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AN ANALYTICAL MODEL FOR PREDICTING THE DEFLECTION OF HOLLOW SURGICAL NEEDLE IN SOFT TISSUES

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ABSTRACT

Considerable research efforts have been devoted for studying the interaction between surgical needles and soft tissues which can be used to measure the deflection of a bevel-tip needle inside a tissue. The development of such a model, which predicts the steering behavior of the needle during needle-tissue interactions, could improve the performance of many percutaneous needle-based procedures. In this study, we use the Euler-Bernoulli beam elastic foundation theory to model the needle as a cantilever beam moving along its longitudinal axis and undergoing various external loads. These external loads are the result of the interaction between the tissue and the needle during insertion and they can be modeled as the following: a tissue cutting force acting at the needle's beveled tip modeled as a concentrated force, and needle-tissue interaction forces acting along the needle length which include: (i) tissue reaction forces as the result of its deformation caused by needle bending and modeled as a transverse distributed load; and (ii) friction forces between the needle shaft and the tissue modeled by an axially distributed load tangent to the needle shaft. The accuracy of the analytical predictions offered by the model are verified by comparing them to the experimental data. The difference between the analytical model and the experimental results was between ~15% to ~33%.

1. INTRODUCTION

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 perform and can lead to significant complications if performed incorrectly [1].

There are three steps where physicians can make mistakes during the needle insertion procedures: determination of the insertion location, needle orientation, and needle movement into tissues. The needle's path may cross some sensitive tissues such as nerves, bones, arteries, or organs. Adverse damage to these tissues may lead to many side effects. Therefore, it is crucial that the needle does not cause any damage to these vital tissues [2].

In previous studies, the Euler-Bernoulli beam theory was used to model the needle as a beam that is subjected to forces inflicted by the tissue to predict the deflection of the needle [3]. In the Euler–Bernoulli thin beam theory (beams in which the length is much larger than the depth with an 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.

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 a fourth-order differential equation 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.

2. MATERIALS AND METHODS

In the early models, we assumed: (i) the deflection of the needle is small compared to its length; (ii) the needle is a long and thin cantilever beam with a constant cross-section clamped from one side and free on the other side; and (iii) the tissue is homogeneous. 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 Euler-Bernoulli beam is used to acquire a governing equation to estimate the deflection from the needle-tissue interaction.

We can characterize the needle-tissue interaction forces during needle steering in soft tissue into three groups [4]:

- *Tissue stiffness force* is a result of the viscoelastic response of the tissue when the needle punctures its surface. The needle tip deforms the tissue and cracks/punctures the tissue when the maximum contact force is reached.

- *Cutting force* is the force that occurs after the crack propagates into the tissue in response to the needle tip displacement [5].
- *Friction force* is the force on the interface of the tissue and the needle surfaces.

Since we are working on a single layer of homogeneous tissue, the tissue stiffness force will show only at the beginning of the needle insertion and its contribution to the forces affecting the steering of the needle in the tissue will be assumed to be minor compared to the other two forces and will not be added to the force model until future advanced models with more complicated tissue textures.

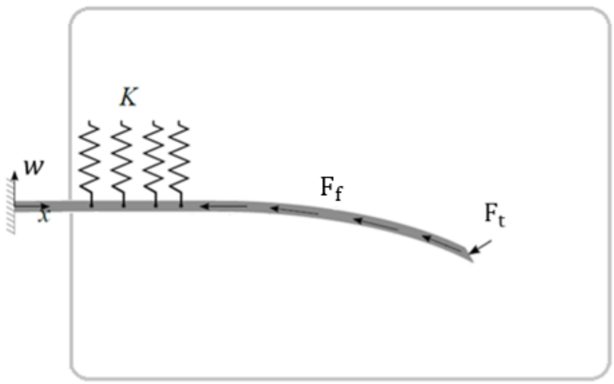

FIGURE 1: The deflection and the forces acting on the needle during the insertion.

From [5] and [6] the friction and the cutting forces can be theoretically estimated and expressed as:

$$\text{Friction force } F_f = \frac{\mu D}{2}\frac{0.65E_2}{1-\nu_2^2}\sqrt[12]{\frac{E_2(\pi D)^4}{E_1 I(1-\nu_2^2)}}\,l \qquad (1)$$

$$\text{Cutting force } \mathrm{F_t} = \frac{4\tan\left(\frac{\alpha}{2}\right)\mathrm{E_R^2}}{\mathrm{E_T}} \qquad (2)$$

where E_1, E_2 and ν1, ν2 are the Young' s modulus and the Poisson ratio of needle and soft tissue, respectively. μ is the friction coefficient between the needle and the soft tissue. I, α and D are the moment of inertia, the bevel tip angel, and the outer diameter of the needle, respectively. E_T is the tissue stiffness per unit length. E_R is the long-time or reference modulus of stiffness. Finally, the Euler-Bernoulli equation used to model the needle as a cantilever beam on elastic foundation is shown in Equation (3):

$$\frac{\partial^2}{\partial x^2}\mathrm{EI}\left(\frac{\partial^2 w(x)}{\partial x^2}\right) + \mathrm{K}w(x) = \int_0^{\mathrm{L}} \mathrm{F_f}\sin\theta + \mathrm{F_t}\cos\varphi \qquad (3)$$

where K is the load per unit length of the beam that causes the foundation to deflect by a unit amount, F_f is the friction force, F_t the cutting force on the tip of the needle, w is the displacement of the beam on the vertical direction, θ is the angle of rotation of the beam, and φ is the sum of θ and α (bevel tip angel)

3. RESULTS AND DISCUSSION

The model is based on realistic measurements of the needle and the tissue. To measure the needle deflection experimentally we used a linear actuator with a motor (Figure 2) to insert a needle into phantom tissues. We used PVC gel phantom blocks to mimic a real soft tissue (such as the liver). To make a gelatin phantom block, gelatin powder was mixed with de-ionized degassed water in a sanitized container before adding Vyse defoamer solution (Vyse Gelatin Co., Schiller Park, IL, USA). The mixture was then heated to 80º C and added into gelatin/water mixture where it was stirred and left to cool to 40º C before transferring into a phantom block [7]. The phantom tissues made have estimated stiffnesses of 2 and 8 kPa. The needle is 180 mm long with a 1.62 mm diameter, a bevel-tip angle of 30 degrees and made of stainless steel.

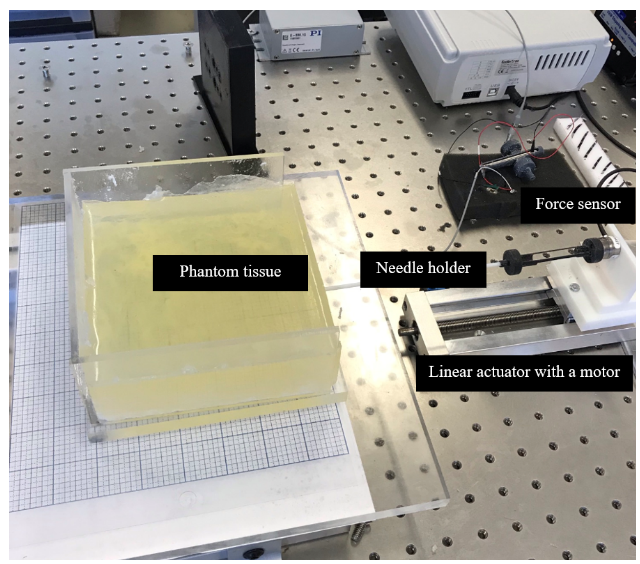

FIGURE 2: Experimental Test Setup.

The needle is attached to a force sensor to measure the forces exerted on the needle. We performed the insertions horizontally with an insertion velocity of 5 mm/sec and an insertion distance of 100 mm. The insertion tests were performed using hollow and solid needles. We performed the experiments on 2kPa and 8 kPa tissues and the number of insertion tests conducted for each needle were three.

The experimental needle deflection is estimated using the image processing program ImageJ. Equation (3) was solved to find the theoretical deflection of the needle and it was plotted using MATLAB R2020b.

Figures 3 and 4 showed the comparison between analytical and experimental data. It is shown that the analytical solution predicts the deflection reasonably well. The analytical solution of the deflection for needle insertion into the 2-kPa tissue (Fig. 3) shows better prediction compared that of the 8-kPa tissue (Fig. 4). Further study needs to be performed to improve our model.

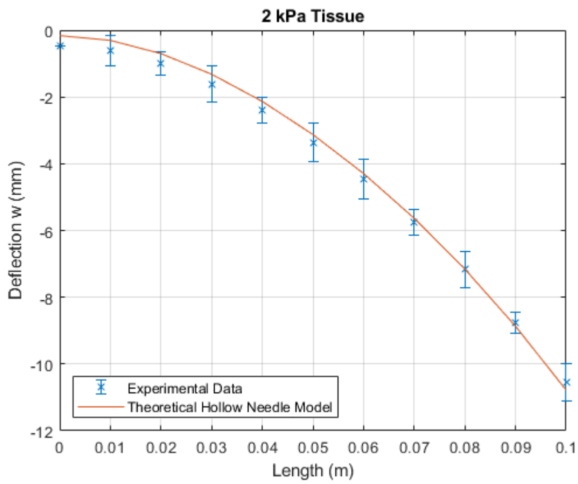

FIGURE 3: Deflection versus needle length comparison for a hollow needle inserted into 2 kPa tissue.

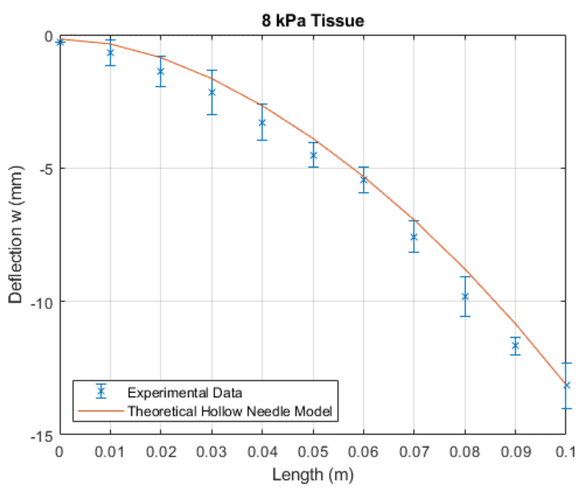

FIGURE 4: Deflection versus needle length comparison for a hollow needle inserted into 8 kPa tissue.

4. CONCLUSION

In this work, we used beam theories to predict the deflection of a needle advancing in tissues. Our preliminary results show that the model predicts the deflection well. The model can be improved by considering viscoelastic properties of the tissues, which is the focus of our current work. Our future work will also include modeling bioinspired surgical needle insertion into multi-layered tissues.

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