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Multi-rate unscented Kalman filtering for pose and curvature estimation in 3D ultrasound-guided needle steering

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ABSTRACT

This paper presents a new method for estimating the needle pose and curvature in the context of robotically steered needles. The needle tip trajectory is represented by a modified unicycle kinematic model. A multi-rate unscented Kalman filter is proposed for the first time in needle steering to fuse asynchronous data coming from 3D B-mode ultrasound images, robot sensors and pre-operative elastography measurements. To demonstrate it, 51 unconstrained 3D insertions have been made in various media. The instantaneous localisation error is smaller than 0.6 mm and the prediction of the final tip position is smaller than 2 mm based on the observation of the first 2 cm of 8 cm deep insertions.

1. Introduction to needle steering

In percutaneous operations, the needle is often inserted by the clinician under ultrasound (US) guidance. Because the clinical efficacy of such procedures is closely linked to the surgeon's ability to reach a specific target with a needle, robotically assisted needle guidance also referred to as "needle steering" has appeared and been the object of intensive research.

There are several ways to steer a needle in soft tissue. In this paper, the focus is put on bevel-tip needles which naturally bend when inserted in soft media. This phenomenon is due to the needle-tissue asymmetric interaction forces. This makes it possible for an expert practitioner to steer a needle to avoid an obstacle and to compensate for the tissue motion during the insertion. Such a deflection is however difficult to predict and to control manually, hence the need for surgical robotics.

One challenge in needle steering is to locate and estimate the current pose of the needle in images, for instance ultrasound volumes, and predict its future deflection. Solving this problem may rely on modelling the needle and developing an observer.

A first kinematic model is introduced in Webster, Kim, Cowan, Chirikjian, and Okamura (2006). It shows that bevel-tip needles follow a circular path when inserted in homogeneous phantoms. This model has been successfully used by many studies (Abayazid et al., 2015; Carriere, Rossa, Sloboda, Usmani, & Tavakoli, 2016; Patil, Burgner, Webster, & Alterovitz, 2014). However, in most studies, the radius of curvature has been considered constant during the insertion with few exceptions. This radius of curvature is often obtained by prior insertions. Such an approach is incompatible with a clinical application.

The curvature of the needle depends on the stiffness of the tissue it penetrates. It has been studied in Moreira and Misra (2015) where the curvature estimation is done both online and offline: online by doing pure insertions and fitting a circle with the acquired planar trajectory; offline with pre-operative elastography measurements. In Carriere et al. (2016) a real-time needle shape prediction is proposed using a particle filter in the case of planar insertions. Unlike these studies, the solution detailed here is an estimator based on a fully 3D-compatible kinematic model. It takes into account the elasticity of the media and does not constrain the insertion to be planar.

Dynamic model-based deflection estimation has also been proposed for 2D (Khadem, Rossa, Sloboda, Usmani, & Tavakoli, 2016) and 3D (Chevrie, Krupa, & Babel, 2016) steering. The approach looks promising and has been validated using 2D ultrasound feedback (Waine, Rossa, Sloboda, Usmani, & Tavakoli, 2016).

With the exception of Chatelain, Krupa, and Navab (2015) and Mignon, Poignet, and Troccaz (2016), there is to our knowledge no recent work about needle steering based solely on 3D B-mode US images. The reason lies in the strong presence of artefacts and noise of such images.

Needles are hard to detect when they are thin or are not perpendicular to the US probe plane (Huang et al., 2007). Many solutions

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(Abayazid et al., 2015, Fallahi, Rossa, Sloboda, Usmani, & Tavakoli, 2017, Moreira & Misra, 2015, and Rossa, Lehmann, Sloboda, Usmani, & Tavakoli, 2017 among others) involve the use of a 2D probe that is translated to follow the needle tip. This solution to observe the needle also called 2.5D aims to keep the needle axis orthogonal to the imaging plane. In this way, the needle is always in the configuration of the maximum visibility and images are acquired at around 30 Hz. Unfortunately, it is not practically feasible for most clinical operations. In addition, the movements of the probe may cause tissue deformation and may affect the needle deflection.

To avoid B-mode US imaging quality issues, Doppler mode imaging has also been exploited in Adebar and Okamura (2014) and in Mignon, Poignet, and Troccaz (2015). Although the needle visibility is improved by this imaging modality, specific needle movements are required and the overall precision of this imaging mode is lower than that of B-mode imaging.

As detailed above, there exists a broad variety of solutions for needle pose estimation and prediction, each of them validated on different experimental setups. Therefore, a direct quantitative comparison is made difficult. An overview of some of the most relevant observers in needle steering can be found in Appendix A.

A clinically-compatible 3D US probe with a 1 Hz refresh rate is used here in B-mode. The limited quality of the images and low frame rate are compensated by the precise estimation of the next needle tip pose. It specifies a region of interest for the next segmentation and decreases the likelihood of losing the needle tip in US volumes.

In this paper, an observer taking all available data as input to estimate the needle tip position and predict its future deflection is proposed. A multi-rate approach has been selected to fuse the needle rotation and insertion information provided by the robot, and measurements deduced form the pre-operative shear wave elastography imaging and intra-operative B-mode US volumes.

This paper is composed of two parts :

– In Section 2, the advantages and limitations of the 3D ultrasound imaging are developed. The model chosen to describe the behaviour of the needle tip is detailed. The proposed observer is then presented.
– In Section 3, the observer is first validated through simulation. Then, it is experimentally validated with 51 insertions made with various 24 gauge (0.51 mm diameter[1]) needles with asymmetric tips. These insertions are made in phantoms and pork tissue samples.

2. Methods

As already mentioned, 3D ultrasound imaging and kinematic modelling of the needle are considered here. Based on these choices, the methods employed in this paper are detailed below.

2.1. 3D ultrasound imaging

B-mode imaging

In this paper, a motorised end fire transrectal ultrasound probe represented in Fig. 1 is set at a fixed position and works in B-mode imaging. 3D volumes are reconstructed from images acquired by rotating an array of piezoelectric transducers located in the probe head. The major drawback of such a setup is the limited image quality it provides. Indeed, when the needle is not orthogonal to the US emission direction, strong artefacts appear (Huang et al., 2007).

The lack of visibility of fine needles in 3D US volumes is due to their shape. The US waves reflect on the needle smooth surface in one direction and do not hit the transducer back. One possible refinement for regular B-mode imaging would involve roughening the surface of the needle so that the US waves diffuse in all directions and reflect back to the transducer.

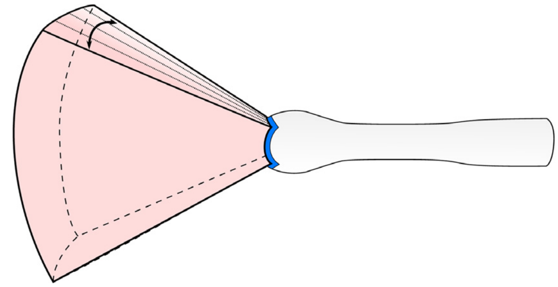
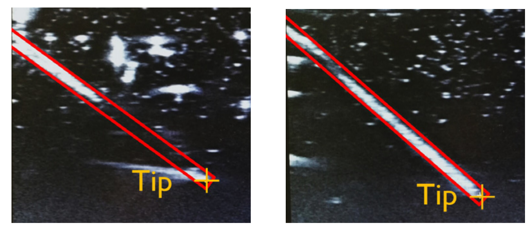


Fig. 1. 3D volume (in red) acquired by a motorised end-fire transrectal ultrasound probe. The sweep direction of the transducer is indicated by an arrow. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)



(a) Without surface treatment (b) With polyurethane coating

Fig. 2. 2D US acquisitions of needles inserted in soft tissue. The needle body is represented by a red frame and the needle tip by an orange cross. The use of polyurethane coating improves the needle visibility.

Clinically compatible solutions to roughen the needle surface exist such as micro laser etching (Mignon, Poignet, & Troccaz, 2018). However, because of the high complexity of this process, a non-clinical approach applied here is to coat needles with polyurethane foam (Huang et al., 2007) (see Fig. 2).

Because the needle visibility depends on the needle position and distance to the transducer and the medium it penetrates, the discernibility of the needle is predictable. The visibility quality can be taken into account by the estimator through the measurement noise covariance matrix.

Shear wave elastography imaging

The second imaging modality used in this paper is shear wave elastography imaging (see Fig. 3). The propagation of an acoustic radiation force impulse is observed and used to compute Young's Modulus of a soft tissue characterising its elasticity properties (Bercoff, Tanter, & Fink, 2004). The radius of curvature of the needle is correlated to the tissue elasticity properties. Therefore, shear wave elastography imaging (SWE) provides a rough measurement of the needle curvature.

2.2. Modified unicycle kinematic model

First introduced in Webster et al. (2006), the bicycle kinematic model has since been validated and embraced by most researchers in needle steering. A bevel-tip needle supposedly traces a circular path when inserted inside homogeneous tissues. The arc direction is related to the bevelled tip orientation. However, when rotated at 180°, the needle follows another arc that is not tangent to the previous path. The so-called cutting angle is defined as the angle between the tangents to the trajectory before and after the rotation (see Fig. 4).

[1] Unit conversion table available at https://en.wikipedia.org/wiki/American_wire_gauge.