Calibration of 2D Ultrasound in 3D Space for Robotic Biopsies

Awais Ahmad
Department of Electrical and
Electronics Engineering
Ozyegin University
Istanbul, Turkey
Email: awais.ahmad@ozu.edu.tr

M. Cenk Cavusoglu
Department of Electrical Engineering
and Computer Science
Case Western Reserve University
Cleveland, OH, USA
Email: cavusoglu@case.edu

Ozkan Bebek
Department of Mechanical Engineering
Ozyegin University
Istanbul, Turkey
Email: ozkan.bebek@ozyegin.edu.tr

***Abstract*—Freehand Ultrasound technique is widely used in intraoperative biopsy procedures for detecting the volumes of interest. Freehand ultrasound probe is faster and flexible with 6 degrees of freedom. Thats why the imaging system must be calibrated in 3D space before integrating it with Robotics Biopsy System. In this paper we present a 3D space calibration method using a multipoint cross-wire phantom. The Ultrasound probe is attached to a robotic manipulator arm which moves it over the phantom in precise steps of distances and angles. The position and orientation of the probe is tracked by an optical tracking system. Optical markers are placed on the probe, phantom tank and the validation needle. The optical tracking system returns the position and orientation of the reference frames attached to these optical markers. The location of threads with reference to the frame of Ultrasound probe is found using this information. These values and the values returned by a mathematical model of the calibration box are used to construct the calibration matrix. The whole system is automated so it can process high number of frames which makes it rapid and more accurate. This process is used to calibrate the space for an automated needle insertion biopsy robot. The accuracy of the system was checked by a validation needle in 3D space. RMS error of the experiment groups on average was 1.67mm.**

***Keywords*—Calibration, Ultrasound Imaging, Autonomous Biopsy Robots, Freehand Intraoperative Ultrasound**

I. Introduction

Ultrasound imaging is widely used worldwide for clinical procedures because unlike x-rays or MRIs, it is low cost, portable and safer to use. Ultrasound is one of the noninvasive imaging technologies and it gives real time images of all types of organs and tissues. In the past couple of decades it has found its place in intraoperative processes like breast biopsies for lesion excision [1]. Ultrasound in the operation theatres has brought revolutionary changes due to its real time imaging capabilities, safety and ease of use for longer durations. Rifkin et. al. in [2] presents a study of liver biopsies with the help of 2D ultrasound and concludes that ultrasound imaging is clinically useful technique. In many instances, it changes the course of clinical procedures.

In recent years the surgical processes are moving towards Minimally Invasive Surgery (MIS). Minimally invasive surgery (MIS) has been implemented in many sophisticated medical processes. Some of the simple applications are biopsies and delivering drugs or inspecting agents inside a specific tissue. There are some commercially developed robots in the market designed to assist in MIS. Needle biopsy is widely used in breast cancer detection. These biopsy procedures have many benefits but it is hard to reach the target tissues because the needle is too flimsy and tends to bend in the dense human tissue especially dense breast tissues. These problems can be solved by developing an autonomous needle insertion robot guided by ultrasound images in 3D space. Researchers have been working on development of robots to assist the physicians in needle biopsy procedures. Kettenbach et. al. in [3] have developed a needle biopsy robot and they have tested their system on gel based phantoms embedded with peas as tumors.

Kaya et. al. [4] has developed imaging algorithm to localize the tip of the needle in 2D US image. In this paper, we will develop calibration scheme that will contribute towards the development of an automated biopsy robot. This calibration process is also used to validate the work done by Kaya et. al. [4] and it will bring the biopsy robot and the target tissue in one universal frame of reference allowing it to accurately track its movement towards the target. This biopsy robot is 5 DOF parallel links robot and an extension of Small Animal Biopsy Robot (SABiR) [5] . 3D ultrasound imaging is an emerging technology and has many benefits over 2D US. 3D US can give direct visualization of 3D visualization of an anatomical structure. 3D ultrasound extends the concept of 2D US so that the area of interest can be acquired as 2D slices data constructing a 3D volume. Measuring volumes using 3D US has many clinical applications for example observing the development of fetus, evaluating the brain shift during neurosurgery or visualizing tumorous areas [6] .

3D visualization of the subject saves the physician from the mental transformation of the US images into understandable situation of the inside and improves diagnostics and decision making. There are four basic techniques to achieve three dimensional US space, Mercier et. al.[7] has briefly discussed these techniques. One way is to use a motor mechanism to move a 2D US probe in space with predefined intervals and constructing a 3D space from the slices. In this method the probe movement is constrained to a predefined path. This path can be a linear path acquiring a series of parallel planes

978-1-4673-7509-2/15/$31.00 2015 IEEE

or a radial movement [8] in which the probe is rotated around an axis perpendicular to the axis of probe itself creating a cylindrical 3D space. This method is simple but, as it uses a constraint motion to grab frames, it contains less accurate information at orientations other than those of input frames.

Another method is the already available 3D imaging probes. These are relatively larger and expensive probes. These probes have the same mechanism of a 2D probe but they use multiple arrays of ultrasound transceivers embedded at the tip with different orientations, typically in conical shape. Due to the limited space on the tip, the resolution of this type of sensor is very low as compared to the 3D reconstruction from 2D images. Another type of sensor uses one array of ultrasound transceivers electronically steered in the space to construct 3D space. Normally the sensor array is steered radially to increase the field of view. Other methods use position sensors to track the motion of the probe and calibrate the US images but some people have also tried sensorless techniques using speckle regression in the images to estimate the calibration but the results were not encouraging [9].

An alternate method of attaining the goal of 3D reconstruction of US is called freehand technique. The movement of the probe is un-constrained and can have 6 degrees of freedom. The probe can be moved even in the hands of an operator hence giving the name freehand technique. A motion tracking sensor is attached to the probe which gives the position and orientation of the probe in space. With this information we can easily find the location of US image pixels in 3D space. The motion can be tracked by optical, mechanical or magnetic position sensors. This method has the flexibility of the intrinsic 3D US probes but has no limitation on trackable volume. Freehand systems are cheaper and can obtain 3D data from any arbitrary position. The results from this method also depend on how the probe is moved in space, irregular motion will construct space with irregular gaps. In this paper we are focusing on the calibration of freehand US motion with crosswire multithread phantom.

II. Methodology

A. Selection of Phantom

Phantom is a coupling material used for the ultrasound waves as a medium. Phantom mimics the properties of tissues in calibration process because they give us liberty to embed predefined geometries in them which is the base of calibration process. Typically phantom consists of a gel type material like gelatin or it can be simply water. In this material, there are 2D or 3D objects whose geometric attributes are known. Mostly the phantoms used contain wires or threads going across the phantom in straight lines or crossed angles. Some phantoms also use 2D shapes of known dimensions [10].
We are using a cross-wire phantom with 8 threads. The calibration tank has holes on two of the sides and the locations of the holes with respect to the reference frame of the tank are known. Fig 1. shows the model of the tank created in Matlab which will be used to calculate the transformation matrix later in experimentation.

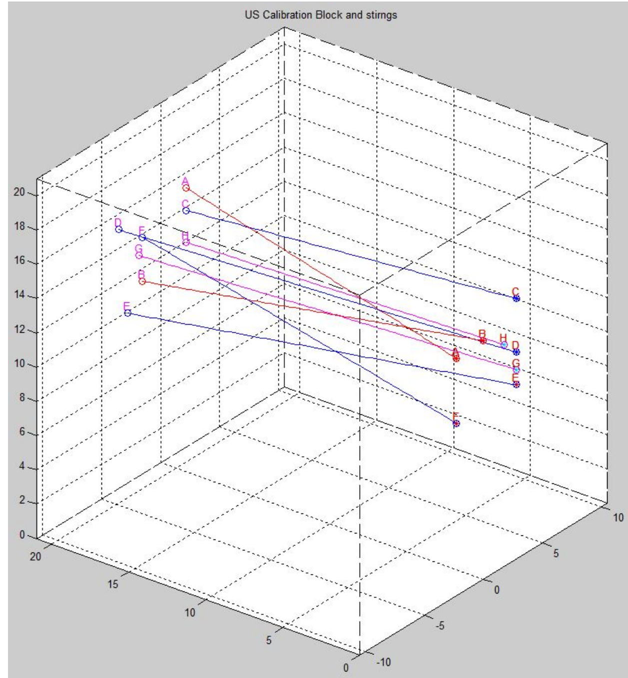

Fig. 1. Model of calibration box with 8 cross-wire threads

The number of threads, or cross-wires, is 8 to achieve higher accuracy of calibration matrix because the results are averaged over all the threads. The holes on both the sheets on the opposite ends do not have symmetric locations, which mean that the threads are woven in the box with wide crossing angles. So, when a cross sectional plane parallel to the sheets is traversed inside this space, the threads would appear as dots on the plane. Due to the angled crossing threads, these dots change location on the cross sectional plane as it move along the direction of threads or even if it rotates in inside the space containing the threads. This allows us to sample a large number of frames from the path along the threads at any location and angle with new readings in each frame. This gives us ability to get critically diverse readings for our method of calibration.

B. Selection of Tracking System

There are four main methods used to track the motion of the components of the system. The first method used was mechanical method in which the probe is attached to a robotic manipulator. The location and orientation of the probe was found by solving the joint angles of the manipulator arm, but the problem was it was only giving the information of the probe and not the calibration tank, which is a crucial part for calibration. Acoustic sensors are also used to locate the probe using Time of Flight to measure the distance between marker and the transducer. As this system uses sound as a medium so the results get affected by the atmosphere.

The most accurate methods are optical tracking and electromagnetic tracking. In [11] and [12] they performed the calibrations using electromagnetic tracking and reported RMS error of 0.4mm to 1.22mm. Optical tracking system is most commonly used method for tracking purposes [13, 14, 15]. In our project we have used the motion capture system Opti-Track developed by Natural Point. This system outputs the 3 positional coordinates and 3 rotation angles of a rigid body in a 3D space. This 3D space or effective volume is created once in the start when the system is calibrated in one fixed location. A frame is attached to the rigid body when multiple numbers of reflecting balls are selected as one rigid body. This frame is initially aligned with world frame of the system and gives the angular rotations with respect to this fixed frame. The origin of this frame is placed at the centroid of the rigid body shape.

C. Calibration Matrix Construction

The purpose of constructing the calibration matrix is to get the location of ultrasound image pixels in real world coordinates. Only than an automated biopsy system would be able to reach the targeted volume of interest and carry out the sample extraction. The accuracy of such application is very important because the subjects are humans and small calibration errors can lead to wrong diagnosis or even internal injury. The ultrasound probe and the calibration tank both are tagged with optical markers. It means that their position and orientation in the base frame of the tracking system, henceforth named as world coordinate system, are available to us all the time.

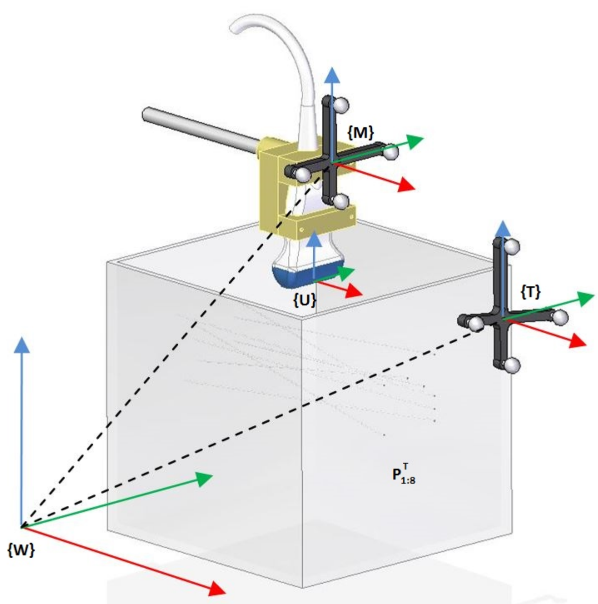

Fig. 2. Frames attached to the US Probe {M} and Calibration Tank {T}

Fig 2. shows our proposed experimental setup. The reference frames of both the probe and the tank are shown with reference to the world coordinate system. {W} is the base frame of the optical tracking system, similarly {M} is the reference frame of the marker attached to US probe and {T} is attached to the calibration tanks marker. Let T^W_M represent the homogeneous transformation matrix of US probe marker with respect to world frame and T^W_T represent the homogeneous transformation matrix of the calibration tank marker. We have both of these matrices coming from the optical tracking system. {U} is the reference frame attached at the base of the ultrasound probe. This reference frame is basically the coordinate frame of the ultrasound image because its origin is at the top center of the US image. The transformation of {U} with respect to {M} can easily be found by measuring the offsets in all dimensions. All the offsets are measured from the 3D models of the system components which are assumed to be accurate.

The mathematical model of the calibration box in Matlab is made with real dimensions of the holes with respect to the reference frame of the optical marker attached to box {T} as shown in figure 2. The model takes the transformation matrix of the calibration tank with respect to the frame of the real world US probe and outputs the 2D coordinates of each thread in the US image plane. We than can compare the results with the location of threads in actual US image and find the unknown parameters of calibration matrix. Eq. 1 shows the relation between the points in both planes through the calibration matrix.

$$P^U_{1:8} = ST^U_T P^T_{1:8} \tag{1}$$

$P^U_{1:8}$ are the 2x1 vectors representing the 2D location of thread 1 to 8 in US image plane and $P^T_{1:8}$ represent the location of threads in calibration tanks frame of reference. $P^U_{1:8}$ has unit in terms of pixels and $P^T_{1:8}$ has the units in millimeters. So we have to add a scaling vector S that will find the scale when pixels in both x and y directions are converted to millimeters. T^U_T is the homogeneous transformation of the calibration tank frame of reference with respect to ultrasound probe. This transformation can easily be found by the given matrix multiplication.

$$T^U_T = T^U_M T^M_W T^W_T \tag{2}$$

T^U_M is the inverse homogeneous of T^M_U which is the homogeneous transformation matrix of US image plane with respect to the US probe optical marker. This matrix is found by physical dimensions of the probe holder and the probe. Similarly T^M_W is the inverse homogeneous of T^W_M. T^W_T and T^W_M are both with reference to the base frame of the optical tracking system so they are calculated by the outputs of the tracking system.Once we have this system of transformations solved, we will have 8 sets of values corresponding to each thread. Each set has 8 parameters of the calibration matrix to be solved. We can find the values of these parameters by applying the least squares on the data of all the threads.

$$f(\alpha, \beta, \gamma, x, y, z, sx, sy) = ST^U_M T^M_W T^W_T P^T \tag{3}$$

$$min \sum_{i=1}^{n} ||f_i(\alpha, \beta, \gamma, x, y, z, sx, sy) P^u_i||^2 \tag{4}$$

$$S = \begin{bmatrix} S_x & 0 & 0 & 0 \\ 0 & S_y & 0 & 0 \end{bmatrix} \tag{5}$$

The values of x, y, z are in the fourth column of the homogeneous transformation matrix and α, β and γ can be found by the reverse mapping of the rotation matrix

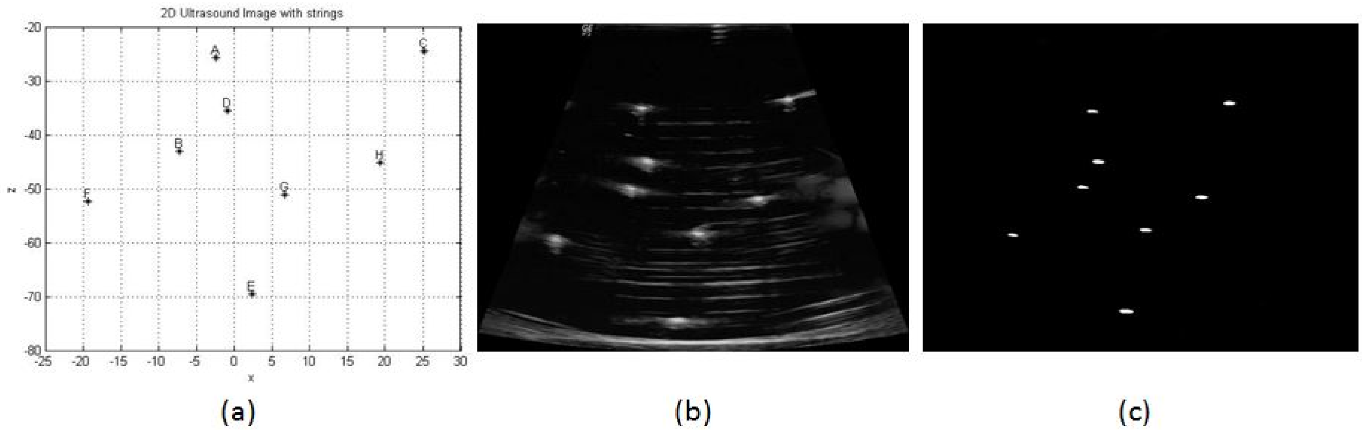

Fig. 3. (a) Thread coordinates from the model for a given position of US Probe (b) US image for the given probe position in the calibration tank. (c) Processed image showing automatically detected thread locations.

taken out from homogeneous transformation matrix. After finding the solution for each image, the whole solution can be found by taking the mean and RMS of all the image results.

Fig 3. shows the comparison of the results of actual ultrasound image of a plane located at distance in tank. Fig 3(a) is the location of the threads with respect to US probe calculated by the mathematical model of the calibration box. Fig 3(b) is the actual US image recorded at the same distance in the tank and Fig 3(c) is the processed image with thread locations segmented using normalized cross correlation. The centroids of the blobs are taken as the thread location in pixels

III. EXPERIMENTAL SETUP

To perform the experiments we have built a tank from transparent acrylic sheets. The holes for the threads were drilled precisely from a CNC. The sheets are joined by screws to avoid the extra glue material between the sheets. Cotton threads are used in this setup due to their high visibility even in deeper distances as compared to plastic wire. Although the size of blob detected in US images in smaller and more precise for the plastic wire but its visibility decreases as the distance from the probe increases. It can be seen in the Fig 3(b) that there are some lines in the page parallel to the US probe at equal distances. This is called the Reverberation artifact [16] which can be removed in future version by making a tilted base tank to reflect back the sound waves at an angle from the base. But the normalized cross correlation matching algorithm ignores this noise in the US image processing. Fig 4. shows the calibration tank with threads and the Optical marker. The tank is filled with deionized water.

The US probe is held in a 3D printed holder with foam paddings. This design of the probe holds the probe in firm position during the experiments which is very important for the calibration process because even a small movement in place can cause high angular deflections on the values at longer distances from the pivot. This holder is than attached at the end of KUKA KR 6 R900 (KR AGILUS) Industrial Robotic Manipulator. This robotic arm can be moved in space with 6 DOFs with minimum step size of 0.1mm and $0.1°$ and less than ± 0.03mm repeatability.

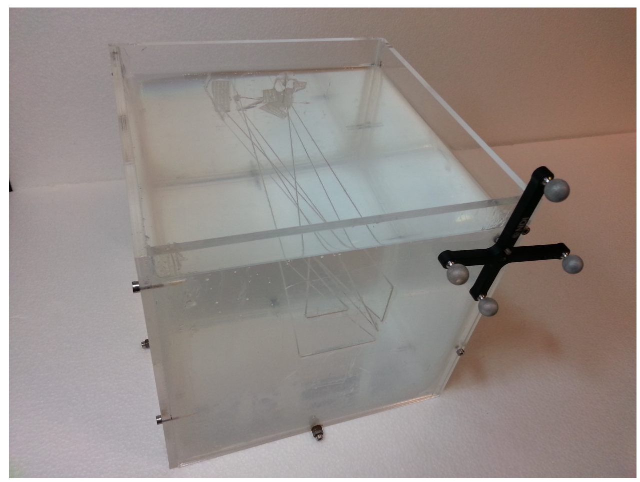

Fig. 4. Calibration Tank with Threads and Optical Marker, filled with water

We are using GE Logic P5 Ultrasound machine for the biopsy robotic system and in the calibration experiments. A Euresys Picolo HD 3G Frame Grabber Card is used for image acquisition purposes. This card connects with the ultrasound via VGA cable and the screen of the ultrasound machine is made accessible into the Matlab for processing purposes. For the calibration experiments, we move the probe in the tank using the robot arm and capturing images. Corresponding to each US image, the position and orientation of the probe is stored in a Comma Separated File. This file can be accessed in calibration algorithm for automatic and faster calculations in calibration process. Fig 5. shows the experimental setup in which the US probe is scanning the threads in water tank.

The optical tracking system uses 6 infrared cameras. These cameras are placed around the experiment arena in a semi-circular arrangement facing the markers of the system to avoid occlusion. The system software runs at the speed of 120 FPS which means the position of each marker is updated 120

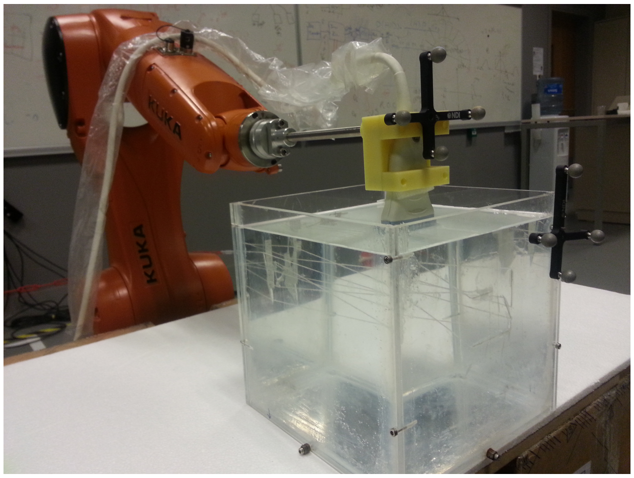

Fig. 5. US Probe on KUKA robot with optical marker scanning the calibration tank

times in a second. The system has RMS error of around 1mm. This error can be reduced by placing the cameras at positions clearly facing the markers. Natural Point has provided us with the API to get all the values into the Matlab in real time. The code can be configured to fetch data at lower rates. The frame rate of ultrasound is 15 frames per second. So we have configured the Matlab code to fetch data from Opti-Track when a new frame from the frame grabber is triggered. The real time data is displayed in a graphical user interface. When required the image frame and the corresponding coordinate values are stored in memory to be fetched later.

IV. RESULTS AND VALIDATION

The values were taken at different locations and angles in the tank. An image processing algorithm detects the dots of threads by cross correlation and extra noise is removed in the US images. When there are only 8 blobs in the binary image, the centroids of these blobs are saved in arrays. These arrays are now used in the calculations with the Optical Tracking data retrieved earlier with the images.

The US probe is attached to KUKA robot so we can move the probe with precise and accurate motion over the tank. 85 images were taken with no angles around any axis. These images are placed at exactly 1mm from each other. Another 30 were taken placed at random locations and 30 images taken at an angle of 5° across x-axis of the image plane. Fig 6. shows the probe modeled at an angle of 30° across z-axis of the tanks frame of reference. The points U_0 through U_5 indicate the plane of the Ultrasound image. In the figure given above, the probe is observing the threads at an angled positing. The location of the threads in the given orientation is found by the mathematical model and used with the original values calculated by US image processing.

S_x and S_y are the scaling factors that scale pixels to millimeters. The coordinates in the US image plane are in pixels while real world coordinates are in millimeters. The units for scaling factors are pixels/mm. The validation of this data is also important in order to make sure that the results

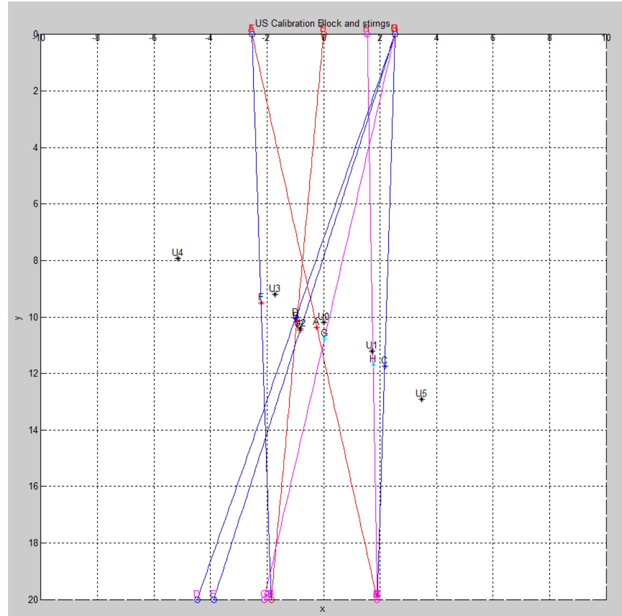

Fig. 6. Matlab Model of US image plane at an angle of 30° around z-axis inside the calibration tank

Calibration Group	Number of Images	S_x	S_y	RMS S_x	RMS S_y
1	85	7.13	7.01	1.53	1.46
2	30	7.02	7.15	1.75	1.68
3	30	6.98	7.09	1.88	1.70

Table 1: Results of Calibration for 3 Groups of Images

are accurate.

Calibration can be validated by four main methods described in [11]. We are using the first method which is to calculate the RMS of the solution. Another way is to calculate the location of a test point which was not a part of the calibration process. It is like checking a test point on data learnt from training data. We have designed a special needle equipped with an optical marker as shown in Fig 7. This needle has a 90° turn at the end which enables us to move it perpendicular to image plane easily. This gives us a nice dot in the US image. This needle is freely moved by hand.

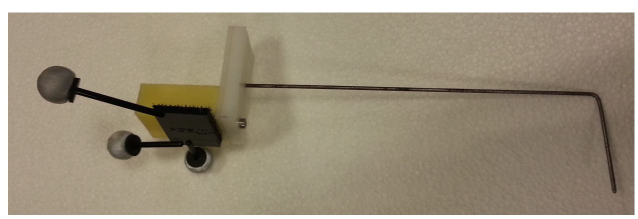

Fig. 7. Validation Needle with Optical Marker

We have the knowledge of the needle tip location with respect to the frame of validation needle marker {V} i.e. PV. We already have information about the transformations of marker {M} and US image plane {U} as shown in the Fig 8. The position of the needle in the US image plane can be found by the given transformations:

For the validation experiments, we took 10 images with needle at different locations and applied the found solution. The true location was found by the built-in measurement options of the US machine. The location of the needle was found within the range of 2mm on average.

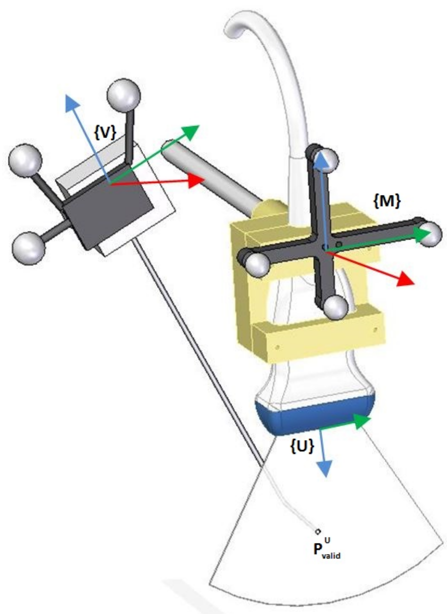

Fig. 8. Validation Needle and US Probe in working position with their respective frames of references

V. DISCUSSION

This system of calibration of 2D US images is developed to assist the development of a 5 DOF parallel robot platform which is aimed to perform needle biopsies on human subjects. For such applications, the accuracy of the system should be within 1mm. Needle biopsies are mostly performed for the extraction of cancerous tissues for diagnosis. These tissues can range in size from 1cm to 6cm. The accuracy of our system is 1.67mm RMS. This calibration process heavily depends on the external sensor used for the tracking purposes, which in our case is Optical Tracking System (Opti-Track). Accuracy of the Opti-Track system varies with camera configuration and is about 1mm RMS. One more factor for causing inaccuracy can be the resolution of the plane slicing in Matlab versus the resolution of the motion of US probe in real space. The current calibration serves our needs in the development of Biopsy Robot and we are limited by the capabilities of motion tracking system.

ACKNOWLEDGMENT

This work was supported by the Scientific and Technical Research Council of Turkey (TUBITAK) under Grant No.112E312.

REFERENCES

[1] LaNette F Smith et al. "Intraoperative ultrasound-guided breast biopsy". In: *The american journal of surgery* 180.6 (2000), pp. 419–423.

[2] MATTHEW D Rifkin et al. "Intraoperative ultrasound of the liver. An important adjunctive tool for decision making in the operating room." In: *Annals of surgery* 205.5 (1987), p. 466.

[3] Joachim Kettenbach et al. "Robot-assisted biopsy using ultrasound guidance: initial results from in vitro tests". In: *European radiology* 15.4 (2005), pp. 765–771.

[4] Mert Kaya et al. "Real-time Needle Tip Localization in 2D Ultrasound Images for Robotic Biopsies". In: *17th International Conference on Advanced Robotics* (2015).

[5] Ozkan Bebek, Myun Joong Hwang, and M Cenk Cavusoglu. "Design of a parallel robot for needle-based interventions on small animals". In: *Mechatronics, IEEE/ASME Transactions on* 18.1 (2013), pp. 62–73.

[6] Thomas R Nelson and Dolores H Pretorius. "Three-dimensional ultrasound imaging". In: *Ultrasound in medicine & biology* 24.9 (1998), pp. 1243–1270.

[7] Laurence Mercier et al. "A review of calibration techniques for freehand 3-D ultrasound systems". In: *Ultrasound in medicine & biology* 31.2 (2005), pp. 143–165.

[8] S Tong et al. "A three-dimensional ultrasound prostate imaging system". In: *Ultrasound in medicine & biology* 22.6 (1996), pp. 735–746.

[9] Pai-Chi Li, Cheng-Yen Li, and Wen-Chun Yeh. "Tissue motion and elevational speckle decorrelation in freehand 3D ultrasound". In: *Ultrasonic imaging* 24.1 (2002), pp. 1–12.

[10] Yoshinobu Sato et al. "Image guidance of breast cancer surgery using 3-D ultrasound images and augmented reality visualization". In: *Medical Imaging, IEEE Transactions on* 17.5 (1998), pp. 681–693.

[11] Hui Zhang et al. "Freehand 3D ultrasound calibration using an electromagnetically tracked needle". In: *Medical Imaging*. International Society for Optics and Photonics. 2006, pp. 61412M–61412M.

[12] Richard W Prager et al. "Rapid calibration for 3-D freehand ultrasound". In: *Ultrasound in medicine & biology* 24.6 (1998), pp. 855–869.

[13] Emad M Boctor et al. "Rapid calibration method for registration and 3D tracking of ultrasound images using spatial localizer". In: *Medical Imaging 2003*. International Society for Optics and Photonics. 2003, pp. 521–532.

[14] Jane M Blackall et al. "An image registration approach to automated calibration for freehand 3D ultrasound". In: *Medical Image Computing and Computer-Assisted Intervention–MICCAI 2000*. Springer. 2000, pp. 462–471.

[15] Thomas Chen et al. "Automated 3D freehand ultrasound calibration with real-time accuracy control". In: *Medical Image Computing and Computer-Assisted Intervention–MICCAI 2006* (2006), pp. 899–906.

[16] Jinlan Huang et al. "Imaging artifacts of medical instruments in ultrasound-guided interventions". In: *Journal of Ultrasound in Medicine* 26.10 (2007), pp. 1303–1322.