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Deflection of a Flexible Needle during Insertion into Soft Tissue

Niki Abolhassani[1,2] and RajniV. Patel[1,2]

[1]Canadian Surgical Technologies & Advanced Robotics (CSTAR), London, Ontario, Canada
[2]Department of Electrical and Computer Engineering, University of Western Ontario, London, Ontario, Canada

***Abstract* - Accurate needle insertion into soft, inhomogeneous tissue is of practical interest because of its importance in percutaneous diagnosis and therapies. The needles used in such procedures are usually long flexible needles with bevel tips that can deflect during insertion. Deflection of the needle can not only cause misplacement of the needle tip at the target but can also cause the needle to deviate from the planned path due to the curvature created along the needle shaft. In order to reduce deflection of the needle from its path and to increase the accuracy of the needle tip placement, we have studied the relationship between the needle base forces/torques and the amount of needle deflection during needle insertion. The model proposed in this paper can be used to estimate the amount of needle deflection during insertion into relatively soft tissues. Such model may be integrated into a trajectory generation algorithm in order to increase needle tip placement accuracy.**

I. Introduction

Accurate needle insertion into soft, inhomogeneous tissue is of practical interest because of its importance in percutaneous ("through skin") local therapies. In robot-assisted procedures, a robot is used to locate a position and insert a needle or a catheter quickly and accurately [1]. The main complication in these procedures is due to tissue deformation and needle deflection, which causes displacement of the target and reduces the effectiveness of the therapy [2]. Needle deflection is caused by flexibility of the long needle and the bevel shape of the needle tip. Tissue deformation is due to the mechanical properties of soft tissue.

During the needle insertion procedure, it is important to measure the location of the needle tip inside tissue and to be aware of the needle shape (track). In procedures such as prostate brachytherapy which involves the permanent implantation of radioactive seeds within the prostate gland, the track of the needle shaft inside the tissue is as important as the needle tip placement. The current dosimetry planning systems are based on straight parallel tracks for the needles. Moreover, intra-mediastinal or intra-abdominal biopsies and abscess draining are examples of medical applications that use the flexibility of the needle in order to create curvatures along the path in order to avoid possible obstacles or anatomical structures [3],[4].

The goal of the research described in this paper is to investigate and analyze the relationship between the force/torque data at the base of the needle shaft and the amount of needle deflection during needle insertion in soft tissue. Eventually the relationship will be integrated into a trajectory generation algorithm for intra-operative needle guidance. This study will improve needle tip positioning in the absence of an imaging system.

The paper is organized as follows. First, related work in the areas of needle deflection and modeling is discussed. Second, the experimental setup is presented. Third, beam deflection models are discussed for needle deflection estimation. Fourth, the results and discussions are given and finally conclusion and directions for future work are provided.

II. Related Work

The problem of needle deflection is one of the reasons for inaccuracy in the needle insertion procedure. During insertion of the needle, the thin needle has to pass through different tissues of varying densities, which makes the net deflection difficult to predict. Okamura *et al.* [5] studied the effect of needle geometry on needle bending. Their results showed that needles with smaller diameters and bevel tips lead to more needle bending whereas conical and triangular tips have less bending that is also more consistent.

Kataoka *et al.* [6] proposed a model for force-deflection of a beveled-tip needle during insertion. In their model, the transverse resistance force per unit length was considered a constant which is different for needles with different diameters. Their result showed that the slope of displacement versus length at different needle diameters was comparable; however, the predicted deflection was smaller than the measured deflection. Webster *et al.* [7] performed needle insertion into a rubber-like simulated muscle. They used needles with different bevel angles. Their results showed that decreasing the bevel angle increases the amount of needle deflection (bending). They also performed insertion with different velocities from 0.5 cm/s to 2.5 cm/s. They found that the velocity of needle insertion in homogeneous, relatively stiff phantom tissue had no discernable effect on the amount of needle deflection.

A few groups have used the flexibility of the needle for steering. DiMaio and Salcudean [8] used a robot with 3 degrees of freedom (DOF) and a needle with a Franseen tip (which was stiff relative to the tissue). In their approach, the tissue was moved by steering the needle base outside the tissue and the tissue deformation helped to avoid the obstacles. Alterovitz *et al.* [9] used a 2D needle insertion simulator for steering highly flexible needles with bevel tips into tissue with obstacles. In their simulation, they presented results for needle deflection in rigid and soft tissues. In rigid tissue, the needle follows a path of constant curvature during insertion while in soft tissues the curvature is not constant and depends on tissue properties. The optimal initial location, orientation and insertion depth was found in their simulation based on predicted tissue deformation. The needle curvature is considered a constant in their work. Abolhassani *et al.* [10], presented an optimization algorithm which required a force model in order to predict the amount of needle deflection. The method proposed in this paper can replace the force model and state-space force/depth-deflection model in [10].

Our approach is different from the previous work as the force readings at the needle base and the angle of bevel are used to predict the amount of needle deflection. In our approach, the needle curvature is not considered constant and the angle of needle deflection which affects the direction of forces acting on the needle is updated intra-operatively.

III. Experimental Setup

A test-bed has been set up in our laboratory for studying needle insertion in soft tissue [11]. This provides needle motion with two degrees of freedom — translation in one (horizontal) direction and rotation about the translational axis. Needle insertion is performed using the translational motion. The axial rotation is added to study its effect on needle deflection, frictional force and tissue deformation while the needle is being inserted into soft tissue. A 6-DOF force/torque sensor is attached to the needle holder to measure the forces and torques acting on the needle. The 6-DOF force/torque sensor is a Nano43 DAQ F/T system from ATI Industrial Automation. In the experiments, an 18-gauge needle with a beveled tip was used. In order to track the needle tip position in 3D during insertion, a sensor coil (see Fig. 1) is inserted inside the needle and is secured very close to the needle tip. The 5-DOF sensor coil is part of the Aurora magnetic tracking system from Northern Digital Inc. (NDI) that is used in the experiments. The magnetic field for this system is metal resistant and the needle material does not have any interference with the magnetic field.

A multi-threaded application for position/velocity control, force reading, and magnetic sensor tracking has been developed using Microsoft® Visual C++. Position data used for motion control is obtained from encoder readings. This application runs on a Pentium 4, 2.4 GHz computer, with Microsoft® Windows 2000 as its operating system. The application is written such that the force readings are obtained at the rate of 1 kHz, the control thread uses a servo rate of 25 Hz, and the Graphical User Interface (GUI) is updated at the rate of 8 Hz. The application has a GUI which enables the user to define the speed and depth of insertion as well as the speed and direction of needle rotation. A proportional-integral-derivative (PID) control scheme is used to control the needle motion in order to track specified trajectories in both degrees of freedom during needle insertion. Magnetic sensors are used instead of an imaging system in order to acquire the needle tip position data at the same high rate as the force readings.

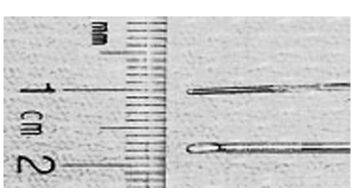

Fig. 1. Sensor coil and the needle.

Experiments were carried out on artificial phantoms (foams) and animal phantoms (excised pig's heart). The test-bed for the experiments is shown in Fig. 2.

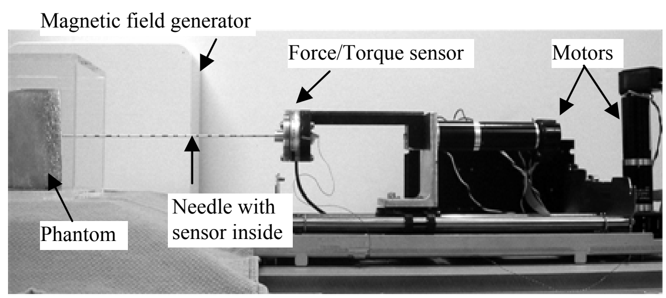

Fig. 2. The experimental setup.

IV. Estimating Needle Deflection

For accurate needle insertion, it is desired to calculate the magnitude of deflection orthogonal to the insertion direction and to compensate for it. It is known that a beveled tip needle deflects towards the direction of the bevel due to the asymmetric forces acting on the tip of the needle [9].This deflection causes the needle to follow a curvature path along the insertion (see Fig. 3).

In this study, we consider the needle as a cantilever beam. To estimate the beam deflection curve, it is required to know the forces and torques acting on the needle and their direction and to solve the boundary conditions. Using the 6-DOF force/torque sensor at the base of the needle, forces and torques are known at each instant during needle insertion. The axial needle force at the needle base is considered to be the summation of the stick-slip frictional

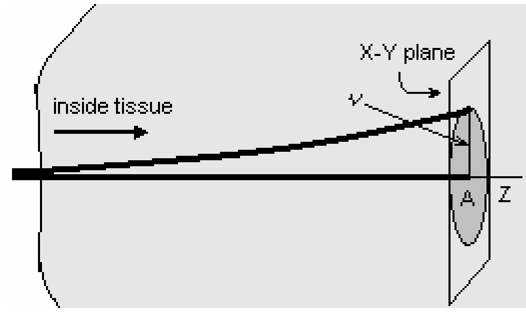

Fig. 3. Curvature of deflection. The amount of deflection orthogonal to the insertion direction at point A along the z axis (insertion axis) is ν.

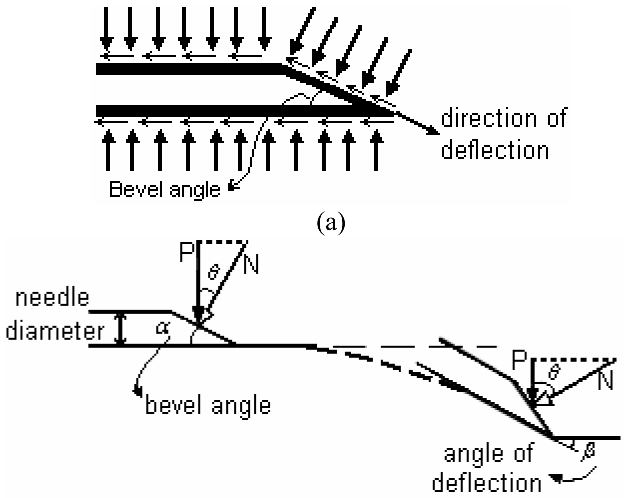

Fig. 4. (a) Forces acting on the needle along the shaft. Thick arrows are the resistance force against compression as the needle is inserted and thin arrows are the stick-slip forces. (b) Force components at the tip.

forces along the needle shaft and the cutting force at the needle tip [5],[8]. The concentrated load at the needle (see Fig. 4) that causes beam deflection is

$$P = F_z / \tan\theta \tag{1}$$

where θ is effective angle of resistance force at the tip, and initially it is equal to the angle of bevel (α), and F_z is the axial cutting force (assumed as a point load). The resistance forces against compression that are orthogonal to the insertion direction along the shaft cancel each other; however these forces may act as a support that holds the needle at the entry point of the tissue. Therefore two models were investigated for estimating needle deflection:

A. Needle with support

When the needle is considered as a cantilever beam of length l with a point load P at the tip and a support at the tissue entry point, we can write the analytical expression of the tip deflection as

$$v_i = \frac{1}{6EI}(3M_i l^2 + R_{1i} l^3 + R_{2i} z_i^{\ 3} + P_i (\frac{d}{2\tan\alpha})^3) \tag{2}$$

where E is the modulus of elasticity, I is the area moment of inertia for the needle and d is needle diameter. At each instant i during insertion, z_i, M_i, R_{1i}, $R_{2i,}$ and P_i, are the length of the needle inside the tissue, base moment, base force, support force and the load force orthogonal to the insertion (z) direction, respectively. M_i, R_{1i} are obtained from the reading of the force/torque sensor at instant i, and from $\sum F_{vertical} = 0$, the support force is

$$R_{2i} = R_{1i} - P_i \tag{3}$$

where the P_i is obtained from (1). It should be noted that the θ is calculated using the tip angle of deflection (β) at the preceding instant, and the bevel angle should be used to adjust the P_i.

B. Needle without support

When the needle is considered as a cantilever beam of length l with a point load P at the tip which is free to deflect along the entire shaft, we can write the analytical expression of the tip deflection as [12],

$$v_i = \frac{P_i l^3}{3EI}. \tag{4}$$

where P_i is obtained using (1) with an adjustment of θ such that it includes the angle of deflection at the preceding instant. In this method, the frictional force component along the needle shaft is not separated from the axial force read by the sensor.

V. Experimental Results

Experiments were carried out on phantoms of artificial tissue (foams with different stiffness) and an excised pig's heart. The cannula of 18Ga stainless steel needles with 22° bevel were used for the experiments. The needle base forces/torques and 3D location of the needle tip were logged using the multi-threaded application explained in section III. The amount of needle deflection at the tip for each insertion was estimated using (2) and (4). The estimated values were compared with the actual amount of deflection obtained from the position of the magnetic sensor installed at the needle tip for each insertion. The results showed that a cantilever model with support can estimate the needle deflection with an accuracy of less than a millimetre for insertions into artificial phantoms. However this model showed very high variation in the estimated amount of deflection. The results of a cantilever model without support showed a more consistent accuracy for both phantoms; however the results for the artificial phantom were worse than those obtained from the model with support. Figure 5 shows the result of one insertion into artificial and heart phantoms. The estimated amount of deflection (shown with dashed lines) for artificial phantom was calculated using (2) while the estimated amount of deflection for the heart phantom was calculated using (4). Table 1 summarizes the result for both phantoms.

VI. Discussion

The source of high errors in the cantilever model with support for insertions into heart tissue could be due to the elasticity and softness of the heart tissue. It can explain that the low resistance forces which were considered as a support to hold the needle were not sufficient to prevent the needle from deflection at the tissue entry point. In addition, it indicates the presence of small tissue deformation which was not detected on our phantom with the naked eye. The results showed that tissue deformation in the direction of deflection is a factor to be considered in estimating the tip position relative to the target. Such microscopic deformation may affect the force readings at the base that lead to the estimation error observed.

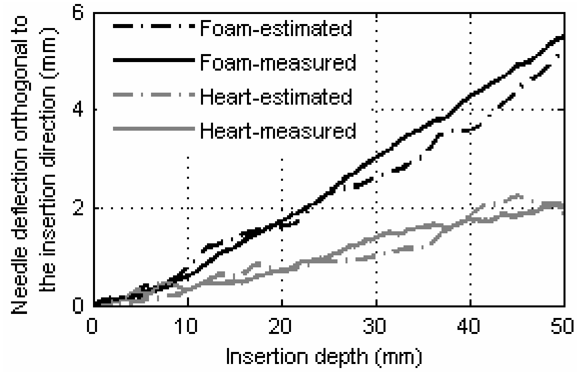

Fig. 5. Needle deflection in two insertions.

To overcome this problem, it is proposed to consider the support as a spring support (see Fig 6). This would allow incorporating tissue properties in the model by defining an appropriate stiffness for the spring depending on the tissue type. An optical marker on the outside of the tissue at the entry point may also help to adjust for the model online by measuring the magnitude of deflection at the tissue entry point. Stiffness of the support could also be obtained if accurate mechanical properties for each tissue are known. It is of interest to study the accuracy of the second model (with no support and the axial forces that include frictional components) with different tissue phantoms and different needle bevels.

Although it is well-understood that considering tissue deformation in estimating overall needle deflection is part of a perfect solution, our intention is to improve needle insertion as much as possible for those cases for which imaging data is not available.

TABLE I
COMPARISON OF THE ESTIMATED NEEDLE DEFLECTION

Phantom	Artificial phantom	Excised pig's heart
Average error of estimation in model with support (mm)	0.3	0.9
Average error of estimation in model without support (mm)	0.5	0.2
Maximum error of estimation in model with support (mm)	0.9	2.3
Maximum error of estimation in model without support (mm)	1.1	1.0
Maximum measured deflection (mm)	5.5	3.1

VII. CONCLUSION AND FUTURE WORK

This research was carried out to find a model which can be used to estimate the amount of needle deflection in soft tissue. Needle insertion was performed on different tissues and the relationship between force/torque readings at the needle base, insertion depth and 3D needle deflection for a needle with a bevel tip was investigated. The estimation of the needle deflection intra-operatively can be used to update needle insertion trajectories by rotating the needle for 180° to compensate for it [10]. The results showed that the needle deflection can be considered as a cantilever beam with a spring support. The stiffness of the support is related to the mechanical properties of soft tissue. This study presented methods that are particularly suitable for situations where needle insertions are to be performed using a robotic system. This study may improve needle insertion in robot-assisted prostate brachytherapy prior to the appearance of the needle tip in the ultrasound field of view. In our future work, tissue deformation will be considered in order to fully predict the amount of needle deflection and needle tip misplacement.

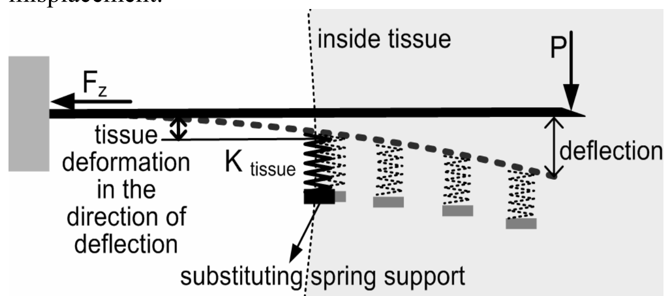

Fig. 6. Needle as a cantilever with spring support.

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