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Modeling of Needle Insertion Forces for Haptics-based Surgical Simulation

Submitted by

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the Faculty of Graduate Studies and Research
in partial fulfillment of
the requirements for the degree of

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Abstract

Virtual reality and haptics-based surgical simulation is a rapidly growing research field motivated by the realization of risk-free and cost efficient virtual environments for medical students to learn surgical procedures and gain hand-on experience.

In minimally invasive surgery, faithful force feedback is crucial for a successful diagnose and treatment especially when little visual feedback is available. Human anatomy includes a variety of fluid filled structures, which are, however, not taken into account in conventional models of force.

In this thesis, a new model of real-time force interaction of needle insertion into soft tissue is developed. Forces are divided into several parts based on the needle tip positions and force characteristics and modeled by different force models. In order to enhance the realism of surgical simulators, the blood effects are taken into account by adding hydrostatic and dynamic fluid pressures into the model. The developed models are compared with real data and the results well confirm the validity of these models. Finally a library of not only the liver insertion force models but also other soft organ insertion force models is established.

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Chapter 1

Introduction

1.1 Motivation

The virtual reality and haptics-based surgical simulator is attracting more and more attention. While there are many outstanding difficulties, for example, the visualization of the operating scene has to be brought to a satisfactory level of resemblance with real medical surgery, realistic force feedback is one of the major obstacles remaining.

The motivation of this work is to increase the realism of the medical haptic simulation by incorporating fluid enclosed structures in target organs into the models. Since many medical procedures occur either in close proximity to, or directly with, fluid filled structures, the addition of the fluid structure to the model is expected to make medical simulations more realistic.

Endoscopy and laparoscopy are excellent examples of the medical procedures that are performed in close proximity and directly with fluid-filled structures. Endoscopy is a safe,

minimally invasive procedure used to evaluate the interior surfaces of an organ. Through the scope, we are able to see lesions, take small biopsies and retrieve some foreign objects. Laparoscopy is a procedure by which we are able to visualize the organs within the abdomen without having to perform surgery. During this procedure, we make two small incisions through which we insert a small camera and other instruments. These treatments have recently become a very popular technique to minimize damage to surrounding healthy tissue normally caused by the process of touching the more inaccessible internal organs during interventions. The small spatial extent of the tissue injury results in a major gain in patient recovery. But the price of these advantages is paid by the surgeon because he loses direct contact with the operation site. The necessary visual information is mediated by a specialized camera and is presented in a screen, distracting normal hand-eye coordination. Due to geometrical constraints posed by the external control of the surgical instruments through the trocar hull, the surgeon loses much of the manipulative freedom usually available in open surgery. Performing operations under these conditions demands very specific skills from the surgeon, which can only be learned through extensive training.

The basic optical and manipulative skills can be learned today by using inexpensive traditional training devices. These training units allow surgeon to learn under monoscopic visual feedback, as well as to acquire basic manipulative components of an intervention, in this way, the surgeon becomes accustomed in completing a particular task. But because the real-life effect is lost, he receives only limited training in dexterity and

problem solving. Additionally, the organs used by these units are generally made of foam, thereby making realistic operative training impossible. While experiments in animals are sometimes used for testing new surgical techniques, practical as well as ethical reasons strongly restrict their use in everyday surgical training.

With the rapid advances in robotics, computer graphics and modeling technologies, virtual reality and haptics-based surgical simulator (VHS) systems offer a very elegant solution to this training problem. VHS systems potentially offer a near-realistic training environment with the possibility for unlimited training over a wide variety of pathologies. However, even with the most realistic visual feedback, a virtual environment cannot be completely immersive, until it does provide haptic stimuli to the user, allowing the operator to feel the virtual structures.

1.2 Background

Haptic simulation of medical procedures is an active area of research in biomedical engineering. Analogous to flight simulators for pilots, medical simulators allow students to learn and practice delicate procedures that are performed on soft tissues in a risk-free virtual environment with both visual and force feedback. Such procedures as suturing [3], bone dissection [4], laparoscopy [5, 6], and needle insertion [7] have been developed and implemented using a variety of architectures.

One of the main challenges in developing medical training simulators is the real-time simulation of soft tissue deformation. Most present surgical simulations [5, 3, 7, 58]

assume that soft tissue is comprised of elastic material with either uniform or varying stiffness. With this assumption, anatomical structures such as blood vessels or cysts are ignored even though their presence has a considerable impact on how the surrounding tissue deforms. By simulating fluid filled structures enclosed in elastic media, this anatomy can be included in surgical simulations, increasing the realism of the simulation.

Real-time force feedback is also very important in medical simulations. It can provide a great source of information for the trainees. It also can allow them to detect transitions between organs and cavities and to identify the nature of the organs and the properties of the tissues.

Before proceeding, a brief introduction to haptic simulation of medical procedures is given to acquaint the reader with the general area of research.

1.2.1 Medical Training Simulation

The goal of medical training simulation is to provide a realistic virtual environment within which students can practice medical procedures. As almost all medical procedures involve hand held instruments like needles, scalpels, or laparoscopic tools, these training simulator systems must allow students to interact with similar tools and feel their effect on the tissue, as well as see it. Figure 1.1 shows two examples of medical training simulations in current medical research. Figure 1.1 (a) shows a needle insertion simulation [7], and Figure 1.1 (b) shows a laparoscopic training simulation [8].

Figure 1.1: Examples of medical training simulators with haptic and graphic feedback (adopted from [7, 8]).

In general, medical simulation systems have three basic components:

1. **Haptic Interface:** The haptic device is the hardware with which the user physically interacts. These devices are instrumented so that the position and orientation of the tool can be calculated, and some or all of the joints are actuated so that force feedback can be displayed as the user interacts with the virtual environment. Because human touch is very sensitive, the force feedback must be updated at a high frequency, with a minimum of 300 Hz [6] for free arm movement. Several different haptic interfaces have been used in a variety of medical training simulators. For example, a 3 degree of freedom (DOF) planar pantograph is used in a simulation of needle insertion as shown in Figure 1.1 (a) for minimally invasive abdominal procedures, a

Laparoscopic Impulse Engine interface is used for the laparoscopic simulator as shown in Figure 1.1 (b).

2. **Real-time Simulation:** The real-time simulation is responsible for computing the interaction physics between virtual tools and virtual objects as the user moves the input device. Computational algorithms for collision detection, deformation, and dynamics are carried out by the simulation for display to the user.
3. **Graphic Display:** Where applicable, visual feedback is shown to the user. Most applications use a mono 2D display. An alternative method for display is to use stereo glasses to provide 3D visual feedback, as used in the Zeus [9] surgical robotics suite.

These three components work together to allow the user to interact with a virtual object, to feel the simulated physical forces through the haptic interface, and to see the motion and deformation of the object on the graphic display. Although graphics and haptics are updated at different rates, they are synchronized so that the touch and visual cues combine to yield a convincing feeling that the user is actually deforming the object on screen.

The flow of information in a virtual environment forms a closed loop [67], as illustrated in Figure 1.2. The model deforms according to the user's input; this deformation results from a deformation force, and the loop is closed when this force is exerted to the user by the haptic interface.

Figure 1.2: Data flow in a typical haptic virtual environment

1.2.2 Deformable Models for Tissue Simulation

Tissue deformation is not covered in this thesis, but in order to make the reader to have a complete idea about typical virtual environment simulation, we will briefly introduce deformable object modeling in the virtual environment for tissue simulation in this subsection.

Several methods for computing deformations are available from the literature. Elasticity based methods such as the Finite Element Method (FEM) [7, 66] and Boundary Element Method (BEM) [69] use continuum mechanics to describe the deformation of an elastic body that has been discretized using either volumetric (FEM) or surface (BEM) elements. Due to its popularity in structural engineering, the FEM has a wealth of research to build upon. These methods are well suited for accurate simulation of tissue because elasticity parameters from experiments can be incorporated into the model easily.

However, these methods are very computationally expensive and require precomputation and numerical optimization for real-time applications [70].

Objects can also be modeled as a mesh of interconnected springs/damper assemblies and point masses. These Spring Mass System (SMS) based methods are quite efficient because of the simplicity of the equations of motion and are very prevalent in computer animation [70]. But the deformations are very sensitive to the spring parameters, and it is difficult to achieve tissue like-behavior.

1.2.3 Haptic Simulation

The work presented in this thesis focuses on modeling of force feedback on the virtual environment. A simplified description of virtual reality simulation for haptic output is given in Figure 1.3.

In order to feel the feedback force, we need a mechanical device, often called haptic display or haptic interface, which interacts physically with the human operator, mediating reaction forces of the virtual environment. Several haptic devices have been proposed, from general force feedback tools [10-14] to instruments specifically adapted for endoscopic surgery [15-17].

Let us consider the systems which simulate the manipulation of virtual objects through instruments. One such haptic interface, representing a virtual instrument, should have the same handle as the device it intends to simulate and should accept the commands imparted by the user, imposing on the human at the same time the forces due

to the interaction of the virtual instrument with the virtual environment. The haptic interface accomplishes therefore a double task. On the one hand, it is an input to the simulation system providing motion and interaction commands, whereas on the other, it provides the user with kinesthetic information about the virtual environment. This reflects the inherent bi-directionality of a kinesthetic interaction, which requires an exchange of mechanical power flow, which is determined through the product of instantaneous

Figure 1.3: Virtual environment simulation.

contact force and velocity. This implies the existence of two substantially different types

of haptic interfaces. In the first type, the positions (or velocities) imposed by the user on the haptic device are measured, and the reaction forces are returned, whereas in the second type, the forces applied by the user sensed and positions are fed back by the haptic device. The first control scheme is called impedance control or force feedback control which has been used in most system up to now [10, 12, 17], and the second one is called admittance control or position-feedback control which has been use in [13].

1.2.4 Fluid Simulation

It is common in medical procedures that one has to interact with fluid as well as solid materials. To model fluid-filled structures, such as blood vessels or glands, fluid physics must be computed using fluid mechanics analysis. Fluid mechanics can be solved using either static or dynamic analysis. Fluid statics is often used in civil engineering for the static analysis of structures like dams or pressure vessels [18]. Static fluid analysis does not include any time-varying quantities, so the only quantity used is hydrostatic pressure. In fluid dynamics, however, fluid motion is considered and quantities such as velocity and hydrodynamic pressure are used to describe the flow of fluid [18].

Both static and dynamic fluid analysis have been used in medical simulations. Any fluid simulation that requires flow, such as bleeding [13] or removal of water and bone paste [19], can be modeled using fluid dynamics. The drawback of using fluid dynamics is that it is very computationally expensive. Due to the computational expense, fluid dynamics is solved only for graphical presentation, which requires a much slower update

rate than haptics does. However, it has been suggested that fluid-filled structures, such as blood vessels or organs, could be modeled using fluid statics to obtain the high update rates necessary for haptics [20]. This thesis will demonstrate that both fluid statics and dynamics can be used for haptic rendering.

1.3 Research Objectives

The primary objective of this work is to develop a realistic model of haptic force interaction between the surgical tool and soft tissue where fluids within structures (e.g. blood vessels) is incorporated for real-time surgical simulation. Different components of insertion forces based upon relative position and motion between the needle and organ are modeled in this thesis. It is shown that the model predicts force feedback in a quite realistic manner.

The second objective is to attempt to extend the models derived from bovine liver data to all kinds of livers and other soft organs such as hearts and prostates.

1.4 Thesis Outline

1. Chapter 1 provides a brief description of haptic virtual environments for medical training, deformable methods, and fluid mechanics. The motivation for and research objectives of this thesis are discussed.
2. Chapter 2 describes the problem that this thesis attempts to solve. A literature

review of the real-time modeling of surgical simulators and haptic rendering of fluids are presented.

3. Chapter 3 provides detailed introductions to six sub-models, which are used in this work.

4. Chapter 4 describes the experimental validation of the models. Two specific models are discussed in detail.

5. Chapter 5 discusses the possibility of building a model library, which contains models simulating real force data in percutaneous therapies.

6. Chapter 6 discusses the conclusions from the work presented in this thesis as well as recommendations for future work.

Chapter 2

Problem Formulation

In this chapter, we will first introduce the Virtual Reality (VR) and Virtual Environment (VE). Then a literature survey of haptic modeling in the medical areas is given and fluid modeling for haptics is presented. In the last, a statement of the problem that we are dealing with is formulated and a solution is proposed.

2.1 Virtual Reality (VR) and Virtual Environments (VEs)

Virtual environments (VEs) provide a unified workspace, which allows almost complete functionality without requiring that all functions to be in the same physical space. According to Ellis [1, 17], VEs can be defined "...as interactive, virtual image displays enhanced by special processing and by non-visual display modalities, such as auditory and haptic, to convince users that they are immersed in a synthetic space." Less technically, a virtual world can be described as an application that lets users navigate and interact with a computer-generated 3-D environment in real time. The system has three

major elements: interaction, 3-D graphics, and immersion [2].

Virtual reality is not only a hardware system but also an emerging technology that changes the way individuals interact with computers [3]. It can be described as "...a fully three-dimensional computer-generated 'world' in which a person can move about and interact as if he actually were in an imaginary place. This is accomplished by totally immersing the person's senses...using a head-mounted display (HMD)" or some other immersive display device, and an interaction device such as a DataGlove or a joystick [4]. However, it is the user immersion in a synthetic environment that characterizes VR as being different from interactive computer graphics or multimedia. In fact, the sense of presence in a virtual world elicited by immersive VR technology shows that VR applications may commonly differ fundamentally from those associated with graphics and multimedia systems [5].

2.1.1 Introduction

According to Bricken [6], the essence of VR is the inclusive relationship between the participant and the virtual environments (VEs), where direct experience of the immersive environment constitutes communication. In this sense, VR constitutes the leading edge of a general evolution of present communication interfaces involving television, computer and telephone [7]. The main characteristic of this evolution is the full immersion of the human sensorimotor channels into a vivid and global communication experience [8]. In fact, VR provides a new methodology for interacting

with information [9]. Since telemedicine is principally involved with transmitting medical information [10], VR has the potential to enhance the telemedicine experience. The two principal ways in which VR can be applied are interface, which enables a more intuitive manner of interacting with information and environment, which enhances the feeling of presence during the interaction [9].

Recently, some research projects have begun to test the feasibility of using VEs in medicine. Applications of this technology are being developed for health care in the following areas: surgical procedures (remote surgery or telepresence [11, 13], augmented or enhanced surgery [14, 15], and planning and simulation procedures before surgery [16, 17]); medical therapy [5, 18, 19]; neuro-psychology [20-25]; preventive medicine and patient education [26]; medical education and training [27, 28]; visualization of massive medical databases [29] and skill enhancement and rehabilitation [30]; and architectural design for health-care facilities [31].

2.2 Surgical Simulators Based on Haptics

Surgery training relies on an apprenticeship model. Students traditionally practice in animals, cadavrs, and patients. Animals do not have the same anatomy as humans, and their use can raise ethical issues; cadavers cannot provide the correct physiological response; there is a risk to patient safety while the student gains competence. Haptics and VR based surgical simulators may address these issues by providing a safe and viable

alternative. Virtual patient models can incorporate realistic human anatomy, while both normal and pathological physiology can be simulated. In addition, simulators can provide a structured learning environment with controlled difficulty levels.

Haptic feedback is essential for realistic training in surgical simulators [68]. Relatively little work has been done on haptic perception. The literature shows in this section a multitude of methods for describing haptic simulation in surgical applications. They are broken up into four classes in terms of their force feedback complexity: needle-based, minimally invasive, open surgery and tele-surgery.

2.2.1 Needle-based Procedure

As the name suggests, needle-based procedures use needles, catheters, guidewires, and other small bore instruments. The Immersion Medical CathSim Vascular Access Simulator [21] was developed to train nursing students in the proper technique for starting an intravenous line. The system used the CathSim AccuTouch device [21] as the haptic interface device to simulate the needle and catheter. The haptic interface consists of an enclosed needle carrier. Sensors on the needle carrier report pitch, yaw, and depth. Encoders on the carrier can detect the depth to which cannula or guidewire has been inserted. As the carrier is pushed or pulled, a variable amount of braking force can be applied. The device can report three degrees of freedom (DOF) orientation data, and provide one DOF haptic feedback along the direction of needle insertion. Real-time visual and haptic feedback was provided during simulation. Liu *et al.* [22] used the same

hardware configuration to develop needle-based trauma procedures such as pericardiocentesis and diagnostic peritoneal lavage.

Needle-based simulators have limited haptic realism. They are useful for teaching relatively straightforward procedures with well-defined algorithms. Their simplicity makes them relatively inexpensive. They are suited for training widely performed procedures at low cost. In situations where opportunities for practice are limited, or where current methods using animal models are not optimal, needle-based simulators can also have an impact.

2.2.2 Minimally Invasive Surgery

Minimally invasive surgery here refers to percutaneous therapies. Percutaneous means performed through the skin, as injection of radiopaque material in radiological examination, or the removal of tissue for biopsy accomplished by a needle. Percutaneous therapies are minimally invasive diagnostic or therapy delivery procedures that involve the insertion of tubular delivery devices (like needles, trocars, bone drills, screws, laser devices, etc.) into targeted locations in the body, typically with the aid of intra-operative imaging. These techniques reduce patient discomfort and decrease recovery time. However, a major shortcoming is reduced visual information transmitted to the physician via the instruments. So the haptic information is particularly interested in this part.

Haptic rendering is the process of computing and generating forces in response to user interactions with virtual objects. The nature of the interaction between the

instruments (needles) and the tissues makes the understanding of deformation, cutting and friction mechanisms quite difficult and deserves further detailed examinations. Different studies have been achieved dealing with the measurement and the modeling of forces involved in percutaneous needle insertion.

Artificial PHANToMs are used by some researchers. Simon P. DiMaio [23] presents a novel approach for estimating needle shaft forces and tissue behavior using a tissue measurement system and soft tissue PHANToM deformation models. Force estimates were used in simulation to compute tissue model deformation, as well as predict target motion and needle trajectories due to tissue deformation. A sensible technologies PHANToM 1.5 desktop unit is adopted by [24, 25] to develop a haptic lumbar puncture simulator that is effective, not cost prohibitive, relatively simple to maintain, and truly usable. Most of the authors are especially interested in the correlation between forces and deformations. Such measurements mainly have an interest to build simulators.

In order to study the behavior of real tissues, animal (human) tissue characteristics are measured by some people. There are two types of tissues available: *ex vivo* and *in vivo*. *Ex vivo* is an artificial environment outside the living organism, *in vivo* means within a living organism (a cell). To my knowledge, very few studies give *in vivo* measurements of forces in surgical procedures. Iman Brouwer *et al.* [26] developed devices to measure *in vivo* tissue properties under extension and indentation, as well as to record instrument-tissue interaction forces. Jeffrey D. Brown *et al.* [27] created an

exponential curve to fit the force data from the *in vivo* organ. Some researchers [26, 28, 29] have developed specialized surgical devices that can acquire *in vivo* data for grasping or probing soft tissues to estimate force distribution along the needle shaft on observed tissue deformations. Tissue properties under *in vivo* conditions provide information about tissue behavior in its physiological state, but none of them specifically deals with percutaneous procedures. Some researchers use *ex vivo* tissue data to build heart model [32], simulating liver microwave ablation [33], monitoring glucose in blood [34] etc..

Among minimally invasive surgery simulators, those for laparoscopy and endoscopy are the most advanced. The interior anatomy generally contains sufficient details and realism for educational purposes. Commercially available laparoscopic trainers (www.surgical-science.com; www.mentice.com) can teach basic skills such as camera navigation, grasping, suturing and knot tying, and cauterization. Laparoscopy simulators have been clinically validated to improve performance in the operating room [35]. Despite recent advances, limitations still exist. Real-time tissue and organ deformation are generally limited to specific organs, or simple structures such arteries, ducts, and other tubular structures. In this thesis, we proposed a methodology to compute and generate forces which can be used for not-one organs in percutaneous therapies.

2.2.3 Open Surgery

Open surgery requires larger incisions in the body. The surgeon often has direct visual and tactile contact with the region of interest. The visual field, range of haptic

feedback, and freedom of motion are considerably larger compared to minimally invasive procedures. Open surgery is thus more difficult to simulate. Early work by O'Toole [36] included a simulator for vascular anastomosis. The system used an SGI Octane for visual rendering. A PC-based system generated haptic feedback using dual Phantom haptic devices. A dedicated 100Mbps Ethernet connection provided synchronization between the PC and the SGI. The system measured the operator's performance, and could distinguish between broad skill levels. Bro-Nielsen [37] described a prototype abdominal trauma simulator with limited haptic feedback. The system was PC-based, haptic effects were rendered using a PHANTOM. Bielser and Gross [38] described the issues involved in simulating incisions and skin retractions. Webster *et al.* [39] described a suturing simulator for incisions and wounds. The system used a dual-processor PC-based system and a PHANTOM Desktop haptic device. Stereographic images provided a 3D rendering of the suture site.

Open surgery remains the holy grail of surgical simulation. Considerable advances in haptics, real-time deformation, organ and tissue modeling, and visual rendering must be made before open surgery can be simulated realistically.

2.2.4 Tele-surgery

During the past few decades, robotics have been applied in industry because of their repeatability, accuracy and the ability to carry out tasks automatically. For those tasks which could not be adequately programmed in advance, tele-surgery was adopted in

which a human operator could interface to control the master when required. Tele-surgery is surgery at a distance which basically needs a master manipulator and a slave manipulator. VR-based telepresence systems are used in tele-surgery to manipulate equipments at remote site. The first modern tele-manipulator was developed by Goertz in 1945. Since then it has been extensively applied in fields such as space investigation, undersea exploitation, nuclear industry, medical therapy, terrestrial mining and construction [40]. The medical applications of tele-manipulators include their development for use in surgery. Green *et al.* [41, 42] have developed a system of telepresence surgery to demonstrate the feasibility. The master and slave units were connected via short cable to perform surgical actions such as grasping and cutting.

The advantages of tele-surgery are clear. For instance, a specialist can help a local surgeon by remote connection, or surgeon to perform complex operations where the local provider is in a rural setting, a ship at sea, an airplane in flight, or even a space station. Besides solving the distance problem, telepresence offers other benefits, such as minimizing the exposure of surgeons to infections and potential trauma [49]. One of the first applications in this field comes from SRI International of Menlo Park, CA. This company designed its Green Telepresence Surgery System (TeSS) to allow surgeons to participate in battlefield operations from sites removed from the front line [50]. The system consists of a remote operative site and a surgical workstation that offers to the surgeon 3-D vision, dexterous precision surgical instrument manipulation, and input of force feedback sensory information. This is done through real-time 3D video vision using

two color CCD video cameras on the remote surgical unit and a stereo display on the surgeon's workstation (the surgeon wears passive polarized glasses). The instrument handles are under the surgeon's display mirror to maintain hand-eye coordination.

The surgeon using this system operates on a virtual image, and a robot on the battlefield reproduces the surgeon's movements. The coupling between the surgeon and the machine has been shown to work from 150 yards away with a fiber-optic connection. The next goal for the tele-surgery is to replace the fiber-optic connection with a wireless signal [2, 50]. A tele-surgery system which was linked through satellites from the United States to Italy to carry out basic tasks such as injecting a piece of pig was tested by Rovetta *et al.* [43]. Because the sensor signals were so unnatural and incomplete, the operator had difficulty in precise, fast control in critical cases such as avoiding cutting through blood vessels or vital organs. Bergamasco [44] designed a system to assist the operator in integrating and interpreting the feedback signals for the specific task of working at a remote site. In order to provide a slave system which reacts to critical situations that occur too fast to be interpreted by the surgeon, Brett *et al.* [45] have demonstrated the benefit of an automatic tool for a stapedotomy to help in detection of breaking through the tissue and automatic withdrawal of the cutter.

However, a challenging issue for all tele-surgical applications is time delay between "master" and "slave" sites. The computation delay ranges from a few milliseconds to 10-100 ms induced by computer networks, to delays of seconds or more induced by multiple satellite communication links [50]. These delays can induce loss of

teleoperator fidelity and task performance as well as dangerous instability of operation [51].

2.2.5 Advantage and Limitation

Remote virtual environments and related technologies add value to health care at least in the areas of improved services, and savings in material resources [76]. First, virtual reality can lead to improved surgical results. Examples include the use of laparoscopic simulators for training [77], the development of applications that simulate human response to medication, simulator systems helping to train anesthesiologists [78], and the development of imaging tools that guide surgical tools through brain tissue to the site of a tumor [79]. Then, the use of simulators saves precious resources such as cadavers and animals. By allowing medical personnel to train using simulators, the demand for non-renewable resources can be drastically reduced. The trainee can practice over and over using a realistic, virtual environment without reducing the supply of non-renewable resources.

However, current virtual environment applications in health care have problems that limit their effectiveness. Specifically, at least three technical problems limit their actual application [75]: the cost, the lack of reference standards, and the non-interoperability of systems. This means that two different departments within the same organization may find themselves having to use two different VR systems because of the difficulty of adapting one single system to their different needs.

In most of medical simulations, haptic feedback coupled with graphical simulation which is a well known bottle-neck in the area of medical robots and computer assisted surgery and training. Haptic systems need high simulation rates (about 1KHz) to obtain realistic force feedback, and the update rates of the physical objects being simulated is normally of the order of 20 Hz to 150Hz. This difference in frequency can cause an oscillatory behavior in the haptic device that can become highly unstable and inflict harm on the operator [46].

Several approaches have been presented as solutions to this problem. Ellis *et al.* [47] solved it by using a numerical approach. They used prediction and correction to extrapolate a force feedback between the simulation updates. However, due to the unknown nature of human behavior, uncertainty in prediction will induce errors that will cause slight vibrations. Cavasoglu and Tendick [48] used balanced model reduction of a lumped element model to obtain the force feedback. Unfortunately, the numerical errors coming from the linearization and from the model reduction are added and can cause unexpected behaviors for non-heterogeneous compliant surfaces. Zilles and Salisbury [49] and Ruspini *et al.* [50] used geometric models which they call god-object or proxy to render haptic forces. Although they did not use deformable models, they treated geometric concavities which are always present in deformable objects.

2.3 Fluid Modeling in Haptics Applications

Most of the present medical haptic simulators, including the commercial products

like Reachin [2] and Surgical Science [51] include fluid simulation only for graphical feedback, to visually indicate bleeding or motion of fluid generated by surgical tools.

For a laparoscopy simulator, Basdogan and Srinivasan [52] described a method of incorporating the simplified 2D Navier-Stokes based wave equations from [53] and [54] with deformable objects using the method of auxiliary surfaces. In this method, a 2D fluid grid is projected onto a deformable object, and the fluid simulation is carried out using the resulting depths from the projection.

In a simulation of mastoidectomy, Agus *et al.* [4] used a combination of the solid mechanics based animation methods described by [55] and [56] to show the motion of water and bone paste. This simulator used a decoupled simulation architecture where haptic force computation was performed at 1000Hz in the “fast” subsystem, and fluid simulation and bone removal were performed at 20 Hz in the “slow” subsystem.

In contrast to the previously mentioned simulation techniques that used fluid only for graphic feedback. De and Srinivasan [57] proposed a nonlinear FEM formulation to model human organs deformation and force feedback as thin elastic walls enclosing incompressible fluid. This method was not implemented in real-time, and its applicability to real-time simulation is based solely on the reduction in dimensionality from a volume to a surface. Laugier *et al.* [58] used the Long Element Method (LEM), developed by Balaniuk *et al.* [59] in a surgical simulation based on principles like incompressible fluid pressure and density to describe haptic simulation. But in real-time simulation, the operator can stop inserting the instrument at any times, the feedback force should change

when the speed changes, so the speed of the operator pushing the instrument should be added in the model. None of the above articles did this.

2.4 Problem Statement

From the literature shown above, there is no work incorporating the dynamics of organ fluids into the display of force feedback in surgical simulation. In order to increase the realism of surgical simulation, the focus of this thesis is to develop a model of force feedback for needle insertion into some soft tissues with the consideration of fluid filled anatomical structures.

2.5 Proposed Solution

In our approach, the insertion force is divided two parts: stiff force and friction force. Different sub-models are used to represent the forces at different stages of insertion. In particular, the spring model, penalty model and exponential function model are used for the stiffness forces; the modified Karnopp model and Elasto-Plastic friction model are employed for friction forces; finally blood vessel models with constant or changing pressure are added in final models.

2.5.1 Limitation and drawback

There are several drawbacks to modeling this force, the first of which stems from

the assumptions for the friction model. The friction data depends on the relative velocity between two connected objects. Because of the softness of liver tissue, it is very hard to calculate the relative speed between the needle and liver, the needle insertion speed is regarded to be the relative velocity. The primary justification for using this assumption is that it can be solved very quickly.

A second limitation is that the blood in the blood vessel is assumed to be inviscid and steady fluid. As a result, the vibration or sloshing of the blood in blood vessels caused either by user interaction or biological forces is not considered. This assumption is made to allow for fast computation time.

Chapter 3

Methodology

In this chapter, we describe in detail the models we used and the rationale for the choice of these models of the needle insertion forces in surgical simulation.

3.1 Introduction

For needle insertion in percutaneous therapies, it is vital that the operator has detailed control of the needle position and velocity. For a method that involves a needle being held and moved by an operator, many factors, such as amount of force applied, are dependent on operator skill. Therefore, operator skill and endurance can affect the results of needle insertion. Since the area surrounding the target location can be easily damaged, plus complicating factors including tissue deformation before puncture, needle bending, and relaxation of the tissue after puncture, these procedures require rapid reaction time and control greater than typical human ability.

The use of cooperative robotic systems for positioning and manipulating therapy

delivery devices presents the option of simultaneous human and computer control. An important justification for utilizing a robot for assistance during needle insertions is the ability to maintain a controlled velocity behind the needle and react to minute changes in force, which a human user is often unable to do. Clearly, a force sensor is thus required between the robot and the needle to allow the force to be measured. The computer controlling robot can determine what speed to move the needle based on real-time force feedback and knowledge of the needle's position. The primary goal of this work is to develop haptic models for percutaneous therapy surgical simulation

The liver was chosen as the initial organ with which to obtain data. The American Cancer Society estimates that 17,550 new cases (12,130 in men and 5,420 in women) of primary liver cancer and intrahepatic bile duct cancer will be diagnosed in the United States during 2005 [60]. Liver biopsy is a percutaneous procedure that is often used to diagnose the cause of chronic liver disease. It is also used to diagnose liver tumors identified by imaging tests. In many cases, a biopsy is used to confirm chronic liver disease, as well as determine the amount of damage to the organ. Liver biopsy is also used after liver transplantation to determine the cause of elevated liver tests and determine if rejection is present [61]. One approach to treating liver cancer is via ablation. Ablation refers to methods that destroy the tumor without removing it, and is also a percutaneous procedure.

3.2 Experimental Equipments

Limited by present lab conditions, direct liver data can't be extracted by experiments, liver data used in this thesis are adopted from a few other papers. A brief introduction to basic experimental equipments used by other authors is given to help the reader understand the whole procedure if one needs to build a real-time force model.

Figure 3.1: Experimental equipments. (a) The Johns Hopkins University Steady-Hand Robot. (b) The delrin needle holder. (c) The Entran[™] force sensor (adopted from [30]).

First, Johns Hopkins Steady-Hand Robot being used by Simon [30] is shown in Figure 3.1 (a). It is a 7-degree-of-freedom manipulator performing one-dimensional translational motion (Axis 5). This axis makes use of a telescoping crossed-roller slide mechanism driven via flexible cable, which provides greater than 150 *mm* of travel. It is known for its remote center of motion capabilities, and has a positioning resolution less

than $2.5 \mu m$. Second, a needle holder shown in Figure 3.1 (b) was needed to hold the needle firmly without slipping and interface with the force sensor. It is comprised of a vise grip and several pieces of delrin for the casing. The needle is first inserted into the vise grip and tightened to prevent any travel along its axis. The vise grip is then clenched between symmetrically cut pieces of delrin designed to clamp the vise grip and leave sufficient space for needle hubs of a range of sizes. Third, an EntranTM force sensor is necessary to capture the needle insertion forces along the axis of the needle as shown in Figure 3.1 (c). The load cell utilizes micro-miniature semiconductor strain sensing elements for unamplified output signals of 100 mV full scale. It is a 10 N capacity load cell with a sensitivity of $12.3 mV/N$ and an overrange limit of 20 N.

3.3 Data Acquisition

B.Maurin *et al.* [62] used Nano-17 6 DOF force sensor from ATI Automation Inc and a special handling tool which can either be mounted as a robotic arm end-effector or held by hand to acquire *in vivo* data with 18 gauges 15.24 cm biopsy needle in a living pig. Two different methods were used by Maurin. In the first case, a proficient radiologist inserts a needle by manually holding the force device attached to the needle. The task executed by the radiologist was to plunge a needle 4 times into the targeted organs, at a constant velocity, constant depth and with as less shaking as possible. In the second case, the needle insertion is done by a robot, with the assistance of a surgeon at constant

velocity of 15 mm/s. The average maximum force obtained for the radiologist manual insertions is 0.7 N with standard deviation 0.29 N, and the average maximum force gotten from robotic insertions is 0.59 N with standard deviation 0.17 N. the difference between manual and robotic insertions is mainly due to the uncontrolled speed and depth of the radiologist. A typical plot of insertion force is shown in Figure 3.2.

Figure 3.2: Needle insertions into a living pig liver (adopted from [62]).

Christina Simone [30] used the equipments we mentioned in section 3.2 to obtain data with a needle at a constant velocity of 3 mm/s inserting into a whole bovine liver. Two typical force vs. position plots are shown in Figure 3.3. From the diagrams, we can easily find that same equipments used to insert into livers coming from different bovines can create different force values at different positions, which means that it is impossible

to build a model to match all liver insertion force data, yet we also find that the overall force shapes are similar. In this chapter, we will build one signal model to simulate average force data which represents the overall shape of the insertion force.

(a)

(b)

Figure 3.3: Two force vs. position diagrams coming from two livers (adopted from [30]).

3.4 Data Analysis

Before proceeding, it is necessary to note the anatomy of the liver. As shown in Figure 3.4, liver regulates most chemical levels in the blood and holds about one pint (13 percent) of the body's blood supply at any given moment. Permeating the entire liver structure is a system of blood capillaries, bile capillaries and lymph capillaries. The force vibrations in the second phase may be caused by collisions with these arteries and veins in the liver organ. Because it is hard to model the liver and its cells, we can assume that the liver is composed of a great number of blood vessels and we only need to model these

blood vessels instead of creating an exact liver model for the structural details.

Figure 3.4: A cutaway of one of the livers revealing large interior vessels.

The above plots (Figure 3.2 and Figure 3.3) can be divided into 2 phases as shown in Figure 3.5. The first phase is called the ascent phase (before point ②). It corresponds to the succession of an exponential-like rise, a sharp rupture, and finally again a rising slope, which we call stiffness force. The second phase is called the shaking phase (from point ② to point ③). It shows the friction as well as the force while needle tip colliding with small interior structures such as blood vessels.

3.4.1 Stiffness force

The stiffness force is due to the elastic properties of the organ and its capsule. Phase one (before point ② in Figure 3.5) is created due to the stiffness of the tissue. In order to model the stiffness force correctly, we divide the phase one into two parts: the first part represents the plot before point ①, and we called it ‘before puncture capsule’;

the second part which we named it ‘before friction occurs’ is from point ① to the highest point (point ②).

Figure 3.5: ‘before puncture capsule’ and ‘before friction occurs’ in first phase.

The general form of the stiffness force on the needle is described by the relationship

[1]

$$F_{stiffness} = \begin{cases} 0 & x_{tip} < x_1 \\ f(x) & x_1 \leq x_{tip} \leq x_2 \\ 0 & x_{tip} > x_3 \end{cases}$$

where $f(x)$ is a one dimensional stiffness model, x_{tip} and x_1, x_2, x_3 are the positions of the needle tip and tissue surface relative to a fixed coordinate system before puncture. The relative locations of x_1, x_2 , and x_3 can be seen in Figure 3.6.

Figure 3.6: Locations of the tissue surface at different stages of needle insertion. (a) Pre-puncture (b) ‘before puncture capsule’ (c) ‘before friction occurs’.

3.4.1.1 ‘before puncture capsule’

So-called ‘before puncture capsule’ means that the needle point has not punctured the surface of the liver but nearly goes through, which refers to the period that the tip of the needle just touches the surface of the tissue to the first sharp rupture just occurs (before point ① in Figure 3.5).

The first hypothesis is to model the tissue as a linear elastic material. This assumption, which considers the stress/strain relationship of the system as always linear

during the simulation, is called mechanical linearity. The linear laws of the viscoelasticity which have been used by many other researchers were put forth by Fung [63]. According to Fung, the material properties can be described using the Hooke's law. But we did not use it in this thesis because (1) biological tissues are more complex especially with large deformations; (2) from the data in Figure 3.2 and 3.3, they are obviously nonlinear.

Biological tissue is linearly elastic for small deformations [63]. Liver is very soft tissue and doing percutaneous therapy in it will result in large deformation (as the experimental data shows). The force must be modeled using a nonlinear method. Equations used by Ottensmeyer *et al.* [29] who is developing devices to acquire *in vivo* data were examined. Ottensmeyer approximates tissue geometry by a semi-infinite body undergoing normal indentation by a right circular punch as illustrated below with the equation:

$$f_z = 8Ea\delta_z / 3K \quad (1)$$

Figure 3.7: Indentation of a semi-infinite elastic body with a rigid, right circular punch (adopted from [29]).

where δ_z and f_z are the displacement and force normal to the surface, a is the cylindrical

indenter radius, E is Young's modulus, and K is unity for a semi-infinite body. This relation is also inappropriate to the work presented here because of two reasons: first, the needle used by Ottensmeyer has a blunt tip, however, the data we used here come from a needle with sharp tip; Second, the boundary conditions assume that the tissue geometry can be approximated by a semi-infinite body with effectively simply-supported end conditions, while in reality the end conditions deform as well.

A nonlinear spring model was tested and finally used in this thesis. As demonstrated by d'Aulignac *et al.* [68], a nonlinear spring model is

$$f(x) = \frac{x}{ax + b} \quad (2)$$

Simone [1] and B. Maurin *et al.* [62] proved that a second order polynomial that relates depth to force can have lower rms error values when fitting parameters to match the deformations. The formula is given by

$$f(x) = a_0 + a_1 x + a_2 x^2 \quad (3)$$

where x is the difference in the length of the spring with respect to its original, resting length. The parameters a_0 , a_1 , and a_2 were fitted to match the experimental data, we assume that before the tip of the needle touches the surface of the liver, there is no force needed, we have: $a_0 = 0$, and a_1 , a_2 can be different values due to different livers or even same liver with different injections. Figure 3.8 shows the typical stiffness data and an average model. These data came from one bovine liver within a 10 cm^2 area, the difference values among stiffness may due to the different internal structures directly

under the membrane of the liver. if we choose a_1 and a_2 properly, we can build a library containing models to fit all liver data, which will be mentioned in chapter 5.

Figure 3.8: Typical stiffness data and a model with $a_1=0.002 \text{ N/mm}$,

$$a_2=0.0023 \text{ N/mm}^2 \text{ (adopted from [1])}.$$

3.4.1.2 'before friction occurs'

It is worthwhile to note that in the one-dimensional insertion, there is a sharp rupture in the first phase, we call it “pre-puncture event” which occurs possibly when the sharp sheath catches on the tissue while the point has already punctured the surface.

There are two models being considered to represent ‘before friction occurs’ part

(from point ① to point ② in Figure 3.5). The first model considered is called exponential function being described by:

$$f(x) = (F_0 + b) e^{a(x-d_0)} + b \quad (4)$$

It is a function of depth x , F_0 , b , a , and d_0 are all parameters needed to be adjusted according to the real data. B. Maurin [62] used this function to model the whole stiffness force (from original point to the point ② in Figure 3.5), and divided the force into two parts and set the vector $\Theta = [a, b, d_0, F_0]$ separately. First part is before point ①, which equals ‘before puncture capsule’ in section 3.4.1.1, and the vector $\Theta = [a, b, d_0, F_0]$ is set to be $\Theta_1 = [0.121, -0.098, 11.45, 0.2]$; second part equals ‘before friction occurs’, and has the vector $\Theta_2 = [-0.031, 1.7, 19.61, -3.39]$. In this thesis, the parameter vector Θ_2 is used to represent the ‘before friction occurs’. In Figure 3.9, a nonlinear spring model is used to model ‘before puncture capsule’ and exponential function with second part parameter vector is used to suit the ‘before friction occurs’.

By adjusting the parameters, the created model can match the real data (Figure 3.3 (b)) correctly. But we want to build not only an average model but also a model library which contains models to represent as many as possible real data by changing the model parameters, since this exponential function has four parameters needed to be adjusted, it is time consuming.

The penalty model is the second model to be considered. The goal of penalty models is to give a continuous representation of the collision phenomenon at a

macroscopic level. It is obvious that a microscopic study should lead to a model of molecular structure, which does not seem feasible in real time. From a macroscopic point of view, the only thing that one can measure is the deformation of colliding bodies.

Figure 3.9: Comparison between a model and real data for the stiffness force.

Almost all materials are deformed during a collision. Empirically, this deformation is close to a linear function of force, which means that the collision force is proportional to the deformation. As this deformation is almost impossible to model, a common hypothesis is that the value of the interpenetration of the two volumes is

equivalent to the deformation so that the contact force between two linearly elastic bodies is not itself linear but is nonlinear and of the form $F(x) = \lambda x^n$ where x is a geometrical measure of the interpenetration, λ is the constant coefficient. A viscous damping term is introduced in order to model visco-elastic phenomenon of the liver and the following equation is proposed by K. H. Hunt and F. R. E. Crossley [64].

$$F(x) = -\lambda x^n - \mu x^n v \quad (5)$$

where v is the relevant velocity between the two collision objects.

Parameters λ and μ are supposed to represent the physical properties of the materials, to determine the parameters, two useful results are given by [64]:

$$\mu = \frac{3}{2} \alpha \lambda \quad (6)$$

$$e = 1 - \alpha v_i \quad (7)$$

where v_i is the relative speed of the two collision objects just before the collision happens, e is called coefficient of restitution which has $e \in [0, 1]$, if $e = 1$, the collision is perfectly elastic, if $e = 0$, plastic collision happens. Combining equation 5, 6 and 7, we got:

$$F(x) = -\lambda x^n \left(1 + \frac{3}{2} \left(\frac{1-e}{v_i}\right) v\right) \quad (8)$$

let's assume that v_i equals v because of this force happening only at the very beginning of the puncture, λ and e should be very small due to the liver characteristics, after data

adjustment , the following diagram (Figure 3.10) has been achieved with $\lambda = 5 \text{ N/m}$ and $\alpha = 250 \text{ sec/m}$.

Penalty method and exponential function are all adopted to create final models as well as to build a model library. Due to the relaxation of the tissues, not surprisingly, even if the needle is motionless, we still have a slow decreasing force applied along the axis of the needle, which is known as a repulsion force applied on the tip.

Figure 3.10: Comparison of a model built by spring and penalty model with real data.

3.4.2 Friction Force

Friction is present in all machines incorporating parts with relative motion. To understand the tribology of engineering surfaces, it is necessary to consider the surface

topography. Early models of friction failed because the surface topography was misunderstood. The interactions at contacting surfaces will be examined by considering progressively smaller contacts [71]. In Figure 3.11 (a), a conformal contact is shown schematically, part A rests on part B, such contacts are identified as area contacts. The apparent area of the contact is determined by the size of the parts.

Parts that do not enjoy matching radii of curvature meet at non-conformal contact, as shown in Figure 3.11 (b). These contacts are called point or line contacts, but this is an idealization. In fact the parts deform to create an apparent area of contact, an area that increases with increasing load.

Crystalline surfaces, or even apparently smooth surfaces, are microscopically rough. The protuberant features are called asperities and shown in Figure 3.11 (c). The true contact occurs at points where asperities come together. In this way, the true area is much smaller than the apparent area of the contact. Contact occurs at asperity junctions, and they have two important behaviors: they deform elastically, giving rise to presliding displacement; and both the boundary film and the asperities deform plastically, giving rise to rising static friction.

Figure 3.11: Different contact surfaces between two objects (adopted from [71]).

From macroscopic view, when a frictional contact is in static, there may none the less be relative motion [71]. This arises with tangential compliance and, because there is no true sliding, is called presliding displacement. A friction model renders presliding displacement if variations in applied force below the breakaway force produce elastic deformation and movement. A friction model renders static friction, or stiction, if, for applied forces smaller than a breakaway force, there is no steady-state relative motion.

3.4.2.1 Modified Karnopp Friction Model

The most commonly used models for friction are viscous damping (Figure 3.12 (a)) and Coulomb model (Figure 3.12 (b)). Viscous damping is rarely encountered in realities, Coulomb model states that the force magnitude is constant when the velocity is non-zero, and when the velocity is zero, the friction can take on whatever value which is necessary, up to the static friction limits, to cancel other applied forces so that the zero velocity condition is maintained. The phenomenon of “stick-slip” is modeled by allowing the static friction magnitude to be higher than the dynamic value (Figure 3.12 (c)). Karnopp’s model (Figure 3.12 (d)) where the static friction condition exists at non-zero velocity if the magnitude of the velocity is less than a small, predefined value was proposed by [80].

With the insertion of the needle, friction force occurs along the length of the needle during insertion due to tissue adhesion and damping. In this thesis, two friction models were tested. First model is a modified version of the Karnopp friction model introduced by Richard [65] as shown in Figure 3.13. This model not only includes viscous and Coulomb frictions but also accounts for the “stick-slip” phenomenon, it also allows for asymmetric friction values for positive and negative velocities.

Figure 3.12: Friction models. (a) viscous damping (b) Coulomb model (c) “Stick-slip” model (d)

Figure 3.13: The modified Karnopp model (adopted from [65]).

The modified Karnopp friction model equation is

$$F_{friction}(\dot{x}, F_a) = \begin{cases} C_n \operatorname{sgn}(\dot{x}) + b_n \dot{x} & \text{For } \dot{x} < -\Delta v \\ \max(D_n, F_a) & \text{For } -\Delta v \leq \dot{x} < 0 \\ \min(D_p, F_a) & \text{For } 0 < \dot{x} < \Delta v \\ C_p \operatorname{sgn}(\dot{x}) + b_p \dot{x} & \text{For } \dot{x} > \Delta v \end{cases}$$

where C_n and C_p are the negative and positive values of dynamic friction, b_n and b_p are the negative and positive damping coefficients, D_n and D_p are the negative and positive values of static friction, $\dot{x}$ is the relative velocity between the needle and liver, Δv is the value below which the velocity is considered to be zero, and F_a is the sum of non-friction forces being applied to the system.

The relative velocity between the needle and the tissue is represented by:

$$v = v_{needle} - v_{tissue} \quad (9)$$

A. M. Okamura *et al.* [31] used an adjustable height platform to put the liver in order to limit the pre-sliding displacement and keep the constant thickness of the liver. Only two velocity values are considered, one relative velocity value is zero, which means the liver and the needle move in unison, it happens at the beginning of the friction occurrence; the other relative velocity value is 3 mm/s, which equals the needle insertion speed, due to the restriction of the platform, the tissue is no longer moving which results in the relative velocity equaling that of only the needle. Table 1 lists the estimated friction parameters (adopted from [31]) and Figure 3.14 shows the friction model.

Parameter	Mean	Standard Deviation
D_p (N/m)	18.45	5.42
D_n (N/m)	-18.23	4.30
b_p (N-s/m ²)	212.31	45.70
b_n (N-s/m ²)	-293.08	30.73
C_p (N/m)	10.57	4.34
C_n (N/m)	-11.96	4.40

Table 1: Estimated friction parameters for bovine liver (adopted from [31]).

Figure 3.14: Modified Karnopp friction model with the parameters in table 1.

3.4.2.2 Elasto-Plastic Friction Model

The modified Karnopp's model only shows the static force or stiction. When needle goes through the liver, due to the softness of the liver, significant presliding displacement occurs. P. Dupont *et al.* [72] used Elasto-Plastic friction model to exhibit both stiction and presliding displacement.

Dahl model [74] introduces the presliding displacement into friction modeling by incorporation tangential compliance. LuGre friction model [73] is an extension of Dahl model, and the Elasto-plastic friction model is an extension of LuGre friction model. The Dahl model introduces a state which represents deformation of a contact asperity and friction force depends on this state. The model is

$$F_f = \sigma_0 z, \quad \sigma_0 > 0; \quad \dot{z} = \dot{x} \left(1 - \frac{\sigma_0}{f_c} \text{sgn}(\dot{x}) z \right)^i \quad (10)$$

where z specifies the state of strain in the frictional contact, and σ_0 is the contact stiffness.

The LuGre model can be written:

$$F_f = \sigma_0 z + \sigma_1 \dot{z} + \sigma_2 \dot{x}, \quad \sigma_j > 0 \quad (11)$$

$$\dot{z} = \dot{x} \left(1 - \frac{\sigma_0}{f_{ss}(\dot{x})} \text{sgn}(\dot{x}) z \right) \quad (12)$$

Where σ_0 and σ_2 are Coulomb and viscous friction parameters, σ_1 is the damping coefficient for the tangential compliance, and $f_{ss}(\dot{x})$ is the steady-state friction force versus rigid body velocity, also called the Stribeck curve. The Stribeck curve describing

friction as a function of steady-state sliding velocity indicates that the steady-state friction force is a decreasing function of velocity magnitude. The Elasto-Plastic model is described by Equation 11 and with $\dot{z}$ given by:

$$\dot{z} = \dot{x} \left(1 - \alpha(z, \dot{x}) \frac{\sigma_0}{f_{ss}(\dot{v})} \operatorname{sgn}(\dot{x}) z \right)^i, \quad \frac{\sigma_0}{f_{ss}(\dot{x})} > 0 \quad (13)$$

where $\alpha(z, \dot{x})$ is used to achieve stiction. The body displacement x is decomposed into its elastic and plastic components z and w , and sticking and sliding can be defined in terms of rate equations:

$$\begin{cases} \dot{x} = \dot{z}, \dot{w} = 0 & \text{for sticking-elastic displacement} \\ \dot{x} = \dot{w}, \dot{z} = 0 & \text{for sliding-plastic displacement} \end{cases}$$

When $\operatorname{sgn}(\dot{x}) \neq \operatorname{sgn}(\dot{z})$,

$$\alpha(z, \dot{x}) = 0 \quad (14)$$

when $\operatorname{sgn}(\dot{x}) = \operatorname{sgn}(\dot{z})$,

$$\alpha(z, \dot{x}) = \begin{cases} 0 & |z| \leq z_{ba} \\ 0 < \alpha < 1 & z_{ba} < |z| < z_{\max} \\ 1 & |z| \geq z_{\max} \end{cases} \quad (15)$$

where z_{ba} is a breakaway displacement such that for $|z| \leq z_{ba}$, all motion of the friction interface consists of elastic displacements. The Equation 15 can be graphically depicted in Figure 3.15.

We choose $\alpha(z, \dot{x})$ is a sin function which is given by

$$\alpha(z, \dot{x}) = \frac{1}{2} \sin \left(\pi \frac{z - \left(\frac{z_{\max} + z_{ba}}{2} \right)}{z_{\max} - z_{ba}} \right) + \frac{1}{2} \quad z_{ba} < |z| < z_{\max} \quad (16)$$

Figure 3.15: Diagram of $\alpha(z, \dot{x})$

and the Stribeck friction curve is given by

$$f_{ss} = f_c + (f_{ba} - f_c) e^{-(v/v_s)^2} \quad (17)$$

Figure 3.16 shows the Elasto-plastic friction model. In the diagram, $\sigma_0 = 0.08$, $\sigma_1 = 0.002$,

$\sigma_2 = 0.045$, $f_c = 0.3N$, $f_{ba} = 0.25N$, $Z_{ba} = 1.5mm$, $Z_{\max} = 0.3mm$, and $v_s = 0.1m/s$.

Figure 3.16: Friction model with Elasto-plastic friction model.

From the real data and the liver structure introduced before, we know that when needle goes through the inside of a liver, there are many collisions between the tip of the needle and liver internal structures such as blood vessels represented by the vibrations between point ② and point ③ in Figure 3.5. In order to create a model which can simulate these impacts, blood vessel models with or without changing pressure need to be built and added to the friction model.

3.4.3 Blood Model

Let us consider the procedure of needle passing through blood vessels. First, the

tip of the needle needs to contact one side of the blood vessel wall from outside and penetrates it, then the tip contacts with the other side of the blood vessel wall from inside and then penetrates it. Because the insertion speed of the needle is slow and the diameter of the blood vessel is small (2 mm), we assume that the tip of the needle does not penetrate the first contact wall (from outside) until just before the tip of the needle touches the other side of blood vessel wall (from inside), the whole procedure can be plotted in Figure 3.17.

In order to model the force encountered during the procedure, we subdivided the whole procedure into two parts. The first part is from the tip just contacts the outside wall of the blood vessel (Figure 3.17 (b)) to the tip just touches the inside wall of the other side of the blood vessel (Figure 3.17 (d)); the second part is from the end of the first part to the tip of the needle fully penetrates the whole blood vessel (Figure 3.17 (f)).

In the first part, the needle encounters not only the stiffness force coming from the blood vessel wall but also the pressure exerted on the tip of the needle by blood. The stiffness force of the wall can be represented by linear spring model $f_s = kx$, where k is the spring coefficient of the blood vessel wall.

Figure 3.17: (a) – (f) shows the whole procedure of the needle passing through blood vessel.

The pressure force along the insertion direction can be calculated as

$$f_p = ps \quad (18)$$

where p is the pressure and s is the contact area. The pressure p only exerts on the tip of the needle, We assume the shape of needle tip is a cone (shown in Figure 3.18), and at any moment between Figure 3.17 (b) to Figure 3.17 (d), we have the height of the cone equals the diameter of base area of the cone inside the blood vessel.

Figure 3.18: Shape of a needle tip where $h = 2r$ at any time during penetrating the vessel wall.

The side area of the cone can be calculated as

$$s = \pi r l \quad (19)$$

combining Equation 18 and 19, we got:

$$f_p = p \pi r l \quad (20)$$

therefore, the force model for the first part can be described by

$$F_1 = f_s + f_p \quad (21)$$

In the second part (from Figure 3.17 (d) to Figure 3.17 (f)), the force needle tip encountered comes from the stiffness of the blood vessel wall and capsule. We assume that the distance for the second part is 1 mm (compared with the 2 mm blood vessel diameter), so the deformation is small, we use linear spring model to represent the second

part force F_2 .

Finally the blood model can be described by

$$F = F_1 + F_2 = f_s + f_p + F_2 \quad (22)$$

3.4.3.1 Constant Pressure Blood Model

Generally speaking, the fluid mechanics can be subdivided into fluid statics, in which the fluid is at rest, and fluid dynamics, in which the fluid is moving. In this section, we will consider the blood in the blood vessels as fluid statics.

In percutaneous therapies, such as liver biopsy, let's assume that needle goes into the liver in x direction, as shown in Figure 3.19. We need to calculate the pressure along the needle tip trajectory which is x axis here.

Figure 3.19: Direction of the needle insertion into the blood vessel.

We assume that the diameter of the blood vessels in a liver is normally 2 mm and regard the shape of blood vessel as rectangular. If the pressure at the center of the blood vessel is designated as p , then the average pressure on the various faces can be expressed in terms of p and its derivatives. We are using actually a Taylor series expansion of the pressure at the vessel center to approximate the pressures a short distance away and neglecting higher order terms. The resultant surface force δF_x in the x direction, δF_y in the y direction and δF_z in the z direction are:

$$\delta F_x = -\frac{\delta p}{\delta x} \delta x \delta y \delta z \quad (23)$$

$$\delta F_y = -\frac{\delta p}{\delta y} \delta x \delta y \delta z \quad (24)$$

$$\delta F_z = -\frac{\delta p}{\delta z} \delta x \delta y \delta z \quad (25)$$

combining the above three equations, we got the resultant surface force acting on the blood vessel in vector form as

$$\delta F_s = -\left(\frac{\delta p}{\delta x} \hat{i} + \frac{\delta p}{\delta y} \hat{j} + \frac{\delta p}{\delta z} \hat{k} \right) \delta x \delta y \delta z \quad (26)$$

which can be expressed in the form of

$$\delta F_s = \delta F_x \hat{i} + \delta F_y \hat{j} + \delta F_z \hat{k} \quad (27)$$

where $\hat{i}$, $\hat{j}$ and $\hat{k}$ are the unit vectors along the coordinate axes. The group of terms in parentheses in Equation 26 represents in vector form the pressure gradient and can be

written as

$$\frac{\partial p}{\partial x} \hat{i} + \frac{\partial p}{\partial y} \hat{j} + \frac{\partial p}{\partial z} \hat{k} = \nabla p \quad (28)$$

Since the z axis is vertical, the weight of the element is

$$\delta W = -\gamma \delta x \delta y \delta z \hat{k} \quad (29)$$

where γ is the fluid weight. The negative sign indicates that the force due to the weight is downward (in the negative z direction). Newton's second law, applied to the fluid element, can be expressed as

$$\sum \delta F = \delta m a \quad (30)$$

where $\sum \delta F$ represents the resultant force acting on the blood vessel, a is the acceleration of the element, and δm is the vessel mass, which can be written as $\rho \delta x \delta y \delta z$. It follows that

$$\sum \delta F = \delta F_x + \delta W = \delta m a \quad (31)$$

which equals

$$-\nabla p \delta x \delta y \delta z - \gamma \delta x \delta y \delta z \hat{k} = \rho \delta x \delta y \delta z a \quad (32)$$

and, therefore,

$$-\nabla p - \gamma \hat{k} = \rho a \quad (33)$$

We already assumed that the blood in the blood vessels is regarded as static, which means the fluid is at rest, so we have $a = 0$ and Equation 33 can be reduced as

$$\nabla p + \gamma \hat{k} = 0 \quad (34)$$

or in component form

$$\frac{\partial p}{\partial x} = 0 \quad \frac{\partial p}{\partial y} = 0 \quad \frac{\partial p}{\partial z} = -\gamma \quad (35)$$

These equations show that the pressure does not depend on x or y . Thus, as we move in a horizontal plane, the pressure does not change. The needle is in a plane parallel to the $x - y$ plane, so during the insertion period, the pressure on the needle tip does not change.

Let's set $k = 5 \text{ N/m}$ and $p = 100 \text{ mmHg}$, combining with two friction models, Figure 3.20 (a-b) are gotten:

In fact, the blood in the blood vessel is flowing, with the forwarding of the needle tip in the blood, the encountered pressure is changing. In the next section, we will first introduce the dynamic fluid characteristics and then build a blood vessel model with changing pressure.

(a)

(b)

Figure 3.20: (a) Karnopp Friction model plus constant blood pressure model. (b) Elasto-plastic friction model plus constant blood pressure model

3.4.3.2 Changing Pressure Blood Model

In reality, blood in blood vessels is not static, in this section, we will use fluid dynamic characteristics to build a blood vessel model with changing pressure.

Before proceeding, we need to introduce two dynamic fluid laws. The first one is *Bernoulli equation* which is a very powerful tool in fluid mechanics. Assuming that the fluid motion is in two-dimension and defined to the $x - z$ plane, which has the following formula:

$$p + \frac{1}{2} \rho v^2 + \gamma z = \text{constant along streamline} \quad (36)$$

where p is the fluid pressure, ρ is the fluid density, v is the velocity of the flowing fluid, γ is the specific weight of the fluid which has $\gamma = \rho g$.

This equation is used based on many assumptions: (1) the motion of the fluid is inviscid. That is, the fluid is assumed to have zero viscosity; (2) the flow is assumed to be steady, which means nothing changes with time at a given location in the flow field; (3) the flow is assumed to be incompressible; (4) the equation is applicable along a streamline. In the derivation of Equation 36, we assume that the flow takes place in a plane (the $x - z$ plane), but in general, the equation is valid for both planar and non-planar (three-dimensional) flows [18], provided it is applied along the streamline. The second one is called flowrate equation, we assume the flow is horizontal, steady, inviscid, and incompressible between point (1) and (2) as shown in Figure 3.21, the *Bernoulli equation* becomes:

$$p_1 + \frac{1}{2}\rho V_1^2 = p_2 + \frac{1}{2}\rho V_2^2 \quad (37)$$

and flowrate equation can be written as

$$Q = A_1 V_1 = A_2 V_2 \quad (38)$$

where Q is the flowrate of the fluid in the pipe along the z direction, $A_{1,2}$ is the outlet area of the pipe, $V_{1,2}$ is the flow velocity crossing the $A_{1,2}$ respectively.

In this section, we assume that the blood in the blood vessel must be steady, inviscid, incompressible, and blood flow must be horizontal. The two dynamic fluid laws can be written as:

$$p_1 + \frac{1}{2}\rho v_1^2 = p_2 + \frac{1}{2}\rho \frac{v_1^2 A_1^2}{A_2^2} \quad (39)$$

from Figure 3.18, we have that $h = 2r$ and $l = \sqrt{h^2 + r^2}$, let diameter of the blood vessel is r_b and $h = x$, taking them into Equation 39, we have

$$p_1 + \frac{1}{2}\rho v_1^2 = p_2 + \frac{1}{2}\rho v_1^2 \left(\frac{r_b}{r_b - x} \right)^4 \quad (40)$$

using Equation 40, we can calculate the encountered pressure at any positions. Combining it with Equation 22, we can get a blood model with changing pressure. Figure 3.22 (a-b) are plotted by setting $p_1 = 100 \text{ mmHg}$, $\rho = 1025 \text{ Kg/m}^3$, and $v_1 = 20 \text{ cm/s}$, $r_b = 2 \text{ mm}$, and $k = 5 \text{ N/m}$.

Figure 3.21: Flowrate in a pipe.

Figure 3.22 (a): Karnopp Friction model plus changing pressure blood model.

Figure 3.22 (b): Elasto-Plastic friction plus changing pressure blood model.

Chapter 4

Comparison and Validation

The models developed in the previous sections are combined and validated by direct comparison of simulated and real data. Based on the real data values, four complete average models of the needle insertion have been created and compared. Models for specific specimens are also built by adjusting the parameters comparing with the real data in order to achieve perfect match. Finally errors and deviations are given in this chapter.

4.1 Four Models

In order to model the forces when needle is inserted into a liver, in the previous sections, we divided the force into two parts according to the positions of needle tip inside the liver. In this section, four models are developed based on the real data. They are shown in Figure 4.1 (a-d). The comparisons between models and real data are plotted in Figure 4.2. Due to significant variations in the data from one liver to the other, the perfect match can not be reached, but the overall shape of these four models are similar to the real data, we can still say that these models may be used for event detection and

haptic simulations.

(a) Spring, exponential, Karnopp plus still blood

(b) Spring, exponential, Karnopp plus changing pressure

(c) Spring, penalty, Elasto-plastic plus still blood

(d) Spring, penalty, Elasto-plastic plus changing blood

Figure 4.1: Four models.

(a) Spring, exponential, Karnopp plus still blood

(b) Spring, exponential, Karnopp plus changing pressure

(c) Spring, penalty, Elasto-plastic plus still blood

(d) Spring, penalty, Elasto-plastic plus changing blood

Figure 4.2: Comparisons between final models with real data

4.2 Two Models for Specific Data

In this section, models for two specific specimens are created by changing some parameters in the previous models to achieve perfect match with the real data (shown in Figure 3.3).

From the previous section, we know that it is impossible to create a single or a few models which are suitable for all the real data. Therefore, we thought whether we could create a single model to match the real data taken from a specific tissue or organ by changing the parameters of the previous average models.

The models created in this thesis are based on the liver insertion force data, so we first choose two real liver insertion force data and make the models fit the real data perfectly by changing the parameters of the previous models.

In order to match the force data in Figure 3.3 (a), we use nonlinear spring model, penalty model, Karnopp friction and still pressure blood model; for the force data in Figure 3.3 (b), the nonlinear spring model, exponential function, Karnopp friction and changing pressure blood model are employed. The two created specific models and their comparisons with real data are shown in Figure 4.3 (a-b). The parameters chosen for the two models are listed in table 2.

Parameter	First Model	Second Model
a (nonlinear spring)	0.005 N/mm	0.002 N/mm
b (nonlinear spring)	0.003 N/mm	0.0013 N/mm
λ (penalty model)	0.003 N/mm	————
e (penalty model)	≈ 0	————
$\Theta = [a, b, d_0, F_0]$	————	$[-0.031, 1.7, 9, -3.39]$
C_p (friction model)	0.0858 N/mm	0.0858 N/mm
b_p (friction model)	0.0070 Ns/mm^2	0.0040 Ns/mm^2
k (blood)	0.005 N/mm	0.005 N/mm
p (blood)	100mmHg	100mmHg
ρ (blood)	1025 Kg/m^3	1025 Kg/m^3
v (blood)	————	20.25 cm/s
r (blood)	2 mm	2 mm

Table 2: Parameters chosen for two models.

(a)

(b)

Figure 4.3: Comparisons of two models with two specific data separately.

It is found that they can be perfectly matched. For Figure 4.3 (a), the mean error is 0.006 N with a standard deviation $\delta = 0.0762\text{ N}$ and the error is plotted in Figure 4.4 (a). For Figure 4.3 (b), the mean error is 0.0051 N with a standard deviation $\delta = 0.035\text{ N}$ and the error is drawn in Figure 4.4 (b). As expected, the errors are low and the models are both of interest.

(a)

(b)

Figure 4.4: Error plots. (a) The mean error is 0.006 N (b) The mean error is 0.0051 N .

Chapter 5

Force Model Library

We found that we can achieve perfect match between a model and a specific force data by changing some parameters of the model, but the real forces are coming from liver insertion experiments and the models are created based on these force. In this section, we want to build a force model library which contains models not only match the different liver insertion data but also represent other soft tissue insertion force during percutaneous procedures.

Matt Heverly *et al.* [75] performed needle insertion tests on *ex vivo* porcine heart samples as well as *in vivo* rabbit hearts. The porcine heart specimens were stored in phosphate buffered saline solution prior to testing to preserve the tissue. Figure 5.1 shows typical force versus needle depth curve at a velocity of 5 mm/s . A small number of rabbit models were used to approximate human fetal hearts. Tests were performed on rabbits with the chest open. During these tests, the heart was still filled with fluid and supported in the chest by the surrounding tissue and organs. Figure 5.2 shows needle force versus needle depth curve.

Figure 5.1: Force vs. needle depth during insertion into porcine heart sample (adopted from [76]).

Figure 5.2: Force vs. needle depth during in-vivo rabbit heart (adopted from [76]).

We adjusted model parameters to match these real data. We choose spring model, Karnopp friction model, and still blood model to simulate these two real data with different parameters (in these two models, we only use spring model to represent stiffness force, because from the real data plots, there are no significant sharp rupture during the increasing period), parameters are shown in table 3 and plots are shown in Figure 5.3 and Figure 5.4.

Parameter	Porcine heart	Rabbit heart
a (spring model)	0.00002 N/mm	0
b (spring model)	0.0013 N/mm	0.0013 N/mm
D_p	0.00858 N/mm	0.00858 N/mm
b_p	0.006 Ns/mm^2	0.004 Ns/mm^2
k (blood model)	0.005 N/mm	0.005 N/mm
P	100mmHg	100mmHg
r	2 mm	2 mm

Table 3: Different parameters for matching porcine heart data and rabbit heart data

Figure 5.3: Rabbit heart data match plot.

Figure 5.4: Porcine heart data match plot.

It is found that they matched very well. For Figure 5.3, the mean error is 0.0186N with a standard deviation $\delta = 0.0469\text{N}$ and the error is plotted by Figure 5.5. For Figure 5.4, the mean error is 0.030N with a standard deviation $\delta = 0.0529\text{N}$ and the error is shown by Figure 5.6. As expected, the errors are also low and the models are of interest.

Figure 5.5: Error plot for porcine hearts.

Figure 5.6: Error plot for rabbit hearts.

Chapter 6

Conclusion and Future work

This thesis presents a method based on real time data to characterize the values of forces during soft tissue insertions. Models for general tissues or organs and for specific individual ones are discussed. A number of models could be created and put into a library to model different soft tissues.

6.1 Contributions

The contributions of this thesis are following:

1. We divided the whole real force into different subparts according to the position of the needle tip and the force characteristics. Within the stiffness period, the forces are divided into two subparts based on the needle positions; the following forces are separated by the reasons for which the forces are created.
2. We proposed a method for modeling fluid structures enclosed in soft tissues. In this thesis, blood vessels are regarded to be the only internal structures of the soft tissues and are modeled and tested.

3. We adopted several existed force models to model the soft tissue insertion force for the first time.

4. We built models based on liver insertion forces, but the created models can represent not only the real liver insertion forces but also many other soft tissue insertion forces in percutaneous therapies.

We developed a new and more life-like haptic medical simulation method by introducing fluid structures enclosed in human organs. The goal of developing a method to model soft tissue interactions with needles has been met. The utility of these models in comparisons with real time force data was demonstrated to be practical and effective. Because the models can be integrated with haptic virtual environments, it can improve the accuracy and reliability of a system which makes use of the models as well as the efficiency of the simulation.

6.2 Future Work

There are several opportunities for the work presented in this thesis to be continued and improved. Some of the possibilities include:

1. Combining the force model with computer graphics. A needle insertion simulation in 3D and a complete model including deformable virtual soft tissue and force feedback could be created.
2. A further extension to the fluid simulation. Easily deformable, low or high pressure blood vessels, glands, and cysts could be added to increase the realism and difficulty

of needle placement.

3. Application to surgical planning. Many surgical procedures are planned using medical images and deformable modeling [75]. Any planning procedure that involves fluid filled organs could be improved by addition fluid interaction.
4. Building a powerful library. Models of force into different soft organs of different person, different age, healthy or diseased, can be built by model parameters adjustment.
5. Future experiments for different needle tips. In this thesis, the shape of needle tip is diamond, in real percutaneous therapy, the tip shape maybe has many kinds, for example: beveled Quincke point spinal needle [30], more experiments could be done including the needles of different tips.

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