A New Sensor Technology for 2D Ultrasound-Guided Needle Tracking

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Abstract. 2D Ultrasound (US) is becoming the preferred modality for image-guided interventions due to its low cost and portability. However, the main limitation is the limited visibility of surgical tools. We present a new sensor technology that can easily be embedded on needles that are used for US-guided interventions. Two different types of materials are proposed to be used as sensor - co-polymer and PZT. The co-polymer technology is particularly attractive due to its plasticity, allowing very thin depositions (10-20 μm) on a variety of needle shapes. Both sensors receive acoustic energy and convert it to an electrical signal. The precise location of the needle can then be estimated from this signal, to provide real-time feedback to the clinician. We evaluated the feasibility of this new technology using (i) a 4DOF robot in a water tank; (ii) extensive *ex vivo* experiments; and (iii) *in vivo* studies. Quantitative robotic studies indicated that the co-polymer is more robust and stable when compared to PZT. In quantitative experiments, the technology achieved a tracking accuracy of 0.14 ± 0.03mm, significantly superior to competing technologies. The technology also proved success in near-real clinical studies on tissue data. This sensor technology is non-disruptive of existing clinical workflows, highly accurate, and is cost-effective. Initial clinician feedback shows great potential for large scale clinical impact.

1 Introduction

Over 20 million needle interventions are performed every year around the world. Although 2D US is an attractive imaging modality due to its portability and cost-effectiveness, it has not yet become the gold standard for interventions. This is due to the fact that interventional needles, which are specular reflectors, cannot be consistently displayed under US (See Fig. 1). Depending on the insonifying angles, the ultrasound signal does not always backscatter to the transducer, resulting in poor needle visibility. Additionally, the needle deviates from the imaging plane during the procedure, further decreasing the clinician confidence. Such limitations hinder the usability of 2D US for interventions.

Various approaches have been proposed to tackle the "invisible tool" problem, which can generally be classified into two categories. The first category focuses

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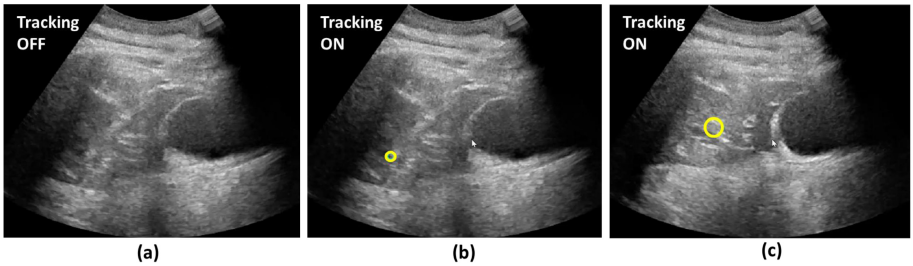

Fig. 1. Screenshots of an *in vivo* experiment where a co-polymer embedded needle was inserted into a swine liver. Fig. (a) is a regular 2D US image. Fig. (b) shows a yellow circle at the tracked sensor position at the needle tip, (c) shows an enlarged circle indicating that the needle tip is shifting out of the imaging plane.

on the enhancement of the interventional needle on the US image. This includes roughening the needle surface to make it more isotropically reflective [1] and modified ultrasound beamforming protocols to better detect straight specular reflectors [2]. These methods have either limited robustness, work only for shallow insertions, or still depend on needle incidence angles. The second category uses external tracking system such as electro-magnetic (EM), cameras or optical tracking to provide quantitative position information. These technologies not only increase the hardware cost but more importantly also make the clinical workflow very complex. This is the primary reason why despite being on the market for many years, clinicians have not adopted these techniques on a large scale.

We propose an alternative direction, where we can exploit the acoustic field of the US probe to achieve needle tracking. The idea is to mount a small US sensor to receive the acoustic signals in order to enhance needle visualization. A small review can be found in [7], documenting early attempts at needle tracking. One early solution saw success for needle tracking [8], but it unfortunately blocked the lumen of the tracked needle. The preliminary system though interesting, was not yet sufficiently stable or cheap, for large scale clinical use. We previously presented a real-time navigation system for hydrophone tracking in 3D US [7]. 3D US, though potent, is infeasible for everyday interventions due to its high cost. Although promising results were obtained for 3D, the technology did not extend to needles for use with 2D US.

Key Contributions. We propose a new sensor technology that works with interventional 2D US systems in a robust, real-time and cost-effective way. Since no change in current workflow is required, it is feasible for large scale clinical practice. The key contributions of the paper are:

- New sensor technology using co-polymer and PZT
- Real-time needle tracking system using 2D commercial US scanners
- Extensive quantitative validation with lab, *ex vivo* and *in vivo* experiments

2 Method

Principle of Tracking. A schematic description of the principle of tracking is provided in Fig. 2. US sensor is embedded at known location at the tip of the needle. A 2D US imaging probe emits ultrasound beams that regularly sample the tissue in an array of N_{AZ} beams in the azimuth direction. The temporal signals sensed by the receiver during the acquisition of one image are formatted in a 2D $N_{AZ} \times time$ "data matrix". In order to retrieve angular position information with a better resolution than that given by the spacing of the US beams,first a maximum intensity projection (MIP) of the "data matrix" over the time dimension is performed to yield a 1D (N_{AZ}) MIP vector on which a Gaussian fit is applied. The Gaussian center is used to estimate the angular coordinates of the receiver in the US coordinate system. The depth information is obtained by finding the time at which the maximum signal arrives at that angle and multiplying by the speed of sound. Thus with angle and depth information, the US sensor location of the needle is computed. Note that though the sound speed varies in human tissue, it does not translate into sensor localization error since such variation is also reflected on the US images. Additionally, by making use of the small elevation of the emitted US beams, the sensor can still be tracked when it is slightly shifted from the image plane.

Tracking Tool Design. Since the quality of the received signal from the US sensor determines the tracking quality, the sensor needs to have 1) high SNR; 2) omnidirectional reception; & 3) cost-effective integration for production. To meet these technical requirements, we designed two different types of sensors.

Co-polymer Sensor: Poly (vinylidene fluoride (VDF) - trifluoroethylene (TrFE)) co-polymer shows good piezoelectricity and is acoustically matched to biological tissue. It is inherently responsive over a wide range of frequencies and

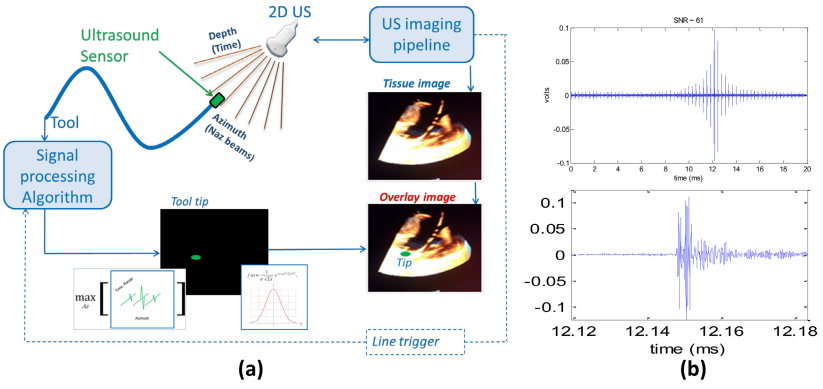

Fig. 2. (a) The scheme of tracking principle. (b) Top row: Signal received by the US sensor along time, from which the time point of the max signal can be found to estimate the AZ position and the depth. Bottom row: Zoomed in at the max signal position.

Fig. 3. Co-polymer sensor needle with (a) different sensor sizes and (b) different gauge sizes. (c): A needle prototype with PZT sensor embedded.

angles. Most importantly, because of its plasticity, it can be deposited in very thin layers on tools with varying shapes & sizes using simple processes (e.g. dipping or spray-coating) resulting in thicknesses as low as 10-20 μm. We successfully dip-coated the co-polymer in 15 $Cook^{\textcircled{R}}$ needles and wire it with 44G microcoaxial cable. Sensors width was as low as 100 μm, increasing to 0.3, 0.6 and 1 mm. The needles were as fine as 25G, increasing to larger 12G (Fig. 3(a)). Note that even on the ultra-fine 25G needle, we successfully dip-coated the polymer on the outside surface of the needle, leaving the inner lumen fully functional for injecting/draining fluids. The technology can be easily adapted to various needle shapes (e.g. Tuohy curve needle) with a low cost at a few cents, adding minimal costs to the needle. In contrast previous acoustic needles used a PVDF strip, which made the needle fabrication very expensive and blocked the lumen [8]. We refer the co-polymer fabrication process can be found in [4].

PZT Sensor: PZT sensors are also piezoelectric and can be used for acoustic tracking. We tested 10 different PZT sensors in varying frequencies and sizes, integrating them in medical devices. For ex. a 30MHz center frequency, axially poled PZT sensors is wired with a microcoaxial cable and threaded through the lumen of 16-gage tubing (Fig. 3(c)). An epoxy formula was injected in the tubing around the sensor to provide mechanical stability and acoustic coupling. The tubing was then cut with a diamond cutting wheel to provide a sharp tip. Due to a lack of space, the details of all sensors/devices are not presented here. Note that since PZTs are solid ceramic sensors (unlike co-polymers), they block the needle lumen and are more complicated to integrate.

System Setup. The proposed navigation system is implemented in Matlab on a HP Z820 workstation connected to a commercial Philips iU22 scanner (Philips Healthcare, Bothell, USA). 2D imaging C5-1 & S5-1 probes (2.5MHz pulse center frequency) are used. US data is streamed to the workstation at the same frame-rate of the scanner and is digitized at 100MHz, 12bits with a Razor data acquisition card (Gage Applied Technologies). Sensor acquisitions are synchronized using the scanner's 'beam out' and 'image out' trigger signals that mark the start of each US beam and each US image, respectively. The display frame rate of the tracking system was 5-10Hz.

3 Experiments and Results

3.1 Co-polymer vs PZT

Setup. We first performed a quantitative analysis characterizing the two proposed sensor technologies. A 4DOF robot in a water tank was used to characterize the sensitivity and omnidirectional reception of different variants of co-polymer and PZT sensors (Fig 4(a)).

Results. Fig. 4 shows that the peak beam SNR intensity of the sensors is stable across all angles. We also see that while the SNR values are comparable throughout different angles between co-polymer and PZT, the co-polymer shows greater stability (Fig. 5). This indicates better omnidirectional performance & robustness. Amongst all the compared sensors, the 0.3mm co-polymer had the best stability. Additionally, as discussed before, the co-polymer sensor is more suitable for needle fabrication due to its exibility and and low-cost. The PZT sensors, while being still feasible, ended up blocking the lumen (Fig. 3(b)), and were more complex to integrate. Therefore, we can conclude that the co-polymer sensor is a better choice for smart needles. The rest of the experiments are consequently based on co-polymer sensors.

3.2 Tracking Accuracy

Setup. The 4DOF robot, together with an S5-1 probe provided high accuracy ground truth for accuracy estimation [7]. A 20g needle was used with a 0.3mm co-polymer sensor. Six 19×17 data-grids have been scanned, where the first scan was used to calibrate the robot and sensor coordinate systems and the rest 5 scans were used to compute the Target Registration Error (TRE).

Results. Fig. 6 (b) and (c) show the mean and standard deviation of the TRE map. Overall, the accuracy is 0.14 ± 0.03mm across the region with no significant biases in any direction. This achieved accuracy is higher than most interventions' requirement.

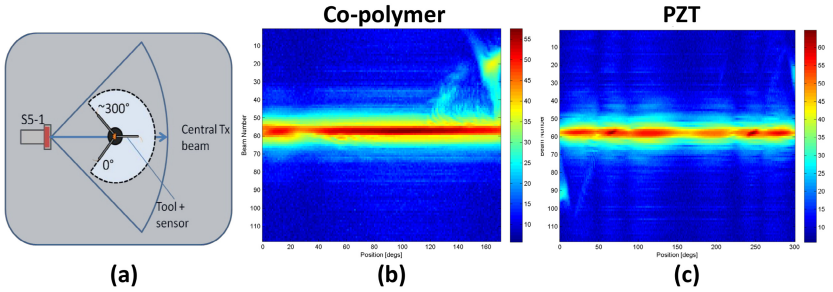

Fig. 4. (a) Rotational measurement setup, (b) and (c) show the SNR of a 0.3mm co-polymer sensor and a 1mm PZT sensor at different angles and beam positions

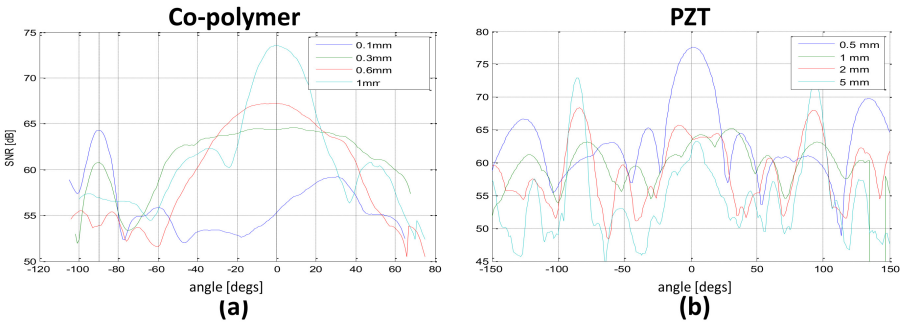

Fig. 5. The SNR of various sensor sizes of co-polymer and PZT at different angles

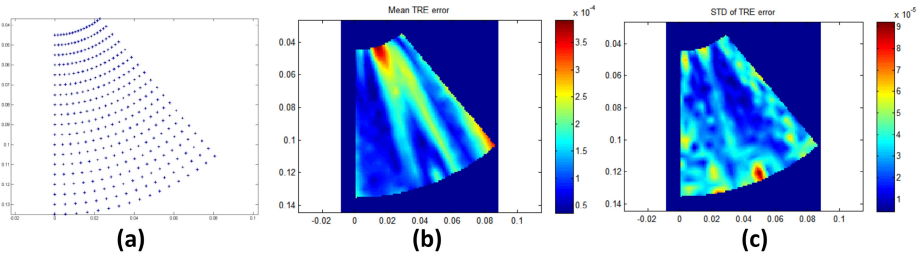

Fig. 6. (a) FOV of the robotic staging plane where data were collected 5 times at a 19×17 grid. (b) mean TRE map, (c) Standard deviation of the TRE map.

3.3 *Ex vivo* Validation

Setup. Two sets of ex-vivo experiments were conducted. A special phantom containing a pig kidney inside a gelatin phantom with graphite powder was designed for the first set of experiments, in which a needle with 0.3mm co-polymer sensor was inserted. The second set tracked the needle in an agar-block through human calf tissue (Fig. 8(a)).

Results. In the kidney phantom experiment (Fig. 7), the proposed tracking system is able to locate the sensor position at the needle tip, which is is invisible in the US image. In the "human" experiment (Fig. 8), the needle tip position is easily tracked through the calf muscle at the depth of more than 10cm.

3.4 *In vivo* Validation

Setup. Three pig experiments were performed by an experienced doctor who routinely performs needles biopsies and catheter drainage. Both qualitative and quantitative performance were obtained, including user feedback as well as the SNR from the sensor.

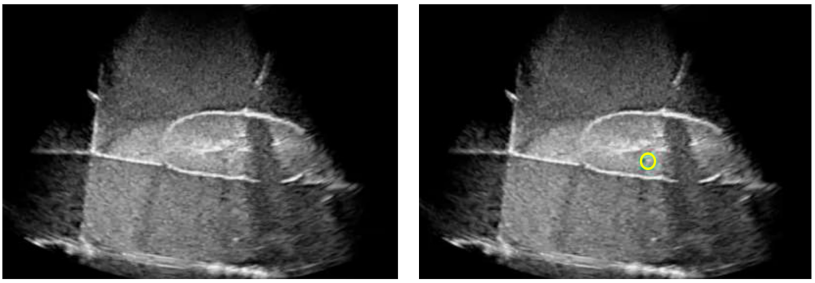

Fig. 7. Left: 2D US image showing an ultrasonically marked needle is inserted in a kidney phantom. (b) Yellow circle shows the tracked position of the needle tip.

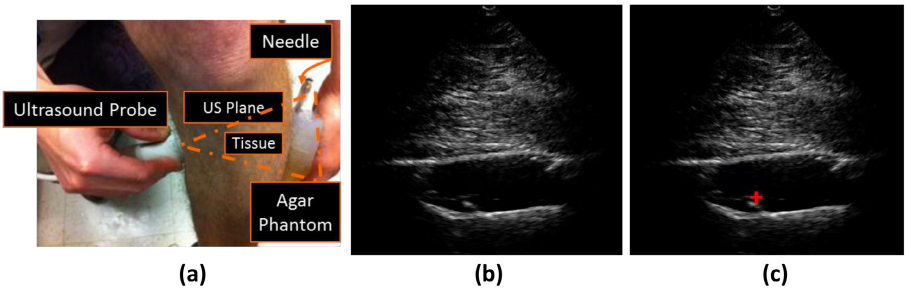

Fig. 8. "Human" validation. (a) The needle was inserted in an agar phantom which received the US signal from the probe through human calf muscle. The red cross in (c) indicates the needle tip location.

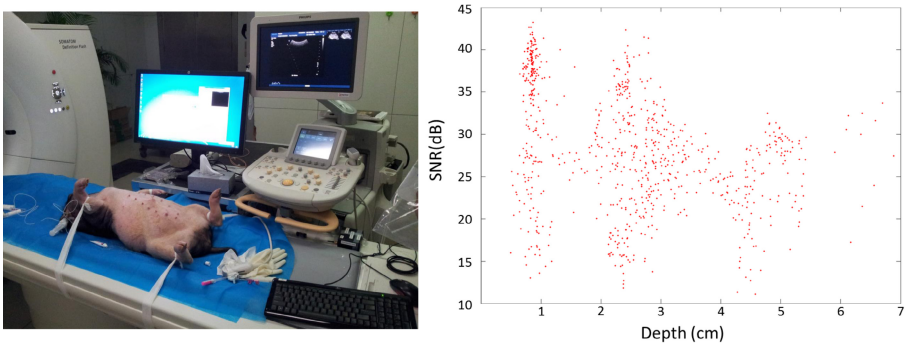

Fig. 9. Left: Pig experiment. Right: SNR values at different depths of tissue.

Results. As seen from Fig. 1, while the needle is almost invisible in the original US images, the position of the needle tip can be tracked using the proposed technology. In addition, the algorithm can inform the clinician when the needle tip is shifting out of the image plane using an enlarged circle Fig. 1(c). The measured SNR in the

experiment ranges from 10 to 45 dB with the majority over 25. Even in deep organ ($>$ 6cm), the SNR remains above 15 which is sufficient for accurate tracking (SNR higher than 6dB is considered good for tracking)[7].

4 Conclusion, Limitations and Future Work

In this paper we present and validate two new sensor prototypes for 2D US-guided interventions. In particular, we propose coating a special polymer on the tip of needles for navigation under US. The polymer is very versatile with very thin coatings possible on any shape/size of the needle. As far as the authors are aware, this is the first paper that investigates the use of co-polymer needles for US guided interventions. Quantitative robotic validation shows an in-plane accuracy of 0.14 ± 0.03mm in 2D US images. This is due to both the superior performance of the co-polymer sensor and the sub-mm wavelength emitted from the US beams. Extensive *ex vivo* and *in vivo* validation demonstrates the feasibility of the technology in near-clinical scenarios, showing high accuracies in tracking the exact tip of the needle compared to other techniques. The method is cheap, self-calibrates, & requires no changes to the workflow/scanner. Clinician feedback indicates high potential for adoption in US-guided interventions.

One potential drawback of the current technology is that the needle needs to be connected to the workstation through a cable. However, the clinicians find it not troublesome for the value that the technology provides. Another limitation is that the pre-clinical study lacks of an accuracy measurements. This remains a challenging task due to the absence of a high accuracy ground truth. A new protocol which makes use of an electrical circuit is being designed to generate quantitative results.

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