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Towards an Augmented Ultrasound Guided Spinal Needle Insertion System

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***Abstract*— We propose a haptic-based simulator for ultrasound-guided percutaneous spinal interventions. The system is composed of a haptic device to provide force feedback, a camera system to display video and augmented computed tomography (CT) overlay, a finite element model for tissue deformation and US simulation from a CT volume. The proposed system is able to run a large finite element model at the required haptic rate for smooth force feedback, and uses haptic device position measurements for a steady response. The simulated US images from CT closely resemble the vertebrae images captured *in vivo*. This is the first report of a system that provides a training environment to couple haptic feedback with a tracked mannequin, and a CT volume overlaid on a visual feed of the mannequin.**

I. Introduction

Medical simulators are becoming more common for teaching students and technicians important clinical procedures [1]. Simulators provide the ability to access a multitude of patient data in an offline environment, especially rare cases that may not be experienced otherwise until further into a clinician's career. *In vivo* training requires the direct supervision of a qualified clinician, the cost of which can be quite high and the available supervising hours limited. Medical simulators (hereafter referred to as training systems) can aid in reducing cost and increasing training availability. In addition, they provide a platform for technical assessment of the procedure via collected multi-sensory data, from visual to position tracking, and the ability to repeat procedures to view and track the progress.

Medical training systems in the literature can be classified into two groups: virtual reality (VR) based, and mannequin based. The VR-based training systems ([2], [3], [4], [5]) allow users to experience and view a three dimensional (3D) scene. In a medical training application such as ultrasound (US) scanning or image guided needle insertion, the user navigates within the scene and interacts with objects such as a patient's body and internal organs using a virtual tool. This interaction and navigation is generally accomplished with the use of a haptic robotic device that is moved, and provides force feedback, in various degrees of freedom (DOF). The end of the haptic device held by the user corresponds to the location of the tool tip in the virtual scene. The visual feedback portion of a VR system is commonly a computer generated image shown on a monitor. Mannequins, on the other hand, are often used to provide a trainee with spatial awareness and visual and physical clues that are otherwise missing with a purely virtual scene.

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An example of a VR-based training system is an abdominal biopsy procedure simulator [2]. Two 3-DOF haptic devices are used to mimic an US probe and needle, with a 3D scene viewed through stereoscopic goggles. An US image is simulated from a slice of a computed tomography (CT) volume. However, the simulated US is not deformed based on the probe pressure or the needle position. A lumbar puncture training system using a 6-DOF haptic device is discussed in [4]. This system is able to provide torque as well as force feedback to the user, but the user can suffer from the lack of visual awareness since the only object visible in 3D is the haptic device, and not the object the user is interacting with. The spinal needle training system described in [5] uses a mass-spring model to deform simulated US slices, based on the needle insertion. It does not seem to model the acoustic shadow effects of the needle, and could suffer from the same spatial awareness issues as in [4] since the only form of visual feedback is a simulated computer image.

A mannequin based US guided needle insertion training system is discussed in [6]. The system tracks a dummy US probe and needle with a motion sensor, and morphs a CT volume onto a mannequin surface. The simulated US image and CT volume are displayed on a monitor. The mannequin provides good spatial awareness for the user while performing the procedure, but the feel of the insertion is restricted to the limited makeup of the inside of the mannequin. In [3], a spine needle biopsy training system is presented which uses a haptic device to simulate needle insertion on a mannequin. A display of the CT volume with needle path is shown on a monitor. This system is a good example of combining VR and mannequins, but does not include an image that is needed for US-guided procedures.

In this paper, we present the basis for an ultrasound guided spinal needle insertion training system that aims to address the shortcomings of both types of previously discussed systems to create a more thorough training and simulation environment. We introduce the first training system to couple the use of a mannequin and haptic feedback with a finite element (FE) tissue deformation model, and to provide an augmented reality visualization by overlaying a CT volume onto a video feed of the mannequin.

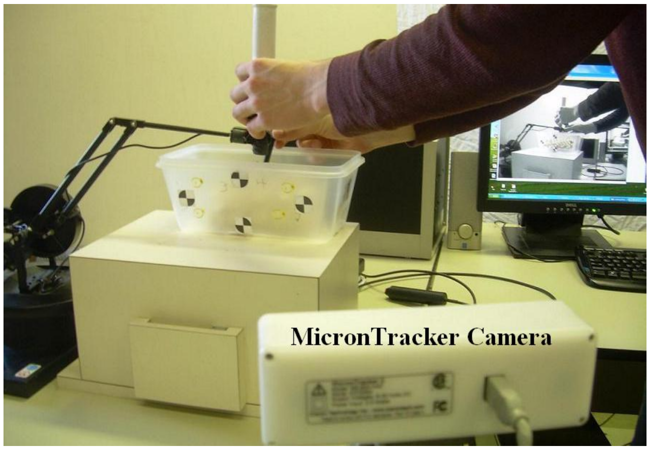

Fig. 1. System Overview: User interacting with mannequin. The MicronTracker Camera is in the foreground

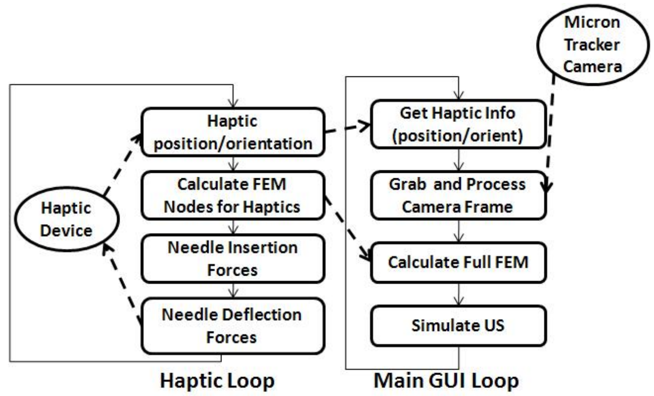

Fig. 2. Program Flow: Main graphics loop shown on right. High priority haptic loop shown on left. Interactions between loops and outside hardware shown with dashed arrows

II. System Overview

The proposed spinal needle insertion training sytem is composed of Sensable PHANTOM Premium 1.5A (Sensable, Wilmington MA) haptic devices, a MicronTracker2 (Claron Technologies, Toronto ON) camera, a PC and a patient-specific phantom with 3D printed spine model in a box. Throughout the paper, the latter component will be referred to as the mannequin, to avoid confusion with the Sensable haptic device. A snapshot of the system is shown in Fig. 1. The haptic devices represent the US probe and needle, and the user moves them to interact with the mannequin which is easily accessible on a table. The mannequin is also in view of the camera system. A PC is nearby, and displays the visual feedback to the user. The visual feedback is composed of two parts: a CT volume overlay on the video feed, and images of the CT and US slices at the current probe location.

Fig. 2 illustrates the flow in the program. The graphics loops (GUI loop) is the main loop of the program, and controls most of the other functions. The loop makes calls to the functions controlling the MicronTracker camera to grab frames from the video feed, and to modify the CT volume overlay based on the location of the mannequin; both the video feed and overlay are shown on the monitor. After this, the ultrasound image is simulated from the CT volume sliced with the given PHANTOM position and angle, and the deformed FE nodes. Finally, the two slices (CT and simulated US) are displayed on the screen (Fig. 3, Right).

Underneath this is the high priority haptics loop. In order to provide a realistic force feedback, this loop must run with a frequency of at least 1 KHz. The haptic loop provides the graphics loop with the necessary position and angle measurements from the PHANTOM device to carry out the rest of the functions. It also calculates the needle insertion and deflection forces based on the virtual needle position.

The system is implemented on a Quad Core Intel Q9550 2.83 GHz computer with 3.25GB of RAM. The C++ libraries of the Insight Toolkit (ITK) [7] version 3.18.0, Visualization Toolkit (VTK) [8] version 5.6.0 and QT version 4.6.2 are also used for the CT volume manipulation, augmented reality overlay, and GUI respectively.

A. Optical Tracking and Image Overlay

A MicronTracker2 camera from Claron Technologies is used to provide a real-time video feed of the simulator, and its marker tracking functions allow for an augmented visual overlay. The MicronTracker tracks visual markers, specifically the contrast between two white and black edges. A facet composed of four markers and two vectors can be tracked to provide location and rotation information. The markers can be transformed into the space viewed by the camera used to output the video feed (in this case the left camera), and from this information the CT volume can be overlaid onto the image. Facets in multiple locations on the desired tracking object can be stored to increase the tracking accuracy. We register the CT volume to the mannequin through the use of the facet markers. The markers define landmarks on the mannequin that we orient the CT volume to match.

The image overlay is performed via a virtual VTK scene. Measurements were performed on the MicronTracker camera to determine its viewing angles, and the virtual cameras properties were changed to match these measurements, and the other properties of the camera. The horizontal viewing angle was measured at 45 degrees. The CT volume is rendered in this virtual scene through the VTK toolkit, and a snapshot of the scene is taken and alpha composited with the image from the MicronTracker camera.

As discussed in [9], the simple overlay of a volume onto video feed results in a depth perception mismatch, as the overlaid image always appears in front of the background image. The paper suggests a three component approach to eliminate these visual errors: the angle of incidence of the volume, the curvature of the surface overlaid on, and the distance falloff. These components are combined and weighted to modify the opacity of the pixels at various locations in the volume. We have implemented a similar opacity function to the distance falloff, whereby pixels become more transparent as they get closer to the 2D projected edge of the volume.

This produces a smoother transition and blend between the overlaid volume and the surface (Fig. 3, Left)

B. Finite Element Analysis

The finite element model (FEM) we implemented is based on the work done by DiMaio et al. in [10] in which a 2D deformation model for needle insertion is presented. The model uses a stick-slip method of node tracking for needle insertion. When the needle is moving past a node it is said to be slipping along the node, while a stationary node along the needle is said to be stuck to the needle. With this distinction it is possible to deform the model based on either force or displacement constraints.

A FEM model for a needle is also discussed in [10]; however, because of the haptic rate constraint the needle is modeled as rigid once it enters the tissue, so no deflection occurs. The FEM is in the simple form of

$$x = K^{-1}y$$

where x and y are the free variables (both force and displacement) and constrained variables respectively. We precompute the inverted system stiffness matrix K^{-1} prior to any deformations being performed. This reduces the amount of calculations per iteration to a simple matrix-vector multiplication. Also, since the needle only acts on a few nodes at once during the insertion, y can be considered sparse, further reducing the number of calculations.

In order to transform between force and displacement constraints, K must be updated. Following the procedure in [10], this is done by an update to the precalculated matrix K^{-1}, instead of re-inverting K, which reduces the number of calculations and as a result allows a larger number of nodes to be used while maintaining nominal haptic rate.

Due to the existing constraint on the haptic sampling rate a 2D FEM model is used. A slice of the volume is taken and meshed to create an in-plane tissue deformation model. Bone is modeled as a rigid structure, so any node in the model that correlates to bone is constrained to be fixed and non-deformable.

C. US Simulation

During initialization of the system, the CT volume model is loaded using ITK and VTK and prepared for slicing. During each iteration of the graphics loop, the CT volume is sliced according to the position and orientation of the haptic end-effector. This slice is deformed based on the FE analysis described in Section *B* and the pixel values are interpolated between the nodes using a bi-linear interpolation method. Following the deformation, the needle is inserted inside the slice based on the location of the tip, and given a high Hounsfield value depending on composition [11]. For this simulation we use a 4000 H value.

To simulate the US image, we cast a ray down each column of the CT slice and calculate the reflection and transmission values from each voxel. Reflection values over a certain threshold are set to be higher to better simulate acoustic shadows from the interaction of the US wave and bone. A hanning window is applied to the resulting intensity map, as well as a logarithmic compression function to boost any minute reflections. Please refer to Shams et al. [12] for more details on the algorithm.

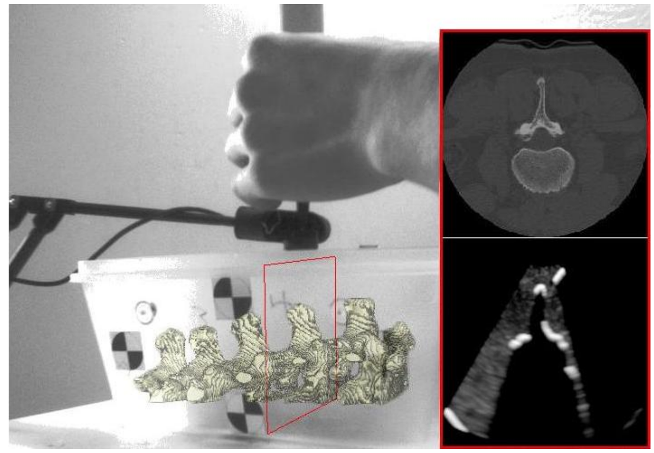

Fig. 3. Left: Screenshot of visual feed with overlaid CT volume and US plane. Right, Top: CT slice. Right, Bottom: Simulated US slice, with inserted needle

D. Haptic Feedback

To provide force feedback from the needle insertion procedure the Sensable PHANTOM premium 1.5A haptic device is used. The PHANTOM provides three active DOF (motion with force feedback) and three passive DOF. Since the device provides no torque feedback, it is necessary to constrain the needle motion along the chosen insertion direction. This constraint is achieved by locking the haptic end-effector to its direction vector upon piercing the surface. Once the surface is punctured, the needle is modeled as a long, slender, bending cylinder with appropriate force per deflection distance calculated using a standard beam deflection method.

The force feedback is calculated from the FE model. The force at the needle base is calculated as the sum of all the tissue nodes acting on the inserted section of the needle. As mentioned in Section *B*, the bone is modeled as a fixed surface in the FE model. Therefore when the virtual needle tip touches the bone a strong force is felt by the user.

To register the mannequin and haptic device, the user moves the tip of the device to a number of locations on the mannequin which have known distances to the marker frame.

III. System Performance

A screen shot of the system running is shown in Fig. 3. The acoustic shadow created by the bone, the change discussed in Section *C*, is properly modelled from the CT slice and shown in the simulated US slice.

We performed measurements on the MicronTracker camera to determine its accuracy. We measured two components - the camera jitter between frames and tracking accuracy. Jitter was measured by fixing a probe facet in a stable position on a firm surface so it would be completely visible and not able to move between samples. From the measured samples the jitter was measured at 0.173 mm per sample, RMS error. The accuracy was measured by recording the position at one location on a fixed table, then at another with known distance to it. The accuracy was measured at 2.1 mm, RMS error.

An iteration speed of at least 1 KHz is needed for smooth and realistic force feedback. In our system, the major time sink in the haptic loop is the FEM, and because of this the number of nodes that can be used becomes restricted. Since the calculation of the free variables involves a sparse vector, it does not contribute much processing time to the loop; the main processing hurdle comes during the change from force constraint to displacement constraint and vice versa. Even with a low-rank update to K^{-1}, a considerable amount of processing time is used. In [10], an iteration period of 512 Hz is reported for a mesh of 361 nodes. This is similar to times we recorded for similarly sized meshes.

For haptic feedback, only the forces at the nodes in contact with the needle need to be calculated, as the other deformations have no effect on the forces transferred to the needle. This fact, along with the sparseness of the constrained nodes y, can be used to drastically reduce the calculations needed for force-displacement constraint swapping. This is done by only calculating the row of the node being swapped, along with the elements in the column being swapped that correspond to the other nodes touching the needle. This procedure allows meshes of over 1000 nodes to be run at frequencies of 1 KHz.

IV. Discussion and Conclusion

We have proposed and reported on the development of an ultrasound guided needle insertion training system that incorporates a FE tissue deformation model, US simulation, haptic feedback and augmented reality interface with an overlaid CT volume. The system is able to provide the user with an appropriate feel of pressing with an ultrasound probe, and inserting a needle into tissue through the use of a FEM that can efficiently handle a large number of mesh nodes. The deformed FEM is also applied to the CT slice found at the location and orientation of the haptic device representing the probe. The US simulation algorithm is applied to this deformed image, after a virtual needle is inserted into it, to simulate images seen *in vivo*. The CT volume overlay provides the user with more detailed patient specific information as the user navigates the mannequin.

Improvements could be made to the FEM used for tissue deformation. The current 2D model could be expanded to 3D which will utilize all three active DOFs of the haptic device. It has to be seen whether this is computationally viable with the speed requirement of the haptic loop. Other additions to the FEM would be multiple layers of tissue for a non-homogenous environment between skin puncture and bone contact. Additional changes to the FEM could reflect modeling some needle bending during insertion, including a bevel tip. It has been reported that bending of needles during spinal insertions is small, yet noticeable up to an insertion depth of 60 mm for common needle gauges [13].

If a mannequin of a human torso was used it would allow the CT to be warped to fit the surface [6]. A torso mannequin is currently being sourced and purchased.

To increase the speed of the graphic a second computer could be added to the system to reduce the computational load, or GPU programming could be used to make more efficient calculations. GPU coding for US simulation is already in progress.

V. Acknowledgments

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