SOFT TISSUE MODELLING FOR HAPTIC RENDERING IN VIRTUAL ENVIRONMENTS

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Abstract

This paper focuses on the development of a haptic recording and modelling system. Currently being evaluated for multiple uses in surgery and manufacturing, this recording system evaluates haptic data captured via a robotic arm coupled with real time high-resolution load cell. This data is then analysed and validated against previous samples and a generated model before being logged for playback during simulation and training of a human operator. 3D models of point force interactions are created allowing unique visuals to be presented to a user. Primarily designed for the medical field, recorded results of soft-tissue cutting have been presented.

1 Introduction

This paper focuses on the challenge of recording live haptic events and analysing data for future reference and playback on force reflecting hardware. Deakin's Haptic Research Facility hereafter known as the HRF encompasses the technology required to record, validate and model soft tissue interaction in real time. Comprising seven main areas the HRF includes two Phantom™ haptic devices, an ABB IRB2400 industrial robot, stereoscopic cameras, three-dimensional projection and headsets, high-resolution data gloves, non-contact position tracking as well as the computational power to drive such devices.

Minimally Invasive Surgery (MIS) has several advantages over traditional open surgery. Patients experience minimised trauma at the site of operation, require reduced hospitalisation and minimal cosmetic distress due to more accurate and smaller incisions [1]. It is therefore obvious that MIS will continue to advance as a stream of medicine.
The accuracy of MIS will increase when surgeons are able to overcome its limitations through training in simulated environments. Currently surgeons performing MIS are inhibited by their inability to directly view the procedure (small cameras are inserted into the operative cavity), the requirement of a higher degree of training and reduced tactile feedback (if any) [2].

Traditionally surgical students have used rubber models, cadavers and animals when learning to perform MIS. It is then necessary for surgeons to refine this operating technique on living people. [3], [2]. The use of rubber models in training does not provide surgeons with an accurate haptic response to incisions and other surgical movements. The disadvantage of utilising cadavers is that physiological changes occur to deceased persons. The mechanical properties of tissue change significantly posthumously.

Temperature, viscosity, hydration and elasticity all change after the loss of air infused blood [2]. Animals suffer from differing anatomical traits to humans and operating on live patients to gain surgical skills raises safety concerns. Medical simulation is an area of great interest with several implementations actively being researched [4], [5], [6]. Training via a computer generated world able to immerse the user into his or her procedure is gaining acceptance for various procedures [7], [8].

Previous work in the topic of soft tissue model validation has focussed on single axis force sensing and conditioning. Simone et al [9], focussed on recording tissue integration with a surgical needle used for performing biopsies. This work, while sufficient for modelling needle insertions is not adequate to explain the interaction of a scalpel consisting of multiple flesh/blade contact points.

To develop realistic training simulations to fully immerse a user in the virtual world, some form of accurate database of soft tissue haptic data is required. From this data, reliable models can be developed. Generally speaking, models of surgical procedures and human anatomical elements are modelled in one of three ways [5]

I... Manual estimation, requiring the adjustment of a surgical simulation by a skilled surgeon utilising a trial and error method to complete a set task. Drawbacks of this method include the dependence of haptic perception of the training surgeon and the time necessary to complete even the simplest tasks. This method is usually only used for needle insertion tasks and other procedures with relative simplicity [2].

II. Structural element measurement. Generally only used in the industrial sector, Structural element measurement relies on a breakdown of a complex mechanism into simplistic parts that can be modelled individually and their interoperability is known. As accurate libraries of internal organs are modelled it will become possible to utilise this method for developing surgical training simulations.

III. Accurate recording of properties in real time is main focus of this paper. One of the most effective methods of transferring force measurement to a surgical apprentice. Especially useful when combined with the above methods to develop an efficient model.

Models are commonly defined by four properties, geometry of organs, kinematics of surgical tools, type of contact with the organ (twist, slice etc) and the mechanical properties of tissue. It can generally be assumed that three of these properties are known (geometry of organs from preoperative scans) with the exception of the mechanical properties of the tissue. Our goal is to improve the way in which the properties of tissue are recorded with the aim of optimising robot assisted surgery and surgical model validation for trainee surgeons.

2 Data Acquisition Hardware

To achieve this goal, we have integrated numerous parts of the HRF with customised software solutions. An ABB IRB2400 industrial robot has been fitted with a JR3 90M31A 6 axis force / torque sensor at its end effector coupled to a customized number 12 surgical scalpel. This sensor is directly interfaced via an 8kHz serial link to a controlling computer. The data from this sensor is then logged at a rate of 500Hz per axis. The robot is controlled remotely from a 3.06Ghz Pentium 4 computer with 1.5Gb of RAM running ABB's WebWare™ interface software. Additionally, a Phantom 1.5 6D.O.F. device is connected to this PC allowing force reflection to the user. The whole system is controlled from a Graphical User Interface (GUI) of our design.

The JR3 90M31A sensor is capable of measuring the force on three orthogonal axes, and the moment (torque) about each axis.

Figure 1. Experimental apparatus for recording data.

3 Procedure

Bovine liver was chosen as the tissue to be tested for this particular experiment. Due to livers relative consistency of internal structure, the forces required for linear cutting were assumed to be relatively constant. While liver contains small internal blood vessels these veins are relatively small compared with those found in ones extremities. Small incisions were modelled via the robotic arm and sensor in multiple samples.

A 750g sample of liver contained in a small plastic container was subjected to a precise 80mm linear cut (X-axis) after initial insertion by a small linear move to a depth of 6mm (Z-axis).

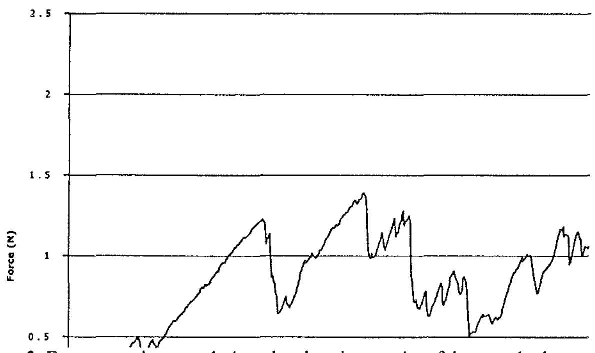

Figure 2. Force vs. robot translation plot showing results of three scalpel passes

This movement was performed three times per piece of liver to differentiate between cutting forces and frictional forces of the blade through the tissue due to the viscosity of the liver membrane (Figure 2). The above experiment was performed several times on multiple samples to aid in development of a generalised model.

Results were then plotted and a quadratic equation for linear soft tissue was derived.

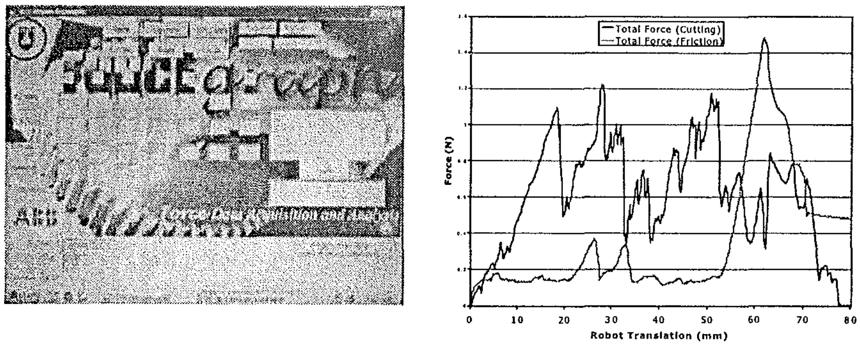

Figure 3. (left) Data recording module. (Right) Force vs. Robot translation plot (*showing initial cutting force and frictional forces*).

4 Results

As the development of an accurate model for soft tissue integration is the main aim of this work, precise recording of results was required. This facilitated a haptic data gathering application capable of capturing sensor data at 500Hz. Six axis of data was saved to a tab delimited text file for later analysis. Figure 3 shows the results of three scalpel passes through liver 1. The blue (higher average force) trace shows the forces in the X-axis required for initial soft tissue cutting.

The subsequent plots (red and green) show the frictional forces of the same blade through the pre-cut liver.
By hypothesizing that after the initial cut, all ensuing forces are frictional, a graph of cutting forces versus frictional forces was produced (Figure 3). It can be further postulated that since the forces on each side of the scalpel blade were equal in magnitude, a model of single sided frictional blade contact could be created.

As the liver is made up of cell systems with a main circulation central artery and a special liver (portal) vein, cutting the vein will result in a larger force vector. This is apparent in Figures 2 and 3 at approximately 62mm into the cut.

Initially it was assumed that the force required to cut liver was relatively linear and proportional to the linear movement produced via the robot. After analysing the data, it was established that this is not the case. A 2nd order quadratic equation was determined to be the most accurate representation of cutting force including frictional forces. This enabled us to produce a model valid for cuts of 80mm in soft tissue (liver). As a result, extrapolating data from this model allows us to achieve an accurate representation of soft tissue/scalpel interaction in linear cuts of any length.

$$f_{cutting} = -5e^{-6}x^2 + 38e^{-4}x + 0.164$$

$$Insertion < x < Cutlengh * 10$$

Where insertion is after the scalpel has entered the liver and is ready to commence linear movement and Cut length is the maximum cutting distance.

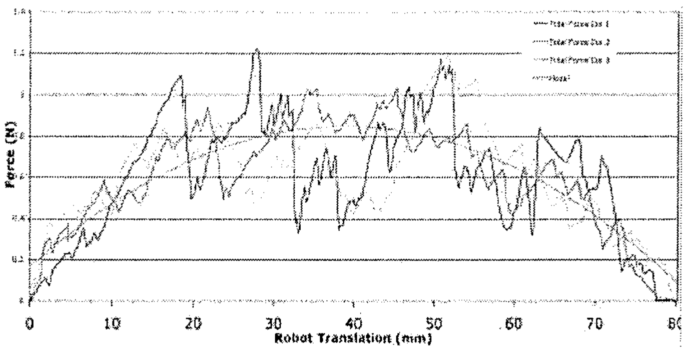

Figure 4. Force vs. robot translation plot
(showing results of three scalpel cuts on different livers and the resultant derived model)

0.607N was established to be the average cutting force applied to the liver over a translation of 80mm, likewise, 0.361N was found to be the average frictional component of the blade during a cutting operation.

4.1 Model Validation

Figure 4 shows the results of 3 cutting operations with frictional forces removed. The model is reasonably accurate and deviation is mainly due to the blade encountering incidental un-modelled internal structures. It becomes obvious from these plots why the assumption of linearity was erroneous.

The reasoning behind the selection of the resultant model is that as the blade was moved linearly through the tissue, uncut liver "rode up" the front of the blade similar to the way in which a ship cuts through the water. This results in more surface area in contact with the blade and higher frictional and cutting forces. The forces abate when the blade draws near the end of its cut due to the liver effectively catching up with the velocity of the blade. The direct comparison between real and simulated data validates the model.
The model only applies to the X-axis and is limited to the one blade geometry. As six axis data was recorded, the HRF is currently in the process of modelling these data sets as well.

4.2 Model Application

This research shows that soft tissue/blade interaction can be reliably modelled over a known cut length and extrapolated for smaller lengths. Developing what others are terming a "tissue library" will result in accurate blade/tissue interaction models available for surgical simulation and robot-assisted procedures. With the use of force reflecting hardware such as the Phantom haptic device in the HRF, we have begun developing simulation packages allowing a user to feel the interaction of a blade through soft tissue (liver). For this research to be applied to real world scenario's, further testing with different blade geometries is required. Furthermore, in vivo data needs to be collected for accurate representation of living tissue modelling.

6. Haptic Rendering

Once tissue has been modelled a procedure can be rendered to a haptic device such as the Phantom 1.5 6D.O.F. Raw data from the liver cutting operation and the model can then be compared for errors. As the data from the force / torque sensor is sampled at 500Hz and the Phantom interface requires 1000Hz for accurate haptic feedback, the data is passed through a high speed averaging loop.

$$F_{axis} = (F_{current} + F_{next})/2$$

This loop performs 6 axis averaging at 1000Hz before transferring the data to the haptic device for playback. Model data is derived directly from the equation at 1000Hz.

A 3D modelling system developed for this project is then utilised. 3DMF files (3D Metafile) are created using a look up table of robot positions and modelled forces. From this data, a text file is produced with the necessary vertices, edges and faces. Tool (scalpel in this case) interaction is modelled as a point force at this stage (shown right Fig 5). Later advances will see multi point tool interaction with the produced mesh. Once the Metafile has been created it is rendered utilising an open source OpenGL library called Quesa. Figure 5 consists of around 102 thousand vertices.

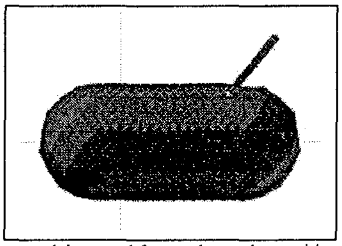

Figure 5. Liver model created from robot point positions and look up table.
(Also visible is the current tool interaction point)

Quesa renders the above mesh utilising multiple light sources and will enable 3D presentation to a viewer using software such as that provided my More3D™.

Figure 9. Manipulation of haptic data using the Phantom 5DOF

The second mode of visual feedback is via a pair of stereoscopic cameras. Video of the initial procedure (model testing phase) is played back in sync with the data trough a 3D capable headset. This video was recorded during the modelling trials that were used to create the data in Figure 5. A user can feel the forces applied to the liver during a cutting motion whilst seeing the relative position of the blade through the tissue. Forces are more constant and noise is reduced in the model where as forces experienced during playback of the raw sensor data can be overwhelming to an operator by providing too much tactile feedback and vibration.

7 Conclusion

The model described in this paper and the method used to develop it has potential in the medical field. As haptic devices become commonplace in surgical training facilities, accurate data of tissues to be operated on becomes a must. Further work must be performed on the described model to make it viable for skilled surgeons to develop a favourable tactile response. This will be aided by work in the in vivo haptic recording field currently being undertaken. Un-modelled deviations from the mathematical model can be simulated in a virtual training session, by introducing variable incidental forces both perpendicular and in the same plane to the cut.

Frictional and total force components of soft tissue interaction can be measured in real time and compared with previous data sets and the model before being uploaded to a force reflecting device controlled by a surgeon in the operating room, or ultimately across the world. We have demonstrated the ability to record haptic data at 500Hz with a high accuracy and repeatability. The evaluation of our standalone 3DMF data files will continue and a high-speed parser and linker will be added to our software library in order to fully appreciate the modified file format. Combining real time haptic data with an accurate model has ultimate benefits to surgeons and patients once applied to medical training simulations and robot assisted surgery.

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