [60]. Other investigations involving reduced degree of freedom tasks include suturing [116] and diagnostic grabbing [124]. Multiple degree of freedom, force reflecting telemanipulators for surgery do exist (e.g.[1]), however, their relevance to surgery is unclear. In most actual surgical tasks, surgeons must simultaneously control the positions and orientations of manipulators in both hands. Consequently, a full analysis of force feedback in minimally invasive surgery must be carried out under conditions that more closely replicate the true complexity of surgical tasks.

Here we investigate the effect of force feedback in a task that realistically replicates the complexity of minimally invasive cannulation. Force feedback can act as an information source (Chapter 4), and in complicated tasks a surgeon must assimilate many information sources to achieve a successful outcome. When information is limited (such as depth information in endoscopically guided tasks), force feedback may provide additional information that improves performance. We investigate a two-handed, six degree of freedom, endoscopically guided, minimally invasive cannulation task (inserting one tube into another tube).

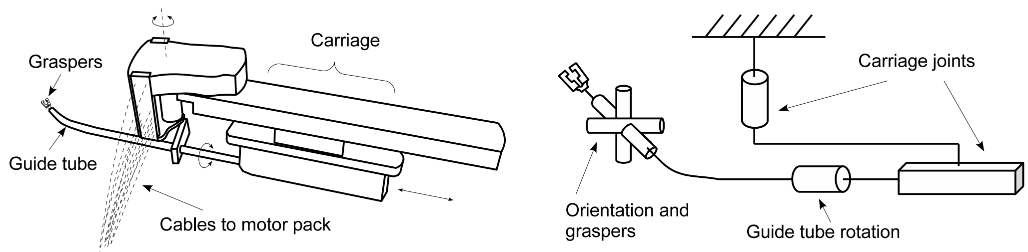


Figure 6.1: Laprotek robot arm and associated kinematic representation. Note that a rotation of the guide tube results in a vertical translation of the graspers.

Vision provides direct information for two positioning dimensions and one orientation direction. The final positioning and two orientation dimensions are more difficult to extract from the two dimensional visual field provided by the camera. We investigate performance of the cannulation task with and without three dimensional force feedback. Our hypothesis is that the addition of force feedback will enhance performance by reducing the time required to complete the task because the subjects gain additional information on position and orientation. We also investigate whether the performance benefit of force feedback depends on the training of the subjects.We compare task performance of subjects with and without minimally invasive surgical experience.

6.2 Methods and Materials

6.2.1 Teleoperation System

We used the Laprotek surgical robot (Endovia Medical, Norwood, MA) as the basis for our teleoperation system [34]. The robot provides two articulated, seven degree of freedom manipulators (three positions, three orientations, one grasping) (Fig. 6.1). The cable-driven, disposable surgical instrument provides two orientation degrees of freedom in the wrist articulating the surgical graspers. The instrument passes through a guide tube to access the surgical environment. Rotation of the instrument along its long axis within the guide tube provides the final orientation degree of freedom. Two joints position a carriage holding the guide tube assembly. A bend in the guide tube transforms rotation of the tube into translational motion, providing the final translational degree of freedom. A passive mechanical positioning arm normally suspends the carriage and guide tube assembly over the patient. All joints are cable driven, with the motors located in a motor pack usually mounted to the surgical table.

Several augmentations were made to the system to allow high fidelity bilateral force feedback. Modifications were required to improve the stiffness of the robot at the instrument tip. We bolted the carriage directly to a rigid mechanical base, replacing the passive mechanical positioning arms. We reinforced each guide tube by passing it through a rigidly mounted spherical joint, which decreased the lever arm between the instrument tip and the mechanical base. Finally, the internal cables of the surgical instruments were retensioned to further increase instrument stiffness at the tip.

The Laprotek system was further augmented by replacing the standard interface with

Figure 6.2: Handle addition to Phantom haptic interface

high fidelity haptic interfaces (Model 1.5, SensAble Technologies, Inc., Woburn, Mass.) [16]. While the original Laprotek system interface does provide some force feedback based on motor torques, the resulting feedback is only sufficient for implementing workspace boundary limits. Further, the standard interface suffered from a low mechanical bandwidth due to its cantilever design. The Phantom haptic interfaces allowed higher force feedback bandwidth. The Phantom haptic interfaces did not, however, provide torque or grasping force feedback, a capability provided by the standard interface. Thus, we achieved high fidelity three-dimensional force feedback at the cost of grasping and torque feedback. We added a lightweight handle (Fig. 6.2) to the standard Phantom interface to command the grasper angles. A low friction potentiometer (CP-UTX, Midori America Corporation, Fullerton, CA) acted as both the axle for the finger joint and an angle sensor.

We attached a custom built, three axis, 14 mm long force sensor to each of the grasper jaws (Fig. 6.3). The force sensors (Chapter 5) provided interaction forces in three dimensions with an accuracy of 0.07 N. We used strain gages epoxied to a metal element to provide a high bandwidth, high resolution force sensor [82]. The metal element consisted of a serial chain of moment rejecting flextures. These flextures, along with calibration in the presence of moments, reduced the sensitivity of the sensors to confounding moments. Two force sensors mounted to each jaw of a grasper could provide grasp force, although this was not used in the current study.

The majority of the teleoperation system, including the haptic interface and the force sensors were controlled by a 2.0 GHz Athlon computer running Windows XP. The teleoperation control software, written in C++ (Microsoft Visual C++ 6.0), updated the haptic interface at 1000 Hz. We used the original Laprotek system software to control the surgical robot, modified slightly to accept position commands over ethernet. The control software uses a standard position feedforward control scheme[107], and runs on the integrated QNX-based system, with an update rate of 95 Hz. As is standard with the Laprotek surgical robot, the position of the haptic interfaces were mapped directly to the position of the gripper.

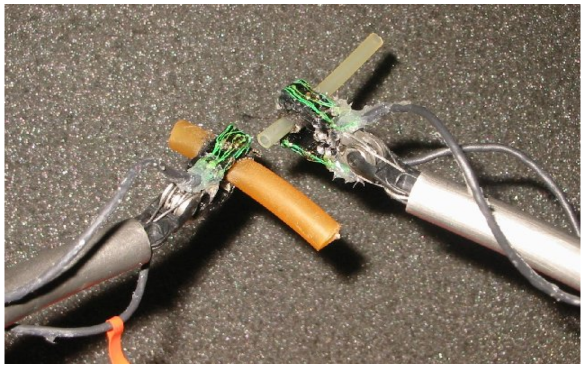


Figure 6.3: Force sensors attached to Laparotek graspers. Larger, compliant tube is shown in the left hand, smaller tube in the right.

We used position and orientation gains of 3 and 1.3, respectively, to provide the subjects with a comfortable range of motion during the experiment. These were also the default gain settings for the Laprotek system.

When force feedback was enabled, the Phantom control computer sampled the instrument forces at 1 kHz and transformed the forces to the haptic interface. A force feedback gain of 1 was used to provide high forces while maintaining system stability.

Teleoperation performance was limited by the hysteresis in the system. While position and orientation resolution was good (mean < 0.5 mm in position, 3 degree angle resolution), instruments suffered from a tradeoff between stiffness at the wrist and hysteresis. Increasing the instrument cable tension, and the corresponding stiffness at the tip, also increased frictional forces and the resulting backlash. The magnitude of backlash depended on the joint. The vertical direction in the visual field was the worst positioning axis, requiring an average motion of 20 mm to cause a direction change. The wrist joint perpendicular to the grasper jaw was the worst orientation axis, requiring an average orientation change of 15 degrees to change direction.

6.2.2 Visual Feedback

A video camera proximal to the robot arms provided visual feedback (Fig. 6.4. The relative orientation between the user and the monitor is approximately the same as the orientation between the camera and the graspers, to minimize the mental effort of relating visual and instrument frames [121]. However, lack of depth perception remained a source of difficulty.

6.2.3 Cannula Insertion Model

Two sections of tubing were used as the cannula model. One tube was a 4 cm section of stiff PVC tubing with an outer diameter of 3.2 mm and a measured Young's modulus of 0.87 GPa. The other tube was a 4 cm section of compliant rubber surgical tubing with an inner diameter of 3.2 mm, outer diameter of 6.4 mm, and modulus of 3.0 MPa. The diameters and materials were chosen so that little force (< 0.5 N) was required to insert one into the other

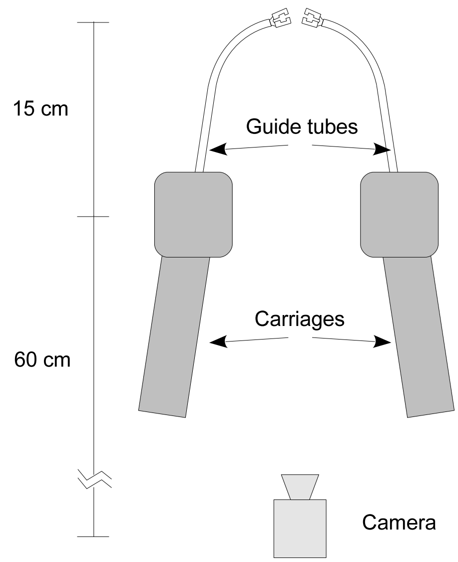


Figure 6.4: Layout of Laparotek arms and camera, top down view. During the task, the major axis of the tubes would be aligned perpendicular to the page.

when the two tubes were axially aligned. High friction grip tape placed around the shaft of the smaller PVC tubing limited the insertion depth to 6 mm. The difference in compliance between the two tubes provided a realistic representation of the salient mechanical interactions in a minimally invasive cannulation task. Examples of procedures involving such a task include intraoperative cholangiography [22, 89] and laparoscopic stented choledochorrhaphy [55].

6.2.4 Protocol

Subjects carried out the cannula insertion task in the presence or absence of force feedback. Subjects were asked to carry out the task as quickly and as gently as possible. No explicit tradeoff between speed and force was advised. Subjects began the task with the tips of the tubes within one cm of one another and with both tubes aligned vertically in the visual view (Fig. 6.6).

Subjects trained for approximately 10 minutes to familiarize themselves with the teleoperation system and to gain a sense of the force necessary to successfully complete the task. By the end of training, all subjects could successfully join the tubes within 60 seconds with and without force feedback.

Twelve subjects participated in the experiment. Six subjects were surgeons, all with backgrounds in minimally invasive surgery (more than 3 years of training). We specifically chose surgeons with minimally invasive surgical training because they have experience with two dimensional visual feedback [105] which creates a disparity between apparent visual motion and proprioceptive hand motion [11]. Six graduate students with no minimally invasive surgical experience provided an untrained population for comparison. Each subject performed 10 trials with and without force feedback, for a total of 20 trials per subject. Trial order (with and without force feedback) was counterbalanced across the 20 trials, with

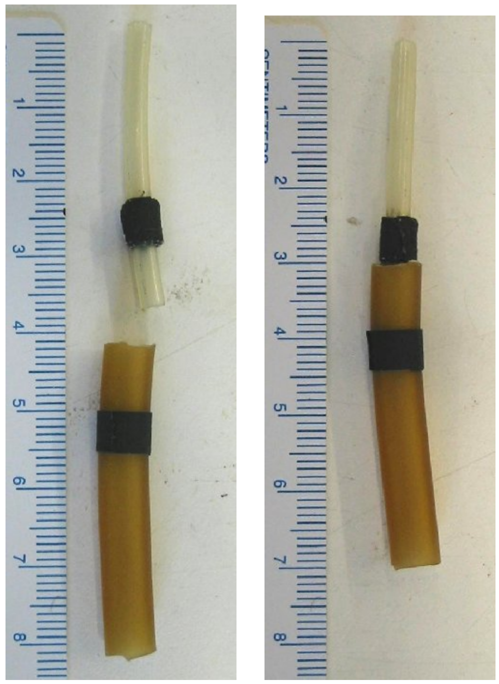


Figure 6.5: Tubes unmated and mated. Larger, compliant tube is the lower of the two. Grip tape bands are used to increase frinction between the tubes and the grasper jaws, as well as provide a mechanical stop.

each subject receiving the same presentation order.

6.2.5 Measures

All forces encountered by the instrument tip, commanded positions, commanded orientations, and trial times were recorded during each experiment at 1 kHz. Each trial was broken down into two subtasks for analysis: the mating subtask and the insertion subtask (Fig. 6.6). During the mating subtask, subjects attempted to position the tip of the smaller tube within the inner diameter of the second tube. Subjects were not required to match the axial orientation to complete this task. Once one part of the smaller tube is within the larger tube, the subject attempts to complete the task by matching the axial orientations of the two tubes and applying the necessary joining force. Note that subjects were not made aware of this task breakdown during the experiment. The trials were separated into these two subtasks because pilot studies revealed that the duration of the mating portion of the task varied significantly between trials. If subjects failed to match the tubes together in their initial attempt, repositioning the two tubes in the depth dimension took a variable length of time. The subtask boundary was denoted by the experimenter during the trial by pressing a key when the mating task was completed. The time of the key press was recorded along with the kinematic and force data.

Two outcome measures were examined for each subtask to characterize the performance of a subject: the RMS force applied during the subtask and the time required to complete the subtask. The RMS force was calculated as the total RMS force on both hands.

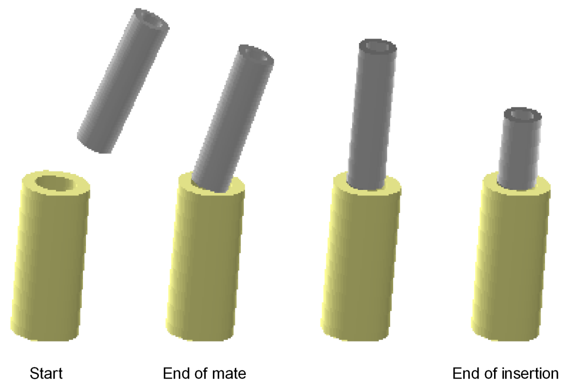


Figure 6.6: Cannula insertion task showing breakdown between the mating subtask and the insertion subtask. Perspective shown is the same perspective seen by subjects, with the major axes of the tubes aligned vertically in the camera view plane.

6.2.6 Statistical Analysis

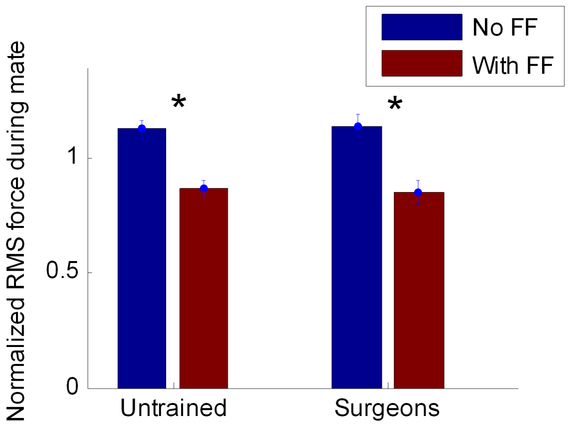
We used a repeated measures ANOVA with a within-subject variable of force feedback condition to test for significant differences in total RMS force and time for task completion. Each subtask was analyzed separately. Because a wide variation in untrained subject performance was observed in pilot studies, all subtask variables were normalized to each subject's mean across both force feedback conditions to remove inter-subject variation. We used the SPSS statistical analysis software (Version 13.0, SPSS Inc., Chicago, Ill.) for all statistical tests. A p-value of less than 0.05 was considered statistically significant.

6.3 Results

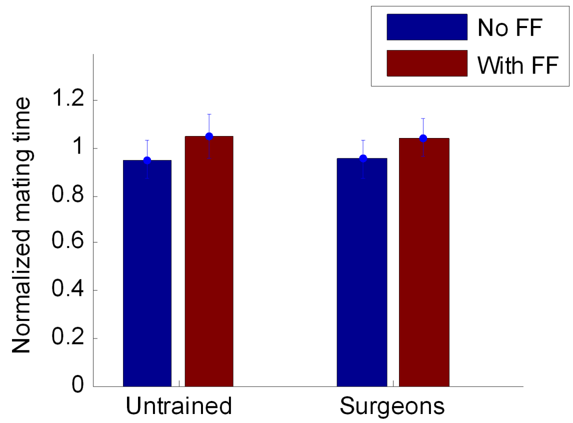
During the mating subtask, the addition of force feedback caused a reduction of RMS force from 0.51 N to 0.39 N in the untrained population, and 0.66 N to 0.49 N in the surgeon population. Both reductions were significant (untrained: $F(1,5) = 55.68$, $p < 0.002$; surgeons: $F(1,5) = 109.28$, $p < 0.001$) (Fig 6.7a). Variation in RMS force between subjects is less than 20% of the group mean force (across both force feedback conditions), with an average 95% confidence interval of +/- 0.09 N (Fig. 6.8).

Untrained subject mating times ranged from 10 seconds to 30 seconds, with a group mean (across both force feedback conditions) of 17.5 seconds. Surgeons completed the mating subtask with a group mean of 12.0 seconds, with average completion times from 4 seconds to 22 seconds.The addition of force feedback caused no significant change in the time required to complete the mating subtask for either population ($F(1,5) = 0.651$, $p = 0.46$; $F(1,5) = 0.587$, $p = 0.478$) (Fig 6.7b). Variation between subjects was high in both populations (Fig 6.9).

With force feedback present during the insertion subtask, RMS force dropped from 0.95 N to 0.74 N in the untrained population, and from 1.40 N to 1.05 N in the surgeon population.



(a) RMS Force



(b) Completion time

Figure 6.7: Performance metrics for mating subtask. Error bars show standard error. Asterisks denote significant difference between force feedback conditions.

This reduction in force due to force feedback was significant for both populations (untrained: $F(1{,}5) = 15.5$, $p < 0.02$; surgeons: $F(1{,}5) = 14.64$, $p < 0.015$) (Fig 6.10a). Variation in RMS force between subjects was higher than during the mating subtask, with an average 95% confidence interval of +/- 0.2 N (Fig. 6.11).

Mean untrained subject insertion times without force feedback ranged from 5 seconds to 22 seconds, with a group mean of 11.2 seconds. With the addition of force feedback, the group mean increased to 24.3 seconds, and individual mean completion times ranged from 9 seconds to 65 seconds. Surgeons completed the insertion subtask with a group mean of 4.2 seconds, with no individual mean completion time above 7 seconds. The increase in time required for the untrained subjects to complete the insertion subtask was significant ($F(1{,}5) = 9.25$, $p < 0.03$) (Fig 6.10b) while the increase in completion time observed for the surgeon population was not ($F(1{,}5) = 0.852$, $p = 0.398$). Variation between subjects was much higher in the untrained versus the surgeon population (Fig. 6.12).

6.4 Discussion

In this experiment we tested the hypothesis that the addition of force feedback to a cannula insertion task can improve performance. Cannula insertion represents a surgical task in which the surgeon must control both position and orientation. A main result of our experiment is that force feedback decreases applied force for both untrained and surgically trained subjects. Our results further suggest that the effect of force feedback on completion time depends on training. The presence of force feedback allowed untrained subjects to carry out the insertion task with lower applied forces but with a longer task completion time. In contrast, surgeons trained in minimally invasive techniques are able to reduce applied forces in the presence of force feedback without a significant decrease in task duration.

The goal of this study was two-fold: 1) to address whether force feedback provides a performance benefit during a cannulation task and 2) to understand the how force feedback

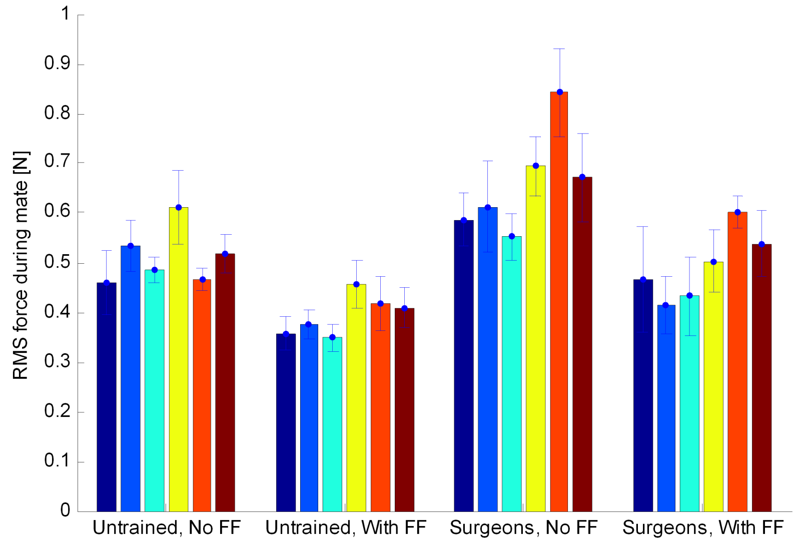


Figure 6.8: Absolute force across subjects for mating subtask by training and force feedback presence. Error bars show standard error.

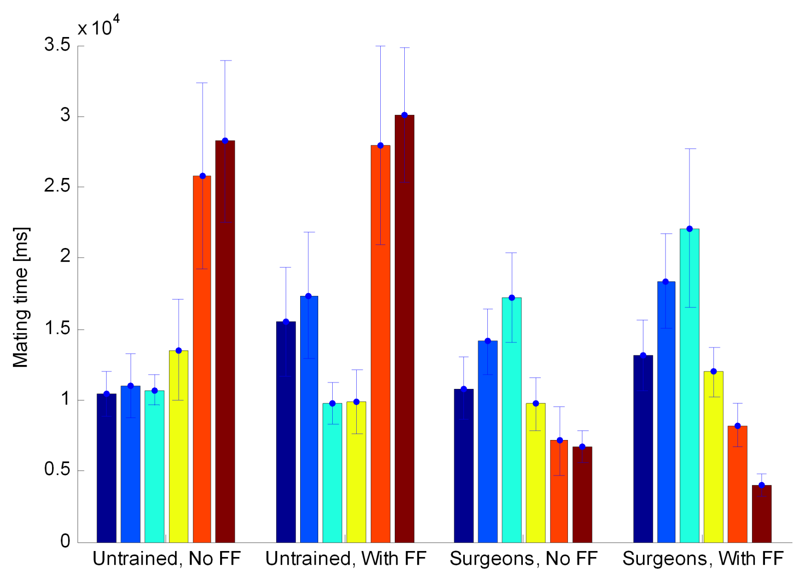
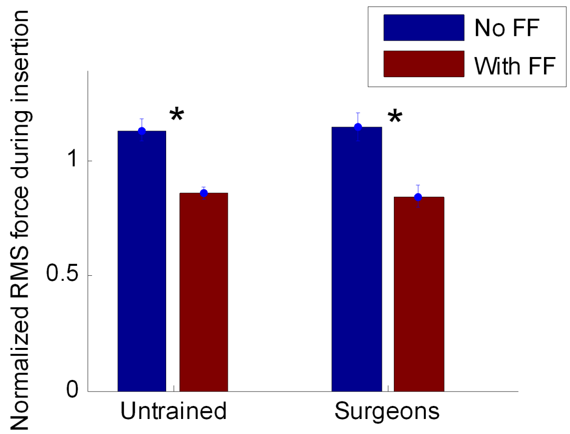
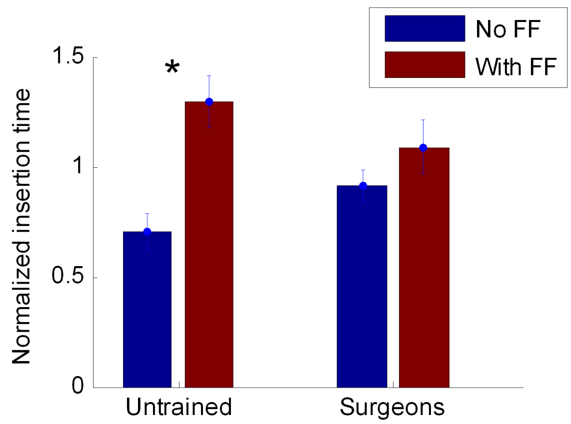


Figure 6.9: Absolute time across subjects for mating subtask by training and force feedback presence. Error bars show standard error.



(a) RMS Force



(b) Completion time

Figure 6.10: Performance metrics for insertion subtask. Error bars show standard error. Asterisks denote significant difference between force feedback conditions.

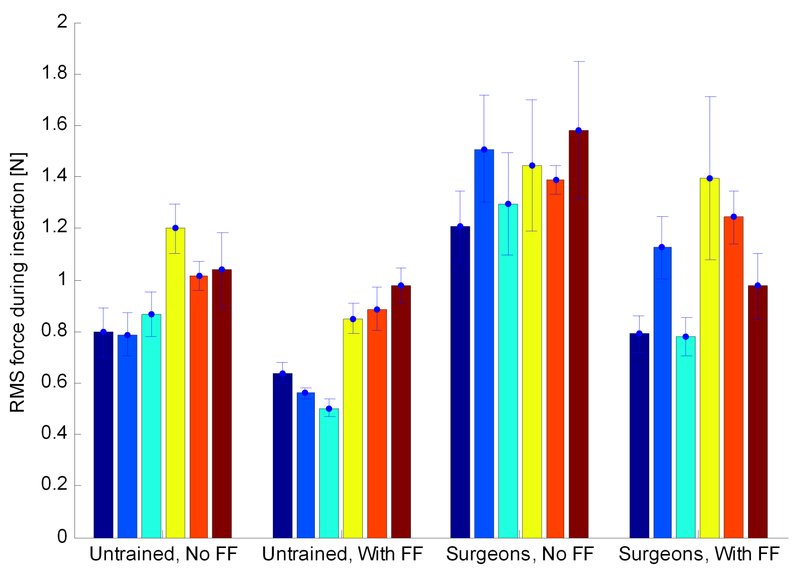


Figure 6.11: Absolute force across subjects for insertion subtask by training and force feedback presence. Error bars show standard error.

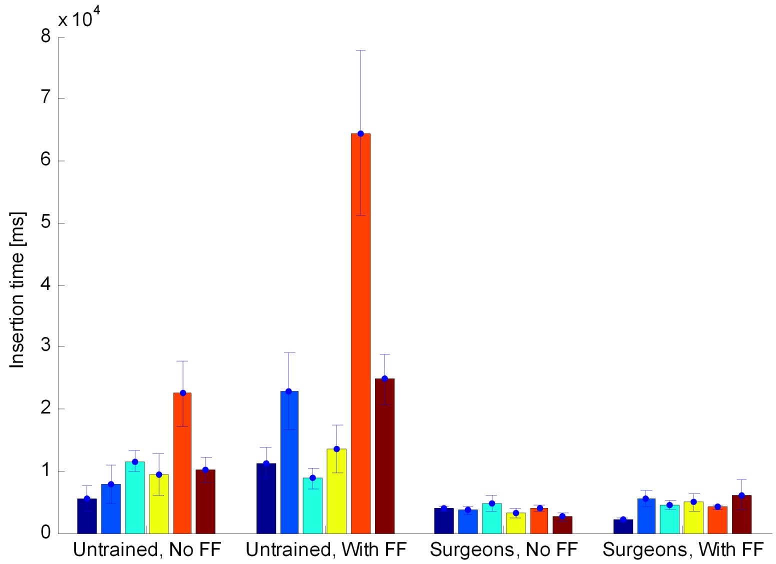


Figure 6.12: Absolute time across subjects for insertion subtask by training and force feedback presence. Error bars show standard error.

provides that benefit. A key factor of the cannulation task was that forces did not always "push" in a direction that led to a successful outcome. The cannulation task was chosen because this property illuminates the role of force feedback as an information source. If the two tubes were mated and the subjects attempted to join them, significant interaction forces were only generated if the tube alignments were mismatched. In the presence of force feedback, subjects would feel the interaction force, signalling an incorrect alignment. If the tube deformations are large enough, a subject could derive this information from the visual field. Presumably this information can be obtained through lower applied forces in the presence of force feedback (Chapter 4).

Force feedback provides a variable amount of information depending the type of alignment mismatch. If there is only an orientation mismatch, force feedback only provides a binary quantity of information: correct or incorrect alignment. Further, this orientation information can only be elicited after a joining attempt. If only the position is incorrect (the tubes are mated and aligned, but the axes are displaced), then the force feedback provides a vector quantity of information, with the force pointing in a direction restoring the smaller PVC tube to the center. The common case was when both the position and orientation were misaligned. In this case, the force feedback provided a sum of the above two information sources.

One key distinction in the analysis of force feedback benefit is the difference between added information versus passive physical constraints that arise when interacting with the environment (Chapter 2). In this cannulation task, do interaction forces serve to aid task performance? The interaction forces push in a direction to minimize forces. With three axis force sensing and recreation, force feedback doesn't constrain the two tubes to match position and orientation. Assuming there is some compliance in the hand, the forces may push the tube to a position that causes lower overall forces, but will never correct the orientation. Thus, there is no immediate physical constraint benefit of force feedback during a cannulation task, except to reduce applied forces.

6.4.1 The Effect of Training

We hypothesize that the training effects observed in this study are due to interactions between the force feedback information source and the visual information source. Because surgeons are trained to overcome the lack of depth perception [105] and hand-eye mismatch [11] encountered in minimally invasive surgery, they can readily integrate additional information from forces into their spatial model. Untrained subjects might lack a sufficient spatial model. Consequently, even though the force information reveals an orientation mismatch, they do not know what motions will correct the orientation. This hypothesis is supported by the known difficulty associated with laparoscopic procedures [4]. Deriving a sufficient spatial model is challenging, as demonstrated by the performance increases observed when using a 3D display versus 2D [119, 52, 30], and the continuing work with artificial enhancement of depth perception [87, 113].

Training is also likely to reduce performance variation between subjects. Performance within the trained surgeon group was consistent, while there was wide variation in the untrained subject group, particularly in the insertion time. This result can also be explained by visuospatial ability. Previous studies have shown that inherent spatial ability correlates well with performance in minimally invasive tasks [43] and there is a wide variation in natural spatial ability [32] among untrained subjects. With training, innate spatial ability ceases to correlate with performance [62], potentially explaining the consistent performance among surgeons.

Our pilot studies during the process of refining the current experimental design emphasized the importance of spatial cognition in untrained subjects. We had to make several revisions to the cannulation task before the untrained subjects were able to successfully complete the task within 60 seconds. One primary change was in camera position and orientation. To make the task easy enough for untrained subjects, the camera was positioned so that the relative orientation between the user and the video monitor was the same as that between the camera and the graspers. This provided an intuitive visual field, because hand motions mapped naturally to observed motions. Consequently, subjects required less training time [130, 121]. We further reduced the required learning time by providing a higher resolution camera, which presumably improved visual feedback quality because of increased visual information. Finally, learning time was also reduced when we increased the length of the tubes so that they occupied a greater fraction of the visual field. This likely facilitated the orientation component of the task by increasing depth cues based on projected areas on the screen. All of the above revisions were present in the final experiment, for both untrained and surgically trained subjects.

Few studies have directly addressed the role of training in surgery with respect to performance with force feedback. Kazi demonstrated a similar reduction in applied force for a single handed telemanipulated cannulation task, but did not address whether subject training influenced the benefit of force feedback [60]. Force feedback has also provided a performance benefit in the related task of peg in hole insertion in a stiff environment (e.g.

[126]). A study of a non-surgical visuo-motor task revealed that training results in a performance increase as subjects learn to incorporate the presence of forces [51]. Nonetheless, a previous study of a blunt dissection task (Chapter 2) showed little difference in the qualitative performance benefit from force feedback among subjects with differing surgical training. This task, however, did not require precise orientation control. Thus, the results of the current study suggest that, although force feedback can improve surgical performance, certain tasks require that the surgeon have prior experience to fully benefit from the presence of force feedback.

Alternative explanations exist for the difference between subject groups. The surgeons consistently applied more force both with and with out force feedback than the corresponding untrained group. This is consistent with surgeon performance in a blunt dissection task (Chapter 2). Surgeons may have completed the insertion subtask in less time simply because they more often exceeded the force threshold necessary to overcome friction. While this certainly contributed to the training effect, it does not entirely explain the difference between groups. Two of the surgeons applied lower forces than three of the untrained subjects, yet still did not require a longer time to complete the subtask in the force feedback condition. In contrast, all three untrained subjects that applied higher forces than these surgeons required more time to complete the subtask with force feedback. Consequently, it appears that there are at least two separate training effects that influence the ability to benefit from force feedback in surgery: 1) familiarity with the force levels required to complete the task, and 2) ability to interpret three dimensional force cues based on a two dimensional visual field.

One effect of the system that could have exacerbated the difference between the two groups was the presence of hysteresis in the robot. Any direction changes would have added a disproportionate amount of motion at the interface, as compared to direct manipulation. If surgeons had a more accurate spatial model, fewer misalignments would occur, requiring fewer direction changes. The presence of hysteresis was perceivable by all subjects.

6.4.2 Application To Surgery

Surgeons were informally questionned after completion of the experiment. All surgeons responded positively regarding the force feedback, reporting that the presence of the force feedback aided performance. The opinions on how it helped, however, varied between surgeons. Some felt that the forces assisted only with the mating portion of the trials, giving quick feedback when the two tubes were near in depth, or feeling the inner edge of the tube, thus knowing the tube were almost mated. Those subjects did not report feeling any benefit of force feedback during the insertion subtask because the force feedback did not help with orientation. On the other hand, some surgeons reported no benefit of force feedback during the mating subtask, but instead reported using the force feedback during the insertion phase to determine whether the tubes were aligned based on a change in compliance. Use of compliance as an information source (such as tugging on the tubes to determine connectivity) reveals that the information content of force feedback is not limited to high frequencies [23]. Some surgeons reported that they could sense when they were pushing the tubes together properly because of the sensation of friction. This apparently provided an informational benefit about proper force direction. Finally, most subjects recognized that force feedback caused them to be more careful because they were aware of the force magnitude.

The above observations illuminate where the addition of torque feedback would be useful. We have discussed ways that three-axis force feedback provides information, and torque feedback can provide similar informational benefits. Additionally, torque feedback is necessary to provide a passive constraint benefit in orientation. Example applications include needle passing (matching orientations between two graspers) and suturing (passively restoring needle orientation once the needle is partially driven). Torque feedback can aid in catheterization training [63]. Naturally, the addition of torque feedback to a telemanipulation system would be associated with a significant increase in system cost and complexity.

A final observation as to the benefit of three axis force feedback in a cannulation task is that, even without torque feedback, force feedback does enable passive strategies that would not be possible with only force information (such as with sensory substitution). Knowing that there is a force threshold that needs to be exceeded for the two tubes to join, a subject can consistently apply that force and vary the position and orientation until the join is successful. This is opposed to the "guess and check" strategy employed by many of the untrained subjects, where they would attempt a join, feel the force rise (implying the position or orientation was incorrect), adjust, then reattempt. Using the passive strategy turns one of the degrees of freedom of the task (positioning along the major axis of the tubes) from a mentally intensive position control task into a relatively easy force control task.

The effect of training on force feedback benefit suggests further investigation. While we have observed a significant effect of training in this study, the trained subjects were familiar with endoscopic camera views, not with force feedback enabled surgical telemanipulators. Since no commercial robots feature force feedback, no surgeons are trained for this aspect of the task. Training is an important factor because it relates to the mental workload of the surgeon. Presumably, reducing the workload of the surgeon improves surgical success and patient safety. While some research suggests that force feedback can reduce mental workload [90, 19, 76], only non-surgical studies have been conducted. Consequently, further investigation of the tradeoffs between force feedback, visual feedback, training, and mental workload is clearly warranted [132].

Chapter 7

Conclusions And Future Work

Force feedback is a compelling feature of surgical robots. Surgeons desire the feature because they normally use their sense of touch in open and laparoscopic procedures. The reduced sense of force perception in laparoscopic surgery has been correlated with an increase in surgical errors. Further, force feedback in telemanipulator systems outside of surgery has been shown to improve performance. Adding force feedback to a telemanipulator, however, is difficult. Interaction forces need to be sensed in the constrained environment of surgery, then recreated against the surgeon's hands in an intuitive manner. The lack of force feedback in robotic surgical systems should be the result of a cost/benefit analysis. While the cost is known to be high, force feedback, though compelling, provides an unknown benefit. This dissertation assesses the benefit of force feedback in surgery and addresses mechanisms for minimizing the cost of incorporating it into telemanipulation systems.

7.1 The Benefit of Force Feedback In Surgery

Force feedback reduces applied forces in surgery. This benefit is strong and consistent, demonstrated across a number of different tasks. Every experiment presented in this dissertation has shown this reduction in forces when force feedback is present. The benefit also exists across all levels of surgical training. Force reduction in surgery is important because forces correlate with trauma. Larger forces increase the likelihood of tissue damage, decreasing patient safety. Exceeding force thresholds lead to patient injury across a variety of surgical tasks, from the accidental transection of an artery, as was simulated in Chapter 2, to the overexertion of force when tightening a suture.

Other benefits of force feedback are context dependent. These benefits not only reduce collateral patient injury, but help achieve a desired outcome. These benefits can be summarized in the context of the mechanisms by which force feedback provides a benefit, discussed below.

7.2 Physical Constraint Versus Information

This dissertation not only demonstrates the benefits of force feedback in surgery, but also reveals the mechanisms for achieving these benefits. We have shown that force feedback provides a benefit in two ways: through a passive mechanical constraint and by providing information to the user. We have demonstrated that both of these mechanisms can help reduce applied force. Understanding these mechanisms also provides a framework for examining other potential benefits of force feedback.

Force feedback can provide passive benefits by turning environmental interaction forces into constraint forces. Physical constraints passively reduce intrusions into environmental structures (and, thereby, forces) due to the interactions of the compliance of the hand and the stiffness in the environment (interacting through the telemanipulator). Because this benefit is completely passive, it can happen without a cognitive response from the user. Accordingly, these benefits occur instantaneously, on the time scale of mechanical interactions. Nonetheless, surgeons only benefit from passive constraint mechanisms during tasks that involve the appropriate mechanical interactions. For a physical constraint benefit, the interaction forces in the task must push the surgeon towards successful task completion, as in the case of minimizing intrusion into a tissue. Any surgical manipulation near sensitive tissues require minimization of intrusion and force into those tissues, suggesting that the benefit of force feedback in surgery could be widespread. Examples include minimizing trauma to the bile duct during a cholocystectomy to prevent bile leakage, or minimizing damage to the internal mammary artery during a coronary artery bypass grafting. A number of other tasks are also likely to involve reaction forces that could serve in this physical constraint benefit of force feedback, such as knot tying or suture tensioning.

The presence of physical constraint forces allows for an additional set of passive manipulation strategies in surgery. Surgeons using force feedback can take advantage of the constraints that exist in the environment to turn mentally intensive positioning tasks into relatively easy force control tasks. An example of how this happens in normal interactions is when attempting to draw a straight line using pen and paper. Drawing a line perfectly straight freehand is difficult. With a ruler, however, the task of drawing a straight line is trivial: simply press the pen up against the ruler and draw. Instead of maintaining the correct position at every point in time, the task is now to apply a positive force toward the ruler. Similar situations occur in surgery, such as during dissection of the gall bladder from the liver, where the stiffness of the organs restricts the motions to lie in the connective tissue plane. Again, these interaction forces for physical constraint must exist in the environment; in this case it arises from the differing tissue stiffnesses. Physical constraint forces can also minimize the degrees of freedom of the surgeon's positioning task. An example is the cannulation task (Chapter 6). With three axis force feedback, users could turn one degree of freedom of the task, joining the two tubes axially, into a simple force control task. By exceeding the force threshold of friction when the tubes were aligned, the users could concentrate on trying a number of different positions and orientations, relying on the mechanics to join the tubes when alignment happened. Note that this manipulation strategy could not happen with sensory substitution or any other information transmission scheme because it relies on the mechanical interaction between the hand and environment.

The surgeon can maximize the passive benefit of force feedback by adjusting hand compliance. Force feedback reproduces the compliance of the surgeon's hands on the instrument tips. This compliance can interact with the mechanics of the environment to achieve a desired outcome. Note that this is something we do naturally in other arenas, such as picking up an object. When reaching to pick up a cup, for instance, we maintain a low compliance of our hands so that small errors in our motion don't cause large interaction forces, potentially knocking over the cup. Another way of looking at the benefit of changing hand compliance is to first classify surgical subtasks as either position control tasks or force control tasks. For instance, needle driving is a position control task, where the task is to cause the needle to follow a certain path. Retraction can be a force control task, where tissues are pulled back to a maximum position without exceeding the force that would cause a tissue tear. Using force feedback, surgeons can address position control tasks as position control tasks by stiffening the hand, and force control as force control directly. Because surgery is made up of many different types of tasks, force feedback may be important in allowing surgeons to use the most appropriate control strategy for a given task.

Because the benefit of physical constraint is mechanical, the benefit can be analyzed using a mechanical model. This model can be used to predict the benefit of force feedback in a certain environment. A mechanical model allows analysis of the tradeoffs between motion speed, applied force, hand compliance, environmental stiffness, intrusion depth, force feedback gain, and reaction time. This has applications to a number of areas of surgical robotics, not simply task based interaction. One area is teleoperator design, of both the controller and the interface. The effect of compliances or delays introduced by the interface and teleoperator controller can be quantified using a mechanical model. Finally, because the physical constraint benefit can be modeled and accounted for, we can use mechanical models to distinguish physical constraint and informational benefits of force feedback in various tasks.

The hallmark of an information benefit of force feedback is that the feedback needs to be consciously perceived then reacted to. Because of this reaction time, the benefit of information cannot happen instantaneously. The surgeon becomes aware of the force levels during the task, and consequently, applies lower forces.

Force feedback as information also assists performance in a number of other ways. For instance, if the task is a force application task, the presence of force feedback increases available information related to task performance. Examples of this include the force application task in Chapter 4, or the task of resection. Information from force feedback is task dependent, and widely varied. For example, surgeons use the vibration of contact to elicit positional cues when passing needles behind visual occlusions. In actual US guided ASD repairs, surgeons distinguish the patch from the heart wall by scraping the instrument and feeling the differing vibrations.

Similar to physical constraint force feedback, a surgeon can gain greater information benefit from force feedback through changes in hand compliance. By reducing the compliance of the hand, for instance, the surgeon can gain contact or vibration information with little force application. In many surgical cases, visual feedback can provide some information about tissue compliance because soft tissues exhibit large, visible deformations at low forces. However, when visual information is limited (such as when using low quality visualization techniques, when movements are primarily in the depth axis, or when tissues are stiff), force feedback might prove to be especially useful, providing information that cannot be gained otherwise.

We have shown that the distinction between information benefit and physical constraint provides a nice framework for analyzing the benefits of force feedback in surgery. This same distinction can also be used to help predict the benefit of force feedback in specific or future surgical tasks. For instance, the physical constraint benefit will only exist if there are environmental interaction forces that aid in task completion. Force information can be useful in tasks where other information is limited, such as depth cues in endoscopically guided surgery. This distinction can aid in the cost benefit analysis of force feedback in surgery, or point to specific tasks where force feedback will provide the most benefit.

7.3 Training and Mental Workload

One important result of this dissertation is that the benefit of force feedback in surgery can depend on training. In the cannulation experiment, force feedback reduced forces, but increased trial time for untrained surgeons. An explanation for this result is that surgeons can easily integrate the cues given by force feedback into a spatial model of their environment to complete the task. Untrained subjects, who have less experience with developing spatial models, cannot easily transform additional spatial information given by force feedback into a performance benefit.

These observations about training shed some light on the hypothesis that force feedback can decrease mental workload. One compelling reason for force feedback is that force interaction is a natural information source, so might be able to provide a benefit without increasing mental workload. However, incorporating force feedback with other sources of information may require training. This might be particularly challenging when the visual field requires the surgeon to perform a relatively complex mapping. Consequently, force feedback reduces mental workload only when other information sources become intuitively processed. Further work is needed to investigate the tradeoff between training, integrating sources of information, and mental workload.

Use of force feedback as a physical constraint can certainly reduce mental workload. Force feedback allows surgeons to take advantage of environmental constraints, using passive strategies to convert difficult position control tasks to intuitive force control tasks. Naturally, training remains an issue because taking advantage of some physical constraints requires incorporation of additional information. Force example, in the cannulation task with force feedback, some surgeons applied a given force, then changed positions and orientations until the tubes joined. This required an understanding of the force required to join the tubes, plus a spatial model for controlling the tube positioning and orientation. No untrained subjects employed this strategy, potentially because they lacked the spatial model necessary to appropriately position and orient the tube based on the available visual field.

We've observed that force feedback depends on training, but no research has been conducted with surgeons who have trained with a force feedback enabled robot. The mental workload benefit may only be realized when surgeons trained with force feedback are compared against surgeons trained without force feedback.

7.4 Future Work

While we have examined both sides of the cost benefit analysis of force feedback in surgery, a resolution has yet to be reached. Force sensing technology continues to be prohibitively costly, along with the associated increase in cost and complexity of the surgical robot. We have demonstrated that force feedback can provide consistent and predictable benefits, but transforming these into direct improvement of patient care and surgical success is difficult. Nonetheless, productive integration of force feedback into surgery should become easier with a continued reduction of cost of force sensing technology and use of the analytical framework developed in this dissertation to assess the physical constraint and informational benefits of force feedback.

One aspect of force technology that remains unexamined is torque sensation and reproduction. The addition of torque feedback would definitely add to the cost of telemanipulation systems However, we also predict important potential benefits based on our analysis. Torque feedback could provide an important benefit by allowing surgeons to use physical orientation constraints. In tasks such as needle passing and cannulation, part of the task requires matching orientations. Furthermore, in both of these tasks, environmental interaction torques assist in this alignment Consequently, addition of torque feedback could further reduce difficult positioning and orientation tasks into simpler force and torque control tasks.

The task of suturing is an interesting case for force and torque feedback. Suturing is nominally a positioning task, where the surgeon must follow a certain path with the curved needle to join two tissue planes with a suture. This task should be carried out with a minimum of applied forces, so the force reduction benefit of force feedback is likely to improve performance. With only three axis force feedback, however, the issue is complex because interaction forces will serve to adjust the position of the needle, not the orientation. Torque feedback could help maintain orientation once the needle is partly driven, because interaction forces and torques serve to keep the needle pointed along a curved path. Finally, other work has shown that proper regulation of force is necessary for successful knot tying when suturing. Force feedback is hypothesized to be useful here, since little visual deformation is likely during knot tying. Thus, force and torque feedback could assist suturing in a number of ways. However, torques dominate the tissue interaction during suturing. Consequently, investigation of force feedback in suturing is difficult from a technical standpoint. A three axis force sensor used in this task must be very insensitive to moments, otherwise the force feedback is incorrect and confusing.

In conclusion, although we have found that the interaction between force feedback benefit, training, and mental workload is complex, we have found some specific benefits and have established a framework for future analysis. Some of these issues will continue to be difficult to resolve due to mental workload being notoriously difficult to measure in a satisfying way. Further, without surgeons trained in the use of surgical systems that have force feedback, the true performance of force feedback in surgery will remain in the realm of hypothesis. However, the work presented here demonstrates how force feedback provides a consistent benefit, and future work can build upon the intuitive strategies enabled by force feedback.

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Appendix A

Deriving The Commanded Hand Trajectory

Restating the force balance for the hand/stylus model in contact with a compliant environment (3.3) yields

$$k_{hand}(x_d(t) - x_a(t)) + b_{hand}(\dot{x}_d(t) - \dot{x}_a(t)) - m_{hand}\ddot{x}_a(t) = F_{wall}(t), \tag{A.1}$$

where k_{hand}, b_{hand}, m_{hand} are the parameters of the second order hand/stylus model, $x_d(t)$ is the desired hand motion from the central nervous system, $x_a(t)$ is the observed trajectory, and

$$F_{wall}(t) = k_e x_a(t) \tag{A.2}$$

is the wall force when in contact with the compliant enviroment, with k_e being the lumped stiffness of the environment and the force feedback gain (see 3.1). Taking the Laplace transform of A.1 is then

$$\begin{aligned} &k_{hand}(X_d(s) - X_a(s)) \\ &+b_{hand}(sX_d(s) - x_d(0) - sX_a(s) + x_a(0)) \\ &-m_{hand}(s^2 X_a(s) - sx_a(0) - \dot{x}_a(0)) = k_e X_a(s). \end{aligned} \tag{A.3}$$

Knowing the initial conditions $x_a(0) = 0$ because we define the wall's position to be at 0, and $x_d(0) = 0$ because we assume the system is in steady state before the wall, we can solve for $X_d(s)$,

$$X_d(s) = X_a(s)\left(\frac{k_h + k_e + b_{hand}s + m_{hand}s^2}{k_h + b_{hand}s}\right) - \frac{m_{hand}\dot{x}_a(0)}{k_h + b_{hand}s}. \tag{A.4}$$

We can then rewrite $X_d(s)$ as

$$X_d(s) = X_a(s)\left(\frac{1}{H(s)}\right) - G(s), \tag{A.5}$$

where

$$H(s) = \frac{b_{hand}}{m_{hand}}\left(\frac{s + \alpha}{s^2 + 2\zeta\omega_n s + \omega_n^2}\right), \tag{A.6}$$

$$\omega_n = \sqrt{\frac{k_h + k_e}{m_{hand}}}, \zeta = \frac{b_{hand}}{2m_{hand}\omega_n}, \alpha = \frac{k_{hand}}{b_{hand}}, \tag{A.7}$$

$$G(s) = \frac{m_{hand}\dot{x}_a(0)}{k_h + b_{hand}s}. \tag{A.8}$$

Solving for $h(t)$ by taking the inverse Laplace transform of A.6

$$h(t) = \beta e^{-\zeta\omega_n t} \sin\left[\omega_n\sqrt{1-\zeta^2 t} + \arctan\left(\frac{\omega_n\sqrt{1-\zeta^2}}{\alpha - \zeta\omega_n}\right)\right], \tag{A.9}$$

where

$$\beta = \frac{b_{hand}}{m_{hand}\omega_n}\sqrt{\frac{\alpha^2 - 2\alpha\zeta\omega_n + \omega_n^2}{1-\zeta^2}}. \tag{A.10}$$

Similarly, solving for $g(t)$,

$$g(t) = \frac{m_{hand}}{b_{hand}}\dot{x}_a(0)e^{\frac{-k_h}{b_{hand}}t}. \tag{A.11}$$

Because division in the frequency domain is the same as deconvolution in the time domain,

$$x_d(t) = \mathrm{deconv}(x_a(t), h(t)) - g(t). \tag{A.12}$$