

IMECE2023-113833

PREDICTING NEEDLE DEFLECTION IN SOFT TISSUE: A COMPUTATIONAL MODELING APPROACH

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ABSTRACT

This study explores the mechanical interactions between surgical needles and soft tissues during procedures like biopsies and brachytherapy. A key challenge is needle tip deflection, which can cause deviation from the intended target. The study aims to develop an analytical model that predicts needle tip deflection during insertion by combining principles from interfacial mechanics and soft tissue deformation. A modified version of the dynamic Euler-Bernoulli beam theory is employed to model needle insertion and predict needle tip deflection. The model's predictions are then compared to experimental data obtained from needle insertions in real tissues. The research aims to deepen our understanding of needle-tissue interactions and develop a reliable model for predicting needle deflection, ultimately enhancing surgical robots and navigation systems for safer and more precise percutaneous procedures. Pig organs are used as a material data source for a viscoelastic model, simulating needle insertion into kidney-like environments and analyzing organ deformation. The modified Euler-Bernoulli beam theory considers the viscoelastic properties of the tissue. Deflection is then calculated and compared to experimental data, with analytical deflection measurements exhibiting a 5-10% difference compared to experimental results.

Keywords: Analytical Methods; Experimentation; Interface Modelling and Characterization; Mathematical Model.

1. INTRODUCTION

In recent years, minimally invasive surgeries have become increasingly popular due to their potential to reduce patient recovery time and the risk of complications [1]. One of the most critical aspects of such procedures is the accurate insertion of surgical needles into soft tissues during biopsies, brachytherapy, and other percutaneous interventions. A significant challenge surgeons and medical professionals face during these procedures

is needle tip deflection, which can lead to deviation from the intended target and increase the risk of complications [2].

Several factors contribute to needle tip deflection, such as the mechanical properties of the needle, tissue deformation, and the forces experienced at the needle-tissue interface [3]. Previous research has explored various models to predict needle deflection and optimize needle insertion, including static models based on the Euler-Bernoulli beam theory [4]. However, these models often need to capture the dynamic nature of needle insertion and the viscoelastic properties of soft tissues, leading to discrepancies between model predictions and experimental results [5].

The mechanical characteristics and parameters of the needle-tissue system can be acquired using a preoperative method. These parameters can be measured analytically or experimentally and then utilized as inputs for modeling the deflection of a needle during its insertion into soft tissues. As a result, real-time feedback of estimated system states for updating model parameters is unnecessary. The present study aims to develop a viscoelastic model based on comprehensive data obtained from porcine organs. This model aims to simulate inserting a needle into a kidney, allowing for an examination of the resulting deformation of the organ. The choice to replicate porcine tissues was motivated by the potential use of pig kidneys for human transplantation.

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tissue-specific parameters and the needle's geometric and material properties. The analytical model does not require real-time feedback from measured system states to update its parameters. Instead, it utilizes computing software such as MATLAB to estimate needle deflection, which is computationally faster than Finite Element methods. This study aims to establish an analytical model capable of predicting the deflection of the needle tip within the tissue. Ultimately, these results can reduce undesirable needle deflection during percutaneous procedures, leading to enhanced safety, accuracy, and better patient outcomes. The potential implications of this research hold promising prospects for improving various aspects of procedural effectiveness.

2. MATERIALS AND METHODS

The adjusted dynamic Euler-Bernoulli beam theory for needle insertion considers the dynamic nature of insertion and the tissue's viscoelastic properties [6,7]. The needle is modeled as a constant, slender beam, while the tissue is represented by a Kelvin-Voigt model combining elastic and dashpot elements. Partial differential equations describe the needle's displacement moment, which is solved numerically. The calculated needle displacement enables needle tip deflection assessment and comparison with experimental data for accuracy evaluation.

2.1 Needle-tissue interaction forces

The study used the Kelvin model to represent tissue viscoelastic properties [8]. The Kelvin-Voigt model is a linear model of viscoelastic material behavior that represents a combination of an elastic element (a spring k) and a viscous element (a damper b) connected in parallel. This model helps understand how certain materials, like polymers or biological tissues, behave under different loading conditions. The interaction between the needle and the tissue was modeled by assuming a constant needle velocity and cross-sectional area. This was represented as a distributed force along the needle, further categorized into viscoelastic and cutting forces, as shown in FIGURE 1.

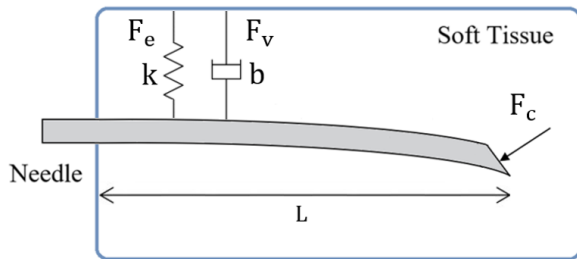


FIGURE 1: THE INTERACTION FORCES BETWEEN THE NEEDLE AND THE TISSUE, F_e IS THE ELASTIC FORCE, F_v IS THE VISCOUS FORCE, F_c IS THE CUTTING FORCE AND L IS THE INSERTION DEPTH.

The porcine muscle tissue was modeled as a Kelvin-Voigt material, including an elastic spring and a viscous damper in parallel to describe the viscoelastic interaction with the needle.

The total resulting viscoelastic force was calculated as the sum of the model's elastic and viscous forces.

$$F_t = F_e + F_v \quad (1)$$

Where F_e is the elastic force and F_v is the viscous force

$$F_e = kw \quad (2)$$

$$F_v = b \frac{dw}{dt} \quad (3)$$

Where k is the stiffness (N/m), which can be determined from the elastic modulus of the tissue and the needle, w is vertical displacement or the deflection (m), b is the viscosity of the tissue (Pa.s)

$$k = \frac{0.65E_2}{1 - \nu_2^2} \sqrt{\frac{E_2(B)^4}{E_1 I (1 - \nu_2^2)}} \quad (4)$$

The moment of inertia of the needle and foundation width are represented by I and B , respectively. Meanwhile, the Young's modulus and the Poisson ratio of the needle and soft tissue are denoted by E_1 , E_2 and ν_1 , ν_2 .

To cut or puncture tissue as the needle advances, a cutting force is necessary, affected by different factors such as the needle tip's shape, the insertion speed, and the tissue's properties. The cutting force per unit length of the needle can be expressed as:

$$F_c = k_c V \quad (5)$$

Where k_c is a constant that depends on the properties of the needle and the tissue which can be determined experimentally, and V is the velocity of the needle. The cutting force was assumed to be constant.

From the Euler-Bernoulli beam theory, the governing equation for the needle deflection in the vertical direction under the effect of the needle-tissue interaction forces can be written as:

$$EI \frac{\delta^4 w}{\delta x^4} = F = F_c + F_e + F_v \quad (6)$$

$$EI \frac{\delta^4 w}{\delta x^4} + kw + b \frac{dw}{dt} = F_c \quad (7)$$

the homogeneous solution by setting $F_c = 0$. The characteristic equation is:

$$EI r^4 = k \quad (8)$$

which has solutions:

$$r = \pm \sqrt{\sqrt{k / (EI)}} \quad (9)$$

The general solution to the homogeneous equation is then:

$$y_h = D_1 \cos(rx) + D_2 \sin(rx) + D_3 \cosh(rx) + D_4 \sinh(rx) \quad (10)$$

Where D_1 , D_2 , D_3 , and D_4 are constants determined by the boundary conditions. For a cantilever beam the boundary conditions are:

$$y(0) = 0, \frac{\partial y}{\partial x}(0) = 0, \frac{\partial^2 y}{\partial x^2}(L) = 0, \frac{\partial^3 y}{\partial x^3}(L) = 0 \quad (11)$$

The particular solution is:

$$y_p = (F/(24EI))x^3 \quad (12)$$

The general solution to the inhomogeneous equation is:

$$y = y_h + y_p = D_1 \cos(rx) + D_2 \sin(rx) + D_3 \cosh(rx) + D_4 \sinh(rx) + (F/(24EI))x^3 \quad (13)$$

By utilizing MATLAB's numerical solution, it becomes possible to calculate the deflection of a needle numerically, considering both the needle's geometry and the material properties of both the needle and the tissue.

2.2 Needle insertions experimental setup

In the experiment investigating needle deflection, FIGURE 2 illustrates the setup used. A Nema 23 CNC Stepper Motor, acting as a linear actuator, is utilized to move the needle during insertion. The needle is connected to a holder that includes a grid. A 6 DOF F-T force sensor (Nano17® from ATI Industrial Automation) is utilized to measure the forces involved, coupled with a data acquisition system from National Instruments. The force sensor captured the forces applied during insertion and extraction. The experiment involved performing five sets of horizontal insertions and extractions at a 2.5 mm/sec velocity. The LabVIEW software is used to collect force data. At the same time, the computer manages the displacement and velocity of the linear actuator, controlling the movement of the needle within the phantom tissue.

The needle insertion force measurements follow a factorial experimental design, incorporating three primary factors: needle size, insertion speed, and bevel-tip angle. A solid needle measuring 180 mm in length, 1.62 mm in diameter, and featuring a bevel-tip angle of 45° degrees is employed for the experiments. The needle is constructed from stainless steel, possessing a Young's Modulus of 200 GPa. The procedure for the needle insertion experiment involves positioning the needle tip at the

starting point on the surface of the outer layer. Subsequently, the needle is inserted into the tissue at a consistent speed until the needle tip reaches the desired insertion distance, at which point the insertion is stopped. To capture the actual needle deflection during insertion, a camera placed above the tested tissue is used for the needle deflection measurements. This enables visual documentation of deflection as the needle is inserted.

Porcine tissue is commonly used in scientific experiments because it closely resembles human tissue in size, structure, and composition. The elastic modulus of pig tissues, such as skin, muscle, and kidney, is comparable to that of human tissues, allowing for accurate replication of stiffness. Additionally, pig tissues replicate the viscoelastic properties of human tissues, allowing for precise simulation of their dynamic behavior and response to varying loads over time. Moreover, pig tissues are ideal for investigating the directional dependence of mechanical behavior due to their anisotropic characteristics. This makes them particularly useful when studying needle insertion techniques or evaluating medical devices and procedures [9-11].

In all experiments, fresh porcine muscles (meat) were sourced from a supplier on the day of each experiment. The muscles were refrigerated and stored at approximately 4°C until use. Before experimentation, the porcine tissue was carefully cut and prepared. The needle, experimental insertion setup, and associated equipment were maintained in a sanitary condition and arranged appropriately. The tissue was securely positioned to prevent any movement during the insertion process. The tissue was immediately disposed of after the experiment concluded, and the necessary data was collected. The length of the porcine tissues was 60 mm.

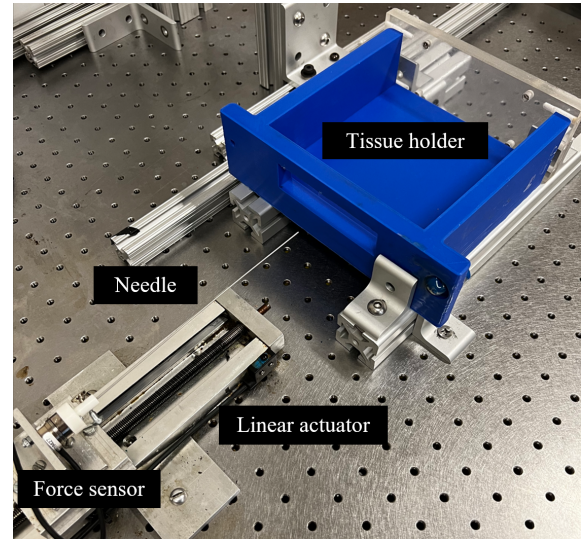


FIGURE 2: NEEDLE INSERTION EXPERIMENTAL SETUP

3. RESULTS

Experiments were conducted on porcine muscles in this section to measure the forces occurring during needle insertion, representing the needle-tissue interaction. The force measurement setup included a linear actuator for needle insertion

and removal, associated with a Nano17® force sensor positioned at the needle's base. The entire equipment setup was computer-controlled, determining needle speed and depth, and measuring the forces in play. Five horizontal needle insertion and extraction series were conducted, each performed at an insertion speed of 2.5 mm/sec and penetrating to a depth of 60 mm. Every experiment lasted 64 seconds. In the initial 8 seconds, the needle made contact with the tissue surface, and an additional 8 seconds were required for the needle to retract to its starting position post-extraction. LabVIEW, a data acquisition software, was utilized to measure the forces involved during the needle's insertion and extraction. Force measurements were taken from 350 samples at a rate of 20 samples per second. Two needle insertion sets were conducted to consider any external forces potentially affecting the measurements. The first set involved needle insertions in the air, while the second set involved insertions into the muscle tissue. The difference between the two sets' results enabled the calculation of the experimental forces during needle-tissue interaction.

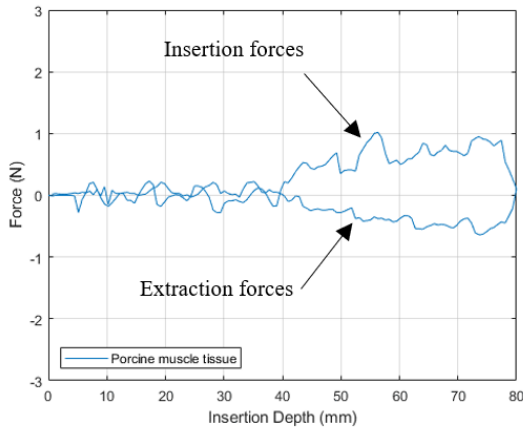


FIGURE 3: THE INSERTION AND EXTRACTION FORCES OF THE NEEDLE-TISSUE INTERACTION FROM FIVE EXPERIMENTAL NEEDLE INSERTIONS INTO A PORCINE MUSCLE TISSUE WITH A STANDARD DEVIATION OF 0.38 N

FIGURE 3 shows the results of a study that measured the forces involved in needle insertion and extraction from porcine muscle tissue. This data is represented by a two-part curve, one for insertion forces and another for extraction forces. The curve's upper part expresses the insertion forces, which start near 0 N representing the point at which the needle first pierces the tissue and peak when the needle is fully inserted. Contrariwise, the curve's lower part represents the extraction forces. The data indicates a significant difference between the forces required for needle insertion and extraction. The average insertion force was greater than the average extraction force. This difference is mainly due to the absence of a cutting force during the extraction process, resulting in lower extraction forces than the insertion ones.

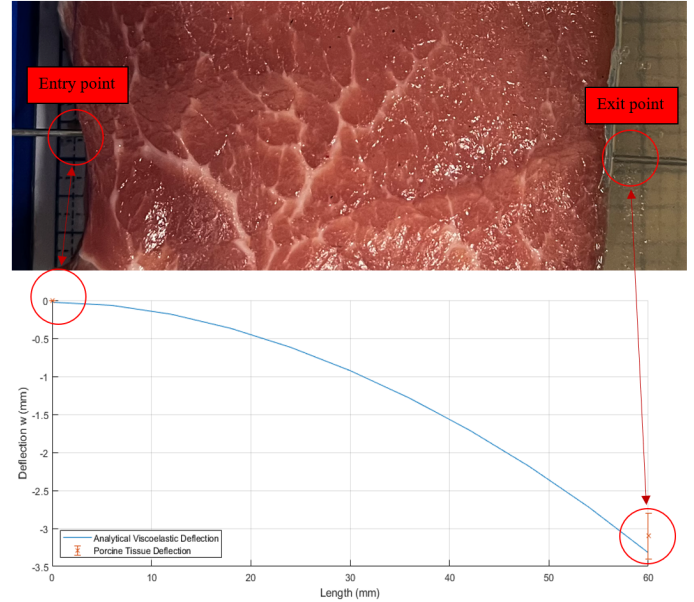


FIGURE 4: COMPARISONS BETWEEN ANALYTICAL AND EXPERIMENTAL DEFLECTIONS DURING THE INSERTION INTO THE PORCINE MUSCLE TISSUE. THE STANDARD DEVIATION OF THE EXPERIMENTAL DEFLECTION ESTIMATIONS IS 0.3 MM.

FIGURE 4 shows the mean values of the experimental needle deflection in porcine muscle tissue alongside the analytical deflection prediction. Three insertion experiments were conducted on the porcine tissue. Each test was performed with an insertion speed of 2.5 mm/sec and a depth of 60 mm. The standard deviation of the experimental insertions at the exit point was found to be 0.3 mm. The analytical deflection was computed utilizing the Euler-Bernoulli beam theory model, as described in the methods section, with a force profile reflecting the characteristics of the muscle tissue. The model's boundary conditions assume that the insertion starts from the air and the needle's tip touches the surface of the first layer, the entry point. As the needle penetrates the tissue and travels the tissue, the entry point becomes the needle's fixed end, and the needle's tip transforms into the free end of the Euler-Bernoulli beam. To begin the insertion process, the needle tip was positioned at the surface of the outer layer and then consistently inserted into the tissue at a constant speed until the desired depth was reached. The needle deflections, both experimental and analytical, were plotted using MATLAB.

4. DISCUSSION

Our study aimed to study the forces involved in needle-tissue interaction during insertion into porcine tissues and predict the needle deflection. In the process of estimating needle deflections, the primary method of comparison between experimental data and numerical predictions is through a combined approach of visual and qualitative analysis. This comprehensive examination involves the creation of plots that

capture the deflection values gathered through experimental methods, followed by visual comparison with the deflection values indicated by the analytical model. The main objective of this comparison lies in evaluating the accuracy with which the needle can reach the intended target while being guided or steered through soft tissue. The comparison tests and examines this critical ability to reach the target accurately in the soft tissue. A 5-10% difference was calculated to quantify the difference between the experimental and numerical datasets. The computational method of determining this difference was quite specific and involved extracting the experimental deflection values from the corresponding analytical or numerical predictions. This subtraction was carried out specifically at the location of the needle tip, a region of paramount importance due to its role in steering the needle accurately toward the target within the soft tissue. This area plays a crucial role in understanding the practical ability of the needle to navigate towards and ultimately reach the intended target in soft tissue with accuracy. The needle tip is the main component that interacts with the tissue and directs the path, making its analysis crucial for any realistic understanding of the process. By visually and qualitatively comparing the experimental and analytical deflection data, we could understand the overall agreement between the two sets of results.

5. CONCLUSION

Precision in needle and catheter placement is crucial for successful targeting within the tissue. This study introduces an analytical model that predicts needle deflection during insertion into soft tissues, considering linear tissue viscoelastic deformation. The model offers a trajectory prediction with a 5-10% modeling error. The utilized linear viscoelastic model, the Kelvin-Voigt model, is relatively easy to implement. However, its ability to accurately represent the behavior of biological tissues may be limited, especially under conditions involving large deformations. Future research aims to focus on nonlinear viscoelastic models. Although these models are more complex, they could offer a more accurate characterization of tissue behavior.

ACKNOWLEDGEMENTS

The financial support received from the National Science Foundation (CMMI Award #1917711) is gratefully acknowledged by the authors.

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