

Modelling of Needle Insertion Forces for Surgical Simulation

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Abstract: A novel force model for needle insertion into soft tissues has been developed for virtual reality and haptics based simulation of percutaneous therapies. The forces are divided into two parts, stiffness force and friction force, for which a blood vessel model is used in order to improve the realism. The validity of the developed model is confirmed by a comparison between the results from the developed model and those derived from real data.

Key words: force modeling, haptics, virtual reality, surgical simulation.

I. INTRODUCTION

Surgeons should be trained sufficiently and properly before they can carry out actual surgery. For instance, for the use of various tests, the sample of the spinal fluid has to be taken from the patient. For this purpose, a needle has to be inserted between vertebrae in the lower back straight into the spinal cord to take the spinal fluid while the patient is conscious. Hence, it is essential to perform a successful lumbar puncture to reduce the patient's pain. This procedure requires very dexterous hand-on skills, which can be only gained by sufficient and proper training. However, it is obvious that the surgeon students can not get the training on real patients.

With the rapid advances in robotics, interfaces, computer graphics and modelling technologies, the virtual reality (VR) and haptics based surgical simulation (VHS) systems are offering a very elegant solution to the training problem since these systems could potentially provide a near-realistic repeatable training environment with the possibility for unlimited training times over a wide variety of pathologies.

However, if these models are not realistic, the training can not be efficient. 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. Actually, realistic force feedback is one of the major remaining obstacles for VHS systems.

Needles are the most common tools used in percutaneous procedures. In this paper, we develop a novel model for the force feedback of needle insertion into soft organs. In order to

increase the realism of the haptic simulation, fluid structures are incorporated into the models of target organs in this paper. 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 the simulations more realistic. In the meantime, the force is divided into two parts according to the different layers of the tissue being injected (skin, muscle, blood vessels, fluid, etc.): first, stiffness force is modelled which is combined with 'before puncture capsule' and 'before friction occurs' parts; second, friction force models with constant blood pressure and changing pressure are created.

II. PREVIOUS WORK

So far, modeling of force involved in percutaneous needle insertion has been studied and developed by a number of researchers. The current work can be classified according to the data acquisition: ex vivo data and in vivo data. In the first category, DiMaio et al [1] focused on the correlation between forces and deformations. Simone et al [2] demonstrated that the forces between tools and needles were composed of three component. They, however, used only one model to approximate these components. In the second category, Brouwer et al [3], Brown et al [4], and Ottensmeyer and Salisbury [5] have developed specialized surgical devices that can acquire in vivo data for grasping or probing soft tissues. Force models are also developed for improving surgeon's skills in procedures where little visual and direct tactile feedback exists, such as brachy therapy [6], catheter insertion [7], lumbar puncture [8], epidural blocks [9], and laparoscopic surgeries [10]. The goal of this paper is, however, to develop a general force model of the needle insertion into soft organs, livers for example, for VHS systems.

III. FORCE MODEL

As mentioned in the previous section, we partition the insertion forces into two parts: stiffness force and friction force (with blood vessel included), in order to model the force accurately. For stiffness force, it occurs before the point of needle punctures the membrane of the liver. According to the

experimental data, it can be subdivided into two parts: ‘before capsule puncture’ and ‘before friction occurs’. For friction force, it occurs along the length of the needle during insertion. We use modified Karnopp friction model and Elasto-Plastic friction model. In order to increase the realism, blood vessels are added to the final model to present the vibration of the force, as shown in Figure 1. It should be mentioned and acknowledged that all the original data used in the paper come from [2] and [11] for which liver is the targeted organ.

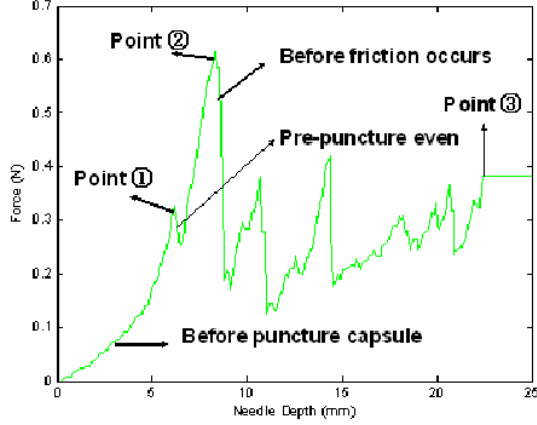


Figure1: Real time force data.

A. Stiffness force

Stiffness force is due to the elastic properties of the liver (from the origin to point ②) as shown in Figure 1. We divide this force into two phases: ‘before puncture capsule’ (from origin to point ①) and ‘before friction occurs’ (from point ① to point ②).

The so-called ‘before puncture capsule’ means the needle tip has not but nearly punctured the membrane of the liver. Simone [2] and Maurin [11] proved that a second order polynomial has lower RMS error values when fitting parameters to match the deformations. The formula is given by $F(x) = a_0 + a_1 x + a_2 x^2$ where x is the difference in the length of the springs with respect to their original, resting length. The parameters a_0 , a_1 and a_2 were fitted to match the experimental data.

In order to present the sharp rupture which is ‘pre-puncture even’ shown in Figure 1, two models were tested to simulate the ‘before friction occurs’, which means that the point of the needle has already punctured the skin but the sharp sheath still catches on the tissue. The model we tested is based on [11] and the force is modelled as an exponential function: $F(x) = (F_0 + b) e^{a(x-d_0)} + b$, where the parameter vector $\Theta = [a, b, d_0, F_0]$ is obtained: $\Theta = [-0.031, 1.7, 19.61, -3.39]$ for average.

The second model used here is a penalty-based model. It is based on a nonlinear visco-elastic force and is simple to

implement. The goal of the penalty model is to give a continuous representation of the collision phenomenon at a macroscopic level. A common hypothesis is that the value of the interpenetration of the two objects is equivalent to the deformation. Hunt and Crossley [12] proposed a formula to calculate the force occurred in the collision: $F(x) = -\lambda x - uxv$, where x is a geometrical measurement of the interpenetration. λ is spring constant by Hooke’s law, and u is the parameter with $u = \frac{3}{2} \alpha \lambda$. In [12], it is shown that the coefficient of constitution e depends on the speed of collision and the relation between e and α which can be approximated by a linear function $e = 1 - \alpha v$, where v is the collision speed. Figure2 shows the average stiffness forces using an exponential function and penalty method.

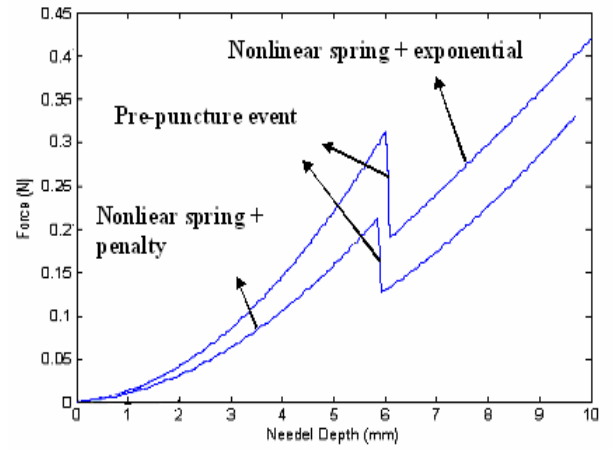


Figure 2: Two stiffness force models

B. Friction force

Friction force occurs along the length of the needle inside the tissue. Two friction models are tested in this paper.

The modified Karnopp friction model is adapted first. The model equation is:

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

where C_n and C_p are the negative and positive values of the 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 tissue, $\Delta v/2$ is the value below which the velocity is considered to be zero, and F_a is the sum of non-frictional forces applied to the system. All these

parameters can be solved from experimental data by using the linear least squares regression methods [2]. But this model only shows the static force or friction. When needle goes through the liver, due to the softness of the liver, significant presliding displacement occurs which is usually not considered in traditional models. In order to incorporate the presliding displacement into the model, an Elasto-Plastic friction model is adopted here.

The Elasto-Plastic friction model exhibits both friction and presliding displacement [14]. The friction model can be written:

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

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

where, z specifies the state of strain in the frictional contact, σ_0 and σ_2 are Coulomb and viscous friction parameters, σ_1 is the damping coefficient for the tangential compliance, $f_{ss}(\dot{x})$ is the steady-state friction force versus rigid body velocity, also called the Stribeck curve, and $\alpha(z, \dot{x})$ is used to achieve friction. After data fitting with the real data, Figure 3 is gotten.

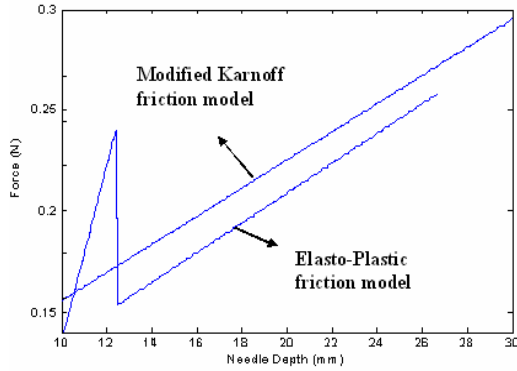


Figure 3: Two friction force models

C. Blood vessel models

We found that from point ② to point ③ in Figure 1, there are many vibrations which are caused by the internal structure of the liver, for example, blood vessels, glands, and cysts. Consequently, a blood vessel model are created in order to increase the realism in our model.

Before we jump into detail of this model, it is necessary to shown the structure of livers. We know that livers regulate 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 nonlinear phenomenon (force vibration) may be caused by these arteries and veins in the liver organ, thus we can assume that the liver is composed of a great number of blood vessels and only these blood vessel need to be modelled. In this paper,

two blood vessel models were adopted and tested: 'constant blood pressure model' and 'changing pressure model'.

Constant blood pressure means the pressure encountered by the needle tip which goes through the blood vessels is constant. The blood is regarded as static fluid and inviscid. We thus use the following model: $F = f_{\text{wall}} + ps$ where F is the total encounter force of needle tip when needle goes through blood vessel, f_{wall} is the force created by the blood vessel wall which can be presented by Hooke law, p is the constant blood pressure and s is the total contacting area of the needle tip with the blood vessel wall.

In fact, the blood in blood vessels is flowing and dynamic, so with the forwarding of the needle into the blood vessel, the pressure encountered by the needle tip is changing. Assuming that the blood is inviscid and incompressible, we can use the Bernoulli equation to characterize the blood streamline and assume a constant volume flow rate $Q = AV$, to create the model of blood dynamics where V is the blood velocity and A is the outlet area of the blood vessel [13]. The Bernoulli equation is given as follows:

$$p + \frac{1}{2} \rho V^2 + \rho gh = \text{constant}$$

where p is the blood pressure, ρ is the blood density, V is the blood velocity, g is the gravitational acceleration and h is elevation.

IV. A MODEL LIBRARY

The final force model is the combination of the previous models. Figure 4 shows the comparison of the final model we obtain and the model derived from real data. From Figure 4., it is revealed that it is impossible to create one final model to match all the real data perfectly, because the data from different livers or even from one liver are different. Consequently, a library of models needs to be established. The advantage of our model is that by changing the parameters according to real data, a very realistic model which is very close to the real data can be always achieved.

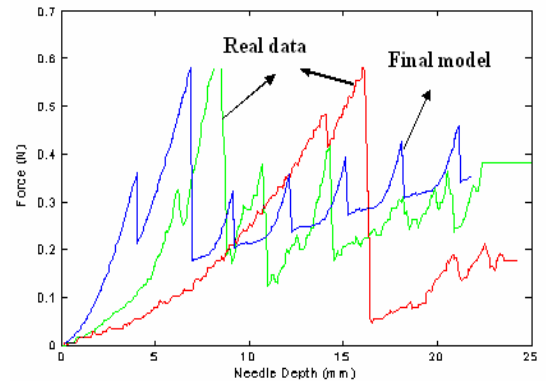


Figure 4: A final model with two real force data.

Figure5 shows two examples of models which match the real data very closely. For one shown in Figure 5(a), the mean error is only 0.006N, and a standard deviation $\sigma=0.083\text{N}$. For

the one shown in Figure (b), the mean error is only 0.0056N, and a standard deviation $\sigma = 0.035$ N. As expected, the errors are quite low compared to the ones of other models that have no consideration of the fluid dynamics inside the tissue structure.

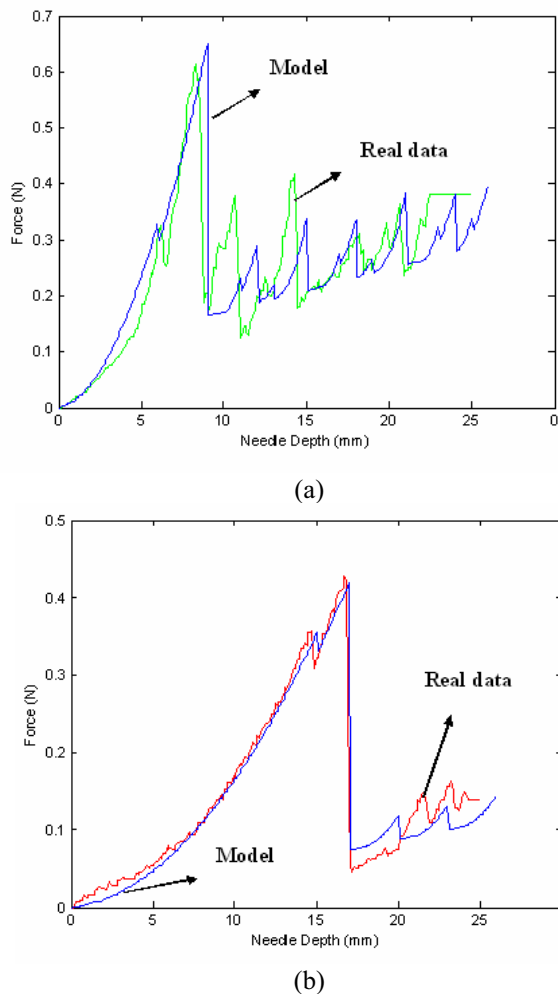


Figure 5: Two perfect match models

V. CONCLUSION AND FUTURE WORK

This paper introduces a novel model for the needle force during liver insertion for surgical simulation. The validity of the presented model is verified by a direct comparison with the real data obtained. Actually, the model represents a spectrum of models in that it can be used to create a library of models for different livers by just changing a few parameters of the master model. It is expected that this model may be extended to describe other soft tissues and organs.

Future work may include different needle shape tests, combining graphics (deformation) with force feedback, modelling the whole force from perforating fasciae of connective tissues, and some muscles and then the liver (just like during real percutaneous interventions), and adding glands, and cysts to increase the realism.

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