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DESIGN, ANALYSIS AND TESTING OF HAPTIC FEEDBACK SYSTEM FOR LAPAROSCOPIC GRASPERS IN IN VIVO SURGICAL ROBOTS

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DESIGN, ANALYSIS AND TESTING OF HAPTIC FEEDBACK SYSTEM FOR LAPAROSCOPIC GRASPERS IN *IN VIVO* SURGICAL ROBOTS

by

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A THESIS

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Laparo-Endoscopic Single Site (LESS) Robotics Surgery is an advanced technology in the field of Minimally Invasive Surgery (MIS). The LESS surgical robots significantly improve the surgeon’s accuracy, dexterity and visualization, and reduce the invasiveness of surgical procedure results in faster recovery time and improved cosmetic results. In a standard robotic endosurgery, the palpation of tissues is performed by laparoscopic graspers located at the end effectors. The master-slave configuration in robotic surgery leads in remote access to the operation site. Therefore, surgeon’s ability to perceive valuable sensory information is severely diminished. Sensory information such as haptics, which is essential for safe tissue and organ palpation, is not possible due to absence of direct access. Therefore, unknowingly excessive grasping forces are exerted by the laparoscopic graspers could lead to tissue trauma and vital tissues and organs damage.

This thesis presents the several aspects of haptic feedback system including a design and analysis of force sensing forearm to measure grasping forces exerting on tissues during palpation tasks, 4-CH bilateral teleoperated Impedance-Impedance based robotic control architecture for haptics and design and developments of Surgical Haptics-User Interface Devices (H-UID). The entire haptics feedback system has been implemented in miniature in vivo surgical robots and tested in animal surgeries at University of Nebraska Medical Center (UNMC). The results of bench-tops and animal surgeries in the presence of haptics were discussed. The haptic feedback system has established the ability to differentiate the different objects of different stiffness, provide appropriate grasping force control and reduce tissues palpation time as a result improve performance of the surgeon.
This thesis is dedicated to
my father, Mr. Charuhas Salvi and Mother, Mrs. Jyotsna Salvi
for their endless support and encouragement.

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Chapter 1: Introduction

Prior to 1990, invasive procedure was widely used as an abdominal surgical procedure. In an invasive procedure, a long incision is prepared for the surgeon to insert his/her hand, instruments in an abdominal cavity and visualizes surgery through incision. The advantage of the invasive procedure is that it allows the surgeon to get the direct access to organs and examine tissues and organs involved in the surgery and provide excellent visualization, while the disadvantage of the open surgery is larger incisions which lead to increase in trauma and longer recovery time of the patient.

Due to the limitations of invasive surgery, Minimally Invasive Surgery (MIS) was introduced. MIS allows a surgeon to manipulate tissues inside a patient using long, slender laparoscopic tools inserted through small incisions. MIS has numerous advantages over open surgery: minimizing pain, tissue trauma and recovery time. A major disadvantage of MIS is that the surgeon uses laparoscopic tools and his/her hands are not directly accessible to tissues and organs. This results in lack of ability to accurately sense the palpation forces which are being exerted on tissues. Surgeons have often reported that the subsequent loss of tactile and kinesthetic information makes many laparoscopic surgical procedures more challenging [1].

The modern era of laparoscopic robotic surgery started when a master-slave combination is implemented in DaVinci system [2]. The various operations are
performed by a laparoscopic approach including: Cholecystectomy, Nephrectomy, Appendectomy, Bowel Resection, Inguinal Hernia Repair and Feeding Tube Placement. There are many advantages of Minimally Invasive Robotic Surgery (endosurgery) over Minimally Invasive Laparoscopic Surgery due to improved accuracy, dexterity and visualization, and reduce invasiveness of surgical procedure results in faster recovery time, less pain, trauma and blood loss.

In robotic endosurgery, the tissue palpation is performed by laparoscopic graspers located at the end effectors. Most of the haptic sensations which are available in open surgery are lost in Laparo-Endoscopic Single Site (LESS) robotic surgery e.g. kinesthetic and tactile sensing, which is essential for safe tissue and organ palpation. This is due to remote access and lack of direct touch to the operation site [3].

During exploration and palpation in MIS tasks, obtaining a tactile and kinesthetic (force) information of tissues and organs is described as a haptics in laparoscopic surgery. Tactile information refers to sensation felt through mechanoreceptors in the skin, including hardness, temperature, pressure, shape where the kinesthetic perception includes discernment of exerting force on tissues.

Many surgical robotics researchers have proposed that a crucial source of information such as haptics assist a surgeon to perform safe MIS procedures as well as preventing tissue damages by applying more forces than necessary [4]. The lack of sensory information diminishes the tactile sensation, making it more challenging for the surgeon to feel tissues and perform delicate operations such as tying sutures. Researchers have discovered that haptic feedback is very beneficial in MIS procedure such as tissue stiffness characterization [5], [6], blunt dissection [7], and shape and
feature identification through palpation [8]. The nature of the MIS procedure requires
the specific haptic information of interest such as tactile feedback for tumor palpation
and force feedback for tying sutures and manipulation of delicate tissues. Most of the
teleoperated MIS procedures become more difficult without haptic information,
leaving surgeons to rely heavily on the visual sense. Manipulation tasks are
particularly difficult to execute without kinesthetic feedback [9].

This thesis presents design, analysis and testing of haptic feedback system for
laparoscopic graspers in in vivo surgical robots. Several versions of force sensors and
surgical Haptic-User Interface Devices were evaluated. Two types of control schemes
of bilateral teleoperation were analyzed and demonstrated that the impedance-
impedance based 4-CH bilateral control architecture is most suitable for the force
feedback enabled laparoscopic graspers application. The complete haptics system has
implemented in in vivo surgical robots developed in Advances Surgical Robotics Labs
at University of Nebraska-Lincoln and tested in animal surgeries at University of
Nebraska Medical Center (UNMC). The results of haptics feedback system during
animal surgeries and bench-tops were discussed.
Chapter 2: Background

Section 2.1. Haptics

Haptics describes as the sense of touch and movements. It is necessary to distinguish between haptics, tactile and kinesthetic feedback. Haptics is a broad term used to describe both tactile (cutaneous) and kinesthetic (force) type of information.

![Figure 2.1. Human Senses and Types of Haptics](image)

The combined sensation of kinesthetic, tactile, thermal and noci reception are known as haptics, which are shown in Figure 2.1. Kinesthetic feedback is defined as the information about the force perception where the tactile feedback, which involves stimulation of cutaneous receptors to distinguish tissue texture and tissue characteristics. Thermal feedback is defined as the information about the temperature
of the skin, muscles and organs, and noci feedback which is based on sensory information received through neurons located in skin.

Section 2.2. The Role of Haptic Feedback System in Robot-assisted LESS Surgery

The robot-assisted surgery improves the ability of surgeons to perform MIS procedures by scaling up/down the motions and adding additional degrees of freedom to the laparoscopic tools for efficacy. Development in the robotic surgery is limited by the unanswered problem which is lack of haptic feedback (in terms of kinesthetic and tactile feedback) to the surgeon.

Most of the surgeons feel that the partial haptic feedback was present in minimally invasive laparoscopic surgery due to the laparoscopic tools but robot-assisted LESS surgery eliminates complete haptic sensation [11]. As these surgical robotic systems are indirectly in contact with the surgical site instrumented tools, surgeons do not receive any force or tactile feedback from controller. It is clearly demonstrated that the sense of touch can provide the primary source of information to the surgeon, which critically guides during surgeries. Current robotic surgical systems equip with endoscopic cameras or stereoscopic cameras can cater an excellent visual feedback but incapable of providing haptic feedback. Therefore, a surgeon cannot perform delicate MIS procedures while relying solely on visual feedback. e.g. A surgeon cannot dissect out the adrenal vein during a laparoscopic adrenalectomy without avulsion it if there is truly no tactile sensation [12].

Haptic feedback can enhance the surgeon’s sense of telepresence, as a result, increases in the performance. Both are essential to create a typical sensations felt by the surgeon’s hand. Improvements in robot-assisted LESS surgery will lead to
reduced trauma and faster recovery time, better patient care and shorter hospital stay as a results lower health care costs.

**Section 2.3. Haptic-Bilateral Teleoperation**

![Figure 2.2. The Flow of Information in the Haptic Feedback based Telesurgical System [62]](image)

Figure 2.2 demonstrated the haptic-bilateral teleoperation, in which the human operator synchronously manipulates and perceives the resulting reaction force through direct force feedback. In other way, the human operator exerts a force to the user interface device (UID), due to which UID displaces and sends signals in terms of displacement to the teleoperator (slave robot), where the slave robot follows same motion as of UID, the resulting force from the interaction between slave and environment is transmitted towards the operator via UID. Bilateral control architecture provides the most natural way of interaction with the remote environment and direct force feedback provides the sense of being exist in the remote environment and improves the ability to perform complex manipulation tasks.
Robot-assisted MIS is based on teleoperation as the surgeon is physically separated from the surgical robot and its workspace. Thus, teleoperation in robotics is a natural tool to extend capabilities in MIS. The surgical instruments can be completely replaced with the surgical robotic systems which can be directly controlled by the surgeon via control systems for teleoperation.

The aim of the telesurgical workstation is to restore the sensory information to the surgeon during manipulation which was lost due to MIS. Therefore, due to the haptic feedback, the fidelity of the manipulation enhances by the force feedback to the master and the loss of tactile sensation is restored by the tactile feedback or visual feedback to the master.

There are multiple research going on in developing a telesurgical systems e.g. endoscopic telepresence system developed by Green, Shah et al [13], the eight-degree-of-freedom teleoperated surgical instrument named the Black Falcon [14] developed by Madhani and Niemeyer in MIT, the telerobotic assistance system [15] developed for systematic procedure to laparoscopic surgery in order to reduce off-load routine tasks, no. of people required in the operation theatre and at the end, improve performance. Another telerobotic surgery experiment was performed by Rovetta et al [16], where they have described about integration of human capabilities with robotics in laparoscopic surgery, requirement of skillset to perform surgery and benefits of robotic application in MIS field. The similar research was conducted in Japan where intravascular neuro-tele-surgery was performed between Nagoya and Tokyo by using a high speed optical fiber network which reduced the time delay between master-slave configurations considerably [17].

At present, the commercial company named Intuitive Surgical Inc. is manufacturing *Da Vinci* surgical system in the field of MIS [18].
There are quite few telesurgical systems are in market due to difficulty comes while developing a miniaturized manipulator design and acquiring high fidelity operations. The essential task is that the telesurgical manipulators need to be as small as possible e.g. 10mm diameter for laparoscopy and 5 mm or smaller for cardiac surgery, as well as produce significant forces and torques to be able to manipulate tissues.

### 2.3.1. 2-Port Network

![Figure 2.3. General 2-Port Model of a Bilateral Teleoperation System](image)

In Figure 2.3, a general 2-port model of a teleoperation system is presented. The task impedance $Z_e$ is transmitted to the user via the 2-port teleoperation system, which means slave forces and velocities are proportional to master forces and velocities. $Z_h$ is defined as the impedance of the user’s hand. The remote task impedance, $Z_e$ is a function of the impedance $Z_{to}$ which is transmitted to the user. The relation of these impedances characterizes the transparency of the teleoperated system.

For complete transparency, the following transparency condition needs to be satisfied:

$$Z_{to} = Z_e$$

(2.1)
Therefore, transparency is a function of the teleoperated system, neither of only the task impedance $Z_{t_o}$ nor the impedance of the hand, $Z_h$.

### 2.3.2. Teleoperation Control Schemes

In general, teleoperation systems are classified according to the devices based on admittance or impedance types, depending on whether they behave like velocity or force sources respectively. This behavior of these devices can be decided by the structural design and actuation utilized by the manipulator.

According to the definition of teleoperation system, an impedance device receives the force command and transmits the same to its environment in response to its measured position [19] e.g. SensAble’s phantom controller developed by Salisbury et al. [20] and 6 DOF magnetically levitated wrist developed by Hollis et al. [21] are the impedance based devices which hold high-back-drivability and low impedance. Instead, an admittance device receives a position/velocity command and transmits a position/velocity to its environment in response to its measured contact forces [19]. e.g. Industrial robot : PUMA robot developed by Clover et al [22], [56] and hydraulic robot : Excavators explained in [23] are admittance based devices and possess low back-drivability and low passivity.

There are 4 types of bilateral teleoperated control schemes are described in [24] are as follows:
Figure 2.4. Control Signal Flow Diagrams for 2-Channel Control Architecture

In Figure 2.4, the paths which are shown in solid lines are the control paths have more significant effect on the system and the paths which are shown in dashed lines have less significant effect on the system.

The LTI equations are used for impedance/admittance types of master and slave manipulators, where $Z_m$, $Z_s$, $Y_m$, $Y_s$ are master and slave dynamics and $F_{cm}$, $F_{cs}$, $V_{cm}$ and $V_{cs}$ represent their control inputs. $Z_m$, $Z_s$ and $Y_m$, $Y_s$ are impedance and admittance dynamics respectively.

$$Z_m V_h = F_h + F_{cm} \quad \text{For Impedance Master} \quad (2.2)$$

$$Z_s V_e = -F_e + F_{cs} \quad \text{For Impedance Slave} \quad (2.3)$$
\[ Y_m \cdot F_h = V_h + V_{cm} \quad \text{For Admittance Master} \quad (2.4) \]

\[ Y_s \cdot F_e = -V_e + V_{cs} \quad \text{For Admittance Slave} \quad (2.5) \]

On the basis of the choice of the network input I and output O variables, Impedance ‘z’, Admittance ‘y’, Hybrid ‘h’ and Inverse Hybrid ‘g’, network matrices are defined by Hannaford et al [19].

\[
\begin{bmatrix}
F_h \\
F_e
\end{bmatrix} = O_z := zI_z =
\begin{bmatrix}
z_{11} & z_{12} \\
z_{21} & z_{22}
\end{bmatrix}
\begin{bmatrix}
V_h \\
V_e
\end{bmatrix}
\quad (2.6)
\]

\[
\begin{bmatrix}
V_h \\
-V_e
\end{bmatrix} = O_y := yI_y =
\begin{bmatrix}
y_{11} & y_{12} \\
y_{21} & y_{22}
\end{bmatrix}
\begin{bmatrix}
F_h \\
F_e
\end{bmatrix}
\quad (2.7)
\]

\[
\begin{bmatrix}
F_h \\
-F_e
\end{bmatrix} = O_h := hI_h =
\begin{bmatrix}
h_{11} & h_{12} \\
h_{21} & h_{22}
\end{bmatrix}
\begin{bmatrix}
V_h \\
F_e
\end{bmatrix}
\quad (2.8)
\]

\[
\begin{bmatrix}
V_h \\
F_e
\end{bmatrix} = O_g := gI_g =
\begin{bmatrix}
g_{11} & g_{12} \\
g_{21} & g_{22}
\end{bmatrix}
\begin{bmatrix}
F_h \\
-V_e
\end{bmatrix}
\quad (2.9)
\]

\textit{Section 2.4. Force Sensor}

To find a solution for lack of force feedback, sensing elements need to be integrated inside graspers mechanism. Size and complexity of such sensing elements based on types of laparoscopic graspers.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{safe_area.png}
\caption{Safe Area Limited for Slip and Damage Forces Occurred Due to Laparoscopic Graspers, De Visser et al [25]}
\end{figure}
Grasping is defined as a pinch and pulls combination. In order to safely grasp tissue, the combination of the pinch and pull force levels applied by the instrument on the tissue should be within a certain safe area which is shown in Figure 2.5. This safe area is different for each instrument and tissue combination as they each have their own mechanical properties.

Many researchers had developed state-of-the art surgical tools to accurately measure the graspers-tissue interaction forces during surgical procedure. The research involved in deriving methods to integrate sensors such as strain gages, force/torque sensors or custom designed sensors into on the shaft or graspers jaws of current laparoscopic tools to measure grasping forces on tissues. Rosen, M. MacFarlane, C. Richards [26] and Bicchi and Canepa [27] incorporated strain gages on the laparoscopic tool shaft that provided indirect grasping force information. Parmeswaran and Payandeh [28] developed Micro-Electronics-Mechanical System (MEMS) based piezoelectric tactile sensor which can be placed on the laparoscopic graspers jaws to measure the direct grasping force.

Several researchers had also focused on the development of new laparoscopic instruments and systems with sensors located on the jaws or tools shaft to measure the surgical manipulation forces in one or more directions [29]. Okamura et al [30] developed a 2 DOF force sensing sleeve to fit concentrically over endoscopic tool shaft to measure indirect grasping forces and bending forces in 5mm Laparoscopic instrument. Also Peirs, Clijnen and Brussel developed a tri-axial force sensor for MIS; the force sensor was based on flexible titanium structure through which deformations were measured through reflective measurement with three optical fibers [31]. However, cost and sterilization issues were not discussed for a tri-axial force sensor.
There are different types of force sensing methods were present in the field of MIS [32]. Few of the sensors are mentioned below:

A displacement-based sensor such as potentiometer, LVDT-based force transducer, which can provide accurate displacement, if the stiffness of the spring was accurately known and the force value can be easily calculated by the Hooke’s law. Rosen et al. [33], [57] developed a force sensing teleoperated endoscopic graspers by using a flat coil actuator and LVDT based force transducer. The forces can also be measured by measuring current flowing through the DC motor by using current sensing op-amps. Tholey et al. [34] developed a current sensing method based on Tholey’s principle, a force exerted on the motor shaft can be considered as a disturbance of the motor control system. The magnitude of the force being exerted can be measured by efforts of the controller by providing motor current to compensate or balance the disturbance effect.

Another way to measure the forces was through pneumatic pressure. A 4-DOF pneumatic-driven forceps were developed by Kawashima et al. [35], where the joints were driven by wire rope transmission mechanism actuated by pneumatic actuators.

Resistive based sensing were also common in force sensing devices where the force causes strain in the structure and strain gages were used to measure the strain. The tradeoff between the sensitivity of the resistive based sensor and deformation of structure is explained in [36]. Desai et al. [37] developed a resistive gage based force sensing graspers where the strain gages were stick to the drive shaft of graspers. During interaction between graspers and any objects, the strain in drive shaft can be measured by strain gages and the grasping and twisting forces can be measured by the special arrangement of strain gages.
Kuebler and Seibold et al. [38], [39] developed a distal force-torque sensor for 2-DOF laparoscopic instrument at the German Aerospace Centre (DLR). The six-axis sensor consisted of a hexapod structure known as Stewart Platform [40], where six strain gages were located for measuring forces and torques along all of its six measurement axis, the prototype is shown in Figure 2.6.

Capacitive-Based force sensors were used for detecting extremely minute deflections in the structures. These sensors were performed better than strain gages sensing. A capacitive-based force sensing equipped miniature tactile array sensor for Laparoendoscopic graspers [41] were developed in University of California-Barkley.

Another research was conducted based on capacitance measurement principle in Harvard University were the remote palpation system was designed for assisting the surgeons to examine for hidden arteries and tumors concealed inside tissues during MIS procedures [42].
Piezoelectric-based sensing sensitively produced voltages when their structures were deformed due to compression. An extremely small deformation in a structure triggered a large output voltage. Sokhanvar et al. [43] developed a multifunctional tactile sensor using PVDF (Polyvinylidene fluoride). The sensor consisted of a base, a flexible beam and 3 PVDF sensing elements, which are shown in Figure 2.7. Two PVDF sensing components were sandwiched at the end supports to measure the magnitude and position of exerted forces. The third PVDF sensing component was located on the flexible beam and it measured the softness of the contact objects.

Vibration-based sensors measure tactile information through dynamic responses. At Fraunhofer Institute for Biomedical Engineering [44] in Germany, a prototype of miniaturized vibrotactile sensor [45] was developed which was capable of measuring the dynamic mechanical properties of soft tissues in MIS. Baumann et al. [46] developed a vibrotactile sensor for endoscopic otalarynologic surgery in which the measurement of mechanical tissues impedances were determined by the
resonance frequency in the range of 100-800 Hz. Baumann concluded that the sensor can help in differentiating healthy and unhealthy tissues during MIS procedure.

![Figure 2.8. Optical Fiber Force Sensor to Measure Tissue Forces in 3D [47]](image)

Optical-based sensing strategy was first proposed by Hirose et al [48]. Implementing optical sensing devices is an alternative way to measure forces and transfer force information. Fiber optics cables are much faster than the electrical wires to carry force information in optoelectronic equipment. The basic components of an optical fiber force sensor are a light source, transduction element and optical detector. A light source which generates light that travels towards transduction component via the light transmitting optical fibers. If the transduction component is located at the shaft of the laparoscopic tool, the transduction component modulates the light in proportion to the value of the forces measured during MIS procedure. Clinjen [47] from Leuven University applied this configuration in 5mm laparoscopic tool which is shown in Figure 2.8.

Several techniques of force sensing methods and applications have been discussed. Displacement-based sensors are simple, cheap and easy to be implemented among all the sensors mentioned above. As the force measurement is indirect, good measurement accuracy could not be obtained. Also friction in assembly, backlash in gearheads, gravity and inertia of mechanical linkages could reduce the quality of the force measurement considerably.
These numerous designs can accurately measure the grasping forces but, they have limitations towards integrating them into actual surgical procedures. The implementation of these sensors into existing laparoscopic tools would prove difficult, costly and non-adaptable to current robotic surgical systems due to their non-modular, disposable designs.

**Section 2.5. Haptic-User Interface Device (H-UID)**

Haptic interfaces are force feedback user interface devices which are based on bi-directional master-slave configuration. The H-UIDs are chosen to exert forces and react according to the user’s motions and imitate the interaction forces obtained through the dynamic behavior of the virtual or remote telepresence systems.

**2.5.1. Augmenting Devices and Systems**

An augmenting device enhances the surgeon’s ability to perform an operation. On the basis of their mode of interaction with the surgeon, haptic devices can be classified into the following categories:

1) **Hand-held tools**: The benefit of using hand-held tools is that they do not limit the ability of the surgeon and comprise negligible changes to the operating room. The disadvantage of the purely hand-held instruments is that a physically support of heavier instruments can be possible with the robot. Apart from absence of robotic arm, instruments cannot be locked in desired place and accurately controlled maneuvers (e.g. as required in MIS and during microsurgery) are not possible [49]. The hand-held devices are commonly sensorized and can be classified into several sub-categories on the basis of their functionality.
a) **Master-slave combined instruments:** The master-slave combined instruments consist of a master interface and a slave robot that are placed in universal instrument body and provided the ability to surgeon to operate the instrument near to patient with other conventional surgical tools.

b) **Instruments for measurement purposes:** The hand-held instruments can be sensorized and utilized for measuring the mechanical properties and characteristics of tissues, defining and evaluating desirable surgical procedures by computing interaction occurring between surgeon and surgical instruments.

c) **Instruments for reducing hand tremor:** Hand tremors may occur when the surgeon becomes fatigue while performing delicate surgical procedures such as neurological surgery. Riviere et al [50] determined the hand motions with the aid of sensors. A frequency domain control scheme was applied to detect tremors occur due to the hand motions and tremors were compensated by using piezoelectric actuators.

2) **Cooperatively-controlled tools:** In the case of cooperatively controlled instruments, the surgeon and robot both possess surgical devices. The surgeon delivers motion, control and robot delivers accuracy, sensitivity and guidance.

   e.g. PHANToM Omni of Sensible Technologies, Inc., Freedom-6S of MPB Technologies Inc., Laparoscopic Surgical Workstation of Immersion Corp., Xitact IHP of Xitact SA. These devices are mostly teleoperated and classified into two sub-categories:

   a) **Force controlled devices:** The force controlled devices attempt to minimize the grasping forces acting on the end-effector of a robot by following surgeon’s hand maneuver.
b) **Passive devices:** Passive devices offer the workspace and dexterity needed to perform for surgeons to carry out certain tasks. Such passive devices are used in cardiac puncturing which is discussed by Schneider et al [51].

3) **Autonomous tools:** Robotic instruments can execute certain tasks autonomously. These devices not only reduce the burden on surgeon’s hand but complete the tasks in less span of time. e.g. Autonomous tool for suturing and knot tying [52].

Tan et al [58] focused on human factors involved in the design of force-reflecting haptic interfaces where human capabilities were discussed and new design criteria for user interface device were elaborated. Tavakoli, Patel and Moallem have presented the design of a robotic master-slave system [53] for use in MIS. Tavakoli has used a handle from a laparoscopic tool as a haptic device. They have followed the same mechanism in laparoscopic handle which has been used in PHANToM Omni. The surgical graspers mechanism was presented in slave side which is robust due to linear actuator. Authors have used direct force reflecting control architecture for haptics. Their design was robust and oversized than the user interface device developed for \textit{in vivo} robots at UNL.

**Section 2.6. Terminologies in the Haptic Feedback System**

**Z-width:** Z width of a haptic device is the dynamic range of impedance that can be passively rendered. Therefore, the haptic device with larger Z-width renders better feeling of virtual environment.

**Transparency:** An ideal transparent teleoperator transmits the exact forces of the teleoperated environment to the operator and the slave manipulators should exactly
follow the position of the master manipulator & apply the equal amount of forces on the environment [54, 55].
Chapter 3: Motivation: Haptic Feedback System

Section 3.1. Design Concepts

In Advanced Surgical Robotics Labs located at University of Nebraska-Lincoln, the group of researchers is developing Laparoendoscopic Single-Site (LESS) Surgical robots. One of the best robots has been considered to propose the need of haptics in in vivo surgical robots.

Figure 3.1. TB v2.0 model, courtesy Tyler Wortman’s Thesis [10]

The Figure 3.1 depicts the prototype of 4 DOF TB2.0 in vivo surgical robot, which consists of two arms that can be held together by the insertion rod. Each arm
comprises of Torso (2 DOF), Upper Arm (1 DOF) and Forearm (1 DOF). The end effector is laparoscopic graspers which are mounted on the forearm for tissue manipulation. Torso and Upper arm provide more degrees-of-freedom and higher amount of dexterous workspace to perform MIS procedures. The TB2.0 robot can be separated and inserted individually through a single small incision. Once the robot is fully inserted within the abdominal cavity, both arms can be assembled with the help of insertion rod and the laparoscopic surgeries can be performed. The TB2.0 was based on position-position bilateral control architecture and during the surgery, the sense of touch gets lost due to lack of force sensing capability. In order to restore the sense of touch that is lost in robot-assisted MIS, the design and placement of force sensor inside the forearm of a slave manipulator (TB2.0) should be proposed and it should provide appropriate force feedback to master.

![Figure 3.2. Phantom Omni as a User Interface Device for TB v2.0 [10]](image)

The Phantom Omni from SensAble Technologies Inc. [59] is one of the commonly used haptic controllers in the field of surgical robotics. The Phantom Omni provides 6 DOF input controls but it delivers force feedback only for first three joints.
To attain the force feedback for tissue manipulation via laparoscopic graspers, the pen-shape handle of Phantom Omni, which is shown in Figure 3.2, is typically used for graspers roll and close-open actuation. It should be replaced with the force feedback enabled device.

Section 3.2. Design Requirements

3.2.1. Force Sensor

The design concepts of haptic feedback system prominently influence the design of force sensor. It is desirable to have a force sensor placed in the slave robot for measuring the tissue interaction. This arrangement can provide more stable, reliable and efficient haptic feedback system.

The functional requirements of force sensor are as follows:

1. Placement: In MIS, a force sensor should be mounted on the jaw of laparoscopic graspers or on the section of a graspers driveshaft that actuates the graspers in order to reduce the noise occurring through friction and other disturbances from an actuator. The placement of force sensor on graspers jaw complicates the design and creates sterilization issues and increases the cost of the system.

2. Range of grasping force: The force sensor should measure the grasping forces in the range of 0N-12.5N which is required for standard laparoscopic surgical procedures [60].

3. Calibration: The calibration of the force sensing system should be easy to use and cheap in cost.

4. Size: According to requirements of MIS surgical tools, the diameter of the force sensor should be less than 10mm [61].
3.2.2. **Surgical Haptic User Interface Device (H-UID)**

An ideal haptic feedback device should transmit the same tissue interaction forces between a graspers and remote environments to the surgeon. Therefore, the surgeon should not detect any difference in the interaction with real object. As results, the completely transparent interface to the remote environment needs to be developed. The versatility and transparency of surgical H-UID are affected by a number of design criteria describing its performance. A number of essential design and functional requirements need to be considered for Haptic-bilateral teleoperation and smooth MIS procedures.

The requirements of Surgical Haptic-User Interface device (H-UID) are as follows:

1. **Range of force**: The typical grasping forces applied by the graspers during standard tissue palpation tasks in MIS are in the range of (0N-12.5 N); therefore the H-UID should transmit the same amount of forces to the operator for complete transparency.

2. **Workspace**: The range of the workspace needs to be considered. The standard laparoscopic graspers jaws rotate around 75\(^\circ\) to actuate from complete open to complete close motion. As a result, the surgical H-UID should at least rotate 75\(^\circ\) to provide as a suitable controller for the laparoscopic graspers and maintain 1:1 scale for smooth manipulation.

3. **Modularity, Ergonomical & Efficacy**: A surgical H-UID should be fabricated to fit concentrically over a gimble of an existing Phantom Omni. The Phantom Omni compatible surgical H-UID should be light in weight so that it can be used efficiently and produce less strain & fatigue to the
surgeon’s hand. It should be designed by taking ergonomical approach into account.

4. **Dynamics or Bandwidth:** The dynamics or bandwidth (force feedback communication capacity) of the H-UID should be in a range of 100 Hz to 200 Hz and the haptic feedback loop or virtual walls creation loop should be executed at the loop rate of 500 Hz to 1000 Hz.

5. **Haptic impedance:** Good back-drivability, low inertia and low backlash in the transmission are the significant features of the H-UID for accurate force reflection in low-impedance environment.

It is naturally clear that these requirements are conflicting but one needs to be compensated with others.
Chapter 4: Design of Haptic Feedback System

Section 4.1. Slave Robot: Force Sensing Forearm

4.1.1. A Force Feedback Enabled Motor-Driven Laparoscopic Graspers Mechanism

4.1.1.1. Design Requirements:

The task was to develop an end effector to a Laparo-Endoscopic surgical robot that is sensorized and actuated in accordance with the requirement of endoscopic surgery. Therefore, the force sensing mechanism should measure the interaction between the graspers and tissues. Due to the incision size constraint in laparoscopic surgery, very limited forces or strain sensing sensors are available in market. Additionally, sensors could not be mounted on the graspers jaws due to sterilization issues, and also it is necessary to use the graspers that can be detached and disposed of after use [62].

Two types of standard laparoscopic graspers were selected for analysis.

1) Autosuture 5mm laparoscopic graspers from single use laparoscopic hand instrument manufactured by Covidien.
2) Ethicon Grasp 5mm laparoscopic graspers from Johnson & Johnson Company.

Few assumptions were made to simplify the analysis:

1) The friction in the mechanism and linkages were negligible for all graspers jaw angles.
2) The horizontal component of pulling force was negligible as compared to forces at tip.

Figure 4.1. A Typical Model of Laparoscopic Graspers

In Figure 4.1, the graspers used in endoscopic surgery needs to be actuated by DC motors for grasping tissues where the jaws pivotally moved relative to one another by linear slider motion of a drive pin. For the end-effector considered here, the graspers actuation assembly consists of two laparoscopic graspers which are hinged to grasper yolk and drive shaft for actuation of graspers (close/open motion).
4.1.1.2. Force Analysis of Laparoscopic Graspers

1) Autosuture Laparoscopic Grasper Force Analysis:

To control the jaw’s angular motion, it is necessary to find the relation between the grasper’s jaw angles with respect to the linear displacement of the driveshaft. The FBD of an Autosuture 5 mm grasper in two different positions are shown in Figure 4.2 and Figure 4.3 demonstrates the FBD of the Autosuture graspers. Where, $A_1$ is the distance between hinge of the grasper to tip of the grasper and $A_2$ is distance between hinge P of the graspers to the center of drive pin. $\Theta$ is the grasper jaw angle w.r.t. horizontal and $\beta_0$ is the angle of slider w.r.t. horizontal at position 1. $x$ is the displacement of the driveshaft during angular displacement of the graspers from position 1 to position 2.

Here, the slider angle at position 2 with respect to the horizontal is $\beta = \Theta + \beta_0$
Figure 4.3. FBD of Autosuture Laparoscopic Grasper

Considering Figure 4.3,

Summing up the moments at pivot point P:

\[ F_{\text{tip}} \cdot A_1 = F_{\text{pull}} \cdot \sin \beta \cdot [(A2 - x) \cdot \cos \beta] \]  

(4.1)

\[ \therefore F_{\text{tip}} = F_{\text{pull}} \cdot \frac{\sin \beta [(A2 - x) \cdot \cos \beta]}{A_1} \]  

(4.2)

2) Ethicon Laparoscopic Graspers Force Analysis:

Figure 4.4. Force analysis and FBD of Ethicon Laparoscopic Grasper

The FBD of Ethicon grasper is shown in Figure 4.4. Summing up the moments at hinge:

\[ F_{\text{tip}} \cdot A_1 = A_2 \cdot \frac{F_{\text{pull}}}{\cos \theta_3} \]  

(4.3)

\[ F_{\text{tip}} = \frac{A_2}{A_1} \cdot \frac{F_{\text{pull}}}{\cos \theta_3} \]  

(4.4)
Where, $A_1$ is the distance between the tip of the jaw to the pivot point of grasper and $A_2$ is the distance between pivot point to the center of link. $\Theta_3$ is the angular distance between axis of link to the horizontal.

The accurate 3D model of Ethicon graspers had been developed in SolidWorks 2010 and simulations were created to derive the empirical relation of $\Theta_1$ and $\Theta_3$ and displacement of the driveshaft, $x$ and Graspers Jaw angle, $\Theta_1$. The results were as follows:

The empirical relation between graspers jaw angle and $\Theta_3$ has been found as below:

![Figure 4.5. The Empirical Relation between Graspers Jaw Angle, $\Theta_1$ and $\Theta_3$](image_url)

The empirical relation between displacement of driveshaft and graspers jaw angle, $\Theta_1$ provided an essential relationship between the grasping forces w.r.t the graspers jaw angle.
Figure 4.6 The Empirical Relation between Displacement of Driveshaft and Graspers Jaw Angle, $\theta_1$

The role of measuring linear displacement of the driveshaft assisted in measuring the angular displacement of the grasper jaw. Hence, this empirical relation in Figure 4.6 used to estimate the grasping forces w.r.t. to the distance travelled by driveshaft.

4.1.2.3. Comparison on the Basis of Grasping Force and Pulling Force

<table>
<thead>
<tr>
<th></th>
<th>Grasping Force (N)</th>
<th>Pulling Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Autosuture Graspers</td>
<td>12.5</td>
<td>178.14</td>
</tr>
<tr>
<td>Ethicon Graspers</td>
<td>12.5</td>
<td>53.6</td>
</tr>
</tbody>
</table>

Table 4.1. Comparison of Autosuture and Ethicon Graspers based on Pulling Forces and Grasping Forces

The forces shown in Table 4.1 were generated from the simulation developed in SolidWorks. Comparison demonstrated that the Ethicon graspers had lower ratio of pulling force to grasping force. Therefore, Ethicon grappers were selected in designing force sensing mechanism. Ethicon grappers also have negligible slop in the mechanisms over Autosuture grappers for accurate force estimation.
4.1.2. **Force Sensor v1.0**

4.1.2.1. **Force Analysis by Strain Gages**

The Figure 4.7 represents the force measurement through strain gages located on driveshaft. Strain gages (model # EA-XX-031EC-120, Vishay Micro-Measurements) were used to measure the strain inside driveshaft. Due to limited amount of space, two strain gages were glued to opposite sides of grasper driveshaft. It is only capable of measuring forces in one dimension. The strain gages placed inside forearm picked up unwanted noise from miniature motor and graspers assembly and caused distortion in the strain measurement which is shown in Figure 4.8.

![Figure 4.7. Location of Strain Gages on Graspers Driveshaft in Force Sensor v1.0](image)

**Figure 4.7. Location of Strain Gages on Graspers Driveshaft in Force Sensor v1.0**

![Figure 4.8. Noise from the Strain Gage](image)

**Figure 4.8. Noise from the Strain Gage**
4.1.3.  Force Sensor v2.0

4.1.3.1.  Specification

The forearm in \textit{in vivo} robot consists of motor driven close-open actuation of laparoscopic graspers mechanism. As the arm of robot which passes inside an abdominal cavity through 1” diameter incision, it should be as compact as possible. Surgeons and researchers related to laparoscopic surgery have concluded that magnitude of grasping forces exerting on tissues ranges from 0 N to 12.5 N during a typical palpation task \cite{60}. The precautions had been taken as per the surgical rules and regulations to ensure force sensor should be biocompatible, sterilized and robust.

4.1.3.2.  Design

As per the analysis shown in section 4.1.2.3, the Ethicon laparoscopic graspers were selected and they incorporated advantages such as low backlash, compact design and low ratio of pulling forces to grasping forces. In addition to these characteristics, the Ethicon laparoscopic graspers are also cheap in cost. The force sensing motor driven laparoscopic graspers mechanism consists of two major components: the actuation mechanism and the force sensor which are shown in Figure 4.9.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{force_sensing_mechanism}
\caption{Force Sensing Mechanism}
\end{figure}
To actuate a linear positioning of the graspers driveshaft in order to open & close graspers, the assembly of DC motor utilized a leadscrew and spring driveshaft to convert the rotational motion to translatory motion. This translatory motion generated the displacement in the spring.

Figure 4.9 shows the layout of the force sensor. The force sensor which consists of an ultra-precision extension spring (part # 9044K203, McMaster-Carr) with known spring constant, a set of special infrared LED (part # L1939-04, Hamamatsu Photonics) of $\phi 300 \, \mu\text{m}$ emission spot and position sensitive detector (part # OD6-6-SO 16, Pacific Silicon Sensor) at one end of the spring and motor optical encoder to other end of the spring to measure the linear displacement of the spring from both sides.

The force sensor would be coupled to the graspers driveshaft by threaded joint of 5mm diameter instrument. By measuring the displacement in spring, the grasper pulling forces can be calculated and generated pulling forces would be sufficient enough to exert appropriate grasping forces on the graspers. The equation (4.4) was used to measure the grasping forces w.r.t. pulling forces measured with the help of spring based force sensor.
The Figure 4.10 plot demonstrated that the spring coefficient of the extension spring can be calculated by deriving empirical relation of hanging weight vs. deflection in spring. According to Table 4.1, due to the higher pulling forces act on the spring, hence selected spring was suitable for measuring pulling force application.

4.1.3.3. Principle of Force Measurement

The exact spring constant can be identified by a method of hanging weights which is explained for Figure 4.10. To measure the pulling forces acting on the spring, the displacement at both sides of the spring need to be measured. The rotary motion of the DC motor was converted into translatory motion by using a threaded shaft and a pair of spur gears. The displacement of the spring, $x_1$ can be measured by finding a distance traveled by a spring drive shaft can be determined by no. of rotations and pitch selected for threading for a spring shaft. The infrared LED which
was attached to the spring emitted the infrared light on the position sensitive photodiode which measures the displacement of the spring, $x_2$.

The pulling forces can be calculated by using Hooke’s law or the spring equation,

$$F_{pull} = -K_{spring}(x_1 - x_2) + I.T. \quad (4.5)$$

Where, $F_{pull}$ = Pulling force (in Newton) of the grasper driveshaft to close the Laparoscopic graspers

$K_{spring}$ = Spring constant in N/m

$x_1$ = Linear displacement calculated by no. of rotations of spring drive shaft and pitch of threads, and no. of rotations were counted by optical motor encoder in mtr

$x_2$ = Linear displacement measured by position sensitive photodiode and infrared LED in mtr.

I.T. = Initial tension in the spring in Newton

The force sensor was capable to exert and sense 0N to 12.5 N of forces and the force range of the sensor can be altered with the help of LabVIEW signal processing module & the resolution can be varied from 0.05N to 0.5N by changing the sampling rate in data acquisition system.
4.1.3.4. Force Sensor Calibration

Figure 4.11. The Calibration System

The force sensor calibration system is shown in Figure 4.11. The state-of-the-art calibration with the help of external sensors developed for force sensor assembly was required for accurate estimation of graspers-tissue interaction forces. The preloaded ultra-precision spring and the manufacturing tolerance of the motor driven graspers assembly made it necessary to calibrate force sensor once it was coupled onto the graspers. To perform this calibration, round force sensor of 5lb (22.29 N) load limit (Interlink Electronics FRS 402 Series Round Force Sensing Resistor) was used. In order to exert the equal amount of force on the active region of the force calibration sensor, specially developed graspers caps were used.
Figure 4.12. The Plot of External Force Sensor (N) Vs. Grasping Force (N)

The graspers were closed from open position and the data from external force sensor and pulling force were recorded and the least-square method was used to find a line that best describes these data points in the grasping forces versus pulling force, the derived the linear empirical relation which is shown in Figure 4.12 provided the accurate force estimation of grasping.

4.1.4. Comparison between Strain Gages vs. Spring based Force Sensor

The force sensor v1.0 (discussed in Section 4.1.2) and force sensor v2.0 (discussed in Section 4.1.3) were compared after detailed inspection and testing. The scores were based on best is 5 and worst is 0. The following comparison will provide comprehensive evaluation:
### Design Aspects

<table>
<thead>
<tr>
<th></th>
<th>Strain Gages on Driveshaft</th>
<th>Spring based Force Sensor attached to Driveshaft</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maintenance</td>
<td>2</td>
<td>5</td>
</tr>
<tr>
<td>Strength</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>Connectivity</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>Accuracy</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>Sterilization</td>
<td>2</td>
<td>4</td>
</tr>
<tr>
<td>Cumbersomeomeness</td>
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<td>5</td>
</tr>
<tr>
<td>Feasibility</td>
<td>3</td>
<td>5</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>19</strong></td>
<td><strong>31</strong></td>
</tr>
</tbody>
</table>

#### Table 4.2. Trade-off between Strain Gages vs. Spring based Force Sensor

In Table 4.2, as compared to strain gage force sensors, no interference signals were generated in spring based force sensor. Therefore, it is clearly demonstrated that the ultra-precision spring based force sensor was suitable for surgical haptic applications.

#### 4.1.5. Benchtop: Force Sensor v2.0

![3D Model of Benchtop Assembly of Spring based Force Sensor](image)

**Figure 4.13. 3D Model of Benchtop Assembly of Spring based Force Sensor**
Figure 4.14. Prototype of Assembly of Force Sensor with PSD

The Figure 4.13 represents the 3D model of force sensing assembly designed on SolidWorks and Figure 4.14 demonstrates the assembly was set up for the benchtop. The benchtop provided an ability to test the force sensing mechanism and to develop force sensing programs on LabVIEW.

The spring based force sensor was coupled to the graspers driveshaft through threaded joint and Ethicon graspers were hinged to grasper yolk and driveshaft. For safety of the DC motor, spur gear of ratio 2.4 was selected for maximum center-to-center distance between motor shaft and axis of force sensor for safety of motor. Maximum diametrical pitch of 64 was chosen for particular set of spur gears for precise motion. For spring driveshaft, ANSI 6-32 internal threads class 3A was chosen to get maximum linear displacement of driveshaft per rotation. Jig arrangement had been developed to prevent spring driveshaft from rotation in order to maintain spring constant by preventing rotational motion in spring. The special LED was powered up to 1.5V and current was regulated to 120mA in order to illuminate IR
of wavelength in the range of 900-950 nm. This set-up helped in deciding the location of position sensitive detectors, position of special LED and actual no. of active coils of extension spring to generate 12.5N of tip forces on graspers and finally estimating time required for entire closing and opening action of graspers by this mechanism.

This prototype was connected to the haptic paddle via bilateral teleoperation which is explained in Section 4.3.1.

4.1.6. Force Sensing Forearm

Figure 4.15. 3D Model of Force Sensing Forearm

Figure 4.16. Actual Prototype of the Force Sensor Enabled Forearm
The TB2.0 compatible force sensing forearm which is shown in Figure 4.15 and 4.16 was developed from the force sensing mechanism explained in Section 4.1.2 and due to the poor estimation of displacement of the driveshaft, the position sensitive detectors were replaced with the linear magnetic encoder, which is shown in Figure 4.17.

**Figure 4.17. Printed Circuit Board: Linear Magnetic Encoder with Filter Board**

The Figure 4.18 presents the inside view of force sensing mechanism for grasper close-open actuation and grasper roll. The driveshaft displaced 1.25mm for 50°- 0° of graspers jaw angle during closing actuation. Therefore per 1° of angular displacement of graspers, driveshaft traveled only 0.025mm of distance. The set of 25um resolution linear magnetic encoder (Part #AS5304, Austria microelectronics) and 0.5mm pole length magnet were used to measure precise motion of driveshaft for accurate grasping force to graspers jaw angle, which provided precision in force measurement system.

**Figure 4.18. Graspers Roll and Close/Open Actuation Mechanism Inside Force Sensing Forearm**
As the graspers driveshaft displaced, the magnet which was mounted on driveshaft moved, this triggered the magnetic encoder and displacement of the driveshaft was measured in terms of counts from the magnetic encoder. This sensor provided precise position and velocity estimation of driveshaft than the position sensitive detector and aided in higher control ability to the haptic system. The special printed circuit board which was shown in Figure 4.18 were developed to mount magnetic encoder and placed inside the forearm.

For roll, 10mm motor was used to actuate in a cylinder in cylinder configuration. The 10mm motor actuated a set of spur gears rigidly attached to the bearings and a pre-load nut. The bearings were fixed to the housing; therefore as the 10mm motor rotated, the entire housing with force sensing mechanism rotated along the axis of rotation.

![Figure 4.19. Force Sensing Forearm Assembly to TB2.0](image-url)
The Figure 4.19 provided a clear demonstration of TB2.0 with a force sensing forearm mounted on the upper left arm. The forearm can be attached to the upper arm of the TB2.0 by link which can be seen in Figure 4.18. Figure 4.19 provided the idea about the kinematical changes in robot when the force sensing forearm was connected.

Section 4.2. Master: Haptic User Interface Device

In the force-reflective master-slave based bilateral teleoperation, the operator transmits forces and receives forces simultaneously via the force feedback routed master, while the slave robot mimic the operator’s hand maneuver over the physical environment. The same situation is considered while developing the haptic feedback system to the in vivo robots.

In order to develop the H-UID which works as a controller to manipulate laparoscopic graspers as well as creates the perception of interaction between tissues and graspers to the operator, haptic paddle was adopted to use as a controller.

4.2.1. Haptic User Interface Device v1.0 (Haptic Paddle)

It was a single axis, 1 DOF, force reflecting user interface device which was low cost, easy to manufacture and durable device. As compared to high end commercial haptic interfaces such as Impulse Engine [63] by Immersion Corporation and PHANToM Omni from SensAble Technologies [59], the functionality and mechanical and dynamic designs of haptic paddle were closely related. But these commercial controllers were much more expensive due to more degrees of freedom, more stable due to exclusive electric hardware. The above haptic interface devices were developed to emulate the interaction forces that take place when user came into contact with the physical systems.
4.2.1.1. Modeling and Identification

The haptic paddle which is shown in Figure 4.20 consists of a capstan pulley, position sensor, a brushed DC motor and a low stretched high bending radius thread to engage the capstan pulley to threaded shaft of the DC motor. The hall effect sensor (Part # A1301EUA, Allegro Microsystems) was used to measure the angular position of the handle. The mechanical hardware which were capstan pulley, support plates and base plates, as seen in Figure 4.20, were designed and manufactured in Type 1 PVC material for light in weight and long term reusability and robustness.
The Figure 4.21 demonstrates the arrangement of hall effect sensor in haptic paddle to measure the position and velocity of capstan pulley. The haptic paddle was based on second order mechanical system equations. The dynamic model of the 1 DOF haptic paddle was closely related to the classical inverted pendulum problem. As the operator tried to rotate the handle, the position and velocity of the handle were sensed and based on virtual environment algorithm, various amount of forces were exerted back to the operator.

The haptic paddle dynamic equations were explained in Appendix B. After modeling, the next step was to derive the numerical values of variables used in the governing equations. The governing equations of motion of paddle position and velocity were derived by energy method. The equivalent inertia of the paddle was determined by using bifilar pendulum method explained in Appendix C. The estimated inertia of the system was calculated by measuring the frequency of the oscillation of the handle and center of mass calculated from bifilar pendulum method. The detail information regarding the variables and derived equations were discussed in Section 4.2.3.

4.2.2. Haptic User Interface Device v2.0

Before designing a Haptic User Interface device, the following requirements need to be considered.

4.2.2.1. Requirement

1) Dynamic Properties: In Impedance-Impedance bilateral teleoperation, the H-UID should be able to provide sufficient range of impedances. In situation where the task involved a highly dynamical interaction from the surgeon on the device, it was significant that the back-drivability and stiffness can be
provided at a large bandwidth. Therefore, H-UID required a good back drivability to reduce unconstrained motions and efficient control architecture to mimic surgeon’s hand maneuver. To suppress the disturbance forces occurred by surgeon’s motion input and the natural dynamics of the device, the control system needed to render the low impedance.

2) **Output Capability:** The output capability of the device can be denoted by maximum force, position, velocity and acceleration which define limits for the haptic interaction that can be rendered. Hence, to achieve the transparency of the control system, the device at least should transmit the same range of forces as of the force sensor.

3) **Workspace:** The size of the workspace can be decided based on range of the interface by which it applied for. In this case, the range was 0N-12.5N.

4) **Compatibility:** The device needs to be compatible with the current surgical robot controllers e.g. PHANToM Omni. It should not reduce the performance of the main controller. The most significant factor of H-UID was that it should be designed based on ergonomical point of view.

### 4.2.2.2. Selection of Motor

For any haptic applications, brushed DC motors were selected than brushless DC motors due to reluctance in cogging and torque ripple phenomena in brushless motors. The prominent factor of choosing a motor as it should be capable of free run as well as produce sufficient torque to transmit the forces towards the operator. The Maxon DC motor (Motor # 339150 and Encoder MR # 225778 (detail specifications are in APPENDIX C).

\[
\tau_{stall} = \frac{F_m d}{\eta}
\]  

(4.15)
Where,

\[ \tau_{\text{stall}} = \text{Stall Torque} \]

\[ F_m = \text{Maximum Force} \]

\[ d = \text{Distance from force applied to the center of shaft} \]

The motor produced 26.2mNm of max. continuous torque which was sufficient for haptic application.

### 4.2.2.3. Choice of Transmission

The force sensor located at the forearm had a range of 0-12.5N. Therefore, the H-UID needed to transmit maximum \( F_m = 12.5 \)N of force towards the user.

The Berkley® fishing line (Part # BGQS30C-81, Appendix C) was chosen to engage the capstan pulley and threaded capstan drive shaft; it had incredible strength, higher bending radius and less slippage due to bit sticky surface area. The nominal diameter of the fishing line was 0.022”. Therefore, capstan drive shaft of 3/8-24 external threads were used to hold the fishing line in between screw threads.

Hence the nominal diameter of the capstan drive shaft is, \( D_{cs} = 3/8 \times 25.4 = 9.525 \) mm.

Considering Figure 4.26, the point A where the output torque, \( \tau_{\text{out}} \) applied to the operator which is 23mm away from the hinge H. therefore, the output torque obtained at the handle can be determined by,

\[ \tau_{\text{out}} = F_m \cdot l_x \]

(4.16)

\[ \tau_{\text{out}} = 12.5 \text{ N} \times 23\text{mm} = 287.5 \text{ mNm}. \]
4.2.2.4. Capstan Drive Gear:

The gear ratio was determined by considering a 12.5N of force being applied to the operator.

The capstan radius can be determined by,

\[
G_t = \frac{\tau_{out}}{\tau_{motor}} = \frac{287.5}{26.2} = 10.97
\]

Capstan drive transmission ratio, \( G_t = \frac{\tau_{out}}{\tau_{motor}} = \frac{287.5}{26.2} = 10.97 \) (4.17)

\[
\text{Capstan Radius} = D_{cs} \times G_t = \left(\frac{9.525}{2}\right) \times 10.97 = 52.24 \text{ mm} \approx 50 \text{ mm due to size constraint}
\]

(4.18)

4.2.2.5. 3D Model and Prototype of H-UID v2.0:

Figure 4.22. Haptic User Interface Device V2.0
The Figure 4.22 presents the H-UID v2.0 developed in SolidWorks 2010 and the Figure 4.23 demonstrates the prototype of H-UID v2.0 compatible to phantom omni. During the designing, all the requirements (discussed in Section 4.2.2.1) were taken into account. It is 1 DOF force feedback controller and capable of providing a max force of 12.5N at it handle. This H-UID has a good back-drivability and can be efficiently used with the Phantom Omni. It has same arrangement of capstan drive as haptic paddle.

The capstan drive design allowed the H-UID to be small in size and could provide a low-friction, zero-backlash drive for reduction in speed and amplification in torque [64]. The capstan joint consists of a pre-tensioned cable clamped at two ends of the capstan pulley and wrapped several times around the treaded shaft of the DC motor. This assembly is also known as cable-capstan transmission. Another advantage of capstan joint was reduction in the vibration in the system.
4.2.3. Virtual Walls Creation

Creating virtual walls was the common method in developing a local control system for a haptic controller. A virtual wall is built by virtual spring, damper and mass combination. In the Impedance-Impedance based control architecture, creating virtual walls based on dynamic properties of the device, which are shown in Figure 4.24. The Figure 4.25 shows the mass, spring and damper combination implemented in the control system.
The differential equation of mass-spring-damper system is as follows:

\[ M_{sys} \ddot{\theta} + B_{sys} \dot{\theta} + K_{sys} \theta = 0 \]  

(4.19)

In which, \( M_{sys} = \) The mass of moving components in the H-UID,

Equivalent spring constant of the system, \( K_{sys} = \):

\[ K_{sys} = K_{Virtual} + K_{Controller}, \]  

(4.20)

And, equivalent damping coefficient of the system, \( B_{sys} = \):

\[ B_{sys} = B_{Virtual} + B_{Controller}, \]  

(4.21)

Where, \( K_{Virtual} \) and \( B_{Virtual} \) are the spring and damping constants of the virtual system respectively.

\( K_{Controller} \) and \( B_{Controller} \) are the dynamic properties of spring and damping constants of the haptic controller respectively.

Therefore,

\[ F_{Virtual Walls} = (B_{virt} + B_{controller}) \dot{\theta} + (K_{virt} + K_{controller}) \theta \]  

(4.22)
4.2.3.1. Deriving $K_{\text{Controller}}$ and $B_{\text{Controller}}$

Consider Figure 4.26, assume if the capstan pulley rotated from position 1 to position 2 by angular distance of $\Theta_p$ and the handle displaced by distance $x$. Due to the rotation of capstan pulley, the motor shaft which was engaged to capstan pulley by means of fishing line rotated by $\Theta_m$. $J_{\text{rotor}}$ and $B_m$ are the polar moment of inertia of rotor and viscous damping of the brushed DC motor respectively. $J_p$ and $m_p$ are the polar moment of inertia and mass of capstan pulley respectively. The relations were as follows:

$$R_{cp} \Theta_p = R_{cs} \Theta_m$$  \hspace{1cm} (4.23)

Where, $R_{cp}$ = Radius of Capstan Pulley
\( R_{cs} = \text{Radius of Capstan drive shaft} \)

\[
\theta_p = \frac{R_{cs}}{R_{cp}} \theta_m \quad \text{(4.24)}
\]

But, \( \tan \theta_p = \theta_p = \frac{x}{l_x} \) \hfill (4.25)

Substituting the equation of (4.24) into (4.25), we get,

\[
\frac{R_{cs}}{R_{cp}} \theta_m = \frac{x}{l_x} \quad \text{(4.26)}
\]

\[
\theta_m = \frac{R_{cp}}{R_{cs} l_x} x \quad \text{(4.27)}
\]

By Rayleigh’s energy method:

\[
\frac{1}{2} m_{eq} \dot{x}^2 = \frac{1}{2} J_p \phi_p^2 + \frac{1}{2} J_{rotor} \phi_m^2 \quad \text{(4.28)}
\]

Differentiating equation (4.25) and (4.27) twice, we get,

\[
\ddot{\theta}_p = \frac{\dot{x}}{l_x} \quad \text{(4.29)}
\]

\[
\ddot{\theta}_m = \frac{R_{cp}}{R_{cs} l_x} \ddot{x} \quad \text{(4.30)}
\]

Substituting \( \ddot{\theta}_m \) and \( \ddot{\theta}_p \), in equation (4.28), we get,

\[
m_{eq} \dot{x}^2 = J_p \left( \frac{\dot{x}}{l_x} \right)^2 + J_{rotor} \left( \frac{R_{cp}}{R_{cs} l_x} \ddot{x} \right)^2 \quad \text{(4.31)}
\]

\[
m_{eq} = J_p \frac{l_x^2}{l_x^2} + J_{rotor} \left( \frac{R_{cp}}{R_{cs} l_x} \right)^2 \quad \text{(4.32)}
\]

For, \( b_{eq} \dot{x}^2 = B_m \theta_m^2 \) \hfill (4.32)
Therefore,

\[ b_{\text{Controller}} = B_m \left( \frac{R_{cp}}{R_{cs} \cdot l_x} \right)^2 \]  \hspace{1cm} (4.33)

For \( k_{eq} x = m_p g R_{cm} \theta_p = m_p g R_{cm} \left( \frac{x}{l_x} \right) \)  \hspace{1cm} (4.34)

Therefore,

\[ k_{\text{Controller}} = \frac{m_p g R_{cm}}{l_x} \]  \hspace{1cm} (4.35)

4.2.3.2. Deriving \( K_{\text{Virtual}} \) and \( B_{\text{Virtual}} \)

![Figure 4.27. The Connection of Virtual Haptic Paddle and Real Haptic Paddle](image)

The virtual dynamic properties dealt with stability of the controller. The Figure 4.27 shows the mass-damper-spring combination between the real and virtual haptic/kinesthetic paddle, where position and velocity of the virtual kinesthetic paddle could be transferred to the real kinesthetic paddle. The values of \( K_{\text{Virtual}} \) and \( B_{\text{Virtual}} \) could be decided based on connecting the virtual haptic paddle to the real haptic paddle. This helped in stabilizing the system in bilateral teleoperation.
This virtual haptic paddle was created in LabVIEW where the pointer slide provided the position and velocity estimation algorithm converted the position into velocity and these values were transferred to the real haptic paddle.

There were different ways of choosing the right values of $K_{\text{Virtual}}$ and $B_{\text{Virtual}}$:

1) Selected values of $K_{\text{Virtual}}$ and $B_{\text{Virtual}}$ in such a way that the real haptic paddle would achieve critically damped behavior. Therefore, when input position and velocity were given to real haptic paddle, the paddle did not oscillate but settled immediately.

2) Plotting the values of positions and velocities of virtual and real haptic paddles. The procedure is explained in Appendix C.

In the Impedance-Impedance Bilateral teleoperation, the values of $K_{\text{Virtual}}$ and $B_{\text{Virtual}}$ were chosen as 400 N/m and 37 Nm/s.

### 4.2.4. Relation between Virtual Walls Forces to Voltage provided to DC Motor

After providing an appropriate virtual wall force values, the voltage provided to the DC motor can be calculated as follows:

![Figure 4.28. Schematic of DC Motor used in H-UID](image)
The schematic of DC brushed motor is shown in Figure 4.28. The motor armature rotated at right angle to a magnetic field B by which the voltage was generated at the terminal of the armature and was equal to \( e = Blv \), where \( e \) is the voltage, \( l \) is the length of the armature coil wire and \( v \) is the velocity of the armature normal to the magnetic field.

Therefore, the current carrying armature which was rotating in a magnetic field, its voltage, \( V_{emf} \) is proportional to speed of motor, \( \theta_m \). Hence,

\[
V_{emf}(s) = K_v \cdot s \theta_m(s) = K_v \cdot \omega \tag{4.36}
\]

Where, \( V_{emf} \) is the back electromotive force. \( K_v \) is a back-emf constant. \( R_a \) and \( L_a \) are the resistance and inductance of the DC motor. The relation between the armature current, \( I_{arm}(s) \), the applied armature voltage, \( V_m(t) \) and the back-emf, \( V_{emf}(s) \), were found by writing a loop equation around the Laplace transformed armature circuit which is shown in Figure 4.28.

\[
R_a I_{arm}(s) + L_a s I_{arm}(s) + V_{emf}(s) = V_m(s) \tag{4.37}
\]

\( T_m \) is the torque developed by DC motor, and \( K_t \) is a motor torque constant, which depends on the motor and magnetic field characteristics. The relation was as follows:

\[
T_m(s) = K_t I_{arm}(s) \tag{4.38}
\]

Therefore, \( I_{arm}(s) = \frac{T_m(s)}{K_t} \tag{4.39} \)

From Figure 4.26, \( T_m(s) = F_{virtual \, Walls} \cdot \left( \frac{r_{cs} l_x}{r_{sp}} \right) \tag{4.40} \)

But from equation (4.22),
Therefore,

\[ F_{VirtualWalls} = (B_{virt} + B_{controller}) \cdot \dot{\theta} + (K_{virt} + K_{controller}) \cdot \theta \]

\[ T_m(s) = \left[ (B_{virt} + B_{controller}) \cdot \dot{\theta} + (K_{virt} + K_{controller}) \cdot \theta \right] \left( \frac{r_{cs} l_x}{r_{sp}} \right) \quad (4.41) \]

After substituting equation (4.36) and (4.39) into equation (4.37),

\[ \frac{(R_a + L_a s) \cdot T_m(s)}{K_t} + K_v \theta_m(s) = V_m(s) \quad (4.42) \]

Inductance of the DC motor was negligible, therefore it was neglected. Substituting the term \( T_m \) from equation (4.40), into equation (4.42), the voltage supplied to the motor based on virtual walls values can be determined as:

\[
V_m(s) = \left( R_a \cdot \frac{(R_{cs} l_x)}{R_{cp}} \right) \left( B_{virt} + B_{controller} \right) \cdot \dot{\theta} + \left( K_{virt} + K_{controller} \right) \cdot \theta \frac{r_{cs} l_x}{r_{sp}} \left( \frac{1}{K_t} \right) + K_v \dot{\theta}_m(s) \]

\[ (4.43) \]
Section 4.3. Haptic Control Architecture

4.3.1. Force Feedback System v1.0: Force-Position Control Architecture

Figure 4.29. The Experimental Setup of Force-Position Control Architecture for Bilateral Teleoperation

Figure 4.29 represents the teleoperation system with haptic paddle (discussed in Section 4.2.1.) as a master and force sensing mechanism (discussed in Section 4.1.1.) as a slave robot, where these two devices were connected by Force-Position bilateral teleoperation system. To implement the force feedback to controller, the following Force-Position control architecture which is shown in Figure 4.30 was developed. To find the trade-off between the force estimation through force sensor and force estimation through the position error, the suitable control scheme was chosen. The haptic paddle was used to control the graspers mechanism and behave according to the grasping forces received through the teleoperation system. This control scheme was efficient for force reflection of soft contacts unless the controller dynamics would be robust.
The Figure 4.30 demonstrates the Force-Position control architecture, where a haptic paddle was dedicated to move the force sensor enabled graspers actuation of slave robot. It was required to know the master and slave positions as well as the grasping forces exerted on the remote environment.

In this control scheme, the dynamics of haptic paddle and force sensing mechanism were described as $M(s)$ and $S(s)$ respectively. The slave robot was represented by the gain $K_{ps}$ and the output of the haptic paddle controller was controlled by $K_f$. $K_e$ represents the remote environment. The forces applied to the operator $f_m$ should be proportional to the interaction force between the slave and the environment, $f_e$.

The input force exerted by the operator on the haptic paddle was represented by $f_h$. The master rotated, if the force $f_h$ was not compensated by the force reflected from the master $f_m$. As a repercussion of such action, control loop of slave robot acquired the new force references. In this way, the operator got aware of the exerted forces by guiding the slave. Hence, the position error $e_p$ was calculated by the difference between positions of master and slave, $(x_m-x_s)$ respectively.
The haptic paddle controller gain $K_f$ retained the force reflected to the operator $f_m$ proportional to the position error $e_p$. Similarly, the slave robot controller gain $K_{ps}$ retained the force exerted by slave $f_s$ proportional to the position error $e_p$. As a result, position error between both devices was displayed on the operator as a resistance against his movement.

The Force-Position bilateral control architecture was derived mathematically as follows:

\[ x_m(s) = M(s) \cdot (f_h(s) - f_m(s)) \]  \hspace{1cm} (4.44)
\[ f_m(s) = k_f \cdot f_e(s) \]  \hspace{1cm} (4.45)
\[ e_p(s) = x_m(s) - x_s(s) \]  \hspace{1cm} (4.46)
But, $x_m(s) = x_s(s)$  \hspace{1cm} (4.47)
\[ x_s(s) = S(s) \cdot (f_s(s) - f_e(s)) \]  \hspace{1cm} (4.48)
\[ f_s(s) = k_{ps} \cdot e_p(s) \]  \hspace{1cm} (4.49)
\[ f_e(s) = K_e \cdot x_s(s) \]  \hspace{1cm} (4.50)
\[ x_s(s) = x_e(s) \]  \hspace{1cm} (4.51)

The equation (4.44) represented the forces reflected by the haptic paddle. A block diagram in Figure 4.31 depicts the relation between forces applied by the operator $f_h$ and the master position $X_m(s)$.

In order to obtain the transfer function $H(s) = \frac{f_m(s)}{x_m(s)}$, the control scheme shown in Figure 4.30 was simplified in Figure 4.31 (a) and (b).
The transfer function $H(s) = \frac{f_m(s)}{x_m(s)}$ can be considered as an impedance transmitted to the operator. The equation for $H(s)$ for this control architecture is described as:

$$H(s) = \frac{f_m(s)}{x_m(s)} = K_f \frac{S(s)K_e K_{ps}}{1+S(s)(K_e + K_{ps})}$$

(4.52)

To evaluate the significant parameters of $H(s)$, extreme cases were taken into account.

**Extreme cases:**

**Case 1: When the environment impedance was zero.**

$K_e = 0$, then $H(s) = 0$

Therefore, when environment impedance was null, the controller would reflect zero amount of forces to the operator. Thus, the value of $H(s)$ dropped down with low values of $K_e$. Hence, the soft contacts could be efficiently reflected towards the controller.

**Case 2: When the environment impedance was infinity.**

$K_e = \infty$, then $H(s) = K_f K_{ps}$

This case occurred if the slave was in contact with ideally rigid environment, and no deformation took place other than the exerted force. Hence, maximum impedance was perceived by the operator and this value was decided by $K_f K_{ps}$.
4.3.1.1. Experimental Approach:

![Figure 4.32. The Relation of Forces Acting on Graspers and Position error $e_p$ obtained at Haptic Paddle in terms of the Virtual Walls](image)

Each and every virtual wall was built on the basis of position error, $e_p$ and grasping forces exerting on objects. In the beginning, for fully open position of a graspers, the values of a virtual walls were 0N and for completely closed graspers, values of virtual walls were in the range of 12.5~16N which were based on the position error between the haptic paddle and graspers jaw angles. These virtual walls were generating direct adaptive haptic feedback to operator. If the operator needed to close graspers from fully open position, one needed to break these virtual walls by applying forces greater than force values of virtual walls. This entire real-virtual mechanical system was called Force – Position control architecture. This algorithm was developed on National Instruments’ LabVIEW 2009 software.
4.3.2. **Force Feedback System v2.0: Impedance-Impedance Based 4-CH Control Architecture for Bilateral Teleoperation**

![Diagram of Force Feedback System](image)

Figure 4.33. The Connection Diagram for 4-Channel Impedance-Impedance Control Architecture for Bilateral Teleoperation

Figure 4.33 represents the connection diagram of Impedance-Impedance based 4-CH control architecture for bilateral teleoperated system. In this master-slave configured teleoperation, H-UID v2.0 (discussed in Section 4.2.2) was used as a haptic controller and force sensing forearm (discussed in Section 4.1.5) was used as a slave robot. Communication between master and slave configuration was maintained by connecting both these devices to the computer system running LabVIEW, where the Impedance-Impedance based control programs were executed. This bilateral control system was also known as direct force feedback system, where the operator perceived the same amount of interaction forces involved in force sensing forearm and remote environment.

User applied forces to the H-UID v2.0 where these forces in terms of position and velocity were transmitted to the grappers actuation system located in force sensing forearm. Therefore the grappers followed the same motion in terms of angular
velocity of H-UID v2.0 and resulting grasping forces were measured through the force sensor located in the forearm transmitted to the H-UID v2.0. Hence user experienced the sense of being present (telepresent) in the remote environment.

A 4-CH bilateral teleoperation system consists of the master and slave mechanical system. These mechanical systems were connected with closed control loops around themselves. Overall, position, velocity and forces could be communicated bilaterally via various filtered versions of positions and forces.

Figure 4.34 shows a block diagram of the entire Force-Force teleoperation control system, which consisted of master, slave, bilateral telecommunication, task
(environment) and operator dynamics. This Force-Force bilateral teleoperation control architecture was a modified version of Lawrence control scheme [71]. The external forces $F_e^*$ and $F_h^*$ were independent of teleoperator system behavior. The nomenclature of the subsystem blocks were listed in Table 1. The values of various blocks for the teleoperation were derived on the basis of 1 DOF master-slave combination.

<table>
<thead>
<tr>
<th>Block</th>
<th>Description</th>
<th>Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Z_m$</td>
<td>Master Impedance</td>
<td>Mass, $M_m s$</td>
</tr>
<tr>
<td>$Z_s$</td>
<td>Slave Impedance</td>
<td>Mass, $M_s s$</td>
</tr>
<tr>
<td>$C_m$</td>
<td>Master Local Position Controller</td>
<td>Damper Spring, $B_m+K_m/s$ S</td>
</tr>
<tr>
<td>$C_s$</td>
<td>Slave Local Position Controller</td>
<td>Damper Spring, $B_s+K_s/s$</td>
</tr>
<tr>
<td>$C_1$</td>
<td>Slave Coordinating Force Feedforward Controller</td>
<td>Impedance Filter</td>
</tr>
<tr>
<td>$C_2$</td>
<td>Master Force Feedforward Controller</td>
<td>Scalar Gain</td>
</tr>
<tr>
<td>$C_3$</td>
<td>Slave Force Feedforward Controller</td>
<td>Scalar Gain</td>
</tr>
<tr>
<td>$C_4$</td>
<td>Master Coordinating Force Feedforward Controller</td>
<td>Impedance Filter</td>
</tr>
<tr>
<td>$C_5$</td>
<td>Master Local Force Controller</td>
<td>Scalar Gain</td>
</tr>
<tr>
<td>$C_6$</td>
<td>Slave Local Force Controller</td>
<td>Scalar Gain</td>
</tr>
</tbody>
</table>

**Operator and Environment**

<table>
<thead>
<tr>
<th>Block</th>
<th>Description</th>
<th>Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Z_h$</td>
<td>Operator Impedance</td>
<td>Impedance Transfer Function</td>
</tr>
<tr>
<td>$Z_e$</td>
<td>Environment Impedance</td>
<td>Impedance Transfer Function</td>
</tr>
<tr>
<td>$F_h^*$</td>
<td>Operator Exogenous Force Input</td>
<td>-</td>
</tr>
<tr>
<td>$F_e^*$</td>
<td>Environment Exogenous Force Input</td>
<td>0</td>
</tr>
</tbody>
</table>

Table 4.3. Nomenclature and Description of subsystem in the 4 Channel 1 DOF Bilateral Teleoperation Control Systems described in Figure 4.34.

In Figure 4.34, the model was assumed as Linear Time Invariant (LTI) dynamic model. $F_{cm}$ and $F_{cs}$ are the master and slave dynamics respectively. After substituting the $F_{cm}$ and $F_{cs}$ as a control commands to the Impedance-Impedance bilateral system, the dynamics of the closed-loop system was expressed as:
Therefore, Taking summations at the junction A and writing the dynamic equations in the form of equation (4.53), we get
\[
(1 + C_5)F_h - C_3 e^{-sTd} F_e = Z_{cm} V_h + C_1 e^{-sTd} V_e
\]
(4.55)

Taking summation at the junction B and writing the dynamic equations in the form of equation (4.54),
\[
C_2 e^{-sTd} F_h - (1 + C_6) F_e = Z_{cs} V_e - C_4 e^{-sTd} V_h
\]
(4.56)

Considering, no communication delay in the system,

Therefore, substituting Td = 0 in equation (4.55) and (4.56), we get,
\[
(1 + C_5)F_h - C_3 F_e = Z_{cm} V_h + C_1 V_e
\]
(4.57)
\[
C_2 F_h - (1 + C_6) F_e = Z_{cs} V_e - C_4 V_h
\]
(4.58)

Where,
\[
Z_{cm} = Z_m + C_m \quad \text{and} \quad Z_{cs} = Z_s + C_s
\]
(4.59)

According to Colgate [65], a bilateral teleoperation system is said to be robustly (or absolutely) stable, when coupled to any passive environment. Thus it transmits the passive impedance to operator. To analyze the stability and performance of the 4-CH bilateral system, Master-Slave Network hybrid parameters are derived as:

\[
\begin{bmatrix}
F_h \\
V_h
\end{bmatrix} = \begin{bmatrix}
h_{11} & h_{12} \\
h_{21} & h_{22}
\end{bmatrix} \cdot \begin{bmatrix}
V_e \\
-F_e
\end{bmatrix}
\]

(Section 2.3.2: from equation (2.8))

Solving for \( F_h \) and \( V_h \) in terms of \( F_e \) and \( V_e \),
\[
F_h = (h_{11} V_e - h_{12} F_e). (h_{21} V_e - h_{22} F_e)^{-1}. V_h
\]
(4.60)

We know that, the task impedance \( F_t(s) = Z_t(s). V_e(s) \)
(4.61)

Therefore, we write equation (4.60) in terms of \( Z_t \),
\[
F_h = (h_{11} - h_{12} Z_e). (h_{21} - h_{22} Z_e)^{-1}. V_h
\]
(4.62)

Considering, \( Z_t = (h_{11} - h_{12} Z_e). (h_{21} - h_{22} Z_e)^{-1} \)
(4.63)
\[ h_{11} = \frac{Z_{cm}Z_{cs} + C_4C_1}{(1+C_3)Z_{cs} - C_2C_1} \] (4.64)

\[ h_{12} = \frac{C_3Z_{cs} - C_1(1+C_6)}{(1+C_3)Z_{cs} - C_2C_1} \] (4.65)

\[ h_{21} = \frac{C_2Z_{cm} + C_4(1+C_5)}{(1+C_3)Z_{cs} - C_2C_1} \] (4.66)

\[ h_{22} = \frac{(1+C_6)(1+C_5) - C_3C_2}{(1+C_3)Z_{cs} - C_2C_1} \] (4.67)

The transparency can be defined as the synchronization between the impedance obtained from environment and the impedance transmitted to the user. Therefore, as per two-port master-slave network equations,

\[ Z_{to} = \frac{h_{11}Z_e}{1 + h_{22}Z_e} \] (4.68)

Substituting equations (4.64), (4.67) into equation (4.68),

\[ Z_{to} = \frac{(Z_{cm}Z_{cs} + C_4C_1) + [(1+C_6)Z_{cm} + C_4C_3]}{(1+C_3)Z_{cs} - C_2C_1 + [(1+C_6)(1+C_5) - C_3C_2]} \] (4.69)

**4.3.2.1. The Impedance-Impedance Transparency Optimized Control Law**

The expression for transmitted impedance provided the essential constraint on the design of a teleoperator architecture which was optimized for transparency. In ideal condition, the complete transparency needs to be obtained, i.e. \( Z_t = Z_e \) for all frequencies.

To get complete transparency \( Z_e = Z_t \), there are some fundamental insights to be derived from equation (4.62);

a) Perfect transparency \( (Z_t \equiv Z_e) \) requires that \( h_{11} = h_{22} = 0 \), and \( Z_e.(-h_{12}) = h_{21}.Z_e \).

b) As \( Z_e \to 0 \), the transmitted impedance, \( Z_t \) is insensitive to \( Z_e \) if \( h_{11} \neq 0 \), since \( Z_t \) depends only on the ratio of \( h_{11}.h_{21}^{-1} \).

c) As \( Z_e \to \infty \), the transmitted impedance becomes \( h_{12}.h_{22}^{-1} \), which is insensitive to \( Z_e \) if \( h_{22} \neq 0 \).
Therefore, $C_2C_3 = I$ \hspace{1cm} (4.70)

In order to eliminate $Z_e$ from the denominator of equation (4.62):

\[
C_4 = -(Z_m + C_m) = -Z_{cm} \hspace{1cm} (4.71)
\]

\[
C_1 = (Z_s + C_s) \hspace{1cm} (4.72)
\]

The $h_{11}$ term in the numerator of equation (4.62) will eliminate by making $Z_t$, a linear function of $Z_e$:

\[
Z_t = -h_{12}Z_e h_{21}^{-1} = C_3 Z_t \hspace{1cm} (4.73)
\]

This describes that when slave force feedforward controller, $C_3 = I$, then complete transparency will be achieved.

### 4.3.3. LabVIEW based Bilateral Teleoperation System and Electrical Hardware Configuration and Implementation

![Functional Modules of the Impedance-Impedance based Bilateral Teleoperation System](image)

Figure 4.35. Functional Modules of the Impedance-Impedance based Bilateral Teleoperation System

The entire haptic feedback system was developed on LabVIEW 2009 (National Instruments). Most of the NI developed electric hardware was utilized for
data communication and miniature DC motor controls. Figure 4.35 provides clear representation of functional modules of bilateral teleoperation system.

A LabVIEW based telerobotic system consists of operator environment and remote environment which is shown in Figure 4.35. Both these environments are connected to each other by using communication channels. These communication channels transmit commands from H-UID v2.0 to force sensing forearm and send back information of remote tasks to the surgeon. The operator environment comprises of devices: H-UID v2.0 which is an actuation device, command processor which works for transmitting position and velocity information to the remote environment and feedback processor which works for acquiring the force information from remote system. The task processor conducts tasks from slave in response from H-UID and control architecture and the information processor provides the force information obtain through interaction between slave robot and remote environment to the feedback processor located at the operator environment via communication channel. This entire process completes the bilateral telerobotics system.

![Figure 4.36. The Basic Hardware & Communication Flow Chart of LabVIEW based Bilateral Teleoperation System](image)

Figure 4.36 represents the detailed communication flow chart for Impedance-Impedance based bilateral teleoperation. The responses from surgeon in terms of positions were recoded and signals were passed through low pass filter circuits for noise reduction. These signals were received through NI-6343 DAQ and processed in
LabVIEW software installed in host computer. The software then determined the desired slave motor positions and velocities based on Impedance-Impedance control architecture and transmitted the same positions and velocities through NI CompactRio-9074 with NI 9505 full H-bridge brushed DC servo motor driver towards the miniature DC motor placed in force sensing forearm. Once the slave motor was driven on the basis of the positions and velocities signals received from CompactRio, produced forces on remote environment. These forces were measure with the aid of force sensor placed inside graspers manipulation mechanism. The force sensor transmitted the force information to host computer via CompactRio and NI 6343 DAQ. The LabVIEW software assisted in transmitting the same amount of forces in term of voltages through NI-6343 DAQ to the DC motor located at the H-UID v2.0 and servo drive was used to supply sufficient amount of current to drive the DC motor to behave according to the forces exerting on the remote environment.

4.3.3.1. Bandwidth of the Haptic Feedback System

![Diagram of Haptic Feedback System](image)

**Figure 4.37. The Range of Bandwidth required at each step of the Haptic Feedback System**

The Figure 4.37 depicts the necessary bandwidth required in each and every segment of the haptic feedback system. Theoretically, in order to perceive the feel of stable virtual walls, it is mandatory to transmit the haptic information measured at the
end effectors in slave robot within the bandwidth of 1 KHz to 10 KHz and recapitulate the same forces or positions to the user through H-UID. The range of 1 KHz to 10 KHz of high bandwidth is required because the frequency of human hand is 500 Hz [72]. The user’s motions and reactions can be measured at a bandwidth of 5 Hz to 15 Hz. and can be transmitted to the slave robot at the same bandwidth to manipulate objects.

To maintain the bandwidth mentioned in [72], Real-Time processor in LabVIEW-RT was considered for haptic feedback application.

4.3.3.2. LabVIEW Application Scheme

![Figure 4.38. The LabVIEW’s Real Time Project Applications used for Adaptive Haptics Control Programs](image)

**Real-Time Processor:**

Figure 4.38 represents the LabVIEW Real module with it’s built in functions and is classified according to execution speed. The NI CompactRio 9074 embedded system consists of an industrial 400MHz freescale MPC5200 processor that deterministically executes LabVIEW Real-Time program for Haptics on the Wind River VxWorks real-time operating system. The built in functions in LabVIEW were
used to transfer data between the FPGA and the real time processor within the CompactRio embedded system. NI 9505 (FPGA Interface) full H-bridge DC brushed servo drive module was used. It had built in encoder interface and current sensor to measure current flowing through miniature DC motor during palpation tasks.

The Impedance-Impedance based control architecture was developed in following segments associated with LabVIEW Real-Time modules:

- **Windows Host VIs:** All data acquisition programs, data record, H-UID velocity estimation algorithm and special features including automatic force sensor calibration programs and autonomous grappers programs were developed and executed in Windows based host computer. These programs were created to measure the analog signals and digital signals from potentiometer, magnetic encoder and force sensor respectively. These programs were executed at the loop rate of 150 Hz [67].

- **Time Critical Interface VIs:** The virtual walls creation algorithm and grappers force measurement algorithm based programs were placed in Time Critical Interface VIs to maintain the loop rate of 1000 Hz ~ 2000 Hz. In which “time loops” were used to regulate the loop rates.

- **LabVIEW FPGA VIs:** The miniature DC motor controlling programs were placed in reconfiguration FPGA segments to execute parallel with other programs. These programs were executed at the loop rate of 1MHz.

- **Normal Priority VIs:** Programs related to defining and initializing variable which were used in haptic application were placed in this segment. The Loop rate was kept at 500 Hz~1000Hz.
4.3.3.3. Graphical User Interface (GUI):

In Figure 4.39, a user friendly graphical user interface is presented, where the user can have an access to all parameters involved in Impedance-Impedance based bilateral control system. Few of the GUI features are discussed below:

- The operator can observe all the parameters involved in the haptic feedback system in form of real-time plots. These parameters are grasping forces acting on tissues, values of virtual walls towards H-UID, voltage provided to the motor located at H-UID, function of graspers jaw angle to grasping force, execution speed of time critical programs and positions and velocities parameters of laparoscopic graspers and H-UID.

- The operator can have an ability to change the range of grasping forces from 0N-12.5N.

- Special features such as an automatic force sensor calibration programs, autonomous graspers programs were included in GUI. The autonomous
graspers program provided ability to the operator to specify certain amount of grasping forces for specific tissues. In special cases, the operator had an ability to perform tissue manipulation tasks with or without force feedback.
Chapter 5: Results

The entire haptic feedback system discussed in this thesis has been tested and analyzed on non-survival animal model at the University of Nebraska Medical Center. All animal surgical procedures and protocols were approved by the Institutional Animal care and Use Committee (IACUC). The haptic feedback system was tested for tissue exploration and manipulation on swine under the supervision of specially trained laparoscopic surgeons. The benefits of haptic feedback on surgeons’ performance and tissue exploration behavior during teleoperated palpation tasks are evaluated and are as follows:

Section 5.1. Surgical Procedure: Tissue Exploration and Manipulation

Figure 5.1. Laparoscopic Graspers Grasping Colon Tissues
The link between forearm and upper arm of force sensing forearm was attached to Iron Intern and graspers were gradually brought near the colon for grasping analysis. The force feedback control system v2.0 (discussed in Section 4.3.2) was implemented, in which the H-UID v2.0 was used as a master and force sensing forearm was used as slave robot. The initial trials were performed to grasp colon tissues without the haptic feedback. Later, the haptic feedback system was activated and graspers movements were controlled by H-UID v2.0. The Figure 5.1 shows the snapshots of palpations tasks of colon tissues. The graspers were providing sufficient grasping forces to lift the colon tissues and open-close actuation was fast enough to acquire real-time force feedback to H-UID v2.0. Operators carried out several palpation tasks with varying levels of force feedback. The entire setup was videotaped and pictures were taken. 3 operators were selected to use the haptic feedback system and 4 times grasping actions were performed by each operator with and without haptic feedback. The sufficient data was recorded and significant parameters of haptics were plotted and discussed further for analysis.
Section 5.2. Tissue Manipulation and Control System Data

Figure 5.2. The Plot of Grasping Force Generated vs. Grasping Time during Several Palpations of Colon Tissues

The Figure 5.2 shows the plot of grasping forces vs. grasping time. The test 1 and 2 were carried out the palpation tasks of colon tissues without force feedback while test 3 and 4 were performed in the presence of haptics. Figure 5.2 demonstrated that users applied high level of forces for longer duration when force feedback was not available. On the other hand, during the trial with haptic feedback was on, less time was spent applying higher forces. The 12 no. of trials were analyzed and the statistics shown in Figure 5.3 depicts that the 24.22% excess time required carrying out the same task when the haptic feedback was absent. The same condition applied to the grasping force as the user provided 16.93% more grasping forces for tissue palpations.
Figure 5.3. Comparison of Grasping Forces and Time Taken to Execute the Tissue Palpation Tasks during Absence and Presence of Haptics

Figure 5.4. The Plot of Grasping Force vs. Laparoscopic Graspers Jaw Angle
Figure 5.4 shows the forces applied by laparoscopic graspers during a palpation of colon tissues. In the above plot, the grasping forces generated as the graspers jaw angle reduced. The exerted forces were sufficient enough to grasp tissues and perform palpation tasks. In this graph, total four tests results were compared. Without haptics, maximum grasping forces produced on colon were in the range of 4.64 N to 5.05 N. With haptics, maximum grasping forces produced on colon were in the range of 3.58 N to 4.18 N. Force feedback significantly reduced the magnitude of the forces applied at the controller during palpation of tissues. The haptic feedback significantly reduced the tissue palpation time and aided user to apply appropriate amount of grasping force on tissues. From 60° to 10° of graspers jaw angles, grasping forces increased linearly and later grasping forces increased exponentially. This phenomenon happened due to the anisotropic characteristic of tissues. The extreme left area of plot where all 4 curves accumulated was the area of haptic region. In this haptic region, the maximum interaction between force sensing robot and H-UID operated by user took place.
Figure 5.5. The Plot of Interaction between Graspers Jaw Angle and H-UID v2.0

In the Figure 5.5, the relation between graspers jaw angle (GJA) and H-UID handle angle was analyzed and the graph assisted in finding exact transfer functions between grappers and H-UID handle. The handle needed to turn 48° to close complete grappers once the haptics was off and handle turns to 38° when the haptics was on. This 10° of excess angular displacement of controller exerted more forces on tissues. The 24° of remaining angular distance required as a space to move handle and experience the haptics to user. Results shown that force feedback improved performance by reducing the overall resultant forces applied. This helped in reducing tissue trauma and prevented vital tissue from damage during palpation tasks and at the end, improved surgical performance.
Figure 5.6. Transparency of the Haptics Control System during Colon Tissues Manipulation vs. Time

The Figure 5.6 shows the plot of transparency which is a ratio of \( \frac{F_{\text{grasper}}}{F_{\text{Virtual Wall}}} \) vs. time, the bilateral teleoperation system is transparent if and only if the operator feels as if one is directly interacting with the (remote) task [70]. Therefore, the graph shows the ideal transparency curve which was a straight line passes through \( Y = 1 \), therefore, \( \left( \frac{F_{\text{grasper}}}{F_{\text{Virtual Wall}}} \right) = 1 \), it represented that the same amount of grasping forces were transmitted to the operator via haptic user interface device. Since the haptic bilateral teleoperation system interacted dynamically with the colon tissues and the operator, therefore the experimental curve came closer to ideal transparency only during the interaction with tissues. The system behaved close to ideal condition during most of the period of interaction with the sponge due to the synchronous motion of the H-UID and laparoscopic graspers. The transparency and versatility were
probably affected due to friction in four bar linkage at grappers, insufficient
transmission tension, gravitational compensation and inefficiency of H-UID DC
motor, slippage in capstan drive and a small time delay in the entire Haptic control
system.

Section 5.3. Experiment using Impedance-Impedance Control Architecture:

Benchtop Procedure:

The haptic system needed to be tested and analyzed further on different
objects of different stiffness. Sponge, sleeve and plastic block were chosen as testing
objects. The experiments were performed with standard procedure where the objects
were grasped by the laparoscopic grappers and H-UID v2.0 was used to operate
graspers. The haptic feedback was activated during the experiment and the effects of
haptics were analyzed for different objects and discussed further. Figure 5.7 shows the
screenshots of the grappers grasping sponge, sleeve and block respectively.

Figure 5.7. Screenshots of Benchtop grasping Sponge, Sleeve and Block
respectively

The forces reflected via a controller on the user were proportional to the forces
acting on objects. In Figure 5.8, angular position of H-UID $\theta_m$, jaw angle of slave
robot $\Theta_S$, grasping force $F_{\text{tip}}$ and force reflected in terms of virtual walls $F_w$ to the user were shown. It was decided to analyze the sponge as a subject due to its anisotropic property which was similar to properties of tissues. By selecting sponge, the flaws in the control system can be examined for further improvements.

Figure 5.8. The Behavior of Inputs-Outputs Impedance-Impedance Control Scheme

There were three different steps in this experiment:

**Downward trajectory:** In this step, master and slave positions were similar. It was necessary to emphasize that a slight force was reflected to the operator. This force was due to the effect of moment of inertia obtained through H-UID and force
sensing forearm connected to the Impedance-Impedance control scheme. However, this force can be compensated by taking into account an equal direction of gravity and selecting appropriate values of virtual spring and damper constants. Drag effect did not exist for this controller; therefore the operator perception was improved.

**Contact steps (Interaction with Sponge):** In this case, the graspers exerted higher forces on the sponge with increase in angular position of H-UID. Besides, during this step, user received higher and prominent forces. The contact with a sponge was thereby perceived as an elastic contact.

Contact steps (interaction with sponge) consist of two stages; first stage involved an elastic contact, where H-UID and graspers positions were very similar and the output from the force sensor increased linearly. Second stage where the surface of the sponge stopped being elastic, therefore H-UID and the graspers were stopped and the torque reflected from the DC motor became constant. When the inner surface of sponge became rigid, graspers could not able to produce grasping forces. However, the user still experienced that the sponge was elastic, which allowed the user to move the H-UID.

**Upward trajectory (Releasing Sponge):** As the angular position of H-UID reduced, forces applied on the user hand reduced faster than the downward trajectory. This phenomenon was due to the effect of gravity, inertia and mechanical inefficiency on the joint which dominated in this step. As a result, the control system responded more slowly.
Figure 5.9. Transparency of the Haptics Control System during Sponge Manipulation vs. Time

The Figure 5.9 shows the plot of transparency vs. time, where the graph shows the ideal transparency curve which was a straight line passes through \( Y = 1 \), which depicts that the same amount of grasping forces were transmitted to the operator via user interface device. Since the haptic bilateral teleoperation system interacted dynamically with the sponge (remote environment) of anisotropic material and the operator, which as results provided a distorted transparency in control system. During the interaction with sponge, the operator’s force on the master and motion of master followed the same relations with the forces applied by the graspers and motion of graspers assembly, therefore synchronous behavior improved the transparency. The transparency and versatility were probably influenced by moment of inertia of H-UID and graspers assembly, slippage in capstan drive and a small time delay in the haptic control system.
Due to the poor transparency in control system which can be seen in Figure 5.9 of the grasping force and virtual walls vs. time, virtual walls curve was distorted as compared to the grasping force curve. Therefore, false force feedback was transmitted to the H-UID. To eliminate the false feedback effect, the local virtual walls creation algorithm was developed.

The local virtual walls creation algorithm is as follows:

According to the complete transparency, \( F_{\text{Virtual Wall}} = F_{\text{Grasping Force}} \)

The value of a virtual wall at the 1st sampling point:

\[
(B_{\text{virt}1} + B_{\text{controller}})\dot{\theta}_1 + (K_{\text{virt}1} + K_{\text{controller}})\theta_1 = F_{\text{Virtual Walls}1} \tag{5.1}
\]

The value of a virtual wall at the 2nd sampling point:

\[
(B_{\text{virt}2} + B_{\text{controller}})\dot{\theta}_2 + (K_{\text{virt}2} + K_{\text{controller}})\theta_2 = F_{\text{Virtual Walls}2} \tag{5.2}
\]

The sampling rate of data acquisition system was 1KHz. Since the sampling rate was high, there were not any significant changes in consecutive virtual terms; therefore, \( K_{\text{virt}1} = K_{\text{virt}2} \) and \( B_{\text{virt}1} = B_{\text{virt}2} \).

Subtracting equation 1 and 2, we get,

\[
(B_{\text{virt}1} - B_{\text{controller}})(\dot{\theta}_1 - \dot{\theta}_2) + (K_{\text{virt}1} - K_{\text{controller}})(\theta_1 - \theta_2) = F_{\text{Virtual Walls}1} - F_{\text{Virtual Walls}2} \tag{5.3}
\]

Ignoring the change in angular velocity, \((\dot{\theta}_1 - \dot{\theta}_2)\) due its negligible value, we get

\[
(K_{\text{virt}1} - K_{\text{controller}})(\theta_1 - \theta_2) = F_{\text{Virtual Walls}1} - F_{\text{Virtual Walls}2} \tag{5.4}
\]

\[
K_{\text{virt}1} = \frac{F_{\text{Virtual Walls}1} - F_{\text{Virtual Walls}2}}{(\theta_1 - \theta_2)} + K_{\text{controller}} \tag{5.5}
\]
By which $K_{v\text{irt}1}$ from equation (5.5) can be calculated and hence, $B_{v\text{irt}1}$ can be calculated by substituting the value of $K_{v\text{irt}1}$ into equation (5.1).

Figure 5.10. The Behavior of Inputs-Outputs Impedance-Impedance Control Scheme with Local Virtual Walls Creation Algorithm

Thus, the virtual walls were created on the basis of the values of grasping forces and complete transparency was maintained. The grasping forces curve and forces of virtual walls curve were very close to each other therefore, the effect of false force feedback was reduced significantly which can be seen in Figure 5.10.
Figure 5.11 shows the Z-width plot of the average stability of the system. Sampling rate was kept as 1 KHz and 100 samples were reading by DAQ at each instant. Low resolution encoder was used during the testing. The Figure 5.11 shows the Z-width of the haptic during manipulation of sponge. Dense dots in plot represented a greater no. of virtual walls, or a larger impedance range that the H-UID can behave stably. This plot provided a clear demonstration that virtual damping can play a significant role in increasing the Z-width. The Z-width of the system: the virtual damping was in the range of \(-20 < B_{\text{virtual}} < 20\) and virtual stiffness was in the range of \(-80 < K_{\text{virtual}} < 400\). Z-width of the device is the ratio of the highest impedance interaction to the lowest impedance interaction that the system can stably render [68]. That is, if in a virtual environment, the free areas feel really free, and walls feel really solid/stiff [69]. Therefore, the plot shows that the range of virtual terms where the
high performance from the force reflecting user interfaces can be obtained, which means more realistic walls were judged by the user.

**Figure 5.12. The Plot of Grasping Force Generated vs. Grasping Time during Several Manipulation of Sponge, Sleeve and Block respectively**

In Figure 5.12 of grasping forces vs. time, the results of subject characterization experiment had been plotted to evaluate the force measured by the graspers when grasping 3 different objects of varying stiffness. Sponge, sleeve and plastic block were selected for the experiment that had a significant variation in stiffness and would be easily differentiated with one’s fingers, the magnitude of grasping forces and time taken for complete grasping were recorded. The plot demonstrated that transmission of grasping forces w.r.t. time changed according to the hardness of objects.
In Figure 5.13, the results show that the laparoscopic graspers differentiated between different objects on the basis of their stiffness. The objects were grasped for three different jaw angles, which depend on the thickness of the objects. The Figure 5.7 shows the pictures of grasping three different objects with the same laparoscopic graspers. Thus, the user was experiencing the same deformation while grasping sleeve and sponge but a significantly different grasping force for both objects.

The sponge shows varying forces which depend on the anisotropic nature of the sponge. The sponge sample showed a maximum grasping force of 11.9N while the soft sleeve shown a maximum grasping force of 11.4N and hard block was grasped by a grasping force of 13.1N in the range of 0° and 44° of graspers angle respectively. Thus it proved that the grasper’s ability of identifying the objects of different stiffness had been validated.
Figure 5.14. Transparency of Control System vs. Time

The transparency improved after the local virtual walls creation algorithm was implemented. The results of exploratory experiments show that the transparency improved during interaction with objects. The Figure 5.14 clearly depicts the range of the transparency limits to 0.9 to 1.1 during the manipulation. The degradation of the transparency occurred in case of sponge due to its anisotropic material characteristics. But transparencies were limited to the safe range for plastic block and sleeve respectively. The deviations in transparencies were caused due the effect of varying time delays and control system took time to stabilize the system from moment of inertia of H-UID and graspers assembly.
Figure 5.15. The Plot of Relation between Graspers Jaw Angle and H-UID v2.0

In the Figure 5.15, for different subjects of varying stiffness, the relation between graspers jaw angle and H-UID handle angle was evaluated and the plots were plotted to find exact transfer functions between graspers-UID handle. The handle of H-UID needed to turn less to grasp block due to its thickness and geometry shape and handle of H-UID needed to turn close to $15^\circ$ to get the feel of haptics in the case of sponge and sleeve due their shape geometry and nature of the material. These results show that force feedback improved performance of grasping different types and shapes of objects.
Chapter 6: Summary and Conclusions

This thesis presented several aspects of the haptic feedback system for laparoscopic graspers in \textit{in vivo} robot. The entire haptic feedback system comprised of force sensing forearm, Haptic User Interface Device and its bilateral teleoperation based control architecture. The thesis provided a clear demonstration that the haptic feedback was assisted in reducing the overall grasping forces applied to the tissues and less tissue trauma. It also reduced the tissues palpation time and improved the efficacy of surgeons in performing standard laparoscopic procedures.

The several aspects of design and development of force sensor were discussed and brief research work on laparoscopic force sensing methods was provided. The design of single directional grasping force measurement forearm was evaluated and tested in animal surgeries. The state-of the-art calibration procedure for force sensing forearm was presented. The force sensing forearm established the ability to sense accurate tissue grasping forces and the most efficient graspers actuation mechanism \textit{in vivo} surgical robot till date.

In this thesis, the designs and development of 2 versions of surgical Haptic User Interface Device were discussed. The phantom omni compatible Haptic User Interface Device v2.0 was invented by utilizing the knowledge gained from haptic
paddle; the feature included were compact in size, low inertia and good back-drivability. Initially, 2-CH Impedance-Admittance (Force-Position) bilateral control architecture was implemented and haptic paddle was assigned as a controller for laparoscopic graspers manipulation. The 2-CH control scheme provided a background to build a new 4-CH Impedance-Impedance (Force-Force) based bilateral control architecture. The Force-Force control scheme was most suitable control scheme for \textit{in vivo} surgical robots developed in Advanced Surgical Robotics labs at UNL. A LabVIEW-RT application for 4-CH Impedance-Impedance Control Architecture was discussed. The entire haptic feedback system was tested in couple of animal surgeries. The results were analyzed and overall system capability was evaluated. The entire haptic feedback system established the ability to differentiate the different objects of different stiffness.

In the future, an intuitive, more stable and high back-drivable Haptic User Interface Device will be developed to acquire complete transparent grasping forces towards the surgeon. Improvements in design of force sensing forearm will be continually evaluated to optimize the size and new force sensors will be developed to measure the direct grasping forces. The high sampling rate data acquisition system will be implemented for smooth flow of sensory information in master-slave configuration without time delay.
References


[3] Conversation with Dr. Farritor and Dr. Dmitry, dated February, 2011.


Immersion Corporation. [http://immersion.com](http://immersion.com), accessed July 1, 2010


Conversation with Aaron, member of LabVIEW tech support team


Appendix A. Datasheets of Hardware used in H-UID and Force Sensing Forearm

1. Strain gage: Model #: EA-XX-031EC-120
2. Ultra-Precision Extension Spring : Part # 9044K203

Material: Steel Music Wire

Spring OD. 0.300”

Wire Diameter: 0.055”

Extended Length: 1.9”

Spring Rate: 47.90 lbs./inch
3. Infrared LED : Part # L1939-04
4. Position Sensitive Detector: Part # OD6-6-SO-16

Interlink Electronics FSR™ 400 series is part of the single zone Force Sensing Resistor™ family. Force Sensing Resistors, or FSRs, are robust polymer thick film (PTF) devices that exhibit a decrease in resistance with increase in force applied to the surface of the sensor. This force sensitivity is optimized for use in human touch control of electronic devices such as automotive electronics, medical systems, and in industrial and robotics applications.

The standard 402 sensor is a round sensor 18.28 mm in diameter. Custom sensors can be manufactured in sizes ranging from 5mm to over 600mm. Female connector and short tail versions can also be ordered.

**Features and Benefits**
- Actuation Force as low as 0.1N and sensitivity range to 10N.
- Easily customizable to a wide range of sizes.
- Highly Repeatable Force Reading; As low as 2% of initial reading with repeatable actuation system.
- Cost effective.
- Ultra thin; 0.45mm
- Robust; up to 10M actuations
- Simple and easy to integrate

**Industry Segments**
- Game controllers
- Musical instruments
- Medical device controls
- Remote controls
- Navigation Electronics
- Industrial HMI
- Automotive Panels
- Consumer Electronics

---

**Figure 1 - Force Curve**

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**Figure 2 - Schematic**

---

Interlink Electronics - Sensor Technologies
6. H-UID Motor: Maxon DC Motor # 339150 with Encoder MR # 225778
7. Fishing Line for Capstan Drive: Part # BGQS30C.

![Image of fishing line](image.jpg)

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8. Hall Effect Sensor: A1301EUA

**Continuous-Time Ratiometric Linear Hall Effect Sensor ICs**

**Features and Benefits**
- Low-noise output
- Fast power-on time
- Ratiometric rail-to-rail output
- 4.5 to 6.0 V operation
- Solid-state reliability
- Factory-programmed at end-of-line for optimum performance
- Robust ESD performance

**Description**
The A1301 and A1302 are continuous-time, ratiometric, linear Hall-effect sensor ICs. They are optimized to accurately provide a voltage output that is proportional to an applied magnetic field. These devices have a quiescent output voltage that is 50% of the supply voltage. Two output sensitivity options are provided: 2.5 mV/G typical for the A1301, and 1.3 mV/G typical for the A1302.

The Hall-effect integrated circuit included in each device includes a Hall circuit, a linear amplifier, and a CMOS Class A output structure. Integrating the Hall circuit and the amplifier on a single chip minimizes many of the problems normally associated with low voltage level analog signals.

High precision in output levels is obtained by internal gain and offset trim adjustments made at end-of-line during the manufacturing process.

These features make the A1301 and A1302 ideal for use in position sensing systems, for both linear target motion and rotational target motion. They are well-suited for industrial applications over extended temperature ranges, from -40°C to 125°C.

Two device package types are available: LH, a 3-pin SOT23W type for surface mount, and UA, a 3-pin ultra mini SIP for through-hole mount. They are lead (Pb) free (suffix, -J) with 100% matte tin plated leads.
9. Linear Magnetic Encoder  :  Part  #  AS5304
10. Magnetic Encoder Filtering Board

Schematic Diagram

Printed Circuit Board with all layers

Actual Size of PCB
11. Electric Hardware Schematic for Haptic Feedback System
12. Potentiometer : Part # EVWAE4001B14
Appendix B. Haptic Paddle Dynamic Equations

To activate the haptic paddle, it needed to be analyzed on the basis of dynamic equations:

The governing equations of motion and feedback control are given to understand the dynamic system of the haptic paddle:

\[ J_{eq} \ddot{\theta} + B_{eq} \dot{\theta} + K_{eq} \theta = T \]  \hspace{1cm} (B.1)

Where,

\[ T = \dot{G}(\tau - \tau_f) \]  \hspace{1cm} (B.2)

\[ K_{eq} = -m_c g r_{cs} \]  \hspace{1cm} (B.3)

\[ b_{eq} = b_s + G^2 b_m \]  \hspace{1cm} (B.4)

\[ J_{eq} = (J_s + m_s r_{cs}^2 + G^2 J_m) \]  \hspace{1cm} (B.5)

First order response of motor:

\[ J_m + b_m = -\tau_f \]  \hspace{1cm} (B.6)

\[ \omega(t) = e^{\frac{-b_m t}{J_m}} \left( \omega_0 + \frac{\tau_f}{b_m} \right) \]  \hspace{1cm} (B.7)

Effect of feedback control, after ignoring Coulomb friction:

\[ T_m = K_p \theta + K_v \dot{\theta} \]  \hspace{1cm} (B.8)

\[ J_{eq} \ddot{\theta} + \left( b_{eq} - K_v \right) \dot{\theta} + \left( K_{eq} - K_p \right) \theta = 0 \]  \hspace{1cm} (B.9)
Where,

\[ \theta = \text{Angular Position of handle} \]

\[ \dot{\theta} = \text{Angular Velocity of handle} \]

\[ \ddot{\theta} = \text{Angular acceleration of handle} \]

\[ J_s = \text{Polar moment of inertia of capstan pulley} \]

\[ J_m = \text{Inertia of the Brushed DC motor} \]

\[ r_{cs} = \text{Distance between center of mass of capstan pulley to hinge} \]

\[ b_m = \text{Viscous damping of the brushed DC motor} \]

\[ b_s = \text{Viscous friction of bearing} \]

\[ G = \text{Gear ratio of threaded shaft and radius of capstan pulley (write in ratio)} \]

\[ m_c = \text{Mass of capstan pulley} \]

\[ \tau = \text{Torque applied by the DC motor} \]

\[ \tau_f = \text{Coulomb friction in the motor} \]

\[ g = \text{Acceleration due to gravity} \]
Appendix C.  Haptic Paddle Labs

Courtesy: Rice University

Lab # 1: Haptic Paddle – System Inertia and Sensor Calibration

FURTHER INVESTIGATION OF THE HAPTIC PADDLE SYSTEM

In Part I of this lab exercise, you will characterize the rotational inertia of the Paddle component by the use of a bifilar pendulum. In Part II of this lab exercise, you will learn how to calibrate the system’s position sensor. An Allegro A1322 ratiometric linear Hall-effect sensor, which outputs a voltage proportional to an applied magnetic field, and a permanent magnet are used to determine the Paddle’s angular position. You will notice during the exercise that the magnet has a cylindrical shape. Its poles go from North to South along the cylindrical axis.

A BIFILAR PENDULUM

A bifilar pendulum consists of a mass that is suspended by two vertical strings as seen in Figure 1. The strings have equal length $h$ and are attached to the hanging mass at a distance $D/2$ from the center of mass. The parameter $r_{cm}$ is the distance from the center of mass to the pivot point of the Paddle component. For sufficiently small angular deflections about the axis through the center of mass, the system can be modeled as the second order system:

$$J_{cm} \ddot{\theta} + K_{np} \theta = 0$$

where $J_{cm}$ is the inertia about the center of mass and $K_{np}$ is the equivalent spring constant due to the force of gravity and system parameters. The damping of the system is negligible and therefore ignored. By observing the natural frequency of the system, it is possible to determine $J_{cm}$ for the suspended mass.
Figure 1. A bifilar pendulum with the Paddle component as the hanging mass.

WORKSPACE AND SENSOR CALIBRATION

A robot’s workspace is the space in which it is capable of moving its end effector. In order to move properly, the robot must know where it is within its workspace. Devices called sensors send signals to the robot’s control system that provide information relating to the robot’s position. As stated earlier, a Hall effect sensor and magnet are used to sense the Haptic Paddle’s position. If you wish to learn more about the Hall-effect, information is readily available on the internet. The sensor outputs a voltage in the range of 0-5 volts which varies linearly with the strength of the magnetic field applied perpendicularly to the sensor by the magnet. In order to take full advantage of the analog to digital resolution of the DAQ, the sensor output is input to a signal conditioning circuit (SCC) that uses a series of op-amp stages to linearly amplify the 0-5 volt signal to a ±10 volt signal. The linear relationship between the amplified signal and the strength of the perpendicularly applied magnetic field must be determined in order for the Haptic Paddle control software to properly know the angular position of the Paddle component.
EXPERIMENTAL SYSTEMS AND EQUIPMENT

- Bifilar Pendulum with Paddle component as hanging mass
- Haptic Paddle Sensor with magnet and signal conditioning circuit (SCC)
- Ruler
- Stopwatch
- PXI industrial computer with data acquisition (DAQ) card and connector block

Figure 2. Experimental Set-up

PRE-LAB ASSIGNMENT

1. Derive an expression for the equivalent spring constant $K_{BP}$ in terms of the parameters $m_p$ (the mass of the Paddle component), $g$, $D$, and $h$. Assume that the vertical motion of the Paddle component is negligible. Hint: when you rotate the Paddle about the axis through its center of mass, you can draw a pair of right triangles relating the current positions of the string and the length $D/2$ to their respective equilibrium positions. Note that these two triangles have a common leg.

Result: $K_{BP} = m_p g D^2/4h$

2. Use the previous result to derive an expression for $J_{cm}$ in terms of the same parameters and the natural frequency $\omega$.

3. Use the previous result to derive an expression for the moment of inertia $J_p$ about the axis through the Paddle’s pivot point.
LAB PROCEDURES

Part I: Characterizing the Rotational Inertia of the Paddle Component

a) Place a piece of masking tape on the Paddle (a spare Paddle is provided so that you do not have to disassemble your Haptic Paddle) such that it goes from the handle, over the pivot point, and to the bottom of the sector pulley.

b) Balance the Paddle on the bar of the bifilar pendulum to determine the approximate center of mass. Then mark two points, each at an equal distance (D/2) from the center of mass, at which you will hang the Paddle. Record the distance D. Record the distance r_cm from the center of mass to the pivot point. The mass of the Paddle is 57.2 grams.

c) Tape the yarn to the Paddle at the marked points and use binder clips to adjust the length and position of the strings. The strings should be equal in length and hanging vertically. The Paddle should be flat in the horizontal plane. Measure the length h of the strings from the binder clips to the top of the Paddle.

d) Deflect the Paddle to an initial angular displacement about its center of mass (less than 30° so that the small angle approximation holds) and let it go to allow it to swing freely. Record the time it takes to make 20 oscillations. Repeat this twice more.

e) Remove your Paddle from the string and discard the used tape.

Part II: Position Sensor Calibration

a) Assemble the Haptic Paddle as detailed in the Haptic Paddle document for the Hall Effect Calibration lab.

b) From the signal conditioning circuit (SCC), connect the 3-wire SIP terminal to the sensor pins on the Haptic Paddle. Ensure that the “Up” label on the connector is Up.

c) The channels named in this step refer to the terminal positions on the DAQ connector block. From the SCC, connect the wire with the flag “All” to channel 68 on the connector block and the wire with the flag “GND” to channel 67. These connections provide for the DAQ to read the amplified sensor output. Have the TA come and check your setup. Turn on the power to the SCC.

d) Open the “Hall Effect Calibration Lab” from the “Haptic Paddle Labs” folder on the Desktop. Run the VI by hitting the white arrow in the upper left corner.

e) The VI has a readout for the voltage output from the SCC. Record 10 data points of the voltage readout versus the Paddle position in degrees. A few of the Paddle positions have been marked along the sector pulley (30°, 0° and -30°).

f) Use Microsoft Excel to do a linear approximation of your data by plotting Paddle position vs. signaled voltage (chart data with scatter plot and add trend line, check option to show equation).

g) Enter the values you determined for A (slope) and B (y-intercept) into the VI. Compare positions of your Paddle to the Position readout in the VI to confirm the calibration of the sensor.

h) Hit the stop button to stop the VI. Turn off the power to the SCC. Save your calibration (Excel) file to your group’s folder on the desktop.

i) Please detach the wires from the Hall mount and the DAQ connector block and store your Haptic Paddle.
RESULTS TO REPORT

- Provide the average value of $J_{cm}$ determined by the use of the bifilar pendulum. Use this result to determine $J_p$.
- Provide a plot of Paddle position vs. signal voltage for the sensor calibration. Include the linearized relationship you determined.

ADDITIONAL ITEMS TO ADDRESS IN THE DISCUSSION SECTION

- Why is it necessary that the initial displacement of the hanging mass of a bifilar pendulum not be too large?
- The background information for this exercise explains that the amplified Hall-effect signal varies linearly with the strength of the perpendicularly applied magnetic field. However, you observed that the amplified signal varies linearly with the Paddle angular position. Explain how the perpendicular component of the applied magnetic field is linear with the Paddle angular position. Use figures to assist your explanation. Your figures should include the magnet and its flux lines, the sensor, and show angle of displacement.

REFERENCES

Lab # 2: Haptic Paddle – Actuator Characteristics

THE HAPTIC PADDLE

The Haptic Paddle is a low cost, single degree of freedom force feedback joystick capable of providing a peak force of about 10N at its handle. It is an ideal tool for the demonstration of electromechanical system properties and concepts covered in undergraduate engineering courses. Over the course of the remaining lab exercises, you will investigate components of the system so that you can better understand your interaction with the complete system in the final exercise.

THE HAPTIC PADDLE ACTUATOR

In this lab exercise, you will characterize parameters of the Haptic Paddle actuator. Specifically, you will characterize the actuator’s viscous damping in Part I and torque constant in Part II. Part I of this exercise provides an example of a homogeneous first order electromechanical system. A Pittman LO-COG® 9434 15.1V DC motor provides actuation for the Haptic Paddle system.

BASICS OF A DC MOTOR

DC Motors are one of the most widely used actuators in industry. A DC motor is effectively a torque transducer that converts electric energy into mechanical energy. In electrical circuits, DC motors are often modeled as a voltage source and a resistance in series as shown in Figure 1. The electromechanical component of the motor that is modeled by this voltage source and resistance is known as the armature. Current that flows through the armature from the positive to negative lead will generate a torque on the motor rotor that acts in the positive direction of rotor spin, and vice versa for current that flows from the negative to positive lead. This torque is expressed as
where \( T \) is the motor torque (Nm), \( K_i \) is the torque constant of the motor (Nm/A), and \( i_{arm} \) is the armature current (amperes). \( K_i \) is sometimes listed as \( K_i \) in motor data sheets. The modeled voltage source is commonly called a back Electro Motive Force (EMF) and will have the same sign as the velocity of the rotor. This back EMF is expressed as

\[
V_{emf} = K_v \omega
\]

where \( V_{emf} \) denotes the back EMF (volts), \( K_v \) is the speed constant or voltage constant (V*sec/rad) and \( \omega \) is the rotor velocity (rad/sec) of the motor. \( K_v \) is sometimes listed as \( K_e \) in motor data sheets.

Equations (1) and (2) form the basis of DC motor operation.

**Viscous Damping and Dynamic Braking**

The viscous damping of the motor is due to lubricants in its ball bearings. The equations presented subsequently make the assumption that this damping is the dominant resistance compared to other disturbances, e.g. coulombic friction. To ensure this assumption is valid, we will enhance the viscous damping by applying a dynamic brake. Dynamic braking is applied by connecting a spinning DC motor to a load resistor \( R_l \) so that the load resistor will dissipate the kinetic energy stored in the rotor and the load inertias. Figure 1 shows the dynamic braking circuit used for this exercise.

![Figure 1. Dynamic Braking Circuit](image-url)
The switch shown in the top of Figure 1 toggles between the spin-up (position 1) and spin-down (position 2) states for the motor. The DC motor is modeled as a voltage source and a resistance, $V_{emf}$ and $R_a$ respectively, in series. $R_a$ is sometimes listed as $R_i$ in motor data sheets. $V_c$ is the control voltage sent to the motor by the amplifier. Measuring devices are used to measure the armature voltage and current, $V_{arm}$ and $i_{arm}$ respectively. The armature current is measured only when the switch is in the spin-up position. When the switch is in the spin-down position, the back EMF of the spinning rotor produces a current of

$$i_{arm} = \frac{V_{emf}}{R_i + R_a} = \frac{K_i \omega}{R_i + R_a}$$  \hspace{1cm} (3)

The current in turn causes a braking torque of

$$T = K_i i_{arm} = \left( \frac{K_i}{R_i + R_a} \right) \omega = B_{dp} \omega$$  \hspace{1cm} (4)

Equation (4) shows that the braking torque is proportional to the motor velocity, which is a property of viscous damping. Also, it can be seen from (4) that the damping effect can be easily adjusted by changing the load resistor.

**TRANSMISSION OF MOTOR POWER TO END EFFECTOR**

Robotic machines, such as haptic devices, often use transmissions to transmit power from their actuators to their end effectors. An end effector is a part of the robot that interacts with its surrounding environment or a user. The Haptic Paddle employs a Capstan cable drive to transmit the power of its actuator to affect a force at the Paddle handle where the user grasps. The cable drive effectively acts as a gear transmission to impart a torque on the Paddle component about its pivot point. This torque is then levered about the pivot point to produce the output force at the handle.

**EXPERIMENTAL SYSTEM AND EQUIPMENT**

- The Haptic Paddle Actuator with attachments including a dynamic brake and load mass for part I and the Haptic Paddle transmission and mass attachments for part II
- Components of the Haptic Paddle system including the Power Amplifier (Amp)
- Various tools for assembly of Haptic Paddle
- PXI industrial computer with 6070E data acquisition (DAQ) card and connector block
- Fluke DMM
- Calipers or rulers
- C-clamp

Figure 2 on the next page gives an overview of the experimental set-up for Part I. The experimental setup for Part II is the same with the exception of the actuator and its attachments. Figure 3 gives an overview of the actuator setup for Part II. Note that the Capstan Spool, Paddle pivot point, and the hole in the Paddle handle are all inline.
Figure 2. Experimental Set-up for Part I

Figure 3. Experimental Set-up for Part II
**Pre-Lab Assignment**

**Part I**

1. Consider the dynamic braking circuit. For the switch in the spin-down position, draw the free body diagram of the motor rotor and derive the equation of motion for the system considering the following variables and parameters (coulombic friction is ignored):
   - $J_{spin}$, the sum of the inertia of the cylindrical mass and the inertia of the motor rotor (provided in the motor spec sheet)
   - $B_{spin}$, the sum of the damping due to dynamic braking, $B_{DB}$, and the damping due to lubricants in the bearings, $B_{m}$
   - $\omega$, the angular velocity of the motor

2. Solve the equation of motion to determine an expression for $\alpha(t)$ in terms of $J_{spin}$ and $B_{spin}$, assume $t_0 = 0$.

3. What is $\tau$, the time constant, in terms of $J_{spin}$ and $B_{spin}$? When $t = \tau$, what is the numerical value of the ratio $\alpha(t)/\alpha(0)$?

4. For the switch in the spin-down position, derive the equation for $V_{emf}$ in terms of $V_{arm}$, $R_{in}$, and $R_{s}$.

**Part II**

5. Draw the free body diagram of the Capstan Spool-Paddle coupling considering only moments about the pivot point and the following variables and parameters (variables and parameters not included are considered negligible):
   - $F_w$, the weight of the mass attached at the Paddle handle
   - $T_m$, the motor torque generated
   - $r_{cs}$, the radius of the Capstan Spool
   - $r_{ps}$, the radius of the Paddle sector pulley (center of pulley is Paddle pivot point)
   - $l_p$, the distance from the Paddle pivot point to the center of hole in the handle

6. What is the relationship $N$ describing the effective gear ratio of the transmitted motor torque via the cable drive? Assume no slipping of the cable.

7. Under static conditions where the Capstan Spool, Paddle pivot point, and handle are all inline, derive the equation for $T_m$ in terms of the other parameters and variables listed.

**Laboratory Procedure**

Have your pre-lab assignment initialed by the TA.

It is suggested that you read the Results to Report section before carrying out the lab procedure.

**Part I: Characterizing the viscous damping**

a) Assemble your Haptic Paddle as detailed for this part of the exercise in the Haptic Paddle Kit handout and then C-clamp it to the table.

b) Take appropriate measurements of the provided mass for determining its rotational inertia.
   For the purposes of this exercise, assume that the Capstan Spool, jam nut and mass are a single homogeneous disk. The mass of the Capstan Spool and jam nut together is 18 g.

c) The black dots on Figure 1 represent banana jacks. Use the wire with two banana plugs to connect the brake to the Fluke DMM at the appropriately labeled banana jack on the brake and the “10A” jack on the DMM. Connect the red wire labeled “Motor Power” from the
Power Supply and Amplifier (hereafter Amp) to the appropriately labeled banana jack on the brake. Connect the black wire labeled “Motor Power” from the Amp to the “Com” jack on the DMM. Use the red and black motor wires to connect the motor to the dynamic brake at the appropriately labeled banana jacks.

d) The channels named in this and following steps refer to the terminal positions on the DAQ connector block. Connect the wire from the Amp with the flag “AO” to channel 22 on the connector block, and the wire with the flag “GND” to channel 55. These connections provide for the DAQ to send a control signal to the Amp. Use an alligator clip and jumper wire to connect the red motor banana plug to channel 68. The hole on the side of the banana plug provides a good place to bite with the alligator clip. Be sure that the alligator clip has a good connection to the jumper wire. Similarly, connect the black motor banana plug to channel 34. These connections provide for the DAQ to read $V_{arm}$.

e) To control the system, you will use a LabVIEW virtual instrument (VI). Open “Actuator Characteristics Lab” in the folder “Haptic Paddle Labs” from the Desktop. This VI enables you to control $V_c$ as well as reading and recording $V_{arm}$. Be aware that the input “$V_c$” is not an actual measurement of $V_c$.

f) Before you turn on any power button, make sure the switch on the dynamic brake is flipped to the spin-down position. **Have the TA check your set-up before proceeding.** Then, turn on the Amp.

g) Run the VI by clicking the white arrow in the upper left corner of the screen. Make sure that “$V_c$” is set to zero and press “Motor Power” (it should turn green).

h) Flip the switch on the dynamic brake to the spin-up position and set “$V_c$” to 9.5 volts. Once the motor speed has stabilized, flip the switch on the brake to the spin-down position to engage the brake and observe the response. It should be a first-order decay.

i) Set “$V_c$” back to zero.

j) Now you will record the first-order decay you just observed. Flip the switch on the brake to the spin-up position. Again, set “$V_c$” to 9.5 volts. Once the motor speed has stabilized, flip the switch on the brake to the spin-down position. After the motor has stopped spinning and while the full decay is still in view on the plot, hit “Pause Data”. To save this data in an Excel file, hit the folder icon next to “File Path”, select a location (for example desktop) and give a name to your file with an extension of .xls, and hit OK. Then hit the “Write Data” button.

k) Set “$V_c$” back to zero and turn off the power to the Amp.

l) Check your data file to ensure your data is good (shows exponential decay). Save your file to your USB flash drive or email it to the group members and delete the file from the computer.

**Important:** $V_{arm}$ and time are recorded in the file – it should be apparent which is which.
Part II: Characterizing the Torque Constant

a) Disconnect the motor from the circuit. Remove the C-clamp and then remove the mass. You will need to connect the motor leads to each other in order to remove the mass.
b) Assemble your Haptic Paddle as detailed for this part of the exercise in the Haptic Paddle Kit handout and then C-clamp it to the table. Reconnect the motor to the circuit. The dynamic brake is not used for this part of the exercise; it simply provides a means of connecting the motor to the circuit.
c) Measure and record \( r_p \) and \( l_a \). \( r_e \) is difficult to measure because of the threads on the Capstan Spool – it’s value is 4.3 mm.
d) Use a minimum of 250 gm to apply a known torque to the Paddle by attaching the provided weights (mass of weight is stamped on it) to the hole in the Paddle handle. The mass of the hex bolt with the washer and spacer is 22 g. The mass of the thread and spacer is 12 g.
e) Turn the Fluke DMM to the ammeter setting. Be sure that it is reading DC current. Press the yellow (Fluke 175 model) or blue (Fluke 83III model) button if the display shows AC. If the DMM turns off during the exercise, press this button to turn it back on.
f) Turn on power to the Amp. Flip the switch on the brake to the spin-up position. Determine the minimum amount of current required to balance the weight by adjusting “\( V_c \)” in the VI (current through the motor is directly proportional to \( V_c \) in static conditions). **START WITH A SMALL VOLTAGE \( V \approx 0.5 \text{ V} \).** Record the current reading on the DMM once you balance the Paddle in a horizontal position. Set “\( V_c \)” to zero after you record the current.
g) Collect 4 more data points of weight applied versus the current required to balance it. Remember to use a minimum of 250 g and to set “\( V_c \)” to zero after each recording.
h) Use a scatter plot in Excel to ensure your data is linear.
i) Turn off the power to the Amp and disassemble the circuit. Use masking tape to label your Haptic Paddle with your team’s name and put it away in the proper drawer.

**RESULTS TO REPORT**

- Provide a plot of your spin-down test with time on the x-axis and rotor velocity (rad/sec) on the y-axis. Be judicious in your choice of \( \omega(0) \) so that it is a good representation of the initial velocity. After you select your initial data point, time shift your data so that \( t_0 = 0 \). There are a significant number of data points for this plot, thus it is only necessary to show a sample calculation as to how you achieved your plotted values.
- Use the ratio \( \frac{\omega(t)}{\omega(0)} \), calculated as a part of the pre-lab assignment, and your plot of \( \omega \) vs. \( t \) to mark the point \( t = \tau \) on the plot and give a numerical value of \( \tau \).
- Determine a numerical value of \( B_{spin} \). Include units. Also determine a numerical value of \( B_m \) the damping constant of the motor.
- In Matlab, generate an array of points \( \alpha(t) \), where \( t \) varies from 0 to 3s, incremented at 0.03 seconds. Use the numerical values you determined for \( B_{spin} \) and \( J_{spin} \) to generate this array. Provide a second plot with your new \( \alpha(t) \) over your original data. It is not necessary to include your array of points; a sample calculation for a single point is sufficient.
- Provide a plot showing motor torque (Nm) vs. current for the data you collected. Determine \( K_t \) from the plot.
- Convert \( K_t \) provided in the datasheet to have the same units as \( K_r \). In your discussion, justify the relationship you observe between \( K_t \) and \( K_r \). Hint: consider power.
ADDITIONAL ITEMS TO ADDRESS IN THE DISCUSSION SECTION

• Explain why you were directed to connect the motor leads together to assist in attaching the mass.
• In what ways are the first-order dynamic system and its time response observed in this exercise analogous/dissimilar to the first-order dynamic system and its time response of the RC circuit observed in Lab #2?
• In what ways is the time constant observed in this exercise analogous/dissimilar to the time constant of the RC circuit observed in Lab #2?
• Motors commonly have a 10% tolerance for their voltage and torque constants. Compare your derived value for $K_i$ to the motor datasheets.

BONUS DISCUSSION

• Explain the sudden drop in $V_{arm}$ when you flipped the switch on the dynamic brake.
• With the switch in the spin-up position, how could you experimentally characterize the voltage constant? Present the equations necessary to do this, as well as the outputs that need to be recorded.

REFERENCES

Lab # 3: Haptic Paddle - Virtual Systems and Teleoperation

THE COMPLETE HAPTIC PADDLE ELECTROMECHANICAL SYSTEM

In Part I of this lab exercise, you will explore virtual systems displayed on the Haptic Paddle. One of the virtual systems is a second order system implemented through the use of PD control. You will analyze this system through the use of the logarithmic decrement method. In Part II of this lab exercise, you will experience teleoperation of the Haptic Paddle.

Recall the following parameters and variables of the Haptic Paddle system from previous exercises:

- \( r_{cm} \), the distance from the Paddle component’s center of mass to the pivot point
- \( l_h \), the distance from the Paddle component’s pivot point to the center of the hole in the handle
- \( r_p \), the radius of the Paddle component’s sector pulley
- \( J_p \), the inertia of the Paddle component about the pivot point
- \( m_p \), the mass of the Paddle
- \( r_{CS} \), the radius of the Capstan Spool
- \( J_{rotor} \), the inertia of the motor rotor
- \( B_m \), the damping in the motor
- \( R_a \), the armature resistance of the motor
- \( T_m \), the applied torque from the motor
- \( K_t \), the torque constant of the motor

You will also need to consider the following variables:

- \( \theta_p \), the angular displacement of the Paddle component from vertical. The positive direction is defined to be clockwise.
- \( \theta_m \), the angular displacement of the motor rotor. The zero position and positive direction are defined to be consistent with \( \theta_p \).
- \( x \), the horizontal component of the displacement of the hole in the Paddle component’s handle. The zero position and positive direction are defined to be consistent with \( \theta_p \).
- \( V_c \), the control voltage sent from the motor amplifier to the motor.
Lumped parameter modeling can be used to derive the equation of motion (EOM) for the Haptic Paddle mechanical system as

\[ m_{eq} \ddot{x} + b_{eq} \dot{x} + k_{eq} x = f_m \]  

(1)

where the variable \( x \) is displacement defined above. The parameters \( m_{eq}, b_{eq}, \) and \( k_{eq} \) are the equivalent lumped parameters experienced with respect to \( x \) due to inertial, damping, and stiffness elements of the mechanical system. The stiffness element of this system is due to the force of gravity. The force \( f_m \) is the equivalent lumped force experienced with respect to \( x \) due to the input motor torque. To effect a virtual second order system, a LabVIEW Virtual Instrument (VI) has been programmed to control \( f_m \) such that the EOM becomes

\[ m_{eq} \dddot{x} + b_{vin} \ddot{x} + k_{vin} (x - x_{sp}) = 0 \]  

(2)

where \( b_{vin} \) and \( k_{vin} \) are virtual parameters entered into the LabVIEW VI. A set-point value, \( x_{sp} \), is input to the system and the control scheme outputs \( f_m \) such that it is equal to the sum of a force proportional to the difference of the position and the set-point, a force proportional to the derivative of the position, and a force to balance the restoring force of the system. This control of \( f_m \) is known as proportional-derivative (PD) control and is commonly used for position control. The VI has also been programmed to display haptic virtual environments in which the user moves the Paddle to input a position and the VI controls the motor to output a force corresponding to the interaction with the virtual environment.

**TELEOPERATION**

Teleoperation is the control of a machine from a remote location. Often times, the machine being teleoperated is a robot. In cases where the teleoperator, or input device, is also a robot, the teleoperator is known as the “master” robot and the teleoperated robot is known as the “slave” robot. The teleoperation scheme implemented in this exercise is depicted in the following figure. The master is the ground reference for the slave. The virtual spring connecting the slave to the master is relaxed when \( x_r \) equals \( x_{sp} \).

![Figure 1. Master – Slave Teleoperation](image)

If the master is a force feedback manipulator, such as the Haptic Paddle, then the user can haptically feel the remote environment of the slave via the master. Haptics enables this unique bimodal communication (communication goes both ways) and both robots are in effect a master and a slave grounded through their environment or user. The haptic teleoperation scheme implemented in this exercise is depicted in the following figure. A VI has been programmed to enable you to explore teleoperation of the Haptic Paddle.
LABVIEW REAL-TIME OPERATING SYSTEM (RTOS)

Fast, even-timed calculation loop rates are critical to maintaining proper haptic environments. This exercise employs the LabVIEW RTOS to run loop rates faster than that achievable within LabVIEW for Windows. You will use a host computer running Windows to open and download a VI to the client RTOS machine. The VI will then run on the client machine and communicate with the host machine. The client machine also performs the data acquisition.

THE LOGARITHMIC DECREMENT METHOD

The logarithmic decrement method allows you to experimentally determine the values of $\zeta$ and $\omega_n$ for an under-damped system. The procedure is as follows:

a) Choose a section of your graph to work with. This section should go from one peak to another and include several cycles.

b) The logarithmic decrement is found using the equation $\delta = \frac{1}{n} \ln \left( \frac{x_i}{x_{i+n}} \right)$, where $n$ is the number of cycles between peaks while $x_i$ and $x_{i+n}$ are the amplitudes of the peaks.

c) The damping ratio is then given by $\zeta = \frac{\delta}{\sqrt{(2\pi)^2 + \delta^2}}$

d) Find the damped natural frequency, $\omega_d = \frac{2\pi}{p}$, where $p$ is the period (time/cycles)

e) Then use $\omega_d = \omega_n \sqrt{1 - \zeta^2}$ to find $\omega_n$.

PRE-LAB ASSIGNMENT

1. Derive the linear relationship for both $\theta_p$ and $\theta_m$ and in terms of the system parameters and the variable $x$ listed on the first page of this handout.
2. From equation (1), derive relationships for $m_{eq}$, $b_{eq}$, and $k_{eq}$ in terms of the system parameters. It is helpful to use energy methods in determining these relationships. Also derive the relationship for the input $f_m$ in terms of the system parameters and the input $T_m$.
3. Use these relationships to calculate numerical values for the equivalent parameters. Use metric units.
BONUS PRE-LAB ASSIGNMENT

1. Derive the expression for the controlled \( f_m \) used to achieve equation (2) in terms of the equivalent parameters, the virtual parameters, \( x \) and its derivative, and \( x_{sp} \).

2. Use this expression to derive the relationship for \( V_e \) in terms of the equivalent parameters, the virtual parameters, the system parameters, \( x \) and its derivative, and \( x_{sp} \).

EXPERIMENTAL SYSTEM AND EQUIPMENT

- Virtual systems displayed on the Haptic Paddle (x2 for Part II)
- PXI industrial computer running Windows
- PXI industrial computer running LabVIEW RTOS with 6070E data acquisition (DAQ) card and connector block
- Haptic Paddle Power Supply and Amplifier (x2 for Part II)
- Signal conditioning circuit (SCC) (x2 for Part II)

EXPERIMENTAL PROCEDURE

Part I: Virtual Systems

a) From the SCC, connect the 3-wire SIP terminal to the sensor pins on the Haptic Paddle. Ensure that the “Up” label on the connector is up. Also connect the following wires to the connector block: flag “A1” to channel 66, flag “A2” to channel 68, flag “GND” to channel 67. Use a short wire to make a connection from channel 67 to 34. These connections allow for the DAQ analog input channels 0 and 9 to read the amplified sensor voltage and its derivative, respectively.

b) From the Amplifier, connect the following wires to the connector block: flag “AO” to channel 22, flag “GND” to channel 55. These connections allow for the DAQ to send control voltages to the Amplifier. Also connect the red and black banana plugs to their corresponding banana plugs on the Haptic Paddle.

c) Have the TA check your setup. Turn on the power to the power supply and the NI-ELVIS protoboard.

d) On the host machine, open the “Virtual Systems Project” project in the “Haptic Paddle Labs\Virtual Systems Lab” folder from the Desktop.

e) In the project viewer, open the file “Virtual Systems Lab” for the appropriate remote system. Your client machine is labeled with its IP address.

f) Run the VI by hitting the white arrow in the upper left corner. It will take a few seconds to download the VI to the client machine.

g) Swing the Paddle to the extents of its workspace to make sure that you do not have saturation of the DAQ. Adjust the magnet cap and sensor position as necessary.

h) Ensure that your calibration of the Hall Effect sensor done in Lab #5 is correct. If not, recalibrate (5 points are sufficient).

i) Enter values for the equivalent system parameters \( b_{eq} \) and \( k_{eq} \) that you determined as part of the Pre-lab assignment. Also enter the value of the torque constant (in Nm/A) that you characterized in Lab #4.
j) Enter values into the controls for $k_{\text{ctrl}}$ and $b_{\text{ctrl}}$. Values of 200 N/m and 5 N*s/m are good initial values. Turn on the “Force-feedback” button.

k) Make sure no one is holding the Paddle and change the set-point control “s_sp (cm)” to a different value. Observe the response of the Paddle.

l) Experiment with different values of $k_{\text{ctrl}}$ and $b_{\text{ctrl}}$ to achieve underdamped behavior which has at least a few oscillations before the motion settles to the set-point. Control the Paddle’s position so that it moves from a position of 1.5 cm to −1.5 cm, and then back to 1.5 cm. You may observe different behavior when the Paddle moves left-to-right as opposed to when it moves right-to-left. Once you have achieved adequate underdamped behavior, adjust $b_{\text{ctrl}}$ to achieve critically and then overdamped motion. For each type of response, note your values of $k_{\text{ctrl}}$ and $b_{\text{ctrl}}$ as you discover them. Do not attempt undamped motion!

m) Data files containing various observed responses will be emailed to you by the TAs.

n) Explore the other haptic environments. The controls for each environment are listed in the VI.

Part II: Teleoperation

a) The Haptic Paddle of the team at the teleoperation station will be the Paddle A. The setup for this Haptic Paddle does not need to change.

b) From the Paddle B SCC, connect the 3-wire SIP terminal to the pins on the slave Haptic Paddle. Ensure that the “Up” label on the connector is up. Also connect the following wires to the Paddle A connector block: flag “A11” to channel 63, flag “A12” to channel 65, flag “GND” to channel 64. Use a short wire to make a connection from channel 64 to 31.

c) From the Paddle B Amplifier, connect the following wires to the Paddle A connector block: flag “AO” to channel 21, flag “GND” to channel 54. Also connect the red and black banana plugs to their corresponding banana plugs on the Paddle B. Have the TA check your setup. Turn on the power to the Paddle B power supply.

d) In the project viewer, open the file “Teleoperation Lab” for the appropriate remote system. The client machine is labeled with its IP address.

e) Enter the previously determined calibration constants for Paddle B. Recalibrate Paddle B.

f) Turn on the “Paddle A Dynamics Cancellation” button. Enter 5 N*s/m for Paddle A $B_{eq}$ value. Move Paddle A back and forth once its power supply is on, and once it is off. Keep increasing $B_{eq}$ by increments of 5 N*s/m until it feels approximately the same for power supply on and off cases. Don’t exceed 35 N*s/m.

g) Repeat step f) for Paddle B

h) Enter values for $k_{\text{ctrl}}$ and $b_{\text{ctrl}}$ and explore teleoperation of the Haptic Paddle. 50 N/m and 5 N*s/m are good initial values. Turning on the “Paddle A Master” button with the “Paddle B Master” button off will make Paddle A the master and Paddle B the slave. Turning on both master buttons enables Haptic teleoperation.

RESULTS TO REPORT

- Produce graphs for each type of system response including the values of $k_{\text{ctrl}}$ and $b_{\text{ctrl}}$ that were used to produce each response. Include both the left-to-right and right-to-left
motions with a time shift so that $t_0 = 0$ for both motions. Remove the extraneous data points from before motion was initiated and after the motion settled.

- For the under-damped system, calculate $\zeta$ and $\omega_n$ from the graph using the log-decrement method.
- In the discussion section of your report, discuss haptic systems, haptic environments you explored with the haptic paddle in the first part of the lab and both modes of teleoperation you have experienced in the second part of the lab. Also, if you have observed a discrepancy between the theoretical and experimental results, discuss the possible reasons.

REFERENCES: