Tracked 3D Ultrasound in Radio-Frequency Liver Ablation

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ABSTRACT

Recent studies have shown that radio frequency (RF) ablation is a simple, safe and potentially effective treatment for selected patients with liver metastases. Despite all recent therapeutic advancements, however, intra-procedural target localization and precise and consistent placement of the tissue ablator device are still unsolved problems. Various imaging modalities, including ultrasound (US) and computed tomography (CT) have been tried as guidance modalities. Transcutaneous US imaging, due to its real-time nature, may be beneficial in many cases, but unfortunately, fails to adequately visualize the tumor in many cases. Intraoperative or laparoscopic US, on the other hand, provides improved visualization and target imaging. This paper describes a system for computer-assisted RF ablation of liver tumors, combining navigational tracking of a conventional imaging ultrasound probe to produce 3D ultrasound imaging with a tracked RF ablation device supported by a passive mechanical arm and spatially registered to the ultrasound volume.

Keywords: liver ablation, RF ablation, 3D ultrasound, calibration, tracking, registration, mechanical linkage

1. INTRODUCTION

In spite of recent advances in cancer therapy, treatment of primary and metastatic tumors of the liver remains a significant challenge to the health care community worldwide. Hepatocellular carcinoma is one of the most common malignancies, accounting for more than 1 million cases per year worldwide. Metastatic disease to the liver, primarily from colorectal cancer, is the most frequent hepatic malignancy in the United States. The principal potentially curative treatment for patients with liver cancer is resection, which results in a 5-year survival rate between 25 and 55%. Unfortunately, most patients with primary and secondary liver cancer are not candidates for resection, due primarily to tumor location, number, or underlying liver disease. For this reason, an increasing interest has been focused on ablative approaches for the treatment of unresectable liver tumors. Radiofrequency (RF) ablation is one of the most promising minimally invasive techniques (Figure 1) for the treatment of unresectable hepatic tumors.[1] Recent studies have reported favorable survival rates and excellent rates of local control after such ablation, especially in patients with hepatocellular carcinoma[2,3].

One common feature of current ablative methodology is the necessity for precise placement of the probe tip within the volumetric center of the tumor in order to achieve adequate destruction. The tumor and a 1-cm zone of surrounding normal parenchyma can then be ablated. RF ablation can be performed percutaneously, laparoscopically, or during open laparotomy. When performed operatively, tumors are identified by preoperative imaging, primarily CT and MR, and then localized by intraoperative ultrasonography (IOUS). Current methodology requires manual IOUS in conjunction with free hand positioning of the ablation probe under ultrasound guidance (Figure 1).

The major limitation of ablative approaches is the lack of accuracy in probe tip placement within the center of the tumor. Whether performed percutaneously or operatively, the current method of freehand simultaneous positioning and tracking of both the US and RF probes also make this approach cumbersome[4]. This is particularly the case when the tumor size is larger and multiple overlapping zones of ablation are required for therapy[5]. Inaccurate targeting, as opposed to inadequate energy deposition or thermal convection, is probably the primary reason that tumors are undertreated. IOUS provides excellent visualization of tumors and provides guidance for RF probe placement, but its 2D-nature and dependence on the sonographer's skills limit its effectiveness.

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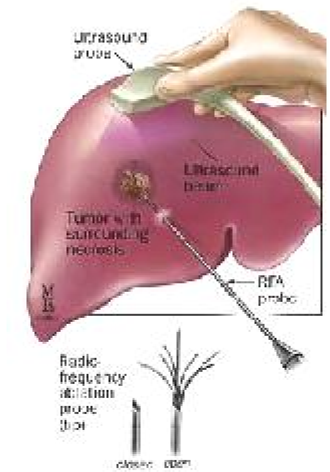


Figure 1: Radiofrequency ablation of a liver tumor

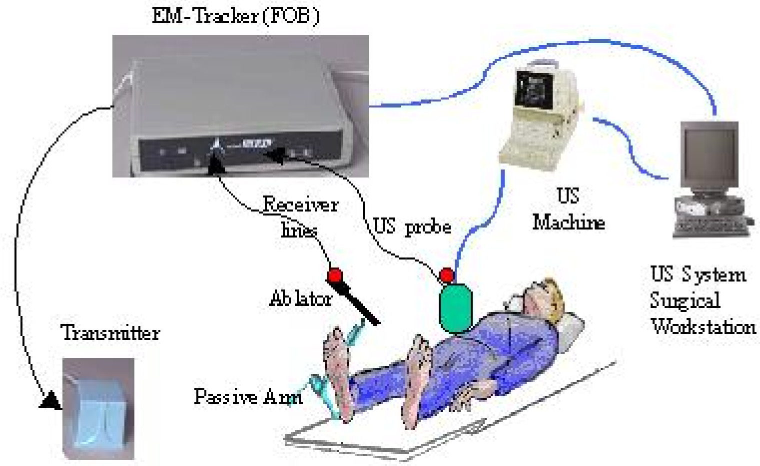


Figure 2: Components of the proposed system with passive arm and freehand US

Ultrasound-guided liver ablation is an ideal setting to make use of tracked ultrasound technologies. Improved real-time guidance for planning, delivering, and monitoring the ablative therapy would provide the missing tool needed to allow accurate and effective application of this promising therapy. It is believed that image-guided tracked ablation can ultimately make these procedures significantly more accurate, more effective, safer, and less expensive. This project demonstrates that the efficacy of liver tumor ablation could be significantly enhanced with the use of tracked 3D ultrasound and a tracked RF ablator attached to an adjustable rigid arm.

Spatial tracking has become a straightforward task in computer-assisted surgery during the last few years. Several thousand tracking systems are currently being used in a wide variety of medical applications in the U.S. and worldwide, typically utilizing some form of optical or electromagnetic technique. A few applications of encoded mechanical linkages are also known, such as the BAT® system by Nomos Corp. (http://www.nomos.com) for US-guided target setup for external beam radiotherapy. A new frameless stereotactic system for radiotherapy guidance is presently being developed by CMS-IGD. The overall engineering literature of spatial tracking is so extensive that we do not cite specific works here. Among a number of research systems, a few commercial solutions also exist for combining 2D ultrasound with tracked surgical instruments: Sononav® by Medtronic Sofamor-Danek (http://www.sofamordanek.com) and UltraGuide® (http://www.ultraguideinc.com) both project the trajectory of a tracked needle-like instrument on real-time 2D ultrasound. In the academic domain, the Galloway-group at Vanderbilt University has made sustained contribution to the subject[10], among several others whom cannot be quoted here individually due to lack of printing space.

Currently, no commercial surgical navigation system applies volumetric assembly and visual analysis of spatially registered 2D ultrasound images. Such systems only exist in the academic domain[5,6,7,8]. Alternatively, one could achieve 3D US volume by moving an encoded device (robot) carrying a 2D US probe along some known or predefined trajectory[11]. We also present a companion paper at this conference about our initial findings in robot-assisted ultrasonography.

2. MATERIALS AND METHODS

2.1 System overview

The key component in the proposed solution is tracking both US and RF probes for better accuracy in targeting the tumor, reducing the time required for the operation, and minimizing the dependency of the surgical experience. This proposed solution would be applicable for all three different liver ablation approaches: percuteanous, laparascopic, and intraoperative. However, in this project, we test the concept in open intra-operative setting. In our proof-of-concept study, an adjustable passive arm (with 6 degrees-of-freedom or DOF) holds the RF ablator and a tracked freehand B-mode ultrasound probe provides data for computer-reconstructed 3D ultrasound (Figure 2).

We use the following component: (1) a commercial B-mode ultrasound scanner (Aloka SSD1400, Japan), (2) a pulsed magnetic field position and orientation measurement system (Flock of Birds, model 6D FOB, Ascension Technology Corp., Burlington, VT, USA); (3) an RF ablation probe (XL, RITA Medical Inc.), adjustable mechanical support arm developed at Johns Hopkins, and (4) graphic workstation for capturing ultrasound (US) images, recording data from the FOB tracker, building the 3D US volume and performing the analysis, planning and monitoring of the delivered treatment.. The particular scanhead used in this study is a 3.5 MHz curved array transducer. Full-frame ultrasound images from the video output of the ultrasound scanner are digitized by a frame grabber board attached to the graphic workstation (Matrox, MeteorII).

2.2 Tracked RF probe with the 6-DOF passive arm

The term “radio frequency” here refers to the alternating electric current that oscillates in the range of high frequency (200-1200 KHz), rather than to emitted wave. Essentially, a generator, a large dispersive electrode (grounding pad), a patient, and a needle electrode are assembled in a closed-loop circuit. Thus, an alternating electric field is created within the tissue in order to generate heat, which in turn causes coagulation of lipids, and finally cell death. The success of this process depends on the accuracy of planning, targeting, delivering, and monitoring thermal energy. The factors affect one another, and this fact makes intra-operative monitoring, re-assessment, and optimization particularly important. The aim of this project is to improving on targeting and planning accuracy, which has been reported as the primary technical problem and procedural reason for failure of treatment.

The RF probe is supported by an adjustable rigid mechanical structure comprising the following components (Figure 3): (1) A 6-DOF passive arm provides moving the needle to the point of insertion. (2) A 5-bar remote center of motion (RCM) linkage, attached to the passive arm, provides two independent rotations for the needle to pivot around the point of insertion. (3) A passive un-encoded linear stage holds the ablator needle and provides insertion into the liver. (4) Friction brakes hold the rotation and linear motion stages in place. (5) A Lego attachment holds interchangeably the Flock of Bird and Polaris sensors. (6) A plastic extension for the RF probe that allows for accurate pivoting calibration around both ends of the needle.

Figure 3: Tracked passive arm

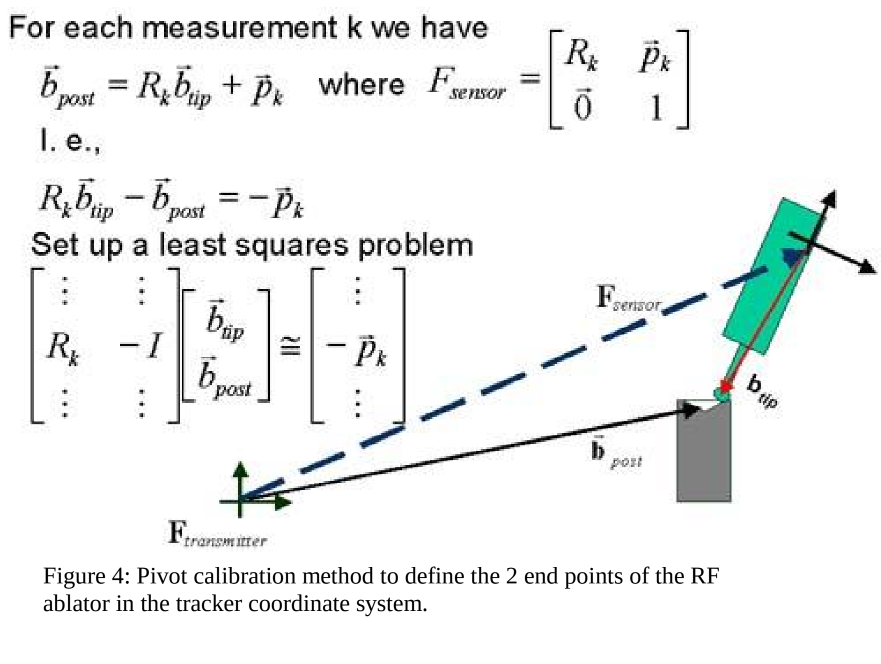


Figure 4: Pivot calibration method to define the 2 end points of the RF ablator in the tracker coordinate system.

Calibrating the RF device means being able to tell the location of the ablator end points with respect to the tracker frame or sensor frame. The location and orientation (F_{sensor}) of the FOB's sensor with respect to the base frame or FOB's transmitter ($F_{transmitter}$) are captured real-time. In order to get the unknown position vectors for the two end points, a Pivot Calibration method was used as shown in Figure 4.

2.3 Tracked 3D ultrasound

The generic tracked US system utilizes a scanning method (freehand or robotically guided) and some sort of tracking device, typically optical or electromagnetic. Figure 5 depicts the basic idea behind such generic tracked US system.

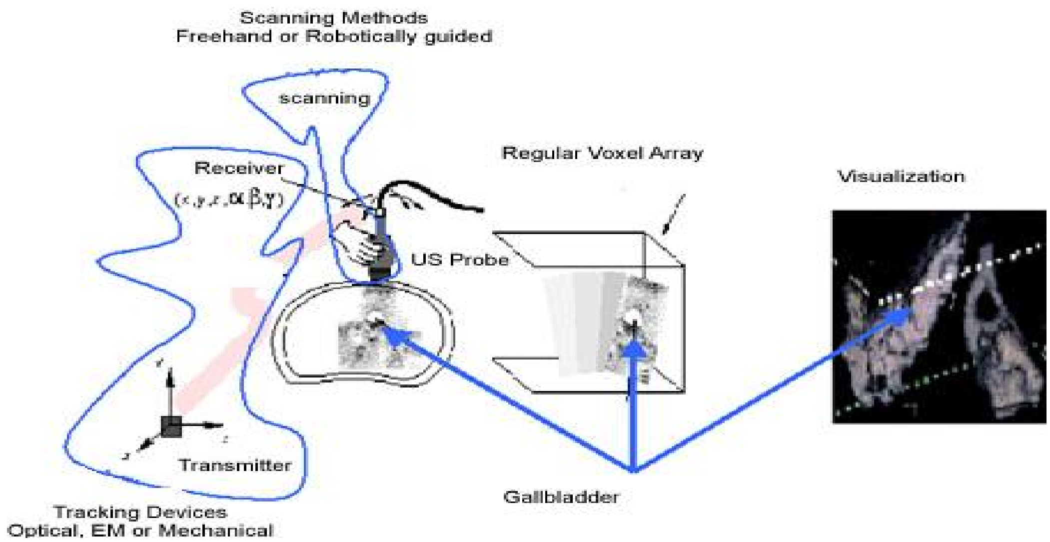


Figure 5. Generic Tracked US System

The following modules represent the main components of the Tracked US system:

- *Acquisition and Synchronization Module*: Capturing the two streams of US data and tracker information in a synchronized way along with setting the various parameters of both units.
- *Reconstruction Module*: Converting the scattered arrays of images into a structured 3D voxel array for the scanned organ by selecting the appropriate algorithm (Shepard, Compounding etc…).
- *Processing and Visualization*: Enhancing the US images by applying smoothing, anisotropic diffusion and/or morphological operators.
- *Calibration Module*: Calibrating the US beam, i.e. to determine the transformation between the US image frame and the tracker coordinate system or the world coordinate system.

There are two calibration processes that are central to the accuracy of the 3D ultrasound System (3DUS): temporal calibration and spatial calibration, both discussed below.

2.3.1 Temporal calibration

Temporal calibration involves estimating the latencies of the position sensing device and the ultrasound machine. The goal is to synchronize two processes essential to 3D ultrasound: one process acquires US images from the US machine and places them in reserved memory. The other acquires readings from the position sensor and stacks them in memory. Synchronization calibration is performed once per hardware/software configuration. A series of experiments of reading the two signals of US capture and position sensor reading on an oscilloscope helped in designing optimal and minimal setting for these two synchronized processes as well as determining the inner delay of the measurement cycle of the position sensor device.

A fundamentally important feature of this calibration process is the use of a real-time monitoring and debugging tool. When either a US capture or tracker reading occurs, a signal is generated to indicate the starting and ending, through the use of a parallel port. The time taken to output to the parallel port by the help of low level commands is on the order of a few processor's cycles. Reading these two signals during normal operation mode on a real-time digital oscilloscope assists in estimating the latency and determining optimal design for this real-time module.

2.3.2 Spatial calibration

The position sensor indicates the position of the small receiver mounted on the probe, with respect to a fixed tracker frame. What is really missing is the position of each B-scan pixel with respect to the tracker frame. The spatial calibration process involves deducing the 6-DOF rigid body transformation between the position sensor and the corner of the B-scan plane, and also the x and y dimensions of the pixels (in mm/pixel). This is a fairly complex process is depicted in Figure 6. Spatial calibration is usually performed by scanning and reconstructing some known object and then, using the discrepancy between the reconstructed shape and the known shape, re-estimating the calibration parameters. The 3DUS system employed uses a novel calibration technique discussed separately in a companion paper[15].

The mathematical framework for the calibration pipeline is shown in Figure 7. The idea behind the calibration process is to calculate very accurately (i.e. as accurately as the tracker allows) the frame of transformation between the US image coordinate system and the sensor coordinate system. Knowing this transformation matrix, one can place every pixel in the image coordinate system into the construction coordinate system (which could be the operation room coordinate or the patient coordinate system).

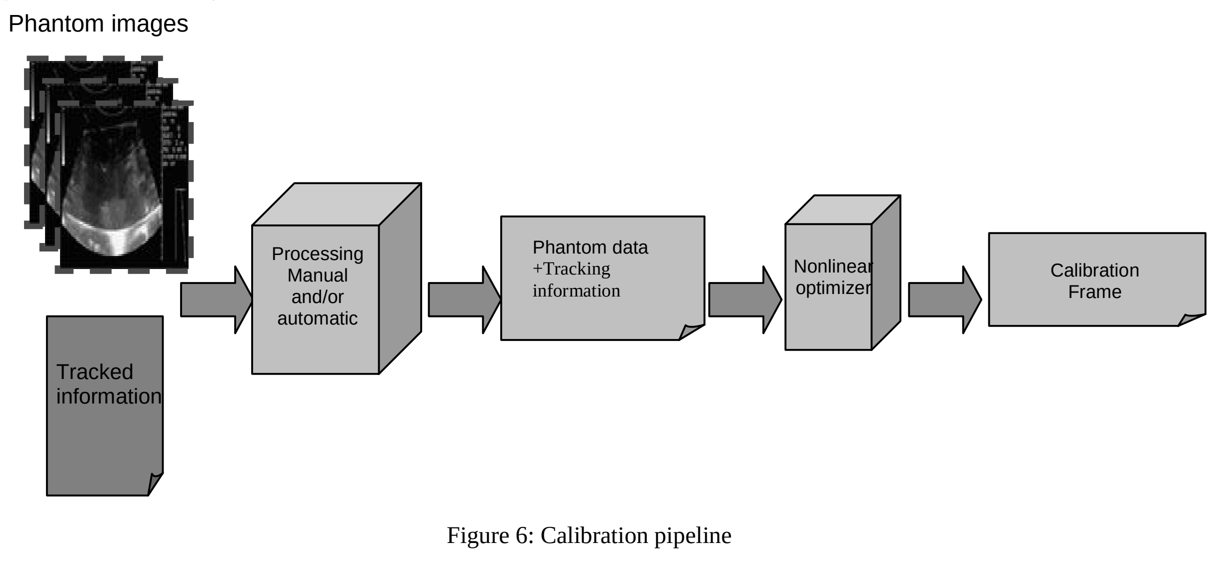
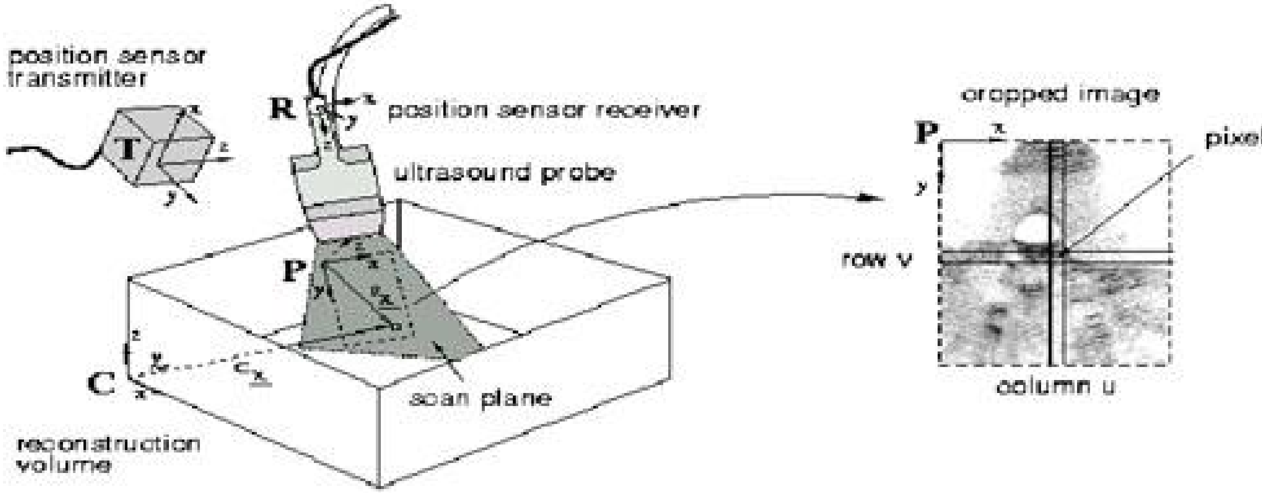


Figure 6: Calibration pipeline



$$ {}^{C}\mathbf{x} = {}^{C}\mathbf{T}_{T}\ {}^{T}\mathbf{T}_{R}\ {}^{R}\mathbf{T}_{P}\ {}^{P}\mathbf{x} $$

$$ {}^{P}\mathbf{x} = \begin{pmatrix} s_x u \\ s_y v \\ 0 \\ 1 \end{pmatrix} $$

Figure 7: The relation between the image coordinate P to the receiver R, from the R to the Transmitter and from T to the Construction volume C and notice the pixel ratio multiplied by u and v, Sx and Sy repectively. (Upper figure Courtesy of R. Prager)[16]

2.4 Planning and visualization software

For graphical user interface, we adapted the 3D Slicer medical data visualization package[13]. This is a public domain open source system primarily developed by the Surgical Planning Laboratory at the Brigham and Women Hospital (http://www.slicer.org), with sustained contribution from our group. The main strength that we added to Slicer is a generic 3D US processing module, providing synchronized real-time capture of 2D US data and position information, and robust assembly of a spatially registered 3D volume from sparse and irregular 2D information. The software also provides a customizable real-time overlay display of the US volume, RF ablator tip with its current ablating range, and the current 2D ultrasound slice, as shown in Figure 8. Tumor coverage can be planned with single-spot and overlapping multiple ablations. Beside the aforementioned visual overlay, are other visual and metric tools help the user in placing the ablator. The system reposts the current insertion depth that can be double-checked against the ruler on the passive unencoded insertion stage, it shows the distance between the actual and planned RF ablator, and also provides coordinate systems and landmarks to ease navigation in the virtual space.

3. EXPERIMENTS AND RESULTS

We report the results from initial experiments using mechanical phantoms to verify and determine the accuracy of the system. Two different phantoms were used. The first phantom is 1.0 cm rubber ball immersed in a transparent water tank. 2D US images incorporating the region of interest were captured in order to build the 3D volume, with methodology discussed earlier in Section 2.3. The reconstructed 3D volume was then combined with the real-time tracked RF ablator in the dual 3D view visualization windows to help in planning and targeting, as discussed in Section 2.4. After planning, we moved the ablator to target. The method of verification was visual observation in the tank under water, on graphical display, and in live 2D US. We evaluated whether (1) the ablator correctly touched the rubber ball in the tank, (2) the system correctly overlaid the rubber ball in the reconstructed 3D US with the real-time picture of the tracked ablator, (3) the rubber ball and ablator show up correctly in live 2D US images. Potential target motion was assessed by finding the dislocation of the image of the rubber ball between the pre-insertion 3D US volume and the live 2D image. As no target motion was noted, the experiment was indeed appropriate to assess absolute accuracy of our system. (None: in the actual needle placement, we set the insertion depth 0.5 cm shorter, in order to avoid dislocating the 1.0 cm rubber ball.) The accuracy of insertion depth with respect to the surface of the rubber ball was on average 4-5mm. On all occasions, the rubber ball was touched with sufficient angular accuracy. This only be estimated in oblique 2D ultrasound, which it appeared to be clinically sufficient.

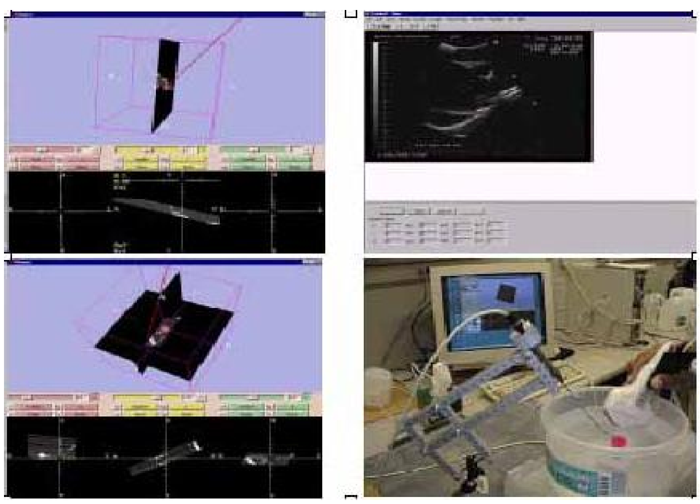


Figure 8: Rubber ball phantom experiment. LR: experimental setup. L: path planning with 2D and 3D US data. UR: confirmatory 2D US snapshot.

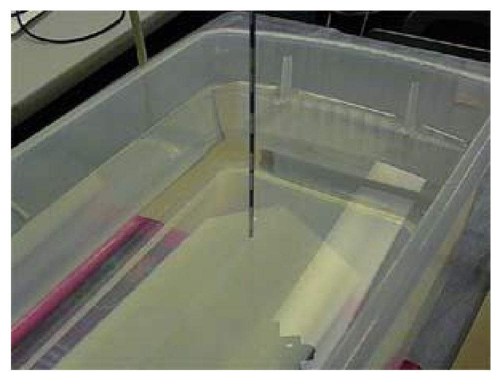


Figure 9: Plastic triangle phantom used to verify the accuracy of placing at three pre-selected landmarks.

Results from these experiments show the ability of the method to accurately reach a target and adopt proper planning accordingly. Both 2D and 3D US guiding schemes proved able to hit the target through visual and US confirmation; however the planning based on single 2D US is not sufficient in covering the needed volume as compared to the 3D US planning. The need to verify placement accuracy of the system led to more phantom tests using triangle shape phantom as shown in Figure 9.

A second series of experiments were performed using the triangle phantom. The average targeting accuracy was 1.5 mm, and all accuracy readings were consistently in the order of the tracking device itself (FOB, Flock of Bird by Ascension Tech., Inc.). Figure 10 depicts the physical setting and shows the images taken for these experiments. This experiment was repeated five times as shown in the left images of Figure 10. The same verification schema was applied here, using visual observation to check the targeting of single point rather than 1cm rubber ball, using US verification, as shown in the right images of Figure 10.

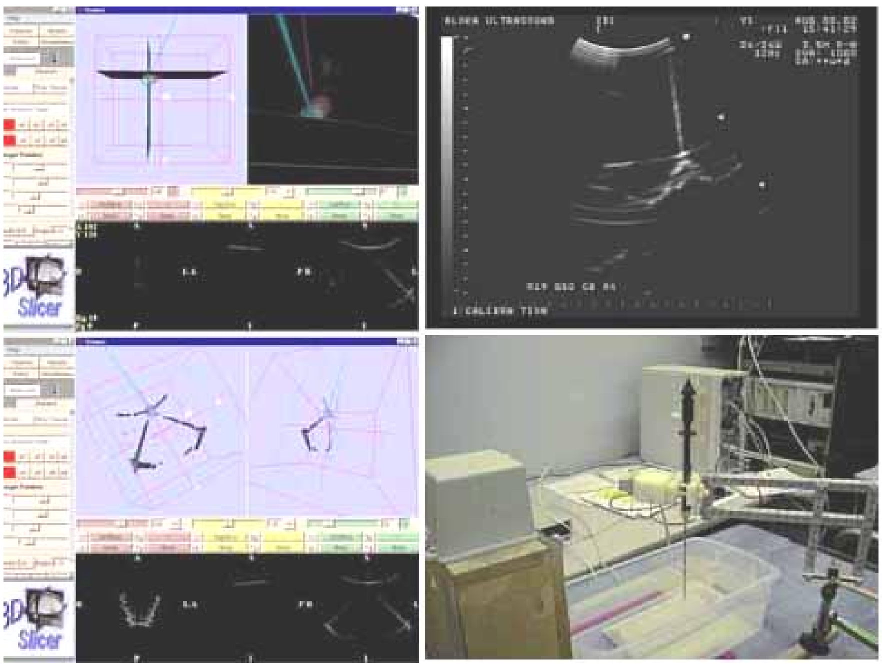


Figure 10: Triangle phantom experiment. LR: experimental setup. L: screen captures from planning and monitoring. UR: confirmatory 2D US snapshot.

4. DISCUSSION AND CONCLUSION

The project demonstrates that the accuracy and efficiency of targeting of an end-effector device such as this RF probe can be aided by the use of simultaneously tracked freehand 3D ultrasound and adjustable passive mechanical linkage. Real-time surveillance of the needle placement in the virtual space on screen, combining live 2D US and reconstructed 3D US was most helpful.

Ideally the RF ablator should be placed perpendicularly to the surface in order to avoid the slippage and minimizing deflection of the needle during insertion. In order to reduce targeting time and enhance accuracy, we consider the use of motorized rotation and needle insertion stages, previously developed at the Johns Hopkins University Urobotics Lab[14]. Simultaneous spinning and advancing the needle could reduce tissue resistance and subsequently reduce deformation and displacement of the target tissue. This, however, requires more sophisticated needle drivers, currently under development at the Johns Hopkins University.

Modeling the liver motion during the breathing cycle would help in overcoming this artifact. Currently we are investigating the use of the robotic arm to handle the US scanning, which could be an effective way to overcome the unstable pressure pattern induced by the manual scanning. (We present our initial results in a companion paper at this conference.)

In the presented phase of the project, we concentrated on the overall mechanical accuracy of the system, assuming stationary target in an ideal setup. As these results were most satisfactory, we move toward more realistic setups. In the near future, we will be preparing for ex-vivo on animal models, and eventually human cadaver tests, in order to address uncertainties in real-time tracking and deformation and mechanical motion artifacts. Next, we will perform in-vivo animal tests to account for respiratory motion artifacts and other in-vivo uncertainties, in realistic operating room setup.

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