Modeling and development of haptics device for therapeutic gastrointestinal endoscopy

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Modeling and Development of Haptics Device for Therapeutic Gastrointestinal Endoscopy

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Abstract

Haptics can improve robotic surgery performance by providing force feedback to the surgeon. Its importance and necessity have been proven by many researchers as they look into providing force feedback to the robotic surgery system. Especially in the **MIS** (Minimally Invasive Surgery) and **NIS** (Non-invasive Surgery) robotic surgical system, it is essential to provide haptics information to the surgeon for a better performance. The goal is to let the surgeons feel as though they are directly operating on the patient rather than having a robot mediate the interaction. To incorporate force feedback to the machine, one way is to design a physical force sensor on the robotic tools to determine how much force is being applied; the other way is to develop a mathematical force model to estimate the forces between the patient and the robot. This project focuses on providing force feedback to develop no scar robotic gastrointestinal endoscopic surgery. The size of the robotic manipulator and the nature of the GI tract bring difficulty in building a proper sensor system. This report presents a mathematical force model for the viscoelastic behavior of the stretchable stomach under large deformation. Experiments have been performed to characterize and validate the model. This model is developed to provide force feedback in the master-slave robotic system. A one-degree-of-freedom master-slave system is built to prove the feasibility of the application. Furthermore this model can be used in the design of virtual environment for simulation of therapeutic gastrointestinal endoscopy. A simple VR (Virtual Reality) model is also presented in this report.

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List of Glossary

MIS — **Minimally Invasive Surgery** should have less operative trauma for the patient than an equivalent invasive procedure. It may be more or less expensive. Operative time is longer, but hospitalization time is shorter. It causes less pain and scarring, speeds recovery, and reduces the incidence of post-surgical complications.

NIS — **Non-Invasive Surgery** means the surgical procedure does not penetrate or break the skin or a body cavity, i.e. It doesn't require an (invasive) incision into the body or the removal of biological tissue.

Endoscopy — it means looking inside and typically refers to looking inside the human body for medical reasons using an instrument called an endoscope. Endoscopy can also refer to using a borescope in engineering and technical situations where direct line-of-sight observation is not feasible.

GI Tract — **Gastrointestinal tract** is also called the digestive tract. It is the system of organs within multicellular animals that takes in food, digests it to extract energy and nutrients, and expels the remaining waste.

Chapter 1 Introduction

1.1 Background

To explore more things in the world that human beings cannot reach due to our physical limitation, various systems have been invented to provide human with the sensation of sight and hearing. These systems have been perfected for decades and people can acquire audio and video information from them. Nowadays, system existing currently and others are being further developed with a third sensation: haptics, which means the sense of touch.

Haptics has undergone rapid development in the past 10 years. It has been applied in many fields, like Virtual Reality, Robotics, Computer Graphics, Neurosciences, Systems and Control, etc. It relates to the ability to receive proprioceptive, tactile, and force feedback from the simulation in response to the user's action. Such feedback is mediated by haptic interfaces[1]. Haptic interfaces provide information to people by manipulating them; the information is a reflection of sensed forces by a remotely controlled slave robot or by a computer simulation. In recent usages the definition has been broadened to include any force application to any part of the body by a robot device[2]. From the usage of the application, haptic system can be generally categorized into three groups: virtual reality simulation, tele-presence and human-assistive. These three groups of systems were discussed in the book “Biologically Inspired Intelligent Robots” 4th chapter[3]. Until recently, haptics systems only existed as demonstrations in research facilities. However, while research is still continuing, consumer-level haptics systems have been introduced. For example, force feedback gaming devices, such as joysticks and computer mice, have become available, while in the medical field, tele-surgery or surgeon directed robotic surgery has been gaining recognition[3].

From another aspect, robots have been widely used in many industrial fields, but only till late 1990's, the idea of robots performing surgery on men was materialized[4]. Robots surgery can overcome some natural limitations of men. It can perform in some operations which seem impossible for human beings. For example, robots can improve the accuracy and stability of surgeons by scaling large movements made by the surgeons down to micro motion[5]; end-effectors and actuators of robots can be made very small allowing complex and agile movements; surgeons may not even be physically present in performing the surgery, thus promoting tele-operation[6]. Since the data processing and decision making power that a human is capable of far surpasses even the most advanced computer system, a human operator can control a robot in unknown or difficult situations more effectively than any computer program[3]. Here haptics information is helpful to the dexterity enhancement for the surgery, because it is the robot interacting with the patient instead of surgeon's hands. The use of force feedback would allow more accurate instrument positioning, so that the visual acuity, dexterity and tactile sensitivity would help the surgeon to define the limits of the microsurgical procedures[7]. Surgical robotics has already been implemented in laparoscopic, thorascopic, endoluminal and arthroscopic interventions[8]. Researchers all over the world continue to develop more robots to assist surgeons. The implementation of endoscopy, allows **Minimally Invasive Surgery** to be achievable since the doctors have access to real time patient image, or video of the operating site from endoscopes[9]. One of the few commercialized **MIS** systems is the famous da Vinci Surgical System from Intuitive Surgical Inc. However no practical surgical robot systems embedded with haptics have been utilized in operation theatres. This application is still under research level, and still requires improvement. Several surgical robot prototypes with haptic system used for **MIS** have been used for research [7, 10-12].

1.2 Motivation

Endoscope acts a very important role in **MIS**. It is made up of optic fiber and coupled with a miniature camera in a thin rigid or flexible tube. One particular type of endoscope for monitoring the stomach, throat and large intestines is the gastro-intestinal endoscope. This flexible endoscope can be used to inspect, diagnose and treat various pathologies in the gastrointestinal (GI) tract. Due to its flexibility, it can pass through the natural narrow and winding openings in the human body such as the colon or esophagus to provide the visual pictures and proper treatment inside the human body. A typical endoscope used in the gastrointestinal (GI) tract is shown below:

Figure 1-1 A Layout of Typical Gastrointestinal Scope

As shown in the figure above, a typical endoscope includes an ultra compact CCD camera, light source and a channel for infusing or withdrawing liquid or gas from the patient's body. Also there are multiple tool channels that can be used to pass instruments used for electrosurgery, cautery, and for cutting and grasping. The use of such devices has enabled inspection and treatment of abnormalities within the GI tract to be achieved without the need for open surgery. Therefore, this inspires people that

gastrointestinal scope can be seen as a potential platform to perform NIS (Non-Invasive Surgery). Greater than MIS, NIS does not even require penetration into the body through any cuts or incisions to implement the treatment. Because there is no incisions on the patient body, it shortens the recovery time and has no scar left.

Figure 1-2 Endoscopy Instruments

However, current manual manipulative tools available for the endoscopist have limited maneuverability, as some examples from Figure 1-2 shown above. As an innovative solution, a multi-degrees-of-freedom master-slave robotic system for no scar robotic Gastrointestinal surgery has been developed by our group[13]. The developed master slave robotic system employs the tendon-sheath actuation method to control the slave manipulator. The degree-of-freedom of slave manipulator is designed from human arm. There are 4-DOF for each slave manipulator, 1 more DOF for the gripper on one manipulator. The degrees-of-freedom can be seen from the Figure 1-3, and it also clearly shows that the design origins from human arm anatomy. The master controllers are designed to follow the kinematic structure of the slave as closely as possible without any compromise in comfort and user-friendliness.

Figure 1-3 Slave Manipulator Designed from Human Arm

Figure 1-4 Slave Robot Design Layout and Prototype

Figure 1-5 Master Robot Design Layout

The slave and master robots are shown in Figure 1-4 and Figure 1-5 respectively. The slave robot is attached to the tip of the endoscope, such that two manipulators could be seen at two sides on the view of the endoscope visual feedback. The endoscopist can control the movement of the slave manipulators inside the patient GI tract through the master robot outside under the guidance of the endoscope view. This master-slave robotic system will allow the endoscopist to have more maneuverability to perform more complex operation as suturing for example; it also can make tasks which can be done with current tools much easier and faster to be performed. To achieve this, other than the visual feedback from the endoscope, haptics information becomes a necessary factor that can improve the surgeon’s performance.

However, the size of the slave manipulator is as small as 7mm in diameter each; there is no available force transducer small enough to be mounted on to it. Besides, the slave is attached to a flexible body; there is no way to mount the force transducer onto it to measure the force at the tip. Even with self-made strain gage sensor or thin-film force sensor, the size remains to be a problem; other than that, with sensor directly in contact with soft tissue, it is difficult to ensure the sensor works properly due to the quasi-incompressible and other properties of soft tissue, furthermore the sterilization and biocompatibility also need to be taken into consideration.

Instead of looking for suitable miniature force sensors, an alternative solution is proposed as shown in Figure 1-6. When the slave robot touches the GI tract, stomach wall for example, it will force the wall to deform. The distance the manipulator makes the wall deforms will be calculated by the tissue force modeling simulator, then proper force could be set to the encoders of motor of the master robot. Therefore the master side could “feel” the force the slave side is feeling. In such a way, the sensing problem reduces from sensing the force interacting with soft tissue to sensing the contact.

Figure 1-6 Sketch map for haptics information collection and feedback

Moreover, the dashed frame portion can be replaced by virtual environment. Then this model can be applied in an endoscopist training simulation program, which could be

very helpful for the endoscopist learning to use this system. Surgical simulation is a topic many researchers have been working on ever since computer graphics technology blossom. And nowadays, haptics is required to be involved in such simulations. Due to tissue deformation, inhomogeneity and opacity, many possible sources of force could apply to the surgical tool (i.e., stiffness, friction and cutting)[14]. Besides, in order to provide force feedback to the master robot, usually motors are used to provide certain force restriction, then relation between the force and the motor torque is also inevitable to be investigated.

1.3 Objective and Scope

1.3.1 Objective

The objective of this project is to develop a 1-DOF haptic device for the master-slave robotic system developed for Therapeutic Gastrointestinal Endoscopy. This is to provide force feedback to the endoscopist, to improve maneuverability and the capability to perform more complex tasks.

1.3.2 Scope

For the accomplishment of this master of engineering degree, the scope is listed:

- To study soft tissue properties and to develop a force model to obtain a fine simulation of the force response of GI tract soft tissue. *Ex vivo* experiments on pig's maw¹ are designed to validate and characterize the force model.
- To built up a simple 1 degree-of-freedom master-slave system with haptics to prove the feasibility of this application.
- To implement this force model into a simple virtual reality environment as a trial of endoscopy surgery simulation, with force feedback.

¹ Pig's stomach, appears as pig's maw selling in the supermarket.

1.4 Organization of Report

This report contains six chapters. They are organized in the following manner:

Chapter 1 introduces the background and motivation of the project, and states the objective and scope, together with the organization of the report.

Chapter 2 provides the literature review for this project. It consists of robotics application in surgery, role of force feedback, introduction to the Gastrointestinal Tract and gastrointestinal endoscopy, haptic devices and haptics in endoscopy, and soft tissue modeling.

Chapter 3 shows the force model for the soft tissue, and experiment results for characterization and validation for the model.

Chapter 4 provides some analyses on the model, and more experiment results. Discussions on the reliability and limitations of the model are also included.

Chapter 5 shows the application of the force model in a 1-DOF master-slave system. System setup is shown, and its potential application in multi-degree-of-freedom master-slave system is also discussed.

Chapter 6 describes the application of the force model in a simple virtual reality simulation with haptics.

Chapter 7 concludes the whole project.

Chapter 2 Literature Review

This chapter provides the literature review for this project. Section 2.1 reviews robotics application in surgery. Section 2.2 elaborates the role of force feedback in robot surgery. This explains why haptic information is needed in the robotic system. Section 2.3 focuses on the nature of the GI tract, since these organs are the targets that surgical tools will interact with. Following section 2.4 gives a brief description of diagnosis procedure in gastrointestinal endoscopy. In section 2.5, the haptic devices available and efforts on the training and simulation as well as adding haptics for endoscopy are introduced. Lastly section 2.6 introduces the soft tissue properties and efforts on force modeling for soft tissue.

2.1 Robotics Application in Surgery

Robotics has been used in industry for everything from arc welding to assembling complex electronic devices. Applications for these devices have reached beyond the industrial arena into areas such as agriculture, space exploration, military, oceanographic exploration, etc. However, the idea of bringing robotics technology into the operating theatre was oppugned by most people when it was first time proposed. It was difficult for people to accept the concept of the presence of a no-feeling, cold machine in the operation room to save patient's life. As the development in robotics progresses, advantages of using robots to surgery become obvious. Once aware of the possibilities, surgeons became the ones who started pushing for this technology to appear in the operation theatre[15]. In 1985 a robot, the PUMA 560 was used to place a needle for a brain biopsy using CT guidance. In 1988, the PROBOT, developed at Imperial College London, was used to perform prostate surgery. The ROBODOC from Integrated Surgical Systems was introduced in 1992, and is a robot to mill out precise fittings in the femur for hip replacement surgery[16]. In surgery, the robot moves a high-speed cutting tool to form the precise shape

specified in the pre-surgical plan. The result is a far better fit between the bone and replacement joint than has been possible with conventional hand-held cutting instruments [17]. With the development of minimally invasive surgical techniques in the late 1980s, concept of using robotics in surgery becomes well accepted.

	Surgeons	Robots
Advantages	Task versatility Judgment experience Hand-eye coordination Dexterity at millimeter-to-centimeter scale Many sensors with seamless data fusion Quickly process extensive and diverse qualitative information	Repeatability Stability and accuracy Tolerant of ionizing radiation Diverse sensors Optimized for particular environment Spatial hand-eye transformations handled with ease Manage multiple simultaneous tasks
Drawbacks	Tremors Fatigue Imprecision Variability in skill, age, state of mind Inability to process quantitative information easily Ineffective at submillimeter scale	Expensive Cumbersome Large Inability to process qualitative information Not versatile Technology still in infancy

Table 1 Advantages and disadvantages of human and robot capabilities: a balance of these elements results in the most useful technologies[18]

To understand the advantages of using robots in surgery, it is helpful to consider the differences in human and machine characteristics (summarized in Table 1); many promising applications are based on unique robotic capabilities. One key difference is precision and accuracy, or more generally, the ability to use copious, detailed, quantitative information. Another important difference is that specialized manipulator designs allow robots to work through incisions that are much smaller than would be required for human hands or to work at small scales, where hand tremor poses fundamental limitations. Humans are superior, however, at integrating diverse sources of information, using qualitative information, and exercising judgment. Humans have unexcelled dexterity and hand-eye coordination, as well as a finely developed sense of touch. Unlike interaction with robots, interaction with human members of a surgical team for instruction and explanation is straightforward. These differences in capabilities mean that current robots are restricted to simple procedures, and humans must provide detailed commands, using preoperative planning systems or by providing explicit move-by-move instructions. Even in the most sophisticated systems,

robots are specialized to specific tasks within procedures; humans must prepare the patient, make many of the incisions and sutures, and perform many other functions. Robotic systems are best described as “extending human capabilities” rather than “replacing human surgeons.”[17]

2.2 Role of Force Feedback in Robotic Surgery

Ever since the robotic surgical systems advancing the field of minimally invasive surgery, they have become a staple in operating rooms. But none of these systems provide haptics information to the surgeon. Even the most famous da Vinci system, which is the only robot approved by the US Food and Drug Administration for conducting surgical procedures, can only provide partial haptics information. The surgeon can receive some force sensation, or haptic feedback, from the da Vinci system, but this haptic feedback is currently limited to interaction with rigid structures, such as tool-on-tool collisions, and not soft tissues. Haptics is always something the surgeons long for. "It always helps to be able to feel what you are doing, to feel the tissue tension and to feel the force when manipulating a suture," says Domenico Savatta, chief of minimally invasive and robotic urology surgery at Newark Beth Israel Medical Center. "Haptics would make it easier to learn robotic surgery, operate on things that are very delicate, and be an overall advantage to have in the system."[19]

Though force feedback is widely assumed to enhance performance in robotic surgery, its benefits were experimentally assessed by Wagner *et al* in 2002 [7]. Performance was compared between force feedback gains of 75% and 150% and no force feedback in a blunt dissection. A master-slave robotic system consisting of two PHANTOM (Model 1.5, SensAble Technologies, Inc.) haptic interface devices was used. One acts as the surgeon master controller and the other acts as the surgical robot. The surgical robot is attached to rigid shaft with a right angle hook at the end. The schematic picture is shown in Figure 2-1.

Surgical model used in their experiment simulated a vital structure such as an artery embedded in its surrounding tissue. Subjects were instructed to expose the artery, clearing away tissue 2mm on each side of the artery as well as removing any tissue on top of the artery. The measurements on the applied force, the number of errors, the length of dissection and the area of tissue affected were taken under two conditions, one with visible artery, the other with an obscured artery.

Figure 2-1 Surgical Robot Set-up

As the result, the absence of force feedback increased the average force magnitude applied to the tissue by at least 50%, and increased the peak force magnitude by at least a factor of 2, the number of errors that damage tissue increased by over a factor of 3. However the rate and precision of dissection were not significantly enhanced with force feedback. From these results, they drew the conclusion of that force feedback allows more precise dissection with lower applied forces and fewer errors. The force feedback at the master side makes it difficult to move the instrument into a damaging configuration because a large force on the hand will oppose any motion that involves contact between the instrument and the tissue. In addition, this constraint not only acts as a safety barrier, but can also act as a guide to the surgical instrument.

The force and torque sensor used in Wagner’s experiment, is the mini transducer manufactured by ATI Industrial Automation, which is built into the instrument shaft. This sensor is applicable to measure six-axis force/ torque. In this way, it results in the user feeling the forces that would be experienced if the master robot was attached directly to the proximal end of the instrument shaft. As found in a method of construction of force sensor promoted by Wagner and Howe, it suggested where force sensor on a surgical robot can be put.

Figure 2-2 Positions for Force Sensor to Be Considered

Figure 2-2 shows the force sensors can be put onto the da Vinci Surgical System. This manipulator provides 7 DOFs (6 DOF for the positioning and orientation of the end-effectors, 1 DOF for grasping). However, to provide all the force information for all these degrees of freedoms is not necessary. Most important haptic information should be the force feedback near the end-effector and the tip force sensing for the grasper.

Later in 2004, task performance with partial force feedback was discussed by W. Semere *et al* [10]. They promoted that it may not be possible to match the number of degrees of freedom of position sensing and control with the degrees of freedom of force sensing and feedback in practical application of force feedback in robot-assisted surgical systems.

Experiments with three conditions were tested: 3-D force feedback, force feedback without the axial forces on the slave tool and without force feedback. They were compared in performance of two tasks: (1) a task to push a cup through a series of poses and (2) a blunt dissection task. The results showed that peak force and RMS force without axial force feedback closely approximated performance with full force feedback in both tasks. And once again, it also showed that there was a significant difference (less average peak force and RMS force) in magnitude of applied forces when comparing no feedback to any force feedback. Same as the results shown in Wagner’s paper, there was no significant difference found between the task

completion time and the length of artery dissected, across the varying force feedback conditions. This finding suggested that efficiency (in terms of force and energy expenditure), but not the overall speed was improved with force feedback. However, successful execution of delicate surgical procedures is more likely to depend on efficiency rather than speed.

The results shown in Semere's paper provided promising evidence that simple force sensing systems can be used to create bilateral tele-manipulation systems that improve surgical performance. In this way, an inexpensive method for measuring force at the tip of a shaft (such as an endoscopic tool) is to attach a pair of strain gauges to each side and measure bending forces. This type of measurement neglects the forces along the axis of the shaft, thereby only providing 2-D force feedback. According to the statistically data shown in Semere's paper, it indicates that teleoperation with a 2-D force sensing device would result in performance comparable to a system with full (3-D) force feedback, which in turn closely approximates the natural feedback from manually holding an instrument. This is just a hypothesis, and need future investigation.

2.3 Gastrointestinal Tract

The Gastrointestinal (GI) Tract is a continuous tube that extends from the mouth to the anus. In short, it is the passage in the human body where it has a direct contact with the food as it travels from being consumed to excretion of it as waste. Normally it is said to be consist of two parts: upper and lower GI tract. The upper GI tract consists of the mouth, pharynx, esophagus, and stomach. The lower GI tract comprises the small intestines, big intestines and anus. There are also accessory organs to the alimentary canal include the liver, gallbladder, and pancreas. The physical layout or the organs along GI tract can be found in Figure 2-3.

Figure 2-3 Upper and Lower Gastrointestinal Tract (from online resource: Wikipedia)

The focus of the study of the GI tract will be on the esophagus, stomach and large intestine, areas where the gastrointestinal scope is design to be more commonly used on. The small intestine is not included in the study since it is too small and too far in to allow any gastrointestinal scope to pass through.

2.3.1 Esophagus

The esophagus is a straight 23-25cm long muscular tube and about 2.5cm in diameter that connects between the stomach and pharynx. Its main role is simply to carry the food from the pharynx to the stomach using contraction and relaxation of its smooth muscle. Bolus, or food that was chewed and mixed with saliva, travel through the esophagus by using the process peristalsis, which is the muscle at the top part of the bolus contracting itself, pushing it downward while the muscle below the bolus relaxing, allowing it to move through.

Figure 2-4 the Esophagus and the Process Peristalsis (from online resources)

2.3.2 Stomach

Figure 2-5 Anatomy of the Stomach (from online resource: Britannica)

The stomach is an expanded section of the digestive tube between the esophagus and small intestine. It is *J* shaped with the right side of the stomach shown above is called the greater curvature and that on the left the lesser curvature. The greater curvature is about 4-5 times longer than the lesser curvature. The most distal and narrow section of the stomach is termed the pylorus - as food is liquefied in the stomach it passes

through the pyloric canal into the small intestine. When the stomach is empty, the stomach walls will form folds called rugae, and when it is full it will stretch and become straight. The shape and position of the stomach are so greatly modified by changes within itself and in the surrounding viscera that no one form can be described as typical [20] On average however the stomach can hold 2-3 liters of liquid, 30cm long and 15 cm wide at the widest point.

Figure 2-6 Shape and Position of Empty, Moderately Filled, Distended Stomach (from online resource: Gray's Anatomy of the Human Body)

The wall of the stomach consists of four coats: *serous*, *muscular*, *areolar*, and *mucous*, together with vessels and nerves[20], as shown in Figure 2-5.

The **serous coat** is derived from the peritoneum, and covers the entire surface of the organ, excepting along the greater and lesser curvatures at the points of attachment of the greater and lesser omenta; here the two layers of peritoneum leave a small triangular space, along which the nutrient vessels and nerves pass. On the posterior surface of the stomach, close to the cardiac orifice, there is also a small area uncovered by peritoneum, where the organ is in contact with the under surface of the diaphragm.

The **muscular coat** is situated immediately beneath the serous covering, with which it is closely connected. It consists of three sets of smooth muscle fibers: longitudinal, circular and oblique.

- The *longitudinal fibers* are the most superficial, and are arranged in two sets. The first set consists of fibers continuous with the longitudinal fibers of the esophagus; they radiate in a stellate manner from the cardiac orifice and are practically all lost before the pyloric portion is reached. The second set commences on the body of the stomach and passes to the right, its fibers becoming more thickly distributed as they approach the pylorus. Some of the more superficial fibers of this set pass on to the duodenum, but the deeper fibers dip inward and interlace with the circular fibers of the pyloric valve.
- The *circular fibers* form a uniform layer over the whole extent of the stomach beneath the longitudinal fibers. At the pylorus they are most abundant, and are aggregated into a circular ring, which projects into the lumen, and forms, with the fold of mucous membrane covering its surface, the pyloric valve. They are continuous with the circular fibers of the esophagus, but are sharply marked off from the circular fibers of the duodenum.
- The *oblique fibers* internal to the circular layer, are limited chiefly to the cardiac end of the stomach, where they are disposed as a thick uniform layer, covering both surfaces, some passing obliquely from left to right, others from right to left, around the cardiac end.

The **areolar** or **submucous coat** consists of a loose, areolar tissue, connecting the mucous and muscular layers.

The **mucous membrane** is thick and its surface is smooth, soft, and velvety. In the fresh state it is of a pinkish tinge at the pyloric end and of a red or reddish-brown color over the rest of its surface. In infancy it is of a brighter hue, the vascular redness being more marked. It is thin at the cardiac extremity, but thicker toward the pylorus. During the contracted state of the organ it is thrown into numerous plaits or rugae, which, for the most part, have a longitudinal direction, and are most marked toward the pyloric end of the stomach, and along the greater curvature. These folds are entirely obliterated when the organ becomes distended.

2.3.3 Large Intestine

Figure 2-7 Layout of the Large Intestine (from online resources)

The function of the large intestines is mainly of absorption, production of certain vitamins, formation of feces and the expulsion of feces from the body. The large intestine is part of the GI tract where it started from the ileum to the anus. It is about 1.5 meters long and is largest at the cecum (6.25 in length 7 cm in breadth) and gradually diminishes in size to as far as the rectum.

A further dilation of considerable size occurs just above the anal canal. The normal large intestine went through a series of bends and can be divided into a few different sections. It started from the cecum, the shape of a large pouch, and its ends opens up to the ascending colon where it ascends up the right abdomen and reaches the undersurface of the liver. After which it bends to the left and traverse across the abdomen to form the traverse colon and bends again downward to the pelvis region, forming the descending colon. Finally it will form another from the end of the descending colon and bend toward the direction of the anus and this form the Sigmoid

colon (about 50-55cm long). Similar to the esophagus, the large intestine is also capable of peristalsis. It uses this action to push the colonic contents from the cecum into the rectum. When the endoscope is inside the colon, the colon performs peristalsis in an attempt to expel out the endoscope.

2.4 Gastrointestinal Endoscopy

2.4.1 Diagnosis Procedure

With the procedure known as gastrointestinal endoscopy, the endoscopist is able to see the inside lining of patient's digestive tract. This examination is performed using a gastrointestinal scope—a flexible fiber optic tube with a tiny CCD camera at the distal end. The camera is connected to either an eyepiece for direct viewing or a video screen that displays the images on a color TV. The endoscope not only allows diagnosis of gastrointestinal (GI) disease but treatment as well.

Figure 2-8 Upper GI Endoscopy and Lower GI Endoscopy (from online resources)

As the structure of the GI tract, the endoscopy also categorizes into two types: upper GI endoscopy (also known as EGD, esophagogastrroduodenoscopy) and lower GI endoscopy (also known as colonoscopy). In both procedures, the patient lies on one side; with lubrication on the endoscope, the endoscopist inserts it into either the

mouth (for upper GI endoscopy) or the anus (for lower GI endoscopy) and advances it under direct vision from the screen. The patient may need to change position during the procedure to assist moving the endoscope. The doctor will study the GI tract and reinspect them as the endoscope is withdrawn. If necessary, biopsies and removal of polyps may be performed.

Figure 2-9 Photos from GI Endoscopy (upper left) mid-esophagus (upper right) mucosal folds ("rugae") in the gastric body (lower left) postbulbar duodenum (lower right) Typical triangular folds characteristic of the transverse colon [photos copied from website of Atlas of Gastrointestinal Endoscopy] [21]

2.4.2 Treatment

In the diagnosis with gastrointestinal endoscopy, these disorders or problems can be investigated, such as colon cancer, polyps, colitis, bleeding lesions, ulcers etc. Out of these, bleeding and removing of polyps can be done with endoscopy tools.

- **Bleeding** — The degree of bleeding can range from nearly undetectable to acute, massive, life-threatening bleeding. Gastrointestinal bleeding is a cause for

hospitalization and can cause death to the patient if it is not treated immediately. Successful treatment to bleeding not only should stop the bleeding but also prevent rebleeding from the same site. A common approach to stop bleeding now is the use of a combination of injection followed by cauterizing to stop the bleeding. First epinephrine injection is given to temporary stop the bleeding and provide a clearer view of the bleeding site to the endoscopist. After which the endoscopist will apply the heater probe to the appropriate area and permanently seal the vessel by cauterizing.

- **Polypectomy** —— Polypectomy can be defined as the removal of polyps (abnormal like growth that protrude into the lining of the bowel), which can be performed from an endoscope. This is regarded as a major breakthrough in gastroenterology since simple treatment can be administered without the cost, delay and potential danger of an open surgery[22]. Polyps can occasionally bleed, discharge and seep rotten oily yellow-brown waste and can develop into cancer and as such, they are usually removed if spotted by an endoscopist. The sizes of the polyps determine the method to use for the removal of the polyps. Basically it is divided into 3 categories: small polyps ranging from 1-3 mm, medium polyps ranging from 4-6mm and large polyps from 7-9mm. Generally, cold/hot forceps were used to remove smaller polyps while hot snares were use for larger polyps. After the removal of polyp, the surgeon may have to stop the bleedings.

Figure 2-10 Remove Polyps with Forceps and Snare Wire (from online resources)

2.5 Haptic Devices and Haptics in Endoscopy

In human-computer interaction, haptic feedback means both tactile and force feedback. Tactile, or touch feedback is the term applied to sensations felt by the skin. Tactile feedback allows users to feel things such as the texture of surfaces, temperature and vibration. Force feedback reproduces directional forces that can result from solid boundaries, the weight of grasped virtual objects, mechanical compliance of an object and inertia [2]. Several haptic devices in these two categories available are introduced in this section, and research on providing haptics to endoscopy or its simulation are described also.

2.5.1 Haptic Devices

Tactile feedback, as a component of virtual reality simulations, was pioneered at MIT. In 1990 Patrick used voice coils to provide vibrations at the fingertips of a user wearing a Dextrous Hand Master Exoskeleton. Minsky and her colleagues developed the “Sandpaper” tactile joystick that mapped image texels to vibrations (1990). Commercial tactile feedback interfaces followed, namely the “Touch Master” in 1993, the CyberTouch glove in 1995, and more recently, the “FEELit Mouse” in 1997.

For force feedback devices, Goertz at Argonne National Laboratories first used force feedback in a robotic tele-operation system for nuclear environments in 1954. Subsequently, the group led by Brooks at the University of North Carolina at Chapel Hill adapted same electromechanical arm to provide force feedback during virtual molecular docking (1990). Burdea and colleagues at Rutgers University developed a light and portable force feedback glove called the “Rutgers Master” in 1992. There clearly has been a resurgence of research interest and haptic interface products. In addition, research on haptic feedback has been aggressively pursued in several countries outside the U.S., notably in Japan, UK, France and Italy. Many works have been done as a survey on haptic devices[2].

In the book of Biologically Inspired Intelligent Robots[3], it presents several current devices categories that are in development: Exoskeletons and Stationary Devices, Gloves and Wearable devices, Point-sources and Specific Task Devices, Locomotive Interfaces, Feedback input devices/ Force feedback devices and Tactile displays. Examples from different categories are presented followed by basic concepts behind the devices in the categories.

Figure 2-11 Iowa State Force Feedback Exoskeleton [VRAC Website] (left) and Master Arm from Southern Methodist University [Systems Lab Website] (right)

Besides these ground-based and body-based haptic devices, another kind of haptic device using tension of cable is also under research. Instead of applying force through links, cables are connected to the point of contact in order to exert a force. Encoders determine the length of each cable so that tension is formed.

A brief comparison is given between linkage-based and tension-based haptic device in Berkley's paper [2]. It states that a tension-based device requires less work space, and the cables used in a tension-based system have little mass, this essentially eliminates inertial effects and increases the accuracy of the applied force. And cables could be safer and unobtrusive. It also declares that such devices are only now appearing in commercial markets, and surgery simulation is the arena where these devices are first making an appearance, with laparoscopy, trans-urethral resection of the prostate and suturing simulation serving as some of the initial applications. Several pictures about tension-based haptic device are shown below[23]:

Figure 2-12 A Schematic of Tension-based Gripper [Tokyo Institute of Technology]

Figure 2-13 Different Degree-of-freedom Tension-based Force Feedback Devices

However, the most sophisticated haptic devices designed for industry, medical and scientific applications are linkage-based device PHANTOM™ series from SensAble Technologies and body-based device Cyber series from Immersion Corporation, which are shown below:

Figure 2-14 PHANTOM™ desktop [SensAble Technologies] (left) and CyberGrasp Exoskeleton [Immersion Corporation] (right)

2.5.2 Haptics in Endoscopy

So far, no research has been done to add force feedback to the endoscopic tools, e.g. biopsy forceps or snare wires. Since only simple treatment can be done due to the low maneuverability of reduced DOF of the endoscopic tools, haptics information is not absolutely necessary. Studies mainly focused on haptic display² for endoscope simulation[24-26], adding force feedback to the MIS system using endoscope[7, 10-12] or to the virtual reality based surgery simulators.

In Olaf Körner and Reinhard Männer's paper [24], design and implementation of an active haptic display for endoscope tube and the tip are proposed. Based on virtual reality techniques, a simulation system is set up. By moving the flexible endoscope inside a pipe, force on the bending endoscope's tip is simulated and fed back to the navigation wheel.

Figure 2-15 Haptic Display for the Endoscope's Tube

In this implementation, the tube of endoscope is inserted into a sliced pipe which represents the patient; the tip of the endoscope is fixed to the carriage, which is driven by toothed belt actuated by the motor; wires are used to simulate the tension on the navigation wheels. Forces which reduced to four degrees of freedom are generated: Translation of tube, rotation of tube, bending of tip in local X direction and in Y direction.

² Haptic display implies that human sense of touch to be simulated by certain tool handle.

Figure 2-16 Force Feedback on the Tube and the Navigation Wheel

Though this idea is only about the haptic display for the simulation system instead of real-time haptic device for real endoscope manipulation, necessary algorithms for a real-time force calculation by investigating the mechanical behavior of the endoscope in the colon are also presented in the paper. In addition, the method using wires to provide force feedback also broaden the outlook in this topic.

It was mentioned in Wagner's paper [7] that an ATI force and torque transducer was used as the sensor in their MIS master-slave robotic system. As shown on the ATI company website, the smallest transducer available is 17mm in diameter. This is much too big compared with the diameter of the endoscope, which is typically a 70-180 cm long flexible tube of only about 11mm in diameter. The manipulator that can be mounted onto the endoscope is even smaller. In addition, this kind of transducer may be applicable for the rigid shaft, but for flexible and multi degrees of freedom manipulator, it might not be able to work properly to give the force feedback information. Even if the tool is fixed at the distal end of the flexible endoscope, the force sensed by the transducer could be different from the force at the end effector, since the bending part of the endoscope won't transfer the exact force to the transducer. If the tools have more degree of freedom, then that is even difficult for this method.

Figure 2-17 (a) Master Subsystem (b) Slave Subsystem for MIS with Force Feedback

Besides Wagner's work, some other researchers also investigated into the haptic prototype providing force feedback to the MIS system. Figure 2-17 shows a typical prototype of master-slave robotic system for MIS with force feedback [11]. Master subsystem and slave subsystem used two PHANTOM haptic devices which are controlled using the position error based scheme. The movement of master subsystem was sensed to control the slave subsystem; the force at the slave side could be sensed with the Phantom device then simulated by the Phantom device at the master side. M.Tavakoli *et al* declared some findings in their experiments. It is said the position error between the master and slave robots is small due to the compliance of the low-stiffness object. And since the force reflected to the user is proportional to the position error, (in order to have a perceivable force) the corresponding gain should be high. They also mentioned that if the gain is not high enough, the user may damage the tissue by incurring excessive deformation because insufficient forces are being transmitted to the surgeon's hand. However, a high gain may cause some problems as well. For example, some force is reflected to the hand even when the slave robot is moving in free space due to any control inaccuracies. Therefore, there should be a tradeoff on the force feedback gain for the dynamic range of perceivable forces. This provides a good guide for this project: dealing with the low-stiffness soft tissue. Both MIS and NIS are facing the same situation.

In Tavakoli et al's work [11], they also pointed out that due to the constraint on incision size in endoscopic surgery, the diameter of the portion of the end-effector that enters the body including all required sensors/actuators should be less than 10 mm. This is quite similar to the requirement for our purpose. And they also mentioned that currently the available sensors that measure forces and torques in all six degrees of freedom are all wider than this requirement. A novel robotic end-effector is introduced in their paper, which is capable of non-invasive measurement of its interaction with tissue in all the degrees of freedom available during endoscopic manipulation (pitch, yaw, roll, insertion and grasping/cutting/dissecting). Strain gauges are used to measure the axial forces and torsional moments. And without using sensors on the jaws of gripper, the end-effector can measure the tip interactions with tissue (grasping force, etc). This could be an important issue for the haptics information collection portion. And also their work provides useful information for the interaction with soft tissues.

Other than the application of haptics on the surgical tools, haptics is also applied in the VR-based surgical simulators. Surgical simulator acts an important role in the training for surgeons. Indeed, most inexperienced surgeons and students need practice to be familiar with the operation. With the development of the virtual reality technology, it provides a more cost-effective and efficient alternative to traditional training methods[27, 28], such as practice with a phantom or some other pick-and-place toys using an endotrainer.

Figure 2-18 (left) VR-based Surgical Trainer (right) a simulated scene with smoke effect [26]

Virtual Reality can create a realistic scene based on geometric representations of anatomical structures and physical nature. S.Cotin and H.Delingette in their work [29] showed a fine illusion of creation of the virtual reality model for the organ. They also presented that the force sensing depends on the deformable nature of the virtual model and the realism of the force feedback is highly correlated to the model's physical realism. Therefore a soft tissue modeling for its physical properties is required here to implement the force feedback in the virtual reality simulations.

Figure 2-19 Creation of the Virtual Reality Model [28]

The following section will give more information on soft tissue modeling.

2.6 Soft Tissue Modeling

Soft tissues are clearly quasi-incompressible, non-isotropic, non-homogeneous, nonlinear viscoelastic material in large deformation. To model soft tissue, the biomechanics of soft tissue need to be studied. In this section, soft tissue properties are described, they will affect the force response. And also, the current researches on force modeling for soft tissue are introduced.

2.6.1 Soft Tissue Properties

Nonlinear Elasticity. The main property of soft tissues may be outlined as being their nonlinear elasticity. Under uniaxial tension, parallel-fibered collagenous tissues exhibit a nonlinear stress-strain relationship characterized by an initial low modulus region, an intermediate region of gradually increasing modulus, a region of maximum

modulus which remains relatively constant till the tendon's failure. Various models have been proposed to describe the nonlinear stress-strain behavior [30]. A typical tensile curve is shown in Figure 2-20 (a). From a functional point of view, the first parts of the curve are more useful since they correspond to the physiological range in which the tissue normally functions

Viscoelasticity. When measured in dynamic extension, the stress values appear higher than those at equilibrium, for the same strain. The resulting tensile curve appears steeper than the one at equilibrium (Figure 2-20 (b)). When a tissue is suddenly extended and maintained at its new length, the stress gradually decreases slowly against time, this phenomenon is called *stress relaxation*. When the tissue is suddenly submitted to a constant tension, its lengthening velocity decreases against time until equilibrium, this phenomenon is called *creep*. Under cyclic loading, the stress-strain curve shows two distinct paths corresponding to the loading and unloading trajectories. This phenomenon is named *hysteresis*. As a global statement, the stress at any instant of time depends not only on the strain at that time, but also on the history of the deformation. These mechanical properties, observed for all living tissues, are common features of a physical phenomenon named *viscoelasticity*.

Figure 2-20 (a) Load-extension curve and (b) Influence of the strain rate of soft tissue

Other Properties. The compressibility of soft tissues has been investigated very little. Soft tissues are however usually assumed to be incompressible materials. Besides, when loading-unloading cycles are applied on the tissue successively up to the same stress level, the stress-strain curve is gradually shifted to the right. After a number of such cycles, the mechanical response of the tissue enters a stationary phase and the results become reproducible from one cycle to the next. This phenomenon is due to the changes occurring in the internal structure of the tissue, until a steady state of cycling is reached. This initial phase of behavior common to all living tissues is usually used as *preconditioning* of the tissues prior to experimentation.[31]

2.6.2 Force Modeling for Soft Tissue

To simulate the deformable motion of soft tissue, a mesh of nodes and elements is usually used to define and discretize the domain geometry; various interpolating functions are then used to define the deformation over an individual element. Based on this concept, many models are built for the simulator. In the works done on the soft tissue models for haptic feedback, computational complexity was mentioned all the time [32, 33]. Besides accuracy, real-time was always emphasized.

In 2001, A.O. Frank *et al* used linear elastic finite element methods to compute haptic force feedback and domain deformations of soft tissue models [33]. Though most soft tissues showed non-linearity in both material behavior as well as large deformations, for small strains, they assumed linear behavior for simplicity. In their work, they suggested three steps to implement the method. First step was to discretize a continuous 3D domain into a finite number of elements connected via nodal points. Second step was material description. Based on the theory of elasticity and linear elastic constitutive material properties, a linear strain-displacement relationship for a single element could be defined. Last step was to obtain a global system of equations. By solving this linear system of equations, the discretized domain deformation could

be obtained. In their work, different computational methods were implemented and tested for the FE equations generated in simulating haptic force feedback. They also suggested some future directions that might lead to real time computations. The FEM approach provides accurate results. However, the high computational cost can be a major problem for real-time modeling. For large meshes, a large system of equations has to be solved. Although the elastic deformation could be assumed to be linear in order to reduce the size of the system equations, only small deformations are accurately modeled.

In recent year, there has been more and more interest in the modeling of forces resulting from tool-tissue interaction [14]. Accurate force model are not only important for haptic feedback, but they also determine the way tissue deforms during contact with a surgical tool. Therefore, data acquired from actual tissues could be used to evaluate the accuracy of the simulators, and it also can provide fine material properties. A number of researchers have developed haptic virtual environments for surgical simulation. These definitely will improve the realism of tissue simulation and performance of robot-assisted surgical systems. Different types of tissue have different properties; the model should be changed accordingly. Interaction with different kinds of tools would be slightly different.

Also using finite element method, a two-dimensional linear elastostatic material model was derived to calculate the force information that is not directly measurable by S.P. Dimaio and S.E. Salcudean in their paper, and based on this model, simulation for needle insertion with visual and kinesthetic feedback was also developed[34, 35]. In their experiment, a 3-DOF, planar robotic manipulator was used to drive an end-effector (blunt probe or needle) into a thin rectangular block of soft elastic material that acts as a tissue phantom, as shown in Figure 2-21. The force on the tool and tissue phantom deformation were sampled. In their work, as a first assumption, it focused on homogeneous, linear elastostatic models. Such models were said to be characterized by two parameters, Young's modulus and Poisson ratio[36].

Figure 2-21 DiMaio et al's Complete Experiment Setup and (a) Instrumented Probe, (b) Instrumented needle

In Okamura's paper [14], a force model for needle is presented. Data was acquired from a 1-degree-of-freedom translation stage to guide a 1.27-mm O.D., 15.24-cm-long surgical needle with a bevel tip into bovine liver. Three possible force sources were considered: 1) capsule stiffness, a nonlinear spring model; 2) friction, a modified Karnopp model; and 3) cutting, a constant for a given tissue. This method is different from the FEM method, the force is a function of the penetration distance, however restricted in 1-DOF. The force expression was presented as

$$f_{needle}(x) = f_{stiffness}(x) + f_{friction}(x) + f_{cutting}(x) \tag{2.1}$$

And in their work, they also characterized the effects of needle diameter and tip type on insertion force using a silicon rubber phantom; concluded that the size and shape of the needle played an important role in determining the forces of needle insertion. They mentioned that the force model developed and knowledge of geometry effects can be used in the design of virtual environment for simulation of percutaneous therapies.

Figure 2-22 Okamura's Experiment Setup for Data Collection under CT Fluoro Imaging

Figure 2-23 Hu et al's Experiment Setup for measuring force and displacement

Different from the needle insertion, T. Hu and J.P. Desai worked on the large deformation analysis with large probe[37]. The large probe can be seen from the experiment setup as shown above. In this large probe analysis, the size of the probe

was much larger than the tissue sample, which ensured the uniform deformation of the sample and consequently simplified boundary condition, since FEM method was employed. This experiment used lead screw to control the probe displacement. A cubic sample of pig liver with the size of 10mm x 10mm x 10mm was placed under the large probe. The probe velocity was selected to be as slow as 6.096mm/min, due to quasi-static analysis. Though a 6axis force and torque sensor was used, this experiment was still considered as a 1-DOF test. The finite element simulation was built up in ABAQUS, and the soft tissue was assumed to have the hyperelastic property.

So far all these works were done based on *ex vivo* experiments or even on tissue phantom. In both Okamura and DiMaio's papers, they mentioned *in vivo* experiments would improve the applicability of the data. Recently, some work was done on this topic. B.K.Tay et al designed the experiments to study the *in vivo* static and dynamic mechanical behavior of the liver and lower esophagus of pigs both in linear and nonlinear regions under compressive and shear indentations[38]. A series of ramp and hold stimuli, as well as sinusoidal indentation stimuli, were delivered to the organs and reaction forces were measured. The conditions for these indentation stimuli were designed such that they were similar to conditions in an operating room. Experiments were also carried out on the organs for *ex vivo* conditions for comparison. Results showed that the breathing and pulse rate significantly affect the measured force responses of the organs. Besides the breathing and heart beat, organ properties may vary from patient to patient. Therefore T.K. Podder et al presented *in vivo* brachytherapy procedures-specific data and patient-specific data to develop statistical models to estimate needle insertion force[39].

However in their work, the viscoelastic property of soft tissue was ignored. Only a few recent papers used FEM method for the characterization of viscoelastic soft tissue properties[40-42]. The models they used are discussed in the next section.

2.6.3 Viscoelastic Modeling

In order to determine soft tissue properties and predict their response under well defined conditions, various stress-strain relationships have been derived from fitting experimental data or based on the continuum mechanics[43]. As in many prior works, the soft tissue was assumed to be linear elastic, within a range of low strain. However, for finite deformations, the nonlinear stress-strain characteristics of the living tissue must be taken into account[44]. From this point of view, Fung developed this QLV (quasi-linear viscoelasticity) theory, in the form of:

$$T(\varepsilon, t) = \int_{-\infty}^t G(t - \tau) \frac{\partial T^{(e)}[\lambda(\tau)]}{\partial \tau} d\tau \quad (2.2)$$

$$\varepsilon(T, t) = \int_{-\infty}^t J(t - \tau) \frac{\partial \varepsilon^{(e)}[T(\tau)]}{\partial \tau} d\tau \quad (2.3)$$

where T represents the Lagrange stress, ε the infinitesimal strain, λ the corresponding extension ratio defined by $\lambda = 1 + \varepsilon$, $G(t)$ and $J(t)$ are known as reduced relaxation function and reduced creep function respectively. $T^{(e)}(\lambda)$ is an elastic response.

With the assumption of $T = \varepsilon = 0$ for $t < 0$, the equation (3.2) can be rewritten as

$$T(\varepsilon, t) = T^{(e)}[\lambda(t)] + \int_0^t T^{(e)}[\lambda(t - \tau)] \frac{\partial G(\tau)}{\partial \tau} d\tau \quad (2.4)$$

Therefore, viscoelasticity may be modeled by adding viscoelastic properties to elastic models. This approach was followed by the researchers to develop various similar formulations in different applications[41, 45, 46].

However, most of the uniaxial structural viscoelastic models have been based on the discrete element combination approach. The elastic and viscous components were modeled by combinations of springs and dashpots, respectively. The elastic components, as previously mentioned, can be modeled as springs of elastic constant E , given the formula:

$$\sigma = E\varepsilon \quad (2.5)$$

where σ is the stress, E is the elastic modulus of the material, and ϵ is the strain that occurs under the given stress, similar to Hooke's Law.

The viscous components can be modeled as dashpots (as known as damper also) such that the stress-strain rate relationship can be given as,

$$\sigma = \eta \frac{d\epsilon}{dt} \tag{2.6}$$

where σ is the stress, η is the viscosity of the material, and $d\epsilon/dt$ is the time derivative of strain.

Different models were developed based on these two elements. For example Maxwell model, Voigt model, Kelvin model (also known as standard linear solid model), Galford model as shown below.

Figure 2-24 Schematic Representation of Different Models

This approach was well accepted by researchers, and the two basic elements were further modified with different formula. In Kim’s work[42], a modified Kelvin model was implemented, with nonlinear springs in parallel.

Chapter 3 Force Modeling and Validation

In this application, the force model focuses on the viscoelastic response of the tissue of the stomach wall. The model is discussed and described in Section 3.1. Then experiment was designed and performed on the pig maw *in vitro*, to characterize and validate the model. Experiment procedures are presented in Section 3.2.

3.1 Force Model for the Stomach Wall Validation

As discussed in the previous chapter, most of prior works focused on the force model of the pig liver or tissue phantom. These objects are solid; with relatively small deformation, the assumption of linear elasticity is tenable. However, for the stomach tissue, the situation is different. It will be inflated to become a hollow cavity during the diagnosis procedure. And because of the existence of the rugae, its volume can be expended, and a large deformation range is expected. Moreover when the stomach wall starts to deform, the rugae will also stretch. With these special characteristics of the stomach, ignorance of the viscoelasticity cannot be accepted. There has been few prior works on the modeling of the force response of stomach. A similar work done was on viscoelastic behavior of pig lower esophagus[42]. However in their work, the elastic response was still considered as linear elastic due to the small deformation of 6mm only.

To model the viscoelastic behavior of the soft tissue of the stomach wall, as described in the Equation 2.4 from the literature review, the model contains two components: one is elastic, the other is viscous.

The elastic component should be nonlinear. Nonlinear elasticity was modeled using either exponential term[42] or second order polynomial term[14, 47]. From a general study on the nonlinear viscoelastic material, it was found that nonlinear elastic

behavior was expended in power series, for the nonlinear viscoelasticity of polymer in uniaxial tensile loading[48]. The effect of truncating the series is shown in Figure 3-1.

For the application in this report, the model is to build the relation between the force and the deformation, in terms of F and x respectively. It can be considered as a uniaxial compression loading as well. After considering these theories from prior works, component of the elastic response is modeled as

$$F(x,t) = \mu_1 x(t) + \mu_2 x(t)^2 + \mu_3 x(t)^3 + \dots \tag{3.1}$$

Figure 3-1 Effect of truncating the series on its ability to represent the experimental data for viscose rayon fibers, at a strain rate of $1.6 \times 10^{-3} \text{ sec}^{-1}$ shown by truncating the series after various terms. A-1st term; B-2nd term; C-3rd term; D-4th term; E-5th term; F-experimental data[48].

For the viscous component, Kelvin model was popularly employed in various research works. The Kelvin model is expressed as

$$F + \tau_\epsilon \frac{dF}{dt} = \mu_0 \left(x + \tau_\sigma \frac{dx}{dt} \right) \tag{3.2}$$

here the constant τ_ϵ is called the relaxation time for constant strain, whereas τ_σ is called the relaxation time for constant stress.

However it was proposed that the viscous component was also a function of the strain itself in [49]. A modified Kelvin model was applied here, but the elastic component was replaced by a 3rd order polynomial term and the term $\frac{dx}{dt}$ by $x \cdot \frac{dx}{dt}$. Overall, the complete force model for this project is developed as

$$F + \tau_{\epsilon} \frac{dF}{dt} = \mu_1 x(t) + \mu_2 x(t)^2 + \mu_3 x(t)^3 + \tau_{\sigma} x \cdot \frac{dx}{dt} \quad (3.3)$$

Experiments were designed and performed on pig maw to characterize and validate this model. Details are described in the following section.

3.2 Characterization and Validation

Experiments were designed to perform tissue compression, with acquisition of the measurement of the compressive force and probe displacement. Section 3.2.1 lists the equipments for the experiment, following by the calibration of the system in section 3.2.2. Data was collected to characterize the developed model.

3.2.1 Experiment Setup

This system consisted of a motion control part, a force measuring part, and a post-data processing part. The configuration of the experimental system is shown in Figure 3-3. Data was acquired using an EZlimo linear motion slider (model No. EZS3-15) to guide a blunt probe, with diameter of the size of the slave manipulator designed in our group[13], to interact with the pig maw. The pig maws were all tested at room temperature on the same day as it packed. A 1-axis 50 N capacity EntranTM load cell (model No. ELFM-B1-50N) was used to measure the interaction force between the probe and the pig maw in the direction of motion. A Windows XP Pentium IV 3GHz computer with Galil DMC1820 Motion control card inside was used to control the linear slider and to collect the force data. ICM-2900 Interconnect Module was used to provide easy connections between Galil controller and other system elements.

The schematic diagram and the real configuration of the system setup are shown below:

Figure 3-2 Schematic Diagram of the System Setup

Figure 3-3 Experiment Configuration

The slider was mounted on the top of a box as shown in the configuration figure. For the connection convenience, the interconnect module for the Galil motion control card and wires were kept in the box. A base holder was designed and fixed onto the slider's

carrier to hold the load cell and probe. A close up look is shown in Figure 3-4.

A Bosch ball bushing with inner diameter of 12mm was used to reduce the friction and ensure the linear motion of the probe. The ball bushing was fixed into the front wall of the base holder. The probe with size of 12 mm can slide inside the ball bushing. For positioning, a shallow circular slot was made on the back wall, which the load cell was embedded inside. The hole for the ball bushing and the slot for the load cell were designed to be concentric to ensure the force on the tip of the force can be loaded right on the bulge of the load cell. This design can prevent the influence by the friction due to the relative motion between the holder and the probe, as well ensure when the probe interacting with the tissue the motion is linear in the horizontal direction. In such a way, all the force at the tip of the probe on the horizontal direction can be fully transmitted to the load cell for measurement. This is very important for this research on 1-DOF force modeling.

Figure 3-4 Close up for the Base Holder

For the tissue holder, it was designed as shown in the Figure 3-5. A plastic rectangular box with similar size to the pig maw was used to hold the tissue. The opening of the box is measured 16 by 12 mm. The pig maw was cut open along the two natural openings, one from esophagus, and the other to the duodenum. The tissue was clipped on the opening of the box, in order to be able to place vertically, but without pre-stressed. In such a way, it can roughly simulate the stomach during the endoscopy diagnosis procedure. However this might not work for extremely large deformation, since the clips constrain the rest of the tissue from stretching. The deformation limit was set to be 3cm. Even larger deformation than this length won't be helpful for this modeling, since it's not likely to happen during the interaction between the tissue and the surgical tools. It is unlikely to simulate the pressure inside and outside the stomach, but to prevent the disturbance caused by the air sealed inside the box, several holes were drilled on the bottom of the box.

Figure 3-5 (a) Pig maw and its holder (b) interior of the pig maw, we can see the rugae (c) The tissue was fixed on the holder.

3.2.2 Calibration of the System

Before the experiment started, the system required calibration, both in the motion control and the force acquisition.

In order to ensure that all indentation displacements were made with respect to the same initial reference position, the probe was moved by slightly increments until it was visually determined to be just barely touching the surface of the tissue. Then a marker was labeled at this position, as shown in figure below.

Figure 3-6 Calibration for Position

For the force acquisition part, the load cell needed calibration also. The specifications of the load cell can be found in the appendix. Since the load cell is small in size and had a bulge on the surface, it is difficult to balance the load directly onto the load cell as other types of force sensor calibration. A linear guide platform was designed for the calibration. The platform was able to move freely along the vertical direction by using two linear guides. This ensured that the bulge of the load cell was the only contact.

Figure 3-7 Calibration of Load Cell

The platform weight was measured, and then the other loads could be placed onto the platform. The friction due to the linear guides could be ignored. From this calibration, the zero load voltage was used to offset the experimental data. The platform weight is 140 grams, and the increment load weights 100 grams each, up to 1040 grams. Record of the voltage and corresponding weight is shown below, the plot shown in Figure 3-8.

Weight (mass/g)	0	140	240	340	440	540	640	740	840	940	1040
Voltage	1.28	1.6	1.73	1.78	1.79	1.8	1.87	2.04	2.55	2.69	2.84

Figure 3-8 Data Plot of Calibration of the Load Cell

From the plot, it can be seen that for this hardware, with a range of 50N (5000 grams) for the starting range, the load cell response is not linear. This may due to the hardware obsolescence problem. However, a few set of calibration tests were performed, it was found that the results were consistent. Thus, the load cell still could be utilized in the experiment. But the relation between the applied load and measured voltage was estimated by the red color lines shown in Figure 3-8. In such a way, by removing the zero load voltage, this relation applied to obtain the corresponding load at the tip of the probe. To further improve the accuracy of the experiment, a proper force sensor with smaller range could be selected.

3.2.3 Experiment Results and Characterization

To find out the viscoelastic behavior of the tissue, ramp and hold³ stimuli was sent to the motion control. The probe moved at a constant velocity to the desired depth of displacement; then it stopped moving while the force information was still collected for some time. Due to the stress relaxation property of viscoelastic behavior, the force should drop then retain steady at a lower value.

For characterization, the moving velocity was chosen to be $0.75mm/s$, the desired depth was $30mm$. The ramp time was therefore $40\ seconds$, while the hold time was $50\ seconds$ to ensure the force reached steady state. The experiment result is shown in Figure 3-9. With the experimental data, least square method was adopted to evaluate the parameters in the model (Equation 3.3), to minimize the sum of squares given by

$$E = \sum_i^m (F_s(t_i) - F_e(t_i)) \cdot (F_s(t_i) - F_e(t_i)) \quad (3.4)$$

where F_s , F_e , t_i and m are simulated forces from model, measured force from experiment, time and total time of data points, respectively.

³ Ramp and hold means the motor moves at a constant velocity for certain time then stops, so the motion track is a slope (ramp) then horizontal (hold)

Then the force model became

$$F + 1.023 \times \frac{dF}{dt} = 0.05333 \times x(t) - 0.0086 \times x(t)^2 + 0.0004345 \times x(t)^3 + 0.06 \times x \cdot \frac{dx}{dt} \tag{3.5}$$

The result is shown in Figure 3-9, as a comparison with the experimental data.

Figure 3-9 Characterization of the Model (velocity 0.75mm/s, indentation 3cm)

By comparing the plot, the model simulation approximately fitted the experimental data. The maximum difference was around the level of 0.5 N. With the parameter fixed, several more experiments were performed with the same settings. The result is shown in Figure 3-10. From the figure we can find the model is quite close to the experimental data. And it can also be seen that as experiments repeated again and again, the response was prone to be smaller in value during the stress relaxation process. And also the maximum forces of experiments are becoming smaller as well. This is because of the phenomenon called preconditioning. Experiments with different velocities were also performed to validate the model, as shown in Figure 3-11. It shows that the shape of curve, maximum and steady state value are very close to the experimental data also.

Figure 3-10 Validation of the Model (velocity 0.75mm/s, indentation 3cm)

Figure 3-11 Validation of the Model (continued, velocity 1.5mm/s, indentation 3cm)

From the experiments above, the model was validated to be useful for estimating the response force during the probe indentation. More on the model and experimental data are discussed in the next chapter.

Chapter 4 Analysis and Discussion

4.1 Analysis on the Experimental Data

Though the proposed model fitted with the experimental data, it is also important to compare the experimental results with those of the literature in order to validate both the results and the testing method.

In Fung’s book[31], a prior work done by him was mentioned in the chapter about behavior of soft tissues under uniaxial loading. The specimen he tested on was a rabbit mesentery. Mesentery is, in anatomy, the double layer of peritoneum that connects a part of the small intestine to the posterior wall of the abdomen. This structure is similar to the stomach wall, a layer of soft tissue rather than other solid organs. In Fung’s test, the specimen was stressed at a velocity of 1.27cm/min. Then the moving head of the testing machine was suddenly stopped to retain the strain constant. The result is shown in Figure 4-1. Comparing with experimental results obtained in this project, the shapes of the curve are similar. Though the force levels with respect to the same indentation displacement are different, that is due to the different properties of two kinds of tissue.

Figure 4-1 Relaxation Curve of a Rabbit Mesentery (Fung 1967)

As mentioned in the previous chapter, as the experiments repeated, the curves of results shifted. It should be due to the phenomenon of preconditioning. This is a very important feature in the stress-strain relationship for soft tissue. It was detailedly introduced in Fung's book. The reason that preconditioning occurs is that the internal structure of the tissue changes with the cycling. By repeated cycling, eventually a steady state is reached. As the cycle repeating, starting from the same load, the relaxation curves will be seen to shift upward. If once reaching certain load, then the probe moves immediately with same constant velocity, the loading and unloading curves will be seen to shift to the right[31, 43]. An example from literature [50] is shown below.

Figure 4-2 Preconditioning of an Anterior Cruciate Ligament (a) Load-elongation (b) Relaxation

To investigate this phenomenon, meanwhile to prove the validity of the experiment data, two groups of experiments were performed. First, 10 sets of experiment were performed on the same specimen. The tissue specimen was loaded to a pre-defined value, and then it was unloaded immediately at the same constant velocity as loading. The force information was collected during both procedures. By analyzing the results, the cycle curve should be shifted to the right. From experiment results #1 and #9, as shown in the Figure 4-3, the curve shifts to the right. However the other experimental results did not show obvious differences comparing with these two, as #5 and #9. This should be due to the relaxation time left in between two experiments performed. An average of 3 minutes time was applied between experiments. According to Fung's

book, his tests were performed 10 minutes after the previous one. Longer resting time should be able to make the curves distinguished. Generally speaking, it still can be seen that there is shift to the right in between the results for my repeating experiments.

Figure 4-3 Cycle of Loading-Unloading Curve Shift

Figure 4-4 Relaxation Curve Shift

Secondly, 10 more sets of experiment were performed without unloading but keeping a constant indentation at certain load F_0 . Three selected experiments were shown as example in Figure 4-4. Not very large differences between experiment #1, #4 and #7, it is clear to see that the curve of later experiments shifts upward.

Overall, a nonlinear stress-strain relationship, a hysteresis loop in cyclic loading and unloading, stress relaxation at constant strain, preconditioning in repeated cycles, from these common features of living tissues, the viscoelastic behavior can be investigated. More important, the experiment data and method are proved to be correct and validated. Therefore the model fits with the data should be validated also.

4.2 Discussion on the Force Model

A good model will provide a guide in determining the force during the indentation process, especially the maximum and the after-relaxation force values. Such information is very helpful for the surgeons in the operation of the surgical tools.

4.2.1 Reliability

By comparing the force model with the experiment data, the force model proposed can approximately estimate the force at certain depth of indentation. It is found that velocity during the indentation will affect the reacting force. Comparing Figure 3-9 and Figure 3-11, the experimental results show that, with the same indentation of 3cm but double indenting velocity, the force values after relaxation are very close to each other, while the maximum forces are different. Therefore due to faster changes in the internal structure, a larger force resulted. After the indentation stopped, the force relaxed to a limiting level based on the real indentation depth. Such effect is also well predicted by the model. It describes not only the local deformation at certain time, but also the history of the deformation.

Although from the experimental results, especially the loading-unloading curves, it shows that as the cycle repeated, the force at the starting point was no longer zero. Although the position was adjusted so that the probe was visually in contact with the tissue surface, assuming contact force started from zero, there might still have some small pre-load force on the load cell. Other than this, as experiments were repeated, the mucus secretes from the tissue might exert certain adhesive force on the probe. These might be the rational source for this offset. And as experiment repeated, due to the preconditioning issue, the experimental results were slightly different from the modeling simulation results. But different from study on the mechanical properties of soft tissue, the tissue in the surgical procedures should not be described as preconditioned. And loading-unloading movement will not happen in the operations. The surgical tools movement should be very purposefully towards finishing certain task. Therefore in the force modeling, preconditioning issue is not designed. That is why in the characterization experiments, only short period left between experiments to avoid preconditioning phenomenon.

Figure 4-5 Comparison between Model and Actual Data on Two Different Specimens

Different specimens were also tested to compare with the model. Results are shown in Figure 4-5. It can be seen that with the same ramp and hold input, the output from two different specimens are similar to the shape of the model. Although the force value from the model cannot exactly match the data, it is because of the different properties of different pigs' maws. Significant variations in stomach geometry and internal structure make a perfect match impossible. But the model can provide an approximate force level which is similar to the real situation to the surgeon.

Overall speaking, the model is able to predict the force profile between the probe and the stomach tissue. It can be reliable and helpful for providing a general idea in event detection and haptic simulations.

4.2.2 Limitations

This model focuses the force on the instrument tool's tip in one dimension, under an assumption that during the endoscopy surgery, the endoscope and tools will always operate perpendicular to the stomach wall. So in this project, we focus on the force feedback in this direction only. However in the real situation, there maybe also exists force in the other two dimensions exerting on the instrument tool's tip, which can bring more complexity to the modeling. Force model in 3D can be the further development direction for this topic.

Other than that, this model can only predict the force when the surgical tool moving forward into the tissue, also known as the loading cycle. During the unloading procedure, the force is not supposed to follow the same cycle. The force model developed in this thesis focus only on this loading cycle because this information appears to be more important for the surgeon during the operations. The surgeons care more about how it feels when pushing the tool into the tissue rather than pulling the

tool out. Of course, some procedures like to grab the tissue with biopsy forceps is out of the scope of this project. After all, to complete the hysteresis loop, and to have more realistic and accurate force feedback when moving the surgical tools backward, unloading cycle modeling is requested.

Moreover, this model is still considered as a phenomenological modeling, which means it mainly consists of fitting mathematical equations to experimental curves. To make this model more robust and physically based, the shape of the probe in future can also be taken into consideration, for example, the size or the shape of the instrument tool's tip. And with different tools, different tasks can be performed. Then corresponding modeling is required. For example, as mentioned above, for the biopsy forceps, the force on the forceps becomes a must.

Chapter 5 Implementation in a 1-DOF Master Slave System

Based on the setup for the validation experiment, a 1-DOF master-slave system was built. The details elaborated in Section 5.1. Section 5.2 discusses the potential of the model implementation in the multi-degree-of-freedom flexible master-slave system developed by our group.

5.1 System Setup

The master-slave system uses the EZlimo linear motion slider in the slave robot. A Faulhaber 3863H024C DC motor is used in the master console. The load cell was still placed at the original position as designed in the validation experiment. The role of the load cell here in the system is functioning as contact sensor, and it can also be replaced by any other contact detector, such as limit switch. User can use the master console to control the movement of the slave robot freely; but once the probe touches the pig maw’s wall, reading from the load cell increases; once the reading of the load cell goes beyond a threshold, it triggers the model to be applied to calculate a corresponding torque for the servo motor. In such a way, the user can feel the “force” at the slave side. Once the probe loses contact with the wall, zero torque will be assigned. Here still utilizing the load cell is because the convenience for the experiment setup in the verification stage, which will be presented in the following paragraph. The flow chart of the system can be found in figures below.

Figure 5-1 the Flow Chart of the System

Figure 5-2 Sketch Map of the Master-Slave System and Labels for Each Components

The sketch map of the whole system is shown in the figure above. For the slave robot side, the configuration remains the same as in the validation experiment. The base holder holds the probe to move together with the slider. The master console employs rack and pinion system to transfer the rotation of the DC motor into translation. There are two mechanisms available for rotation to translation: rack and pinion method and feeding screw method. As shown in the sketch map of the two methods, both of them can provide position and velocity control.

Figure 5-3 Sketch Map of Methods for Rotational Motion to Translational Motion

Although comparing to rack and pinion, the feeding screw method can have more precise adjustment, the friction along the helical screw thread makes it unqualified for this application. The purpose is to move the master in 1-DOF freely; the lesser the

friction, the lesser the influence on the force feedback effect. For the rack and pinion method, to reduce the influence due to friction, the rack gear is mounted on a slide guide as shown in the sketch map of the system setup.

On selection of the rack and pinion, both of them have 20° pressure angle and module 1.0. It's also known that the probability that interference between the tip of the teeth on the pinion and the fillet or root of the teeth on the gear will occur is greatest when a small pinion drives a large gear. Rack gear can be taken as a gear with an infinite pitch diameter. Therefore to ensure this interference does not occur, the surest way to do is to control the minimum number of teeth in the pinion. For the tooth form of 20° pressure angle, involute, full-depth, this minimum number of teeth is 18, as stated in Robert L. Mott's book *Machine Elements in Mechanical Design* [51]. With this number of teeth or a greater number, there will be no interference with a rack or with any other gear. Also the effective moving distance of the rack is about 3cm. Based on these factors, taking the size of the pinion into consideration the pinion is chosen to have 20 teeth.

It is designed that the relation between the moving velocity of the master and of the slave is one-to-one. In such a way, the master console can provide the slave robot position and velocity control. The gear ratio between the DC motor and linear slider is decided through the procedure shown below:

$$v_R = v_t = (D_p / 2) \omega_p$$

The linear velocity of the rack, v_R , must be the same as the pitch line velocity of the pinion, v_t . D_p is the pitch diameter of the pinion, ω_p is the angular velocity of the pinion. Also this velocity must be the same as the velocity of the linear slider, v_s , for a one-to-one relation. The linear slider is a stepper motor, so the velocity depends on the number of pulses, N_p , it receives per second. The resolution of the slider is 0.015mm per one pulse. So

$$(D_p / 2) \omega_p = v_R = v_s = 0.015 \times N_p$$

Also, the angular velocity ω_p is a function of encoder counts per second

$$\omega_p = \frac{2\pi}{N_R} \times N_c$$

where N_R is the number of counts per revolution for the motor encoder, N_c is the number of counts the DC motor encoder detects per second. Therefore the gear ratio, GR, between the master and slave is

$$GR = \frac{N_p}{N_c} = \frac{D_p \pi}{0.015 N_R}$$

With the gear ratio obtained, the probe on the linear slider will synchronize its motion with the rack on the master console. In such a way, the user can feel as he is pushing the probe directly into the tissue. How the system works is shown in the figure below. The user can use his finger to push the rack. The probe will move exactly the same distance as the user's finger. Before the probe touches the tissue, the user will feel no restriction force. Once the probe is in contact with tissue, the model applies; and the user can feel the restriction force according to the distance and the velocity of his finger movement.

Figure 5-4 Sketch Map of the Working Principle of the System

5.2 Measurement of the Force Feedback

As the user controls the master console, when the model is applied, the user should feel the restriction force as if he is in touch with the real soft tissue. The further the user moves his finger into the tissue, the larger the restriction force he should feel. For the viscoelastic behavior, the faster the user moves, the larger the force; and up to certain distance by stopping movement and holding the position, the force he feels should reduce until a steady state. However, the force level is only about few newtons. It may not be so obvious for human to distinguish the difference in these viscoelastic behaviors. If this difference is to be amplified to an obvious level, overall force may be too large. Then it will tend to affect the user's motion rather than providing the force feedback. Therefore in order to have a clear picture about the force feedback, to show the viscoelastic behaviors numerically, meanwhile to prove this method of using torque to provide force feedback really works; previous experiment setup was re-designed to measure the force at the master side.

Another linear slider with the same design as the slave robot was used to provide the input as the finger movement. Because the Galil card used in this project has only two axes and the slave robot will not play any role in this measurement, the slave robot was not connected to the system. The master console was triggered by a switch, once the switch is on, the model will apply. The master console was secured on the base, and adjusted to make the rack at the same height as the probe. For simplicity, the switch was set to be on at the beginning, and the probe was initially in touch with the rack. The same ramp and hold type signal as in the validation experiment was sent to the linear slider. The master console will be pushed up to 3cm at a constant velocity of 0.75mm/s. The force was measured by the load cell. And the torque that was sent to the DC motor was recorded also. By comparing these two sets of data, the viscoelastic behavior can be numerically shown.

Figure 5-5 Experiment Configuration

Figure 5-6 Comparison of the Measured Force and Torque

The experiment configuration and the result are shown in the figure above. Recall that it has already been validated in the previous chapter that the model can be reliable to predict the force at the slave side. From the results shown in the figure above, it can be seen that with this method the force feedback on the master side can follow the

model very precisely. Therefore, this results in that the user at the master side can exactly feel the force change on the slave side. It satisfies the purpose of this application. Also, using this procedure, relation between the force and torque can be obtained. The torque in the motor is set by bias voltage. For different motors, the same magnitude of the voltage set, it provides different torque. This relation is helpful to predict the force on the master side. This can be a reference when designing the force feedback for multi-DOF master-slave system in the future.

5.3 Application in Multi-DOF Master Slave System

The final goal of the project is to understand the forces of the slave robot interacting with the patient and to use motors to create forces that are equal to those being applied to the patient on the master robot, which is controlled by the surgeon. As reported by Brittany Sauser in the article “Surgical Robots Get a Sense of Touch” from the magazine *Technology Review* in December 18, 2006 [19], it says that for the current available systems “the only thing that can be truly felt is resistance in the instruments if they hit an immovable object like bone...”, which is far from the requirement of the working environment for the endoscopy surgery. The force model developed in this thesis can provide a sense of touching “soft tissue”. The force feedback not only related to the tissue indentation but also the speed of the slave robot; so the surgeon can feel the gradually varying force rather than a constant resistance while moving the master. To implement this model into the system, contact detector is required. This can solve the problem of the inaccuracy and size of the force sensor as mentioned in the motivation in Chapter 1.2. A contact detector should be much easier to be built and implemented. Another possible solution to implement the model into the system is to trigger the model on using visual feedback. The surgeon can visually observe the contact from the endoscope camera, and then trigger the force feedback on/off manually. In such a way, though the device cannot tell the surgeon about the contact of the tool and tissue, the surgeon may still be able to feel an approximate force exerting on the tool’s tip, which should improve the performance of the surgery.

Chapter 6 Application in VR Simulation

This chapter presents the work on building a simple VR haptic simulation based on the force model developed. Section 6.1 introduces the haptic device chosen and basic virtual environment setup. Section 6.2 elaborates the algorithm for simulating the deformation of the soft tissue.

6.1 Preliminary Work

Recall the sketch map in Figure 1-6, the dashed line portion can be replaced by virtual environment for the visual feedback; the master robot is also needed to provide the force feedback to the user. Therefore, a PHANTOM desktop OMNI is selected as the haptic interface. This device has 6 degree-of-freedom position sensing, three dimensional force feedback for the point source. And it has compact footprint, with less workspace. What makes it most cost-effective is that this device is affordable. Moreover, the sophisticated development toolkits of SensAble, based on C programming, OpenGL is able to provide the visualization.

Figure 6-1 Six Does of the Phantom^s Omni Unit

As developmental tools, the OpenGL and OpenHaptics toolkits from Sensable are briefly introduced in the following section.

6.1.1 Introduction to OpenGL

OpenGL is a software interface to graphics hardware[52]. It was developed as a hardware-independent version of GL in the early 1990s. It provides a basic library of functions for specifying graphics primitives, attributes, geometric transformations, viewing transformations and many other operations[53].

OpenGL library is specifically designed for efficient processing of three-dimensional applications, but it also can handle two-dimensional scene descriptions as a special case of three dimensions where all the z coordinate values are zero. 3D could be seen as extension of 2D shapes, but in 3D coordinates space. And it also provides sophisticated features as shadows, fog, textures and lighting effects.

OpenGL is streamlined and hardware-independent. It could be implemented on many different platforms, such as Windows, Linux, OSX, and even PDA (Symbian, PalmOS, WindowsCE). Therefore OpenGL could work efficiently on different computers, even if the computer that displays the graphics created isn't the computer that runs it. This allows OpenGL to be used in a networked computer environment where many computers are connected to one another by wires capable of carrying digital data.

With a low level language as C, it allows the user to do almost anything with the hardware as quick as possible. Together with the optimized calculation algorithm and proper hardware acceleration, implementation in real time could be achieved.

6.1.2 Introduction to OpenHaptics Toolkit

The OpenHaptics toolkit is provided together with the SensAble PHANTOM device. It includes the Haptic Device API (HDAPI) and the Haptic Library API (HLAPI) [54]. The HDAPI provides low-level access to the haptic device, enables haptics programmers to render forces directly, and offers control over configuring the runtime behavior of the drivers.

The HLAPI provides high-level haptic rendering. It is built on top of the HDAPI, and it provides higher level control of haptics than HDAPI, at the expense of flexibility in comparison to HDAPI. HLAPI allows the developer to command the haptic rendering pipeline from the graphics rendering loop which makes it more approachable for introducing haptic rendering to an existing graphics loop driven application. However it shields the user from implementing force rendering algorithms and managing the synchronization of servo loop thread-safe data structures and state. HLAPI only deals with the device at the Cartesian space level whereas HDAPI offers access to lower-level control spaces, like the raw encoder and motor DAC values. To allow more maneuverability, HDAPI is a better choice.

6.1.3 Combining Haptics with Graphics

Haptic devices are typically absolute devices⁴, because of the mechanical grounding necessary to provide the force actuation. Therefore, the only way to accommodate non-absolute manipulation is to apply additional transformations to the device coordinates such that the device appears to be moving relative to a given position and orientation instead of its mechanically fixed base. To ensure that the interaction producing force feedback can be effectively visualized in the graphics window, the two scenes which are to be created separately must be mapped onto each other. There is description of some steps for combining the two scenes in Figure 6-2.

⁴ Absolute device is a locating device, that reports its position to the operating system as a set of numbers on a coordinate system

Figure 6-2 Flow Chart for Combining Haptics with Graphics (left) and haptics render (right)

First, the program sets up graphic scene by creating a graphics rendering context and tying it to a window. Then it initializes the haptic scene by creating a haptics rendering context and tying it to a haptic device. Then the program specifies how the physical coordinates of the haptic device should be mapped into the coordinate space used by the graphics. This mapping is to map the geometry specified in the graphics space to the physical workspace of the haptic device.

Next, the application renders the graphic scene using OpenGL. Then the program processes any events generated by the haptics rendering engine such as contact with a virtual object or a click of the stylus button. Then the haptics are rendered, the graphics will be rendered accordingly.

In this process, haptics and graphics rendering loop are typically performed concurrently in separate threads, due to the different refresh rates for two rendering loops. This difference stems from the psycho-physics of human perception. Typically, a graphics application will refresh the contents of the framebuffer approximately 30-60 times a second in order to give the human eye the impression of continuous motion on the screen. However, a haptic application will refresh the forces rendered by the haptic device at approximately 1000 times a second in order to give the kinesthetic sense of stiff contact [54]. It has gained accordant recognition that realistic real-time haptic VR requires feedback at greater than 30Hz for visual sensations and 500-1000Hz for haptic sensations[33, 55, 56]. Due to the refresh rate of the two rendering loops, it's important to consider the time duration of each rendered frame. For graphics rendering, maintaining a 30 Hz refresh rate requires that all rendering take place within 1/30th of second (i.e. 33 ms). Contrast this with the 1-2 ms frame duration of the haptics rendering loop, and it becomes apparent that there's a significant disparity in the amount of time available to perform haptics rendering computations versus graphic rendering computations. Therefore, there are some points need to be considered, for example the state synchronization between the threads should be always treated as a snapshot copy operation versus accessing data in a disjointed fashion using a mutex; and the other common interface as event handling between haptics and graphics loops. The graphics rendering is out of scope for this project, so a brief introduction to the force rendering is presented in the following section.

6.1.4 Force Rendering

For the class of haptic devices supported by the OpenHaptics toolkit, forces are typically used to either resist or assist the user’s motion. There are a variety of ways to compute the forces that are displayed by the haptic device. In this application for this project, the force comes from the interaction between the position of the end-effector and its relationship to objects in a virtual environment. To estimate contact forces, the approach to simulate it through the concept of proxy. Ruspini *et al* in 1997 [57] proposed this modified constraint-based method The method is described below.

Figure 6-3 Sketch Map of Surface Contact Point

The geometry for the proxy is typically a point, referred as SCP (*surface contact point*). SCP is a point that attempts to follow the end-effector position but is constrained to be on the surface of the object. This means that during contact the end-effector can penetrate the virtual objects while the proxy will remain on the object surface. The sketch map is shown in Figure 6-3. In free space the SCP is at the end-effector position as shown in t_0 . When touching an object, the SCP can be calculated by moving the last SCP towards the end-effector position without violating the surface. When the device end-effector penetrates a surface, a transform for the proxy should be computed that achieves a minimum energy configuration between the contacted surface and the device end-effector. In addition, the proxy should respect spatial coherence of contact. Then the force model could be implemented here since it reflects the relation between the force and the position.

6.2 Implementation of the Simulation

For 3-D soft tissue modeling, the Finite Element Method approach is often implemented. In FEM approach, a mesh of nodes and elements is used to define and discretize the domain geometry; various interpolating functions are then used to define the deformation over an individual element. FEM approach can produce very accurate results, but this increased accuracy of FE computations comes at the heavy price of significantly increased computational time. Because it involves with a boundary value problem; and necessarily requires the simultaneous solution for all domain elements. This is computationally expensive, especially for complex geometries and nonlinear behavior [33].

As mentioned in the previous section, the refresh rates for graphics and haptics are very crucial to provide a smooth visual and force feedback. Therefore, the computation time must be efficiently reduced to meet the real-time requirements. In consideration of the scope for this thesis, as a starting point, a virtual “soft tissue” plane is to be created. When the user control the haptic device in touch with the virtual plane, the plane in the graphic window would deform accordingly. Other than the FEM approach, the SurfaceChainMail method [58] was employed for the deformation motion. It is simpler and faster; and it is sufficient to meet the scope to show the application of the presented force model in the VR-based simulator. Details of method and algorithm for the graphical scene were discussed in the sections below. Based on these, a simple VR scene with haptic feedback was built.

6.2.1 Meshing Method

To draw the deformable plane, a mesh of nodes and elements was defined to discretize the surface. First of all, the coordinates of the nodes were defined. This was decided by the dimensions of the elements and the number of the nodes in rows and columns. The elements used in this simulation were selected to be square shape. The nodes

were connected in quadrilateral, and the distances between adjacent nodes were constant.

Figure 6-4 Sketch Map of the Mesh of nodes and elements

Since deformation is involved, the motion is displayed in a 3D environment. To display deformation motion in the dimension other than the plane, depth cue is important. Lighting could be one way to provide the depth cue to the scene. Here normal vector becomes essential. A normal vector is a vector that points in a direction that's perpendicular to a surface. It defines the orientation of the surface in space - in particular, its orientation relative to light sources. The normal vector is used by OpenGL to determine how much light the object receives at its vertices. For a flat surface, one default perpendicular direction normal vector would be assigned for every point on the surface. But for a general curved surface, the normal direction might be different at each point. Therefore without normal vector specified at each node, even deformation occurs, only a flat surface could be observed.

Therefore, the normal vector was required to calculate at each node. To find the normal vector for a node, take any three vertices v_1 , v_2 , and v_3 of the polygon that do not lie in a straight line. The cross product $[v_2 - v_1] \times [v_3 - v_1]$ is perpendicular to the element.

Figure 6-5 Sketch Map of How to Calculate the Normal Vector

For instance, as shown in Figure 6-5, there are four nodes adjacent each node. So for each node, four normal vectors $\mathbf{n}_1$, $\mathbf{n}_2$, $\mathbf{n}_3$, and $\mathbf{n}_4$ could be calculated, since node P is shared by four elements (excluding the boundary condition). To avoid giving too much weight to one of them, calculate $\mathbf{n}_1 + \mathbf{n}_2 + \mathbf{n}_3 + \mathbf{n}_4$ and then normalize it. The resulting vector could be used as the normal vector for node P. For the boundary condition, the average normal vector could be calculated accordingly.

6.2.2 Deformable Motion Calculation Algorithm

As mentioned in previous section, the deformable motion was calculated using SurfaceChainMail method in this scene. This method is illustrated in Figure 6-6, following by the interpretation of the algorithm and the performance of the results.

Figure 6-6 Illustration of SurfaceChainMail Method

Suppose the node A has been moved to the new position. Then its right neighbour, the node E, is a candidate for movement in the chain reaction. The node E is constrained to be greater than *MinDist*, and less than *MaxDist*, in its horizontal separation from A; and to be less than *MaxShear* in its vertical separation. If any of these constraints are violated, then E moves the minimum distance to be within the marked region in Figure 6-6, and therefore within the constraints. Then the same procedure applies to the right neighbour of node E, and so on. Besides the right neighbours, the left, bottom and top neighbours also follow the same rule. The key point is that any node is moved only once. The softness is said to be described by the values of the constraints [58].

However, in this project, the simulation focuses mainly on the deformation along the third axis other than the plane in 3D coordinates space. Therefore, for a flat surface drawn in *x-y* plane, the distance between two adjacent nodes in the mesh is kept constant in *x-y* plane, and the deformation only concerns the one along *z* axis. Moreover, to avoid non-necessary calculation, a maximum influencing range is defined. If the movement of the adjacent node is less than a threshold value, which could be relative small, the nodes beyond these nodes could be considered no movement at all.

Figure 6-7 Deformable Surface in Mesh View

When the end-effector touched the surface, the position of the end-effector may not fall right on the coordinates of a node. An approximation of taking the nearest node applied. The more the nodes used in the mesh, the better and the more accurate the approximation would be. An example of deformation in mesh view is shown in Figure 6-7. The solid sphere into the surface is the cursor. The penetration distance and the speed of the end-effector can be calculated from the Phantom device readings. So the force can be calculated using the model developed. The user can feel the force through the Phantom device as he moves into the virtual plane.

Based on the deformable plane, a simple VR scene was built as shown in the figure below. Manipulator was also simulated with orientation. Although the SurfaceChain method implemented here may not represent a realistic 3-D soft tissue model, due to lack of parameters for viscoelastic behaviors, it is still able to provide a general idea of the concerns about providing haptic feedback in a real-time VR-based simulator. Anyhow, the scope is to show the feasibility of the force model developed in this application. To build a real-time and realistic VR-based simulator is out of the scope of this project.

Figure 6-8 A Simple VR Scene with Manipulator

Chapter 7 Conclusions

This thesis presented the work that has been done for the completion of master degree. To provide haptic information for no-scar robotic surgery inside the GI tract, problems were formulated and analyzed. An innovative method to provide haptics by using a force model to estimate the interaction force based on the movement of the robot was proposed. The focus was on the stomach, and pig's maw was chosen to be the replacement specimen. Therefore haptic device and its role, tissue properties and other related modeling on tissue deformation behavior were studied. To be distinguishable from other's work, due to the large strain of the stretchable stomach tissue, viscoelastic property was taken into consideration. Experiments were performed to characterize and validate the model. The results show that the model is reliable to predict the force profile.

Also, a 1-DOF master-slave system was built. Torques were calculated based on the model and updated to the DC motor to provide the force feedback. The motion for the master and slave were synchronized. Therefore, by applying the model, the force on the slave side can be exactly felt on the master side. Experiment was designed and implemented to collect the force data on the master side. The relation between the force and torque can be obtained through this procedure, and it is helpful for designing the magnitude of the force feedback. It also discussed about the application of this force model in the planning and execution of the endoscopy surgery using multi-degree-of-freedom master-slave robotic system.

The potential of this force model application in the design of virtual environment simulations with haptic feedback was also discussed in the thesis. SurfaceChainMail method was employed to provide efficiency and fast calculation speed, aiming for real time simulation, and a simple virtual reality scene was created based on it. The method is simple and fits the current 1-DOF force model very well.

For further improvement, two main directions can be followed. One direction is to improve the experiment procedures to provide more accurate and realistic data for characterization of the model; the other is to expand the applicability of the model such as modeling in 3-dimensional or for different surgical purposes.

In the actual endoscopy diagnosis procedure, the stomach will be inflated by air to give workspace for the surgical tools. In current experiment setup, unnatural fixtures and holder are used to keep the pig maw under similar condition when it is stretched under inflation. In the future, in-vitro test on an inflated pig maw could be performed to obtain more realistic experimental data. Also in-vivo test are expected to investigate more possible uncertainties in a real environment.

Future work in the second direction should include development of testing and modeling methodologies to measure movements and forces in 3-dimensional (e.g. deformation and friction). This is very helpful to provide more realistic effect in virtual reality simulation. Experiments can be designed and performed to measure the force during the unloading cycle. With the data obtained, unloading cycle can be modeled to complete the hysteresis loop. Also, the tool's geometry may also be a great interest to enrich the model (e.g. probe sizes and tip types). Considering of this, difference surgical procedures with different tools may require different modeling.

In a word, the purpose of future work is to enhance the model's robustness and reliability.

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Appendix

Appendix A Entran ELFM Series Load Cell

Appendix B Faulhaber DC Motor and Encoder

ELFM Series Load Cells

Subminiature - High Stability

CE

ELFM-B1, -B2 & T2

Dim: mm (inches)

ELFM Series

BODY STYLES	Lb RANGES "FS" (± for -T)	N RANGES "FS" (± for -T)	OVERRANGE LIMIT	OUTPUT "FSO" nom. (± for -T)
ELFM-B1 & -T2	5	25	2x FS	2 mV/V
ELFM-B1 & -T2	10	50	1.5x FS	2 mV/V
ELFM-B1 & -T2	25	125	1.5x FS	2 mV/V
ELFM-B1 & -T2	50	250	1.5x FS	2 mV/V
ELFM-B2 & -T2	100	500	1.2x FS	2 mV/V

EXCITATION:	5VDC
BRIDGE IMPEDANCE:	350 Ω
SENSING TECHNOLOGY:	Metallic Foil
NON-LINEARITY:	-B:±0.25%FSO -T:±0.5%FSO
HYSTERESIS:	-B:± 0.25%FSO -T:±0.5%FSO
DEFLECTION AT "FS":	0.025 to 0.075mm (0.001 to 0.003") nom.
THERMAL ZERO SHIFT:	0.02%FSO/°C (0.01%FSO/°F)
THERMAL SENSITIVITY SHIFT (TSS):	0.02%/°C (0.01%/°F)
OPERATING TEMPERATURE:	-50°C to 120°C (-65°F to 250°F)
COMPENSATED TEMPERATURE:	15°C to 70°C (60°F to 160°F)
ZERO OFFSET AT 20°C (70°F):	±2%FSO typ.
MODE:	-B: Compression (+Signal Output for Compression) -T: Tension/Compression (+Signal Output for Tension)
CALIBRATION:	-B: In Compression -T: In Tension
BODY MATERIAL:	Stainless Steel

"Off-the-Shelf" Stocking Program

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Subminiature - High Stability

Smallest Size & Profile

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USA: Fairfield, NJ

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Europe: Les Clayes-sous-Bois, France

SPECIFICATION	ISSUE	PAGE
ELFMS001U	PB2	1 of 2

ELFM specifications continued ...

Options and Accessories:

COMPENSATED TEMPERATURE RANGES:	STANDARD	= 15°C to 70°C (60°F to 160°F)
	Z1	= -20°C to 40°C (0°F to 100°F)
	Z2	= 0°C to 60°C (32°F to 140°F)
	Z4	= 40°C to 90°C (100°F to 200°F)
	Z*	= Non-standard, contact Entran
EXCITATION VOLTAGE:	STANDARD	= 5VDC
	V*	= Non-standard Excitation, contact Entran
SPECIAL LEAD LENGTH:	L00F	= Replace "00" with total length in feet.
	L00M	= Replace "00" with total length in meters.
SPECIAL MODULE LOCATION:	M00F	= Replace "00" with distance between sensor and module in feet.
	M00M	= Replace "00" with distance between sensor and module in meters.
WATERPROOFING LEAD EXIT Not available on ELFM-B1 below 25lbs or 125N:	X	= Short term waterproofing. Limited to 105°C (220°F).
CONNECTOR WIRED TO LEADS:	C	= Microtech type male (w/o mate)
	R	= RJ Telephone type male (w/o mate)
	RQ	= Pins to mate with MM50 screw terminals
WIRING COLOR CODE:	STANDARD	Green = + Output Signal White = - Output Signal
	W+	White = + Output Signal Alt. Color = - Output Signal
MATING CONNECTORS FOR CONNECTOR OPTIONS:	See Cable and Connector Bulletin	
CALIBRATION:	STANDARD	-B: Compression -T: Tension
	Available	Tension & Compression calibration for -T body Compression calibration for -B body

Model Number construction:

ELFM Series	-	B1 Housings	-	50 Range	N Units	-	/Z1/L2M/R Options
		B1 B2 T2E, T2M		(K used for 1000 Ex.: 1K)	N=Newtons L=Pounds		C, R or RQ L00F or L00M M00F or M00M V* W+ X Z1, Z2, Z4, or Z*

"Off-the-Shelf" Stocking Program

Mounting & Wiring:

Entran®	ELFM LOAD CELLS	SPECIFICATION	ISSUE	PAGE
		ELFMS001U	PB2	2 of 2

110 mNm

Series 3863 ... C

Recommended values - mathematically independent of each other								
27	Speed up to	Π e max.	8 000	8 000	8 000	8 000	8 000	rpm
28	Torque up to	M e max.	110	110	110	110	110	mNm
29	Current up to (thermal limits)	I e max.	7,6	4,9	3,8	2,4	1,9	A

Encoders

Optical Encoders

Features:
100 to 1024 Lines per revolution
2 or 3 Channels
Digital output

Series 5500, 5540

		HEDS 5500	HEDS 5540	HEDM 5500	
Lines per revolution	N	100 - 500	100 - 500	1 000 - 1 024	
Signal output, square wave		2	2+1 index	2	channels
Supply voltage	V _{CC}	4,5 ... 5,5			V DC
Current consumption, typical (V _{CC} = 5 V DC)	I _{CC}	17	57	57	mA
Pulse width	P	180 ± 45	180 ± 35	180 ± 45	°e
Phase shift, channel A to B	Φ	90 ± 20	90 ± 15	90 ± 15	°e
Logic state width	S	90 ± 45	90 ± 35	90 ± 45	°e
Cycle	C	360 ± 5,5	360 ± 5,5	360 ± 7,5	°e
Signal rise/fall time, typical	tr/tf	0,25 / 0,25			µs
Frequency range ¹⁾	f	up to 100	up to 100 ²⁾	up to 100	kHz
Inertia of code disc	J	0,6			gcm ²
Operating temperature range		- 40 ... + 100		- 40 ... + 70	°C

¹⁾ Velocity (rpm) = f (Hz) x 60/N
²⁾ HEDS 5540 requires pull-up resistors of 2,7 kΩ between pins 2, 3, 5 and 4 (V_{CC})

Ordering information

Encoder type	number of channels	lines per revolution	For combination with:
HEDS 5500 C	2	100	DC-Micromotors and DC-Motor-Tachos Series 2230, 2233, 2251 2342 2642, 2657 3242, 3257, 3557, 3863 brushless DC-Servomotors Series 2036, 2057, 2444, 3056, 3564
HEDS 5500 A	2	500	
HEDS 5540 C	2+1	100	
HEDS 5540 A	2+1	500	
HEDM 5500 B	2	1000	
HEDM 5500 J	2	1024	

Interlocking connector options: extension cables 300 mm length, on request.

Features

These incremental shaft encoders in combination with the DC-Micromotors and brushless DC-Servomotors are designed for indication and control of both, shaft velocity and direction of rotation as well as for positioning.

A LED source and lens system transmits collimated light through a low inertia metal disc to give two channels with 90° phase shift.

The single 5 volt supply and the two or three channel digital output signals are interfaced with a 5-pin connector.

Motors with ball bearings are recommended for continuous operation at low and high speeds and for elevated radial shaft load.

Details for the Motors and suitable reduction gearheads are on separate catalogue pages.

Output signals / Circuit diagram / Connector information

Output signals HEDS, HEDM with clockwise rotation as seen from the shaft end

Connection diagram HEDS 5540 requires pull-up resistors

Pin Function

Connector suggested connectors
AMP 103686-4/640442-5,
Molex 2695/2759
FCI 65039-032 / 4825x-000