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A model to predict deflection of bevel-tipped active needle advancing in soft tissue

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ABSTRACT

Active needles are recently being developed to improve steerability and placement accuracy for various medical applications. These active needles can bend during insertion by actuators attached to their bodies. The bending of active needles enables them to be steered away from the critical organs on the way to target and accurately reach target locations previously unachievable with conventional rigid needles. These active needles combined with an asymmetric bevel-tip can further improve their steerability. To optimize the design and to develop accurate path planning and control algorithms, there is a need to develop a tissue–needle interaction model. This work presents an energy-based model that predicts needle deflection of active bevel-tipped needles when inserted into the tissue. This current model was based on an existing energy-based model for bevel-tipped needles, to which work of actuation was included in calculating the system energy. The developed model was validated with needle insertion experiments with a phantom material. The model predicts needle deflection reasonably for higher diameter needles (11.6% error), whereas largest error was observed for the smallest needle diameter (24.7% error).

1. Introduction

Needle insertion is a percutaneous and minimally invasive technique that is commonly used in medical diagnosis (e.g. biopsy) and treatment (e.g. brachytherapy) procedures. The success of these procedures depends on the accuracy of needle placement at target locations, as well as the ability of the needle to navigate within the tissue to avoid critical organs and to conform the organ shape [1]. To improve the success of the procedures recently there is an increased interest to design active or self-actuating needles that can bend with the help of actuators [1–4].

Hutapea and co-workers [1,2] proposed a smart needle, see Fig. 1, where the cannula was bent by Nitinol wires attached to the body. These Nitinol wires contract when resistively heated beyond a transformation temperature thereby applying bending forces. Similarly, using Nitinol wires as actuators, Ryu et al. [3] developed a prototype of a magnetic resonance compatible active needle. They increased the temperature of Nitinol wires by optical heating. Ayvali et al. [4] developed a prototype of a discretely actuated steerable cannula, where bending forces were achieved by a Nitinol wire that was initially straight at room temperature and was transformed to a curved configuration when resistively heated. Shape memory alloy is preferred as actuators because of its high power to volume ratio which is critical in surgical needles to minimize the needle size. However, though manageable, a limitation of Nitinol is the relatively high transformation temperature required for actuation that might locally induce thermal damage to surrounding tissue [2]. Alternatively, Webster et al. [5] proposed to steer the needle using pre-curved concentric tubes. This concept was initially implemented with a single active cannula [6] and recently was extended to a multiple active cannula [7]. Several interventions could benefit from these curved needles, such as dosimetric benefits for prostate brachytherapy with active needles [1] and benefits to neurosurgery with the concentric tubes [8].

The maneuverability of active needles can be greatly enhanced using an asymmetric bevel-tip. Several studies including [9] showed that the asymmetry at the needle tip naturally bends these needles. To achieve accurate needle trajectory and targeting there is a need to develop planning and control algorithms of active bevel-tipped needles within tissue [10] based on tissue–needle interaction models. Though these models have already been discussed in the literature for needles with forces applied only at the proximal end to steer the needle (passive needles), such models are yet to be developed for active needles.

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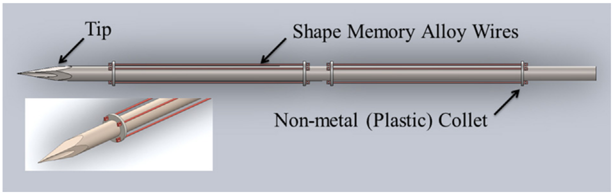

Fig. 1. Schematic of the active surgical needle.

1.1. Related work

Needle insertion with passive needles has been studied both experimentally and numerically which have been summarized by Abolhassani et al. [11] for needle insertion behavior and by Gerwen et al. [12] for needle–tissue interaction forces. Most of the early needle insertion models [13–19] simulated needle insertion without specifically including the tissue–needle interaction forces though experimental work showed the importance of interaction force [20], as pointed out by Misra et al. [21]. However, these interaction forces were considered in the following works [21–26]. Webster [22] developed a nonholonomic kinematic approach to simulate steering of bevel tip needles. Mechanics-based models to predict the needle deflection have also been developed [21,23–26], where tissue–needle interactions are generalized and thereby do not depend on empirical relations developed from previous needle insertion experiments. Kataoka and Washio [23] related the force on the needle to its deflection, while Abolhassani [24] considered the needle deflection as a cantilever beam with a spring support. Misra et al. [21] developed an energy-based model considering classical and fracture mechanics to predict the in-plane bending of bevel tip needles. Roesthuis et al. [25] modified this model to predict in-plane needle deflection with multiple bends by considering tip forces as input and assuming that needle is supported by springs. Later this model was improved by replacing spring support with distributed load [26]. None of the models mentioned above can be applied directly to simulate active needles and therefore have to be modified by including the actuation forces. This paper addresses this issue by presenting a mechanics-based prediction model for deflection of bevel-tipped active needles within soft tissue by incorporating the actuation forces into the model. This mechanics-based tissue–needle model will enable to develop mechanics-based adaptive control algorithms that would be useful for minimizing modeling uncertainties and eliminating surrounding disturbances in real tissue.

An energy-based model has been proposed in this study that was verified with needle-insertion experiments with plastisol gel and bevel-tipped steel needles of 0.38, 0.51, and 0.64 mm diameters. These validation experiments were simplified by replacing shape memory alloy actuation with magnetic forces, which is discussed in greater detail in Section 2.2.1. Finally, comparisons between the predicted and experimentally observed needle tip deflections were made.

2. Methods

2.1. Energy-based model for active needle

An energy-based model was developed to predict the deflection of bevel-tipped active needle during insertion into soft tissue. This model is an extension to the model proposed by Misra et al. [21] for bevel-tipped passive needle. In our model, in addition to the insertion forces, the actuation forces were considered as external forces acting normal to the needle. Considering actuation forces in predicting the deflected shape of the needle enables to improve needle design by helping to optimize actuator size and location, as well as to plan needle trajectories. Friction forces acting on the needle were not considered because their effect on needle deflection was found to be negligible [21], although they significantly increase the insertion force.

In this model, total system energy is expressed in terms of the transverse, $v(x)$, and axial, $u(x)$, deflections of the needle. Functional forms were initially assumed for needle deflections and then the Rayleigh–Ritz method was used to find the coefficients of these functions. The system energy, Π, was calculated by considering the internal energy stored in the system, U, and the work done by the system, W, as

$$\Pi = U - W \tag{1}$$

The internal energy, U, is composed of the energy stored by needle bending, N_E, and the energy of the tissue deformation, S_E, which are given by

$$\begin{aligned} N_E = {} & \frac{EI}{2}\int_0^{l_i}\left(\frac{d^2 v_{i-1}}{dx^2}\right)^2 dx + \frac{EI}{2}\int_0^{l_i}\left(\frac{d^2(v_i - v_{i-1})}{dx^2}\right)^2 dx \\ & + \frac{EA}{2}\int_0^{l_i}\left(\frac{du_{i-1}}{dx}\right)^2 dx + \frac{EA}{2}\int_0^{l_i}\left(\frac{d(u_i - u_{i-1})}{dx}\right)^2 dx \end{aligned} \tag{2}$$

$$S_E = \frac{1}{2}\int_0^{l_i} K_T (v_i - v_{i-1})^2\, dx + \frac{E_T (A l_i)^2}{3(1-2\nu_T)} \tag{3}$$

where, subscripts i and $i-1$ refer to present and previous iterations. E, I, A, and l are the elastic modulus, moment of inertia, cross-section area and length of the needle, respectively. K_T and ν_T are the stiffness and Poisson's ratio of the tissue. Here, N_E is composed of energy due to both transverse and axial deformation of the needle, which are further divided into energy associated in previous step and the incremental increase in energy from previous to current step. Similarly, S_E is composed of axial and transverse deformation of the tissue. Works done on the system include the work of insertion, $W_{insertion}$, and the work of actuation, $W_{actuation}$, expressed as

$$W_{insertion} = P_{insertion} u_i(0) \tag{4}$$

$$\begin{aligned} W_{actuator} = {} & P^n_{actuator}(v_i(x_m) - v_{i-1}(x_m)) + P^a_{actuator}(u_i(x_m) \\ & - u_{i-1}(x_m)) \end{aligned} \tag{5}$$

where $P_{insertion}$ is the insertion force. $P^n{}_{actuator}$ and $P^a{}_{actuator}$ are the normal and axial actuator forces acting at location x_m of the needle. The needle-tip reaction force R_i, shown in Fig. 2, is resolved into

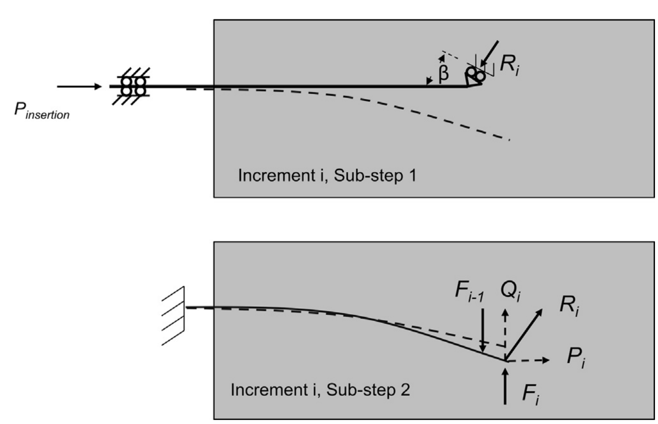

Fig. 2. Illustration showing the forces and boundary conditions acting on the needle in sub-steps 1 and 2 of increment i. R_i, Q_i, and P_i are the reaction force at needle tip and its traverse and axial components, respectively. F_i and F_{i-1} are the actuation forces acting in the current and previous steps.