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Effects of Different Insertion Methods on Reducing Needle Deflection

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***Abstract*— Needle steering in medical procedures has attracted considerable attention in recent years. For example, in prostate brachytherapy, it is desired to insert a flexible beveled-tip needle with minimum deflection. To date, different methods of insertion which incorporate needle rotation about its insertion axis have been proposed in order to reduce needle deflection and target displacement. In this paper, needle deflection resulting from different methods of insertion are compared with our "model-based" method which estimates the amount of needle deflection using Euler-Bernoulli beam equations. Experiments are performed in gelatin phantoms and animal tissue. The results show that the proposed "model-based" method reduces the amount of needle deflection more than other methods. In this paper, some factors for choosing the appropriate method of insertion are also discussed.**

I. Introduction

PERCUTANEOUS needle insertion has several medical applications such as blood sampling, brachytherapy, abscess drainage, regional anesthesia and tissue biopsies. There is not a defined tolerance for the accuracy of needle insertion in clinical practice and in general, insertions with higher accuracy result in more effective treatment or increase the precision of diagnosis. The desired performance depends on the application. In procedures such as biopsy (for prostate, kidney, breast and liver), brachytherapy and anesthetic accuracy of millimeters is required while in procedures involving fetus, eye and ear, accuracy of micro-millimeters is desirable. In addition, the target might be in the millimeter neighborhood of another organ, vessel or nerve; therefore extra caution is required to avoid any damage or spread of the disease.

To date, a number of researchers have explored ways to improve the process of needle insertion in soft tissue using robotic tools, different modeling and simulation systems, and image-guided techniques [1]. One of the problems during needle insertion is needle deflection due to the flexibility and geometry of surgical needles. Needle deflection causes needle tip misplacement and results in an undesired track for the needle shaft [2]. The effects of needle type, gauge and tip on needle deflection has been studied [3]-[6]. Result of these studies show that bevel tip needles deflect more than needles with symmetric tips. Also results show more needle deflection when needles with higher gauge and/or smaller bevel angle are used.

In some medical procedures, it is desired to minimize needle deflection; while in others, avoiding obstacles using the curvature of deflection is desirable [7]-[9]. Nonetheless, to steer a needle in any procedure, it is required to know the model of needle deflection, i.e., needle curvature. Modeling needle deflection is challenging because of the variation of tissue properties in different patients and different organs. Therefore some researchers have suggested using needle spinning (high rate needle rotation) in order to reduce needle deflection. Wan *et al.* [10] performed needle insertion into a phantoms made of agar and chicken. Their results showed that high speed needle spinning and orientation reversal at half of the insertion depth can reduce needle deflection and seed misplacement. Podder *et al.* [11] presented the result of needle oscillation on needle deflection in a PVC phantom. Their result did not show consistent improvement when a different oscillation method was used. They also found that increasing the velocity of axial rotations during insertion could reduce PVC phantom deformation. In past work [2], we have proposed that needle deflection can be compensated for by rotating the needle through 180° when the deflection goes beyond a pre-defined threshold. In this paper, we propose a method to estimate the amount of deflection during insertion, using a model based on online force/moment readings. The aim of this paper is to compare the effect of different insertion methods on needle deflection in different phantoms. A discussion on the requirements and advantages of each method is also presented.

II. Experimental Setup

A test-bed has been set up in our laboratory for studying needle insertion in soft tissue (see Fig. 1). The manipulator provides needle motion with two degrees of freedom, translation in one (horizontal) direction and rotation about the translational axis. A 6-DOF force/torque sensor (Nano43 system from ATI Industrial Automation) is attached to the needle holder to measure the forces and torques acting on the needle. In order to track the needle tip position in 3D during insertion, a sensor coil is inserted inside the needle and is secured very close to the needle tip. The sensor coil is part of the Aurora magnetic tracking system from Northern Digital Inc. (NDI) that is used in the experiments.

A multi-threaded application for position/velocity control, force reading, and magnetic sensor tracking has been developed using Microsoft® Visual C++. 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 developed to control the needle motion in order to track specified trajectories during needle insertion. The magnetic sensor is only used to acquire the needle tip position data for comparing the result of different motions. Experiments were carried out on gelatin phantoms and animal phantoms. In order to validate the performance of the proposed method with regard to tissue deformation, ultrasound images are acquired before and after each insertion. A Philips iU22 ultrasound system is used for studying target displacement.

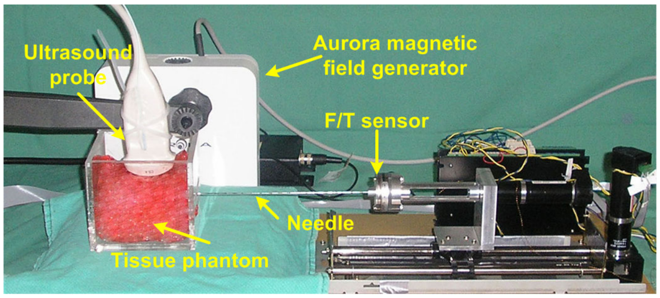

Fig. 1. The experimental setup.

III. Methods

Two groups of experiments were designed to be performed on each phantom. Group one consists of four insertion methods and compares the amount of needle deflection during insertion. The following motions are considered in this group: Insertion without rotation; needle spinning with constant velocity during insertion; needle insertion with a 180° rotation (orientation reversal) at half of the insertion depth (called mid-way rotation); and needle insertion with 180° rotations at certain insertion depths defined by the needle deflection model. In this paper, the last method is referred to as the "model-based" method. Experiments in group two compare the effect of translational and rotational velocity on needle deflection.

To estimate the amount of needle deflection along the needle shaft which is required in the "model-based" method, we use the relationship between forces and moments at the needle tip and the needle base, where a force/torque sensor is attached (see Fig. 2).

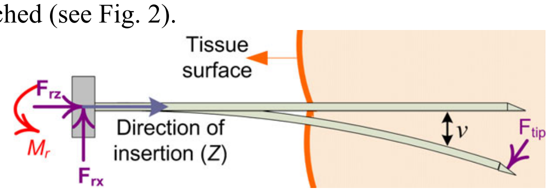

Fig. 2. Needle deflection in 2D plane.

From a study of bending [12], the simplified relation between the needle curvature and the internal bending moment is defined by

$$\frac{M}{EI}=\frac{1}{\rho}=\frac{d^2v}{dz^2} \tag{1}$$

In this study, the effect of vertical tissue resistance force on the needle is assumed to be negligible. The internal bending moment of the needle at a cross section distance z from the base is obtained from the free-body diagram and the amount of deflection at a distance z from the needle base can be calculated by its twice integration. However, the precise amount of deflection is obtained from a small scale model obtained from linearization. This model removes the dependence of the force and moment values on the insertion depth, i.e., they can be considered constant over small insertion depths. Therefore, the amount of deflection at the needle tip is

$$v=\sum_{i=0}^{n}(3\Delta M_{r_i} l^2-\Delta F_{rx_i} l^3)/6EI \tag{2}$$

where n is the number of steps to reach a certain depth inside the tissue, E is the Young's Modulus of the needle, I is the area moment of inertia , ΔM_{r_i} and ΔF_{rx_i} are changes in the amount of moment and force values between two consecutive steps. Using (1) and (2), the radius of the needle curvature can be calculated at each time step. We use the estimation model presented by (2) to find the amount of needle deflection at each time step. When the estimated deflection goes beyond a pre-defined threshold, the needle is rotated through 180˚. After the first needle rotation, the needle tip position is predicted using the uniform motion along a circular segment with curvature ρ that is calculated prior to rotation. Rotation of the needle through 180˚ causes a change in the direction of motion on the circular segment [9]. In group one, we also studied the amount of target displacement resulting from each insertion method. Images are acquired by ultrasound before (I_{bu}) and after (I_{au}) each insertion. To measure target displacement, a set of beads and spacers are placed orthogonal to the path of insertion, at the same height as the needle at a certain depth inside the tissue. For each insertion, the ultrasound probe is adjusted to grab an image from the plane of insertion with one bead defined as the target in the image. Displacement of the bead (p_b) is calculated using I_{bu} and I_{au},

$$\delta_{target}=\|\, T_{us}\, T_c\, I_{bu}(p_b)-T_{us}\, T_c\, I_{au}(p_b)\,\| \tag{3}$$

where T_c is the ultrasound calibration matrix [13] and T_{us} is the transformation matrix of the ultrasound probe.

IV. Experimental Results

Experiments were carried out on phantoms of animal tissue (beef muscle) and gelatin (3% Gelrite gellan gum in water). The gelatin phantom was hard enough to provide visible needle deflection without any deformation. Therefore, we used the model with support presented in [14] for estimating needle deflection in the "model-based" insertion into gelatin phantom. This model was suitable because no deformation was observed at the entry point to the gelatin phantom. The cannula of a 18Ga stainless steel needle (E = 200 GPa) with beveled tip was used for the experiments. Six insertions were performed into each phantom for each method and needle tip positions were measured. Results are presented as Mean ±

SEM (Standard Error of Mean). It is important to note that each group of experiments was performed at different times with different tissues. Therefore, the results of each group should only be compared with the results within that group as tissue properties can vary slightly. The needle deflection model was validated on beef muscle. The average estimation error was 0.23 mm.

Results of insertions in group one are presented in Fig. 3. In these experiments, the needle was inserted for about 80 mm with constant translational velocity of 10 mm/sec, and the 180° rotations were performed with constant rotational velocity of 5 rpm. Needle spinning was generated with the rate of 50 rpm. As found from both experiments, needle deflection in methods with rotation can reduce the amount of needle deflection. The spinning and "model-based" methods reduced the amount of needle deflection more than 80%. Target displacement was also measured for group one. The result of target displacement did not show any target displacement in the gelatin phantom. As shown in Fig. 4, insertion methods which reduce needle deflection did not increase target displacement in beef muscle.

It was also observed that in the "model-based" insertion in beef muscle, the needle motion on the circular segment after rotation was slightly different than predicted. This difference was caused by the softness of the animal tissue and it should

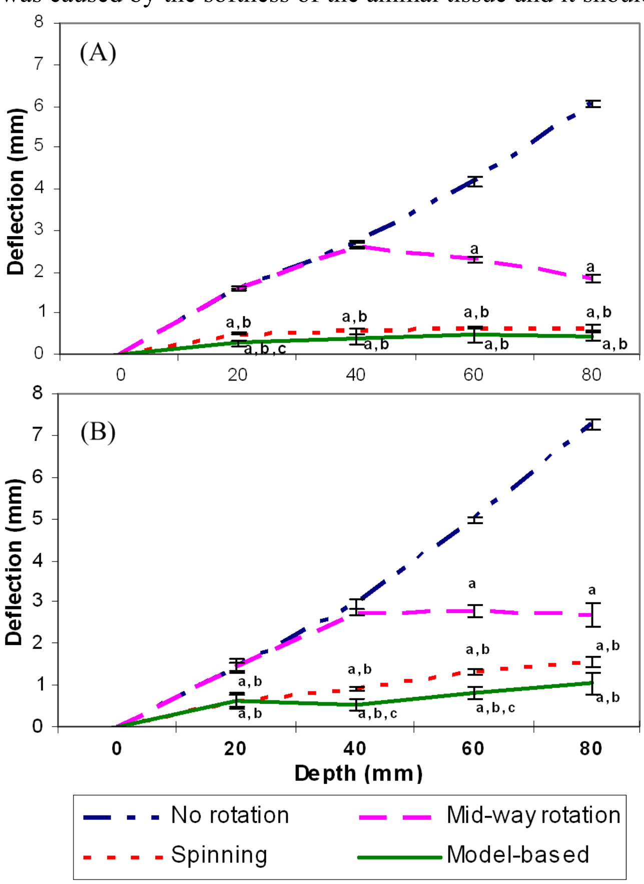

Fig. 3. Needle deflection resulting from different insertion methods (A) in gelatin phantom (B) in beef muscle. Needle deflection in points labeled as 'a', 'b' and 'c' are statistically different from the corresponding points in the method with no rotation, mid-way rotation and spinning, respectively.

be corrected in order to maintain needle deflection along the needle shaft below the desired threshold.

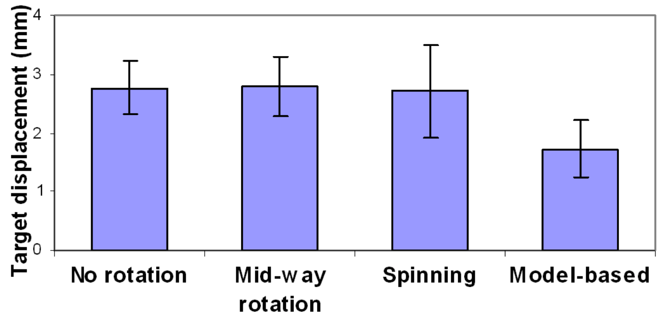

Fig. 4. Target displacement in beef muscle.

In group two of the experiments, the needle was inserted with constant translational velocity of 10 and 20 mm/sec. Needle rotation (spinning) was generated with constant rotational velocities of 10, 25 and 50 rpm. Higher spinning rates were not considered because of the wrap-around of the force senor cable. The results show that needle spinning reduced the amount of needle deflection. The amount of reduction was also found to be dependent on the speed of insertion. The reduction of needle deflection resulting from small increase in the rate of spinning may not be statistically significant as shown in insertions with 10 mm/sec translational velocity in beef phantom (see Fig. 5). Fig. 5 also shows that reduction of needle deflection resulting from increasing velocities may not be the same in different phantoms; therefore there is no clue as to what optimum velocities should be used to minimize needle deflection in certain tissue.

V. CONCLUDING REMARKS

In this paper, a model for needle deflection in soft tissue was presented. The proposed model was used to estimate the amount of needle deflection from real-time readings of force and moment values at the needle base and the insertion depth. The online estimation was used to develop a "model-based" needle insertion method which rotates the needle through 180˚ when the amount of estimated deflection goes beyond a pre-defined threshold. Rotation of needle through 180˚ causes the bevel at the needle tip to point in the opposite direction to the preceding direction of deflection and thereby reduces the needle deflection from the insertion path.

Past research had suggested other insertion methods which can also reduce the amount for needle deflection. Therefore, we decided to compare different methods of insertions: Insertion without rotation; needle spinning (with different rates of rotation), needle insertion with orientation reversal at half of the insertion depth; and "model-based" needle insertion. The presented results should help to choose a suitable method of insertion which can address the needs of a particular medical procedure. All methods presented in this paper can be used to minimize the amount of needle deflection in the absence of imaging feedback. None of the methods increased the amount of target displacement in our

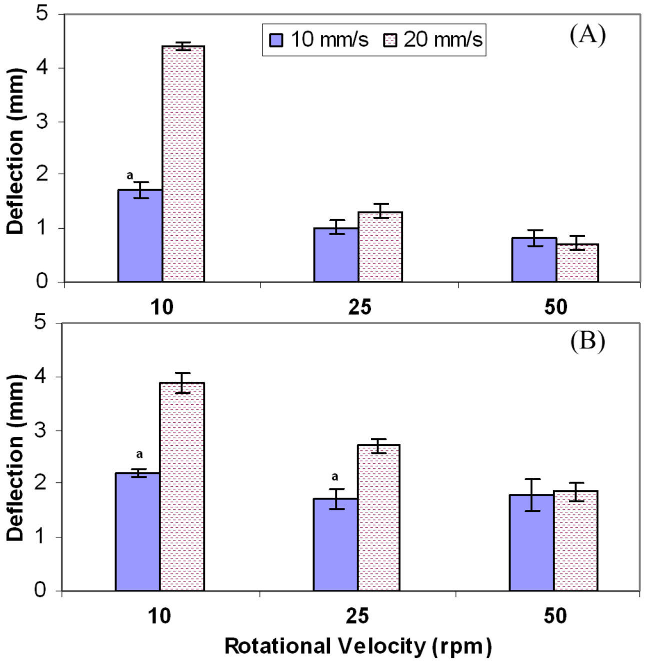

Fig. 5. Needle deflection at 80mm depth (A) in gelatin phantom (B) in beef muscle. Solid columns labeled as 'a' are statistically different from dashed columns.

phantoms. Although needle spinning and "model-based" insertion show quite similar performance, a few factors should be considered when selecting the insertion method:

- Depth of insertion: A "model-based" method could be too complicated to implement, in particular if there are different tissue types during small insertion depth. On the other hand, in large insertion depths, orientation reversal at half of the insertion depth does not reduce deflection to a desired extent; hence it does not provide desired accuracy for needle insertion.
- Tissue damage: Needle spinning may cause tissue damage during insertion in soft tissue [10]. This damage results from any small inaccuracy in the machining of the needle holder or any minor defect in the straightness of the needle. Any of these could reflect more at the tip due to the length of the needle. In such circumstances, increasing rotational velocity would increase the damage.
- Hardware requirements: This is an important factor to be considered. It is found that increasing the spinning rate can reduce the amount of needle deflection to some extent. However, rotating the needle beyond a certain velocity requires removal of force/torque sensor from the needle holder because of the wrap-around of the sensor cable. In some applications, it may be fine to remove the sensor. However, in many medical applications, the presence of a force/torque sensor in needle assembly could improve other aspects of insertion such as detection of transition between tissue layers, parameter estimation and/or controlling tissue deformation [1]. Wireless force/torque sensors should also prove useful.

- Performing the operation manually or robotically: It is clear that high rate needle spinning cannot be achieved in manual needle insertions, therefore needle spinning is only suitable for cases when a robot-assisted insertion is performed. The "model-based" method can be used for manual insertions and can be implemented as a computer-assisted method using a needle holder equipped with a force/torque sensor. During manual insertion, the program defines when rotation is required and feedback is provided to the clinician via a software user interface or a hardware indicator such as an LED on the needle holder.

The above suggests that the "model-based" approach would be a suitable method for prostate brachytherapy in which multiple needles pass through similar tissue types before reaching the prostate gland with minimum deflection.

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