An Analytical Model of the Deflection of Hollow Needle-like Structures Moving in Soft Materials

Samer Al-Safadi, Parsaoran Hutapea PhD
Temple University, Department of Mechanical Engineering, Philadelphia, PA

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Project Objectives and Goals

The objective of this project is to develop an analytical model using the dynamic Euler-Bernoulli beam theory to predict the steering behavior of a needle during needle-tissue interactions, that model could improve the performance of many percutaneous needle-based procedures.

Introduction

Significant research efforts have been focused on modeling the interaction between surgical needles and soft tissues which can be used to measure the deflection of a bevel-tip needle inside soft tissue[1]. One of the most common procedures employed in modern clinical practice is the subcutaneous insertion of needles and catheters. In many cases, such procedures are difficult to plan and to perform, and can lead to significant complications if performed incorrectly [2].

Needle insertion procedures involve three basic steps where physicians can make mistakes: determination of the insertion point, needle orientation, and needle movement into soft tissues. The needle's path may traverse some sensitive tissues such as nerves, bones, arteries, or organs. Adverse damage to these tissues may lead to many side effects. Therefore, it is important that the needle does not cause any damage to these crucial tissues [3].

The Euler-Bernoulli beam theory can be used to model the needle as a beam that is subjected to forces inflicted by the tissue to predict the deflection of the needle [4]. The model to estimate the deflection from the needle-tissue interaction is based on Euler-Bernoulli beam theory. In the Euler–Bernoulli thin beam theory (beams in which the length is much larger than the depth with aspect ratio of, at least, 10:1), the rotation of cross-sections of the beam is neglected compared to the translation and the angular distortion due to shear is considered negligible compared to the bending deformation.

Analytical Setup

The needle is assumed to be a cantilever beam and the equation of motion for the transverse vibration of the beam is in the form of fourth-order partial differential equations with two boundary conditions at each end. The needle-tissue interactions can be modeled by a distributed load perpendicular to the needle shaft acting along the inserted needle portion and a point load acting at the needle tip representing reaction forces caused by the cutting of tissue by the beveled needle tip.

The previous assumptions can be modeled as a single Euler-Bernoulli beam rests on an elastic foundation under the effect of a distributed moving load. This dynamic Euler-Bernoulli beam theory is used to acquire a governing equation for the needle tissue system. The resulting equation is a partial differential equation (PDE):

$$\frac{d^2}{dx^2}\left(EI\frac{d^2\omega}{dx^2}\right)+\rho A\left(\frac{d^2\omega}{dt^2}\right)+k_f\omega=f(x,t)$$

- E, I, ρ, A properties of the needle
- k_f function of tissue elasticity (elastic modulus measurement)
- $f(x,t)$ estimated friction force density (distributed dynamic friction model)

$$f(x,t)=\begin{cases}0, & 0\le x\le L_1\\ F(x,t), & L_1\le x\le L\end{cases}$$

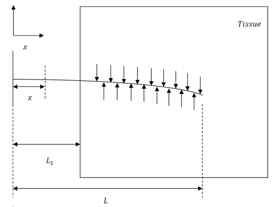

Figure 1. The deflection and the forces acting on the needle during the insertion

Experimental Setup

To measure the needle deflection experimentally we used a linear actuator with a motor (figure 2) to insert a needle into a PVC block. We used a PVC block to mimic a real tissue (such as liver and kidney, the PVC tissue was assumed to have a stiffness of ~ 6-7 kPa). The needle is 150 mm long with a 3 mm diameter and it was manufactured from polymers (such as acrylic monomer, Isobornyl acrylate, and phenol) using a 3D printer, the needle was assumed to have a Young's modulus of ~ 2.5-3 Gpa depending on the materials it was made from, the shear parameter of tissue: ts = 2.6786 N .

The needle is attached to a force sensor to measure the forces exerted on the needle. We performed three insertions horizontally with an insertion velocity of 3 mm/sec and an insertion distance of 60 mm. The maximum insertion force measured was 8.49 N.

Data and Results

We started from a simple model of the needle-tissue insertion which is going to be the base of more advanced and complicated models in the future work of this project. We assumed the needle to be a solid beam with a constant cross-section, Euler-Bernoulli beam on an elastic foundation theory was used to model the needle as a cantilever and the deflection equation was derived analytically:

- *Euler–Bernoulli elastic foundation*

$$\frac{\partial^2}{\partial x^2}EI\left(\frac{\partial^2 w(x)}{\partial x^2}\right)=q$$

- *Winkler elastic foundation*

$$\frac{\partial^2}{\partial x^2}EI\left(\frac{\partial^2 w(x)}{\partial x^2}\right)+k_f w(x)=q$$

- *Pasternak elastic foundation*

$$\frac{\partial^2}{\partial x^2}EI\left(\frac{\partial^2 w(x)}{\partial x^2}\right)-ts\left(\frac{\partial^2 w(x)}{\partial t^2}\right)=q$$

The model is based on realistic measurements of the needle and the tissue, the theoretical deflection is solved and plotted using MATLAB.

The experimental needle deflection is estimated using image processing program ImageJ. The comparison between theoretical and experimental results is shown in figure 3.

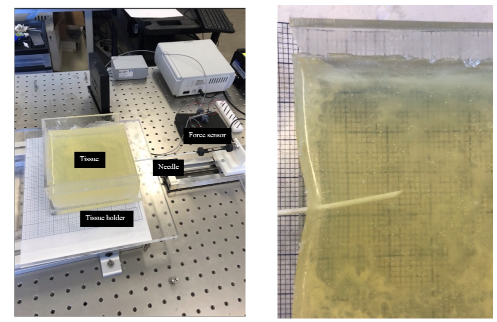

Figure 2 Experimental Test Setup

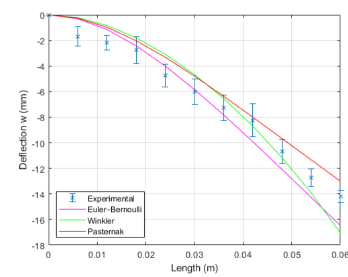

Figure 3 Deflection versus needle length comparison

Conclusion and Future Studies

Although the preliminary results show some promises, the main challenge in this study is to thoroughly model the tissue-needle interaction.

Our ongoing work is to improve the analytical model for studying the mechanics of complicated shapes of surgical needles and predict the deflection of these needles during insertion into multi-layered tissues. We then compare the results using experiments and simulations software.

Acknowledgement

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Publications

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