Phantom validation of optical soft tissue navigation for Brachytherapy

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Abstract. In high dose rate brachytherapy, needles are inserted into soft tissue and subsequently radioactive sources are used to deliver a high dose inside the target region. While this approach can achieve a steep dose gradient and offers a focused, organ sparing treatment, it also requires a careful positioning of the needles with respect to the tissue. We have previously proposed to use an optical fiber embedded in the needle to detect soft tissue deformation. To validate the approach, we have developed an experimental setup to compare the actual needle motion with the motion estimated via the fiber. Our results show a good agreement between actual and estimated motion, indicating that optical deformation detection through the needle is possible.

1 Introduction

In high dose rate (HDR) brachytherapy, needles are inserted into soft tissue and subsequently radioactive sources are used to deliver a high dose inside the target region. On the one hand, this approach can achieve a steep dose gradient and offers a focused, organ sparing treatment. On the other hand, it also requires a careful positioning of the needles with respect to the tissue, as the dose is highest in the direct proximity of the sources [1–3].

Different methods including gridlike templates and robotic needle drivers have been proposed for needle placement. One issue is the soft tissue deformation resulting from the insertion force [4]. One approach is modeling the needle tissue interaction [5], or to study the resulting force [6]. Clearly, image guidance can also help identifying tissue motion. However, sophisticated modalities like magnetic resonance imaging (MRI) are typically not available in brachytherapy settings. Moreover, imaging can be subject to artifacts caused be the needles, e.g., phantom echoes in ultrasound images.

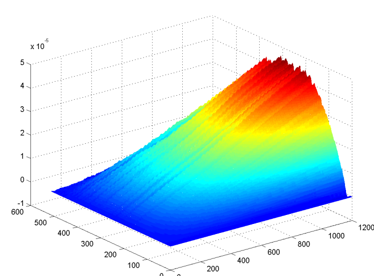
We have previously proposed to embed an optical fiber into a brachytherapy needle to allow for optical coherence tomography (OCT) along the needle path [7]. Moreover, we have studied the feasibility of using Doppler data acquired through the needler to estimate the relative motion between needle tip and soft tissue [8]. However, the actual validation of the approach requires measuring the tissue deformation caused by the needle. We present a phantom setup to measure the deformation and to compare the motion estimated from the Doppler data with the measurements.

2 Material and Methods

Optical coherence tomography (OCT) is an interferometric approach that can penetrate up to 3 mm into tissue, resulting in images with a depth resolution of better than $10\,\mu m$. While the images represent different scattering properties in the probe, OCT it can also be used to obtain Doppler data [9]. As the phase shift between subsequent A-scans is proportional to the relative velocity between fiber tip and scatter source, integrating over the velocities yields an estimate of the relative motion of the needle with respect to the tissue.

We use a bevel point tipped needle with an embedded optical fiber (Figure 1a) connected to an OCT-System (Callisto, Thorlabs) with a 1200 Hz A-scan rate. An industrial robot (Adept Viper s850) with a positioning accuracy of 30 µm is used to move the needle along its axis. Tissue motion is monitored using a Logitech Pro 9000 camera using the vcapg2-plugin for Matlab [10]. The camera's sensitivity was adapted to provide good contrast between needle and phantom structures.

(a)



(b)

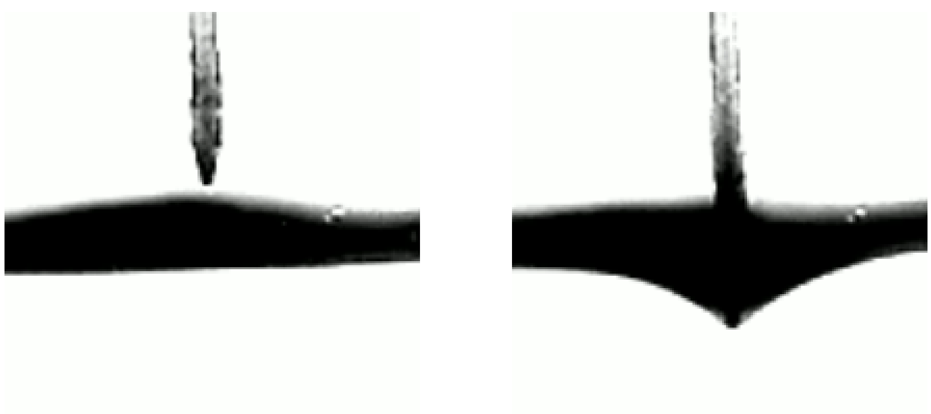
Fig. 1: On the left the modified needle with the embedded fiber is shown (a). On the right an image of the relative motion profiles for different depth accumulated over one second of needle motion (b).

For our experiments we used phantoms made of gelatine. To obtain sufficient scattering, TiO_2 powder was added. However, typically gelatine shows little friction and hence little deformation after the initial penetration of the surface. To induce and detect deformations we added layers of colored gelatine with a different stiffness, such that the needle tip and the layers are well visible in the camera images. The robot was configured to continuously move the needle with a speed of approximately 0.05 mm/s while OCT and camera images were recorded with 1200 Hz and 30 Hz, respectively. In a post-processing step, the camera images with maximum deformation of the gelatine layer were identified and the related time stamp defined the end point of the motion trace. The deformation was then estimated from the camera image by taking the needle as a reference, and the relative motion estimated by the OCT as well as the actual robot motion were determined from the recorded data.

As the OCT Doppler data can be considered as one-dimensional depth profiles of the relative velocity v_{rel} between tissue and needle tip, we need to decide in which depth to measure the motion. To this end we compute the accumulated motion for all depth and a windows of 1 s. Figure 1b shows the resulting motion profiles, which generally show a maximum approximately 300 pixels or 1 mm from the needle tip.

3 Results

Figures 2a and 2b show the deformation caused by the needle. The horizontal layer is gelatine with a higher stiffness, leading to the distinct deformation,



(a) (b)

Fig. 2: The left image shows the situation before the needle moved into the colored gelatine layer visible as a black horizontal bar (a). The right image illustrates the maximum deformation of the layer, this defines the endpoint for the measurements (b). The distance between the lower boundary of the layer in (a) and the needle tip in (b) is measured as an estimate for the deformation, it is approximately 3.28 mm.

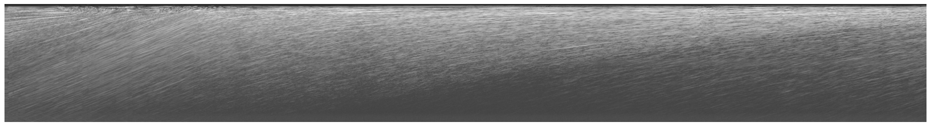
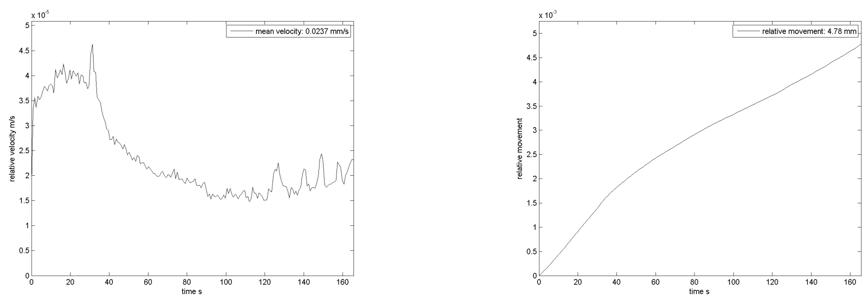


Fig. 3: An overview of the OCT A-scans at different time steps / needle positions, from left to right.



(a) (b)

Fig. 4: The left plot shows the velocity of material in front of the needle over time / needle position (a). The right plot shows the corresponding motion estimate (b).

which is approximately 3.28 mm in this case. In Figure 3 the OCT data recorded during the needle motion is summarized. Note that each column in the image corresponds to an A-scan showing the real part of the signal, i.e., the left hand side represents the situation before entering the layer and the last column on the left corresponds to the A-scan in the situation shown in Figure 2b. A careful look at the image reveals that there is a brighter region that gradually gets smaller on the left, indicating some compression.

The actual velocity of material in front of the needle and the resulting estimate of the relative motion are presented in Figures 4a and 4b, respectively. Note that the total motion through the gelatine is estimated as approximately 4.78 mm. The robot moved 8.16 mm while deformation of 3.28 mm and the estimated motion add to 8.06 mm

4 Discussion

Our experiments are preliminary in that they need to be repeated and the determination of the deformation from the images should be automated. Moreover, the scattering and absorption in actual tissue may affect ability to detection of motion in sufficient depth, where the smaller SNR typically leads to an underestimation of the Doppler signal [11].

However, the results still provide further indication that a precise estimation of the needle tip motion relative to soft tissue is possible using OCT Doppler data. Such information would be valuable when placing needles for brachytherapy or biopsies, where proper placement of the needle with respect to soft tissue is important. The methods is not affected by artifacts from other needles and the high sampling rate allows for real-time control, e.g., of robotic needle drivers.

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