Ultrasound lumbar puncture guidance: a hands-free approach

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***Abstract*—During lumbar puncture for cerebrospinal fluid extraction, needle insertion is a highly skill-dependent task due to the visibility of the bone structure. In order to improve its success rate, we propose a patch-shaped device to guide needle insertion with ultrasound imaging without the need for a hand-held probe or a sonographer. We motorized a phased ultrasound array on a linear rail in plane with a needle guide. The beam steering capabilities of the phased array insonificate the bone in multiple angles and visualize details of bone that are shadowed under a linear array and also improves the visibility of the needle. We apply adaptive compounding to the B-mode images collected along the rail based on bone probability to improve the visibility of the bone structure. We triangulate the tip of the needle through an active photoacoustic ultrasound source located in the tip of the needle. Result of the test bed device showed improved visibility of the bone structure surrounding the needle insertion path as well as the needle in deep tissue.**

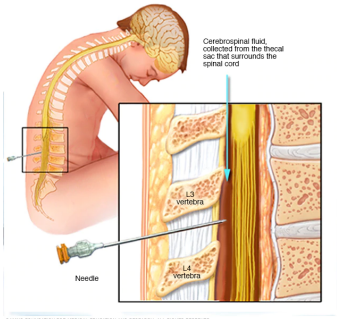


Fig. 1: Illustration of the operational set-up of a typical lumbar puncture. [4]

I. Introduction

Lumbar Puncture, also known as a Spinal Tap, is a medical procedure whereby a needle is inserted into the spinal canal most usually to collect the cerebrospinal fluid (CSF) (Figure 1). Acquiring CSF is valuable for several diagnostic procedures used to identify diseases such as meningitis and hemorrhage. There are approximately 400,000 new procedures performed each year, but an approximately 23.3% fail[1] [2]. This is most common with severely overweight patients, who are reported to have a 50% higher rate of complications[3]. The respective commercial solutions used to assist in this "blind" procedure each have deficiencies in some regard or aspect that prevents them from fully overcoming the challenges inherent to this procedure. For example, some rely on image guidance techniques that require registration between two or more imaging systems, while others produce an image with a single system (ultrasound) but do not allow for real-time visualization of the needle as it is inserted because otherwise, the probe remains in the way. Other more have physical constraints that make it challenging to interface with the needle.

Previous research address this problem by mimicking an ultrasound array with a single ultrasound element on a tracked needle[5][6]. However, the single element is unable to continue to produce image during insertion due to the constraint that the needle has limited motion in tissue.

To minimize the number of insertion trials and insert through the desired direction, what lumbar puncture would benefit most by is an accurate, real-time imaging system that combines both needle tip tracking and high-quality images of the spinal processes. In addition to improving patient well-being and comfort, such a solution has the potential to save hospital administration hundreds of millions of dollars per year by reducing the complications stemming from blind lumbar punctures.

Our goal is to deliver high-quality ultrasound images in real-time in order to help guide physicians to deep targets. Strides have already been made in investigating the feasibility of single-element needle tracking, in studies whereby an ultrasound transmitter was embedded into the tip of a needle. For this project, we propose a different use of an ultrasound system that would address the problems associated with this procedure. We hypothesize the use of a moving phased-array ultrasound transducer can be applied for better image quality and that advanced imaging techniques such as photoacoustic sensing can be used to track the position of the needle for reliable and accurate guidance of needle placement into the spinal canal.

Medical ultrasound is a good technique for imaging the deep bone structures of the body, but it has several limitations that must be addressed before it can be successfully implemented to accomplish this goal. Namely, traditional ultrasound configurations come with a probe encased in a plastic housing attached to the processing machine, which can become expensive. Additionally, the common configuration of ultrasound probes is several elements arranged in a linear ray which fire signals directly ahead of them in space. The creates a problem in which the resulting image shows shadows indicating where a feature could have been obscured by a contoured surface reflecting the ultrasound waves. Additionally, in a linear configuration, scattering results in a decrease in

image resolution and contrast because the information from the reflected waves is lost as they bounce off the bone surface and are refracted through media.

II. Methods

A. Moving phased array

By moving a small phased array in the lateral direction, we can insonificate the bone in different angles and visualize details of bone that are usually in the shadow under a linear array (Figure 2). This achieves better-than-linear-array image quality especially once the data from the probe is compound to create the image from multiple linear positions of the phased array. The phased array generates redundant insonification angles, which helps visualize steep walls and shadowed areas. This also improves the visibility of the needle in deep tissue. As the depth of the needle increase, the phased array moves away from the needle and steers the beam towards the needle to image it at a more direct angle.

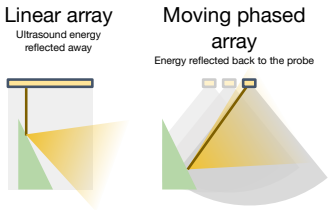


Fig. 2: Comparison of imaging a steep wall with a linear array and a moving phased array

We used an ATL P7-4 with 64 elements for proof-of-concept. We first moved the probe with a modified 3D printer as a three degrees-of-freedom stage. Then we constructed a custom patch-shaped testbed with a linear rail (Figure 3).

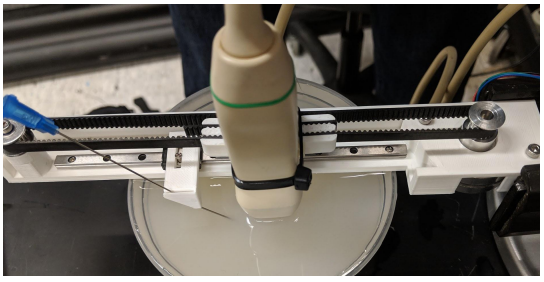


Fig. 3: Test bed for moving a ultrasound phased array

We acquire the envelope detected post-beamform image in polar coordinate from the phased array probe with Verasonics. We program the Verasonics to transmit the content of the mage buffer to an HTTP server on the computer running motor control, then the image data is subsequently tagged with motor position and saved for further processing.

B. Adaptive compounding

We compounded multiple images from different locations filtered by bone probability to increase the SNR of the bone image. The method is based on Foroughi et. al.'s work for bone segmentation [7]. We used the polar coordinate image as input because the shadow of the bone is cast in the radial direction for beams not in the axial direction. We Gaussian blurred the image, then estimated bone probability based on intensity and shadow. The intensity map is generated from adding Laplacian of Gaussian of the blurred image to the blurred image, which detects sharp bright bone features. The shadow-based bone probability is estimated by looking for bright pixels with dark pixels underneath. Figure 4 shows the bone probability map generated from a single B-mode frame.

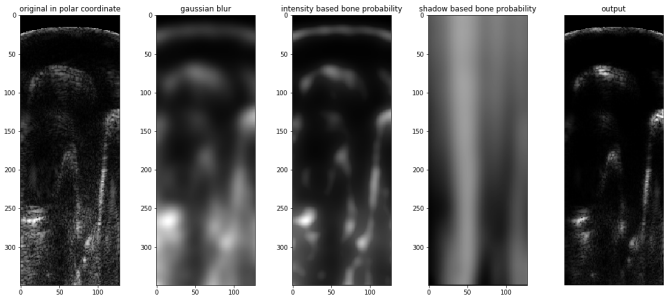


Fig. 4: Intermediate results for the bone probability map

C. Active needle localization

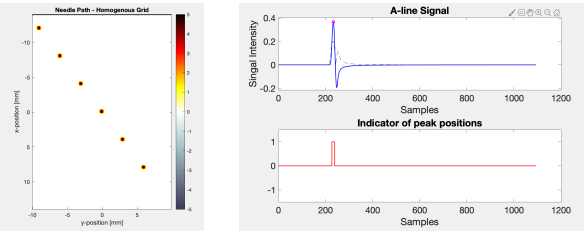
A simple and direct needle tip tracker is necessary for this device to work without requiring complex registration steps and additional systems ultrasonic sensors. On their own, needles reproduce poorly in most ultrasound images and also cause shadowing over the features beneath them. This causes a problem because the addition of equipment causes substantial disruption to the workflow of the lumbar puncture procedure, meaning that our solution must remain as unobtrusive as possible.

Our concept for handling this problem is based on previous initiatives to embed the needle with some form of acoustic transmitter[8]. With a PZT element, we can perform a traditional acoustic localization, or by adding an LED or fiber optic cable to the needle tip, we hypothesize that tracking can be possible by utilizing the photoacoustic effect. The "photoacoustic effect" refers to the generation of acoustic waves by the absorption of electromagnetic energy, such as the light from a pulsed laser beam. Active point sources are generally straightforward to the segment from ultrasound images because they either have a higher intensity than the background or the ultrasound system can be configured such that there is no acoustic transmission and hence no background. By pairing such a sensor at the tip of a needle with an ultrasound transducer, it would be possible to build a system that can both see and hear the needle tip. By spatially combining the geometrical loci from the two sensors using an ultrasound calibration process, using a similar method to how geographical position is determined from GPS, one can uniquely determine the location of the piezoelectric sensor. This data can be acquired using the very same ultrasound

transducer performing the image reconstruction of the spine. Since phased arrays have very good elevational focusing, meaning that they are bad at imaging out-of-plane-targets, our rail needs to be designed to hold the needle within a 1 DoF holster that is coplanar with the sensors. All this holster would do is change the angle at which the needle could be inserted.

In order to simulate the active needle tracking, a test environment for photoacoustics was created in k-Wave Ultrasound software. The ultimate aim of utilizing k-Wave was to measure the performance of the localization algorithm in a heterogeneous algorithm. In order to acquire a sense of the feasibility, baseline data was collected in a simulation using a homogeneous medium for comparison with data collected from those same points in a medium now made heterogeneous.

In simulation (Figure 5), within a homogenous grid of size 28 mm by 20 mm, six point sources of magnitude 5 Pa were one at a time made to pulse a signal to be read by two elements at known locations at the top of the grid. The time of flight for the peak pressure value was extracted from the sensor data and the distance of the target from the elements were calculated using the speed of sound, previously set to the default 1500 m/s.



(a) Simulated needle tip path showing every position from which time of flight data was collected

(b) Example of one of the signals detected by the sensor and the max point

Fig. 5: Needle tip localization simulation set up

III. Custom Electronics

Synchronizing motion of the moving aperture with ultrasound imaging is critical for mechanical ultrasound probe. While clinical ultrasound machines and Verasonics support wobbler control, we need more flexibility in sensing and actuation. We need to build electronics to transmit and receive with a small (32-64) element phased array with full programmability.

On the receive side, we chose AD9671 ultrasound AFE (Analog Devices, USA), which includes 8 channels of low noise amplifier, a variable gain amplifier, filters, and 65 Msps 14-bit ADC on the same package. The transmit side has an 8-channel pulser and transmit/receive switch.

A Zynq FPGA SoC (Xilinx, USA) is used to interface with the transmitter, receiver, motor driver, and sensors.

Four boards can be synchronized to interface with a 32 element array. For systems without a fast frame rate requirement, multiplexing is a common technique for low-cost ultrasound machines [9]. Each channel transmits individually to allow phasing, and the receiving is multiplexed.

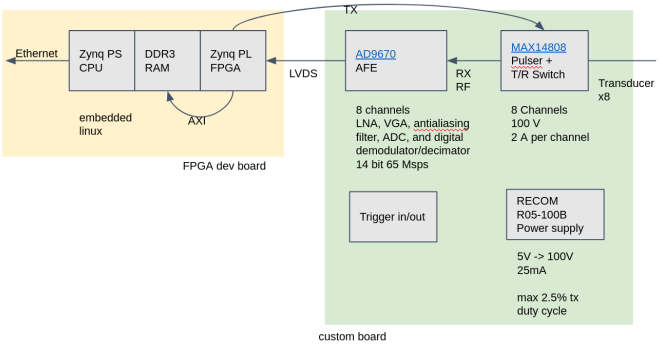


Fig. 6: Architecture of the ultrasound interface

IV. Results

V. Moving Phase Array

Figure 7 shows B-mode images of a spine phantom (PLA plastic in agar) under a moving phased array and an Ultrasonix L14-5 linear array. Moving phased array visualizes the steep and negative-slope structure between the spinous processes, which is critical to determine the angle to insert the needle. The same feature is shadowed under the linear array.

Figure 8 shows that the moving phased array is capable of imaging a needle in deep tissue. The 22 gauge needle is inserted 70 mm into a gelatin phantom. The entire length of inserted needle is visible under the moving phased array.

VI. Active Needle Localization

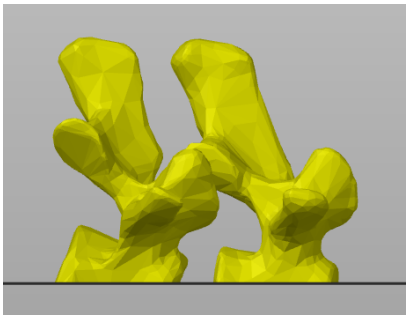
The grid was defined on a coordinate plane of 260 by 200 points spaced with distance between of 0.1 mm. The locations of the two sensor elements were placed at (1,1) and (1, 200).

Once a baseline was established the process was repeated but for a grid with three equally separated layers of media with their own density and speed of sound characteristics. Layer 1 had density and speed of $1000\,\mathrm{kg\,m^{-3}}$ and $1500\,\mathrm{m\,s^{-1}}$. Layer 2 had density and speed of $1100\,\mathrm{kg\,m^{-3}}$ and $1540\,\mathrm{m\,s^{-1}}$. Layer 3 had density and speed of $1200\,\mathrm{kg\,m^{-3}}$ and $1580\,\mathrm{m\,s^{-1}}$. These parameter correspond roughly to the tissue features of the lower lumbar region. The performance (Table I, Table II) had sub-millimeter error, with mean of 0.361 mm, max of 0.699 mm, and minimum of 0.207 mm. As the needle went deeper, the error increased, which is predicted to be a trend as the depth increases. Further testing is required to determine whether this approximation method using time of flight is appropriate for use.

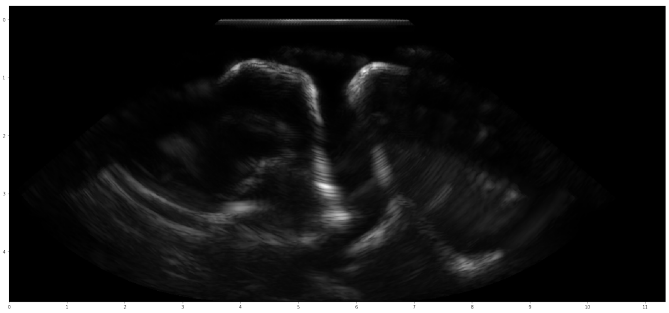
VII. Discussion

The phantom study shows that a moving phased array can reveal the important structure of the spine for lumbar puncture that is hidden under the linear array.

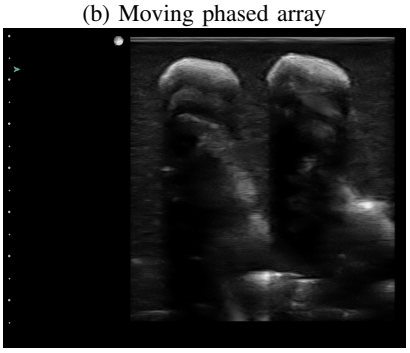
The image quality is sensitive to positioning error. A coherence-based compounding method can mitigate this issue and allows cheaper actuation hardware.



(a) Phantom



(b) Moving phased array



(c) Linear array

Fig. 7: Comparison of image acquired with our moving phased array and Ultrasonix L14-5 linear array

The phased array on linear rail test bed cannot be directly used to guide needle insertion into the spine because the bulky clinical ultrasound probe is unable to reach close to the needle to avoid shadowing of the bone and the needle after insertion of the needle. The visibility of the needle is sensitive to the alignment of the needle guide, which is difficult to adjust.

To address these issues, we believe a custom phased array is necessary. Instead of placing the phased array co-plane with the needle, a better alternative is to place two independently moving phased arrays parallel to the needle plane. The phased arrays should be narrow in elevational direction to increase out-of-plane sensitivity. This configuration allows the bone structure to be visible during needle insertion.

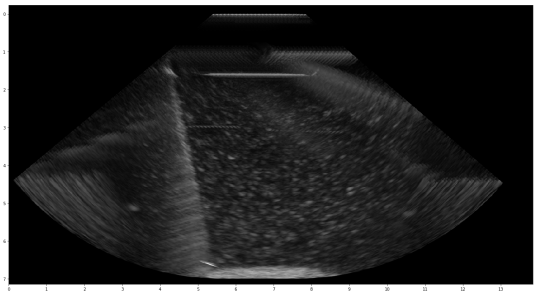


Fig. 8: B-mode compound image of a needle in gelatin under a moving phased array.

The solution to this problem of reconstruction within an unknown medium in needle localization requires an optimization equation not solved during the course of this project. It would require more computational assessment in simulation, but it is predicted to go much slower than this triangulation method. It may be worth the time to continue to treat the lower lumbar region as a homogeneous space.

VIII. Conclusion

We present the concept of an ultrasound patch that enables hands-free image guidance for lumbar puncture. We show that a moving phased array can reveal the important structure of the spine for lumbar puncture that is hidden under the linear array.

IX. Acknowledgements

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	X	Y	Presumed X	Presumed Y	Error (0.1 mm)
Point 1	20	10	17.48	11.42	2.89
Point 2	60	40	57.96	40.99	2.27
Point 3	100	70	98.59	70.32	1.45
Point 4	140	100	138.59	99.99	1.41
Point 5	180	130	178.57	130.2	1.44
Point 6	220	160	218.21	160.36	1.82

TABLE I: Homogeneous Needle Tip Tracking

	X	Y	Presumed X	Presumed Y	Error (0.1 mm)
Point 1	20	10	17.71	11.34	2.65
Point 2	60	40	58.02	40.72	2.1
Point 3	100	70	97.99	70.53	2.07
Point 4	140	100	136.77	100.02	3.23
Point 5	180	130	175.43	129.17	4.64
Point 6	220	160	213.33	157.92	6.99

TABLE II: Heterogeneous Needle Tip Tracking

Coulter Translational Foundation, and the Maryland Innovation Initiative, and the Steven Alexandra Cohen Foundation for their support throughout this project.

X. Management Summary

A. Project Aims

See abstract.

B. Contributions

Keshuai did the moving phased array, adaptive compounding, and ultrasound electronics.

Christian did simulations and active needle localization.

C. Plans vs. Accomplishments

Minimum

✔ image+code+doc - A pair of a B-mode image comparing the image quality of the linear scan and our new phased array synthetic aperture scan on a spine phantom

Expected

✔ video+code+doc - A video showing inserting a needle into tissue phantom (no bones) while maintaining its visibility all time

✔ code+doc - Adaptive compounding algorithm that maximizes information from vertebrae

Maximum

- video - A video showing needle insertion in spine phantom with hands-free ultrasound guidance

✔ code+doc - Acoustic needle localization simulation

- image - Deep tissue photoacoustic imaging

D. Potential Next Steps

See section VII.

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