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| Abstract—Needle insertion procedures are commonly used for diagnostic and therapeutic purposes. In this paper, an imageguided control system is developed to robotically steer flexible needles with an asymmetric tip. Knowledge about needle deflection is required for accurate steering. Two different models to predict needle deflection are presented. The first is a kinematics-based model, and the second model predicts needle deflection that is based on the mechanics of needle–tissue interaction. Both models predict deflection of needles that undergo multiple bends. The maximum targeting errors of the kinematics-based and the mechanics-based models for 110-mm insertion distance using a φ 0.5-mm needle are 0.8 and 1.7 mm, respectively. The kinematics-based model is used in the proposed image-guided control system. The control system accounts for target motion during the insertion procedure by detecting the target position in each image frame. Five experimental cases are presented to validate the real-time control system using both camera and ultrasound images as feedback. The experimental results show that the targeting errors of camera and ultrasound image-guided steering toward a moving target are 0.35 and 0.42 mm, respectively. The targeting accuracy of the algorithm is sufficient to reach the smallest lesions (φ 2 mm) that can be detected using the state-of-the-art ultrasound imaging systems. Index Terms—Computer-assisted surgery, image-guided control, minimally invasive surgery, needle–tissue interactions, ultrasound.  PERCUTANEOUS needle insertion is one of the most common minimally invasive surgical procedure. Needles are often used for diagnostic and therapeutic applications such as biopsy and brachytherapy, respectively. Clinical imaging techniques such as ultrasound and magnetic resonance images, and computed tomography scans are commonly used during needle insertion procedures to obtain the needle and target positions. Needles that are used in clinical procedures often have a bevel tip to easily cut and penetrate a soft tissue. Such needles naturally deflect from a straight path during insertion, which make them difficult to steer intuitively [1]. Moreover, the needles that are used in surgical procedures are often thick and rigid. Such thick needles cause deformation of tissue, and this can result in target motion, which affects the targeting accuracy [2], [3]. Another disadvantage of using thick needles is that they cause patient trauma. Besides needle deflection and tissue deformation, other possible causes of targeting inaccuracy are patient motion during the procedure and physiological processes such as fluid flow and respiration. Inaccurate needle placement may result in misdiagnosis or unsuccessful treatment. Thin needles were introduced to minimize patient discomfort [4]. Another advantage of using thin needles is that they are flexible and, therefore, facilitate curved needle paths. This enables steering the needle around obstacles (such as sensitive tissues) and to reach locations which are unreachable by rigid needles (see Fig. 1). Manually steering thin, flexible needles toward a desired location is challenging [5]. Using a robotic system which automatically steers the needle can assist the clinician. Such a system requires a model to predict the needle deflection to steer the needle to reach a certain location. This paper presents two different models to predict needle deflection. The first is a kinematics-based model, which assumes that the needle tip follows a circular path. This model is based on the unicycle model that was presented by Webster et al. [6], but modifications are made to account for cutting tissue at an angle by bevel-tipped needles. The second model is a mechanics-based model which predicts deflection using needle–tissue interaction forces [7]. The mechanics-based model to predict deflection of needles undergoing multiple bends is presented in this study. Both models are validated using double-bend experiments (see Fig. 1). In this study, image feedback is combined with the kinematics-based deflection model to steer the needle toward a target. Charge-coupled device (CCD) camera images are used for image feedback in the first set of experiments to evaluate the tracking and steering algorithms. Experiments are then performed using ultrasound images to demonstrate that the presented framework is applicable to a clinical imaging modality. To the best of our knowledge, the use of ultrasound images to steer a bevel-tipped flexible needle (φ 0.5 mm diameter) toward a moving target (less than 2 mm diameter) has not been investigated. The study also provides a method that allows the needle to move along a certain path using set points during the insertion into a soft-tissue phantom. The elasticity of the phantom affects the needle deflection [8], [9]. An acoustic radiation force impulse (ARFI) technique is an ultrasound-based noninvasive method that is used to measure the elasticity of the soft-tissue phantom. This paper is organized as follows: Section II presents the related work in the area of flexible needle steering. Section III describes the needle deflection models and the experimental setup used for model validation. Section IV presents the control system that is used to steering the needle during insertion and the image processing techniques that are used for feedback. In Section V, the experimental results are presented, followed by Section VI, which concludes and provides directions for future work.  II. RELATED WORK In the recent years, several research groups have developed algorithms for image-guided needle steering. Some of these algorithms encompass needle deflection models (see Section II-A), and techniques to track the needle tip and target in real time (see Section II-B). In this section, algorithms that were used in previous studies are discussed. The section also concludes by briefly presenting our proposed method for needle steering. DiMaio and Salcudean [10] were among the first to investigate steering needles through a soft tissue. They developed a needle Jacobian which relates needle base motion outside the tissue to needle tip motion inside the tissue. Maneuvering the needle base causes the soft tissue around the needle to deform, and this enabled them to place the needle tip at a desired location. Glozman and Shoham [5] also used base maneuvering to steer the needle. A model was used to simulate the interaction between a needle and a soft tissue. Needle steering was accomplished by solving the forward and inverse kinematics of this model. Neither DiMaio and Salcudean [10] nor Glozman and Shoham [5] used needles with an asymmetric tip. The advantage of using needles with asymmetric tips is that the needle deflection can be used for steering. The direction of deflection (in the planar case) is changed by rotating the needle 180◦ during insertion (see Fig. 1). Several research groups have focused on the steering of flexible needles with a bevel tip, e.g., [1], [6], [11]–[18]. The deflection of a needle with a bevel tip can also be controlled using duty cycle rotation [19].  A. Needle Deflection Models  Webster et al. [6] presented an approach in which they used the kinematics of unicycle and bicycle models to predict the needle deflection. In their work, they assumed that the needle tip moves along a circular path. The unicycle model assumed that the paths followed by the needle before and after rotation are tangent to each other. In the bicycle model, the paths before and after rotation are not assumed to be tangent to each other. They assumed relatively stiff tissue and showed that their model agrees with experiments. The kinematics-based model by Webster et al. is limited since it did not account for needle–tissue interaction along the length of the needle. Several groups focused on a mechanics-based approach to model needle deflection. They used the interaction between the needle and surrounding tissue to predict the needle curvature. Alterovitz et al. [12] presented a planning algorithm for a needle with a bevel tip to determine the insertion point in order to reach a desired target. Finite-element (FE) modeling was used to model the needle–tissue interaction, and this was employed in their planner to account for soft-tissue deformation. FE modeling requires computing power that is not convenient to implement in real-time control. Therefore, analytical needle deflection models were proposed to predict the deflection of needles with a bevel tip during insertion in a soft tissue [9], [13], [20], [21]. Kataoka et al. [20] presented a force-deflection model, where they assumed a constant force per unit needle length. This assumption resulted in discrepancies with the experimental deflection. Abolhassani and Patel [13] described a model that related force/torque data at the needle base to deflection. They did not account for tissue deformation along the needle shaft. This led to errors between measured and predicted deflections. Misra et al. [21] presented a mechanics-based model that predicted needle deflection using the Rayleigh–Ritz formulation. Roesthuis et al. [9] extended this model by adding spring supports along the needle shaft. However, none of these models could predict the needle deflection for the case when the needle is rotated during insertion (i.e., multiple bends). The authors presented a mechanics-based model to predict deflection of a needle undergoing multiple bends [7]. In addition to the mechanics-based model, a modification of the unicycle model is presented in this study. This model requires fewer parameters than the bicycle model to describe needle deflection accurately. This kinematics-based model is compared with the mechanicsbased model. The deflection model and real-time needle tracking are used to develop a feedback control system to steer flexible needles.  B. Needle and Target Tracking  In previous studies, a needle (without a bevel tip) and target positions were tracked in fluoroscopic and ultrasound images using image processing algorithms [5], [22]. In ultrasound images, the needle visibility is affected by the operator’s skill in aligning the needle in the ultrasound imaging plane [22], [23]. Okazawa et al. [24] developed two algorithms that were based on the Hough transform to detect the needle shape in ultrasound images during insertion. There are other segmentation techniques that can be used for needle tracking based on corner detection and subtraction [22], [25]. The main advantage of the subtraction and corner detection techniques is that the required processing time is short, which makes these techniques suitable for real-time applications. The disadvantage of using the subtraction method is that it is sensitive to motion of a soft tissue. Corner detection is immune to such artifacts that may appear in the image outside the processed region. Magnetic tracking sensors [26], [27] and fiber optic strain sensors [28] were also used for real-time needle tracking. Tracking the needle tip slope and the target position is also required to steer the needle. The tip slope changes during insertion due to needle deflection. Target displacements over 2.0 mm have been measured during placement of a biopsy needle in the breast [2], [29]. Target displacement introduces targeting errors [3]. The target position needs to be measured in each frame in order to increase the targeting accuracy of the steering algorithm.  C. Proposed Algorithm for Steering  In this study, two different (kinematics-based and mechanicsbased) needle deflection models are presented. A revised set of experiments are performed to compare the results of the models. The kinematics-based model is used in the proposed control system. The system uses processed images for feedback control. A real-time needle tracking algorithm is developed based on processing camera and ultrasound images. The Harris corner detection technique is used for tracking the needle (φ 0.5 mm) tip position. The algorithm that is used to measure the tip slope is based on image moments [30], [31]. The displacement of the target is detected to reduce the targeting error. Target motion is measured by calculating the centroid of the target shape using image moments [32]. The tracked moving target is of φ 2.0 mm. The proposed algorithms for needle and target motion tracking are applicable for both CCD camera and ultrasound images. The tracking algorithms are suitable for real-time applications. The steering algorithm uses set points to specify a certain path for the needle to follow during insertion. In the control system, it is assumed that the needle follows a circular path during insertion. This assumption was used in previous studies [6], [12]. Deviation of the needle from its planned path due to disturbances or inaccurate assumptions is corrected in real time by the developed algorithm.  III. NEEDLE DEFLECTION MODELS  In this section, two models to predict needle deflection are presented. Both models assume that the needle bends in plane (2-D). The first model uses a kinematics-based approach, while the second model is based on the mechanics of needle–tissue interaction. Both models assume that the needle shaft follows the path that is described by the needle tip. The experimental setup and the soft-tissue phantom used are described. Experiments are presented to fit the parameters of both models. Experiments are performed to validate both models in the case of steering toward a target with a single rotation.  A. Kinematics-Based Model  The idea of using nonholonomic kinematics to describe the needle path of a flexible, bevel-tipped needle has been demonstrated by Webster et al. [6]. The approach assumes that the needle tip follows a circular path. They proposed using a unicycle model with a steering constraint to describe the circular needle path. The unicycle model could not describe the needle path when needle rotation is performed during insertion. This is due to the fact that the circles describing the needle path before and after needle rotation are not tangent to each other [see Fig. 2(a)]. To describe the nontangent needle path, the bicycle method is used. In this study, a modification of the kinematics-based unicycle model is presented, which accounts for the nontangent needle path (see Fig. 2). It has been observed that a bevel-tipped needle cuts the tissue at an angle from the central axis of the needle [12], [21]. We denote this angle as the cut angle β. The cut angle is modeled by placing a frame at the needle tip (Ψt), which is rotated by the cut angle with respect to the central axis of the needle [see Fig. 2(b)]. The needle tip travels through a soft tissue in the direction that is indicated by xt. Rotation of the needle around its central axis results in a change of direction of xt [see Fig. 2(c)]. This causes the needle tip to follow a path which is not tangent to its path before rotation [see Fig. 2(a)]. The needle tip follows the circumference of a circle (center c and radius rt) as shown in Fig. 2(a). The needle tip lies at the origin of frame Ψt; expressed in the global coordinate frame this becomes o0 t = [ xtip ytip ztip ] T . Since planar needle deflection in the x0 y0 -plane is assumed, ztip equals zero. The needle deflection ytip can be expressed as a function of the needle tip x-coordinate xt    where c0 x and c0 y are the x- and y-coordinates of the circle centre expressed in the global coordinate frame (Ψ0 ), respectively. The circle centre coordinates c0 that are expressed in the global coordinate frame are calculated by performing a homogeneous transformation    where ct are the homogeneous coordinates of the circle centre expressed in the tip frame    In (2), H0 t represents the homogeneous transformation from the tip coordinate frame to the global coordinate frame    The rotation matrix R0 t depends on the orientation of the bevel tip (see Fig. 3). In the case of the bevel face pointing up, the tip frame (Ψtu ) needs to be rotated by the cut angle about the ztu -axis to align it with the central axis of the needle    If the bevel face is pointed down, the frame Ψtd first needs to be rotated about the ztd -axis by the cut angle, then a rotation of ϕ about the xtd -axis is required to align it with the central axis of the needle    Finally, a rotation equal to the needle tip slope θ around the z1 - axis has to be performed to align the x1 -axis with the x0 –axis    Thus, for the bevel face pointing up, R0 t in (4) is calculated by    and for the bevel face pointing down this is    Using (2)–(9), needle tip position (xtip ,ytip ), and needle tip slope θ, the centre of the circle ci+1 that describes the next needle path can be determined at each instant during insertion. This allows predicting the future needle path if a rotation is to be made. This is essential to steer the needle, which will be discussed in later sections.  Using (2)–(9), needle tip position (xtip ,ytip ), and needle tip slope θ, the centre of the circle ci+1 that describes the next needle path can be determined at each instant during insertion. This allows predicting the future needle path if a rotation is to be made. This is essential to steer the needle, which will be discussed in later sections.  and hence, shortening of the needle is not considered. When the needle is inserted without being rotated, it has a single-bend shape [see Fig. 5(a)]. The needle deflects due to a combination of the distributed load and the tip force. Needle rotation is performed when the insertion distance equals the rotation distance (xt = xr ). This results in a change of orientation of the bevel tip, and hence, the tip force also changes direction [see Fig. 5(b)]. This causes the needle to deflect in the opposite direction. For a single rotation, this results in the needle having a double-bend shape. To model this, the part of the needle before rotation is fixed by a series of springs. Given a sufficiently small spring spacing Δl, one can approximate an elastic foundation [33]. The stiffness of such an elastic foundation K0 is described in terms of stiffness per unit length and depends on needle and tissue properties. The length of the foundation in Fig. 5(b) equals xr − x0 , resulting in a foundation stiffness Kt of    For a total number of m springs, this results in a spring stiffness Ks of Kt m . The distributed load is applied to the part after rotation and this enables modeling of a needle undergoing multiple bends. To evaluate the deflected needle shape (v(x)) under the action of distributed load and tip force, the Rayleigh–Ritz method is used. Rayleigh–Ritz is a variational method in which equilibrium of the system is established using the principle of minimum potential energy [34]. For a mechanical system, the total potential energy is expressed as    where U represents the energy that is stored in the system, and W is the work done on the system by external forces. To find the deflected needle shape using the Rayleigh–Ritz method, an assumed displacement (shape) function has to be defined. Several shape functions were evaluated, and it is found that a cubic function is a suitable shape function    For complex needle shapes, a single shape function as in (12) is not sufficient to approximate the deflected needle shape. Therefore, the needle is divided [see Fig. 4(b)] into a number of elements n, each described by their own shape function (vi(x))    The unknown coefficients a0,i,...,a3,i are determined using the Rayleigh–Ritz method. For the first needle element (i = 1), xi−1 equals xb , and for the last element (i = n), xi equals xt. Each of the shape functions has to satisfy the geometric boundary conditions of the system. Since the needle is fixed at the base, the needle slope θ(x) and deflection v(x) are zero at the base    Furthermore, the shape functions have to satisfy continuity conditions, meaning constant deflection and needle slope at the boundaries of the elements    For the single-bend case [see Fig. 5(a)], the stored energy equals the strain energy due to transversal needle bending (U = Ub ). Using Euler–Bernoulli beam theory [35], the strain energy due to transversal bending Ub is found to be    where E and I represent the Young’s modulus and second moment of inertia of the needle, respectively. The needle is cylindrical and EI is constant along the length of the needle. For a needle undergoing multiple bends, energy is also stored in the springs [see Fig. 5(b)]. The stored energy is the sum of the energy due to needle bending as defined in (17) and the spring energy for a total number of m springs    where Ks represents the spring stiffness, and v(xk ) is the amount of deflection for the kth spring with respect to the bend configuration, as shown in Fig. 5(b). The work done on the system by external forces is the sum of the work done by the distributed load Wd and concentrated tip load Wc :    The work done by the distributed load is given by    and the work done by concentrated tip load is given by    where v(xt) is the deflection at the needle tip. The shape functions defined in (14) are substituted in the equations for the stored energy [see (17) and (18)] and work [see (20) and (21)]. This results in the total potential energy of the system, defined in (11), to be a function of the shape functions, and hence the unknown coefficients    The equilibrium of the system is found by taking the partial derivative of the total potential energy with respect to each of the shape functions’ unknown coefficients    for k = 0, 1, 2, 3 and i = 1,...,n. The unknown coefficients ak,i are calculated by solving the system of equations obtained in (23). Substitution of the coefficients back into (13) and (14) gives the deflected needle shape. Experimental data are used to evaluate the parameters for both needle deflection models. These parameters are the radius of curvature and the cut angle for the kinematics-based model and the distributed load for the mechanics-based model. In the next sections, the experimental setup is first introduced, and then experiments are presented which are used to evaluate the parameters. With the parameters known, both models are then validated in a series of (open-loop) steering experiments. |  |