A Method for Ultrasound Probe Calibration Based On Arbitrary Wire Phantom

Chunxu Shen, Liushuai Lyu, GuangZhi WANG, and Jian Wu*

Abstract: To create a freehand three-dimensional (3D) ultrasound (US) system for image-guided intervention, US probe calibration process plays an important role. This paper introduces a novel method, based on arbitrary wire phantom, to achieve both spatial and temporal calibration for ultrasound probe. Spatial calibration is realized by solving an optimization problem established by the wires and corresponding intersections in US plane. Next, temporal calibration is achieved by processing US image sequence and the corresponding position of optical localizer mounted to US probe. Also, in order to make up for the deficiency of geometry structure in arbitrary phantom, we develop point recognition algorithm to determine the correspondence between wires in phantom space and intersections in US image space. Extensive comparative experiment is conducted on N-wire phantom and our phantom in 20 independent trials to fully evaluate precision, accuracy, and performance of proposed calibration method. Shallow probe experimental result shows that proposed method improves average calibration precision to 0.896 mm and accuracy to 1.022 mm, compared to 0.938 mm and 1.140mm using N-wire respectively. Further, we also perform 5 independent trials to evaluate the impact of deep image for proposed method. Result shows the precision ranges from 0.740 mm to1.178mm, and the accuracy ranges from 0.939mm to 1.400mm.

Keywords: Arbitrary-wire phantom, probe calibration, ultrasound-guided intervention, speckle tracking.

1. INTRODUCTION

Freehand tracked ultrasound imaging is widely used in several guided interventions such as biopsy, ablation, and computer aided surgery [1-2], due to the characteristics of low cost, safe, high temporal resolution and portable. Tracking is easily achieved by mounting a localizer to probe, and then the probe can be traced by position sensing system (e.g. Optical or electro-magnetic tracking device). However, it is not adequate to determine the position of acquired two-dimensional ultrasound images if we just obtain the position of ultrasound probe. Probe calibration is the task of determining the transformation between the localizer space and the ultrasound image space. Therefore, the transformation, which consists translation and rotation, is a fundamental and noticeable step in ultrasound-guided intervention.

Methods for ultrasound probe calibration have been widely investigated [3-4] to facilitate the calibration procedure. Only a few research groups [5-6] complete probe calibration without calibration phantoms, whereas most researchers need different phantoms with known geometry constrains to provide fiducials for calibration process.

Point phantom, can be formed by a spherical bead-like object [7] or a pair of cross wire phantom [8], is one of the first kind phantom to be adopted for probe calibration. This phantom is easily designed, but the major difficulty is the ultrasound images of the phantom need to be segmented manually, which makes calibration process time-consuming and tiresome. While Hsu et al. [9] proposed an automatic algorithm, recognition of isolated points in ultrasound images are seldom reliable

Plane phantom, typically such as a precision-made Cambridge phantom [10] and its variants [11-12], overcomes the drawback mentioned above. The most

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attractive part is that the method realizes an automatic segmentation algorithm, and then probe calibration can be achieved rapidly. However, scanning from a wide range of angles and positions limits further development in clinical usage. Otherwise, plane reflects away much of the ultrasound energy, which makes automatic segmentation in some position become relative difficult.

N-wire phantom successfully achieves automatic segmentation and rapid calibration. Compared with at least 550 images to calculate acceptable accuracy, N-wire phantom just needs about 6-30 images [13-15]. For N-wire method, nevertheless, only middle line in the N-shaped structure is taken into account and to compute calibration transform. Higher precision can be achieved if all lines of N-shaped structure are used for calibration. Besides, due to the thickness of the US image slice, the localization accuracy in US imaging is usually highly anisotropic. Based on the above reasons, a cost function [16] is proposed to minimize iteratively and get precision starting from a seed which is computed with a closed-form solution based on the middle wires.

Inspired by the method [16] above, we propose a new method that takes all the wires into account for calibration, and then achieve a better result. The difference is that we directly compute calibration parameters based on the collinearity of the wire and its corresponding intersections in ultrasound plane rather than by doing iteration from a seed solution. Meanwhile, because of the lack of known geometry constrains (e.g., N-wire phantom) in arbitrary wire phantom, we develop a point recognition algorithm based on detection and tracking to determine the corresponding between intersections in US image space and wires in phantom space.

The remainder of this paper is organized as follows. First, an overview of probe calibration process is given. Then, the specification of arbitrary wire phantom is present. Further, we introduce our experimental set-ups when acquire ultrasound image sequence and collect localizer tracking information, as well as propose a robust and fully automatic point recognition algorithm. Subsequently, we achieve both temporal and spatial calibration with fully evaluating precision and accuracy of proposed method. Finally, extensive comparative experiment with N-wire method and probe calibration with different deep image are conducted, followed by our detailed discussion.

2. METHODS AND MATERIALS

2.1 Probe calibration overview

This section presents an overview of probe calibration process (see Fig. 1), transform matrix involved, and a description about hardware. Images are generated by an ultrasound machine (A6 B/W HCU, SonoScape, China) with a convex probe (C354, SonoScape, China), then fed to a frame grabber (DIV2USB3.0, Epiphan, Canada) and transmitted to central processor (Precision M6800, Dell, China). Simultaneously, an optical tracking device (Vicra, Northern Digital Inc., Canada) continuously monitors the localizers attached to probe and phantom respectively, and then provides position and orientation of probe-localizer and phantom-localizer in tracker device coordinate system (Tracker coordinate frame, abbreviated as T). So, the spatial accuracy of 3D tracked ultrasound mainly depends on calibration transformation, which determines the points in

Fig. 1. Transformations involved in calibration process. (a) Arbitrary wire intersection points visible in ultrasound images (b) Ultrasound probe with attached localizer (c) Sensor of optical tracking device (d) Calibration phantom with attached localizer
Ultrasound plane (Ultrasound image coordinate frame, abbreviated as $I$) with respect to the probe-localizer (Probe coordinate frame, abbreviated as $P$).

It is noteworthy that ultrasound images are not acquired synchronously with the corresponding tracking recordings, due to which the frame grabber acquisition rate is usually slower than tracking frequency [17]. So it is essential to develop temporal calibration before studying spatial calibration.

Fig. 2 illustrates the design specification of proposed arbitrary wire phantom. To prevent occurrence of reverberance artifacts [13], we ensure all five wires are not parallel to each other.

2.2 Arbitrary wire method

For brevity, let $X_A$ and $X_B$ denote a 3D position in coordinate frame of $A$ and $B$ respectively. Furthermore, the 3D position $X$ in $A$ coordinate frame can be denoted as $X_A = [x, y, z, 1]$ . Then, $T_B^A$ represents a homogeneous transform [18] that maps from $X_A$ to $X_B$, which can be written as:

$$X_B = T_B^A \cdot X_A.$$  

(1)

The probe calibration process showed in Fig. 1, therefore, can be written as:

$$P_c = T_f^C \cdot T_p^T \cdot T_f^P \cdot P_I,$$  

(2)

Where, $P_I$ is the intersections of nylon wires and ultrasound plane in US image space, and we can collect points $P_I$ by point recognition algorithm when scan proposed arbitrary phantom. Similarly, $P_c$ is the intersections of nylon wires and ultrasound plane in calibration phantom space, which must be located in the line equations of nylon wires. Moreover, $T_f^C, T_f^P$ can be obtained using position tracking device by localizers attached to probe and phantom respectively.

Here, we suppose that elevation thickness of ultrasound beam is 0. The intersections of nylon wires and ultrasound plane, further, can be written as $P_I = [u, v, 0, 1]$, where, $u$ and $v$ are the pixel index of row and column respectively in ultrasound image. So the probe calibration matrix, $T_f^P$, can be written as:

$$T_f^P (\varphi) = \begin{bmatrix} a_1 & b_1 & 0 & d_1 \\ a_2 & b_2 & 0 & d_2 \\ a_3 & b_3 & 0 & d_3 \\ 0 & 0 & 0 & 1 \end{bmatrix},$$  

(3)

Where, $\varphi = [a_1, a_2, a_3, b_1, b_2, b_3, c_1, c_2, c_3, 1]$. The procedure of ultrasound probe calibration is just the process to find $\varphi$ based on (3). Then, the intersections $P_p$ in probe coordinate system can be written as:

$$P_p (u, v) = T(u, v) \cdot \varphi^T,$$  

(4)

Where, $T(u,v)$ is,

$$T(u,v) = \begin{bmatrix} u & 0 & 0 & v & 0 & 0 & 1 & 0 & 0 & 0 \\ 0 & u & 0 & v & 0 & 0 & 1 & 0 & 0 & 0 \\ 0 & 0 & u & 0 & v & 0 & 0 & 1 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 1 \end{bmatrix}. $$

In Fig. 2, the equations of nylon wire FB are written as:

$$T_{FB} \cdot P_c = 0,$$  

(5)

Where, $T_{FB}$ is the coefficient matrix of the line equation. and can be determined once the geometry structure of proposed arbitrary phantom is designed by precision machining. Therefore, based on (2)-(3), (5), the following equation makes sense.

$$f(\varphi) = T_{FB}^C \cdot T_f^C \cdot T_f^P \cdot T_f^T \cdot P_I = 0,$$  

(6)
In fact, the errors are unavoidable because of the thickness of ultrasound beam, and the tolerances of machining and point-recognition algorithm. So (7) can be modified as:

\[ f(\phi) = T_{FB} \cdot T'_{F} \cdot T'_{P} \cdot T(u,v) \cdot \phi^T = \varepsilon, \]  

(7)

Where, \( \varepsilon \) presents the errors mentioned above. Then, the calibration parameters, \( \phi \), can be determined by the optimization process as followed:

\[ E(\phi) = \sum_{i=1}^{m} \sum_{j=1}^{n} |\tilde{e}_{i,j} - T_{FB} \cdot T'_{F} \cdot T'_{P} \cdot T(u_{i,j},v_{i,j}) \cdot \phi| \]  

(8)

Where, \((u_{i,j},v_{i,j})\) is the intersections between the \( j \)-th nylon wire with the \( i \)-th ultrasound image. \( T_{FB} \) is the corresponding equations of \( j \)-th nylon wire. \( E(\phi) \) is the calibration error, so we can determine the calibration parameters \( \phi \) by minimum \( E(\phi) \).

2.3 Points recognition algorithm

Point recognition in acquired ultrasound images is a key process for proposed calibration method. Because of without assistance of a special geometric constrains (e.g., N-wire [14]), it is difficult to determine which line equation that intersection points belong to. In this section, we propose an effective point recognition algorithm, which combines point detection and speckle tracking [20], to address the issue and determine the corresponding between intersections in US image space and line equations in calibration phantom space.

Because point detection is just used once in probe calibration process, it relieves the misgiving that detecting isolate points in ultrasound images is seldom reliable. Firstly, we employ ostu [19] threshold segment algorithm to remove abundant low echo signal, and employ morphological operation to clear isolate pixel in US image (see Fig. 3a). Besides, the echo signal from the side face of tank and near crystals would yield adverse effect on point recognition because of high intensity close to intersections. In our experiments, both of them include more pixels (see Fig. 3b). So, we label all connected domains, count the number of pixels, and then remove the larger areas (larger than 500 pixels in our algorithm). Once eliminating these negative factors, detecting points is easily realized by using the large difference of echo signal between nylon wire and water (Fig. 3d).

As a stable feature in ultrasound image, speckle has been wildly used to respiratory motion estimation [20] and echocardiography [21]. Here, we employ speckle feature and best match [23] to track the positions of intersection points when acquiring a consecutive ultrasound images. The initial positions of five intersection points for tracking are determined by the output of point detection part (Fig.3e). We set the search window to 40 pixel × 40 pixel in all experiments. Fig. 3f shows tracking results in 350 consecutive ultrasound images.

2.4 Temporal and Spatial calibration
Before developing temporal and spatial calibration, we should carefully discuss the issue of speed of sound in water. The speed of sound in water is 1480 m/s at room temperature, which is different from an average velocity of sound – 1540 m/s in human body [24]. Therefore, we heat the water to 37°C, where the speed is 1570 m/s, so that the speed of sound in the tank would match the hardware constant of ultrasonic machine closely.

In probe calibration process, ultrasound images and tracking data are from different hardware devices, so the inherent delay is unavoidable. In order to measure the delay correctly, we design an experimental schedule as Fig. 4(a) shows. Meanwhile, ultrasound plane is adjusted carefully so that it is approximately perpendicular to the nylon wire, and then the probe moves up and down with the traction of robot arm (Magician, Dobot, China) periodically. During the motion, both the ultrasound images and the position of probe localizer are recorded simultaneously with the time stamps. In our all experiments, we set the frequency of frame grabber and optical tracking device are 30 Hz.

We then process the acquired ultrasound sequence and extract the speckle (intersection between ultrasound plane with nylon wire) representing the ultrasound image stream. The normalized dominant direction of the speckle in the images and probe localizer in tracking coordinate system are used to recover the delay. The bottom of Fig. 4(b) shows the result of temporal calibration, and the inherent delay between ultrasound image stream and tracking data stream is near 100ms, which usually cannot be ignored in image guided intervention.

In all, what (8) has established an over-determined system for $T^P_I$, which can be solved using least mean square.

3. EXPERIMENTS AND RESULTS

In this section, we first introduce the evaluation methods for precision and accuracy. And then extensive trials are conducted to evaluate the robustness of proposed calibration method, and the response of precision and accuracy in different depth image.

3.1 Evaluation Method

In order to evaluate the precision of proposed calibration method objectively, we employ calibration reproducibility (CR) [16] to measure it.

N independent calibration experiments are performed using the proposed phantom, and then we can get a set of N calibration transformations, written as $T^P_I(i)$ with $i=1,...,N$. Eight special points (center and corner, see Fig. 5) in ultrasound image are mapped to probe coordinate system using these $N$ translations mentioned above. If proposed method is precious, each special point should appear the same or roughly the same.
position in probe coordinate system. So the CR is defined [4] as:

\[
CR = \frac{1}{N} \sum_{i=1}^{N} \| \overline{P}_{P} - T_{P}^{P}(i) \cdot P_{I} \|	ag{9}
\]

Where \( P_{I} \) is the point we select in the ultrasound image. \( \overline{P}_{P} \) is the average position that selected points are mapped by multiple \( T_{P}^{P}(i) \) transformations. Then the \( \overline{P}_{P} \) can be wrote as follows:

\[
\overline{P}_{P} = \frac{1}{N} \sum_{i=1}^{N} T_{P}^{P}(i) \cdot P_{I} \tag{10}
\]

According to (9), CR rules out the errors from point recognition algorithm and tracking device, totally focusing on probe calibration process.

To evaluating calibration accuracy directly is a challengeable task because exact correspondence relation isn’t existent between lines in phantom coordinate system and points in ultrasound image coordinate system. So we employ double-N phantom [14], which has a definite geometric structure and can provides ground truth points, to evaluate proposed calibration method. Further, we reconstruct the results of double-N positions in the physical phantom space using our calibration matrix and tracking device message, and then compare them with these ground truth points. This process, which also called fiducial registration error (FRE) [3], can be expressed mathematically as:

\[
FER = \frac{1}{N} \sum_{i=1}^{N} \| P_{G}(i) - T_{C}^{P} \cdot T_{P}^{T} \cdot T_{P}^{P} \cdot P_{I}(u,v) \| \tag{11}
\]

Where \( P_{G}(i), P_{I}(i) \) with \( i=1,...,N \) are \( N \) identified positions in double-N phantom and ultrasound image space. And \( T_{P}^{T}, T_{C}^{T} \) are rigid transformations from probe space to tracking space and from tracking space to phantom space, respectively. All these tracking message can read from optical tracking device. Fig. 6 shows the error between ground truth and reconstructed points in one calibration trial.

3.2 Calibration precision, accuracy and robustness

The performance of proposed method is impacted by the precision of point recognition algorithm, optimization method, errors from hardwire devices et al. Multiple independence trials maybe a good idea to vilify the robustness of our method. In this section, we firstly perform 20 independence trials (40 images are acquired per trial) to calculate our own calibration matrix. Secondly we employ N-wire phantom to evaluate precision and accuracy of calibration process. Here, we acquire 15 ultrasound images through scanning double-N phantom [14], and then label the intersections between middle line of N-wire and ultrasound plane manually (see Fig. 6a). So the precision, accuracy of
each trial can be calculated. Therefore, it is possible to evaluate the robustness of proposed method quantitatively.

We need to mention that the central frequency of ultrasound probe is 3 MHz, ultrasound focus point is about 5.6 cm, and the data acquisition frequency of frame grabber and optical tracking device set to 30 Hz. Besides, we use random number which range is ±0.5 mm to generate ε, which is difficult to estimate before probe calibration.

The evaluation results of proposed method show in Fig. 8 and Tab. 1. Our proposed point recognition algorithm can identify all intersections and correctly determine the corresponding relationship between points in image space and lines in phantom space. The average precision (CR) of 20 independent trials is 0.896 mm with the standard deviation 0.075 mm and the average accuracy (FRE) of 20 independence trials is 1.022 mm with the standard deviation 0.085 mm. Moreover, compared with N-wire, proposed calibration achieves better results with precision increasing 4.5% (from 0.938 to 0.836, see Table 1) and accuracy increasing 10.4% (from 1.140 mm to 1.022 mm, see Table 1). Substantial improvement in accuracy evaluation indicates proposed point recognition algorithm, based on speckle tracking, is more robust than [14].

3.3 Calibration responds for deep imaging

Resolution in single ultrasound image is different from one area to another area. This factor can be decomposed to axial and lateral resolutions. And the axial does not change with the image deep, but the lateral resolution is depth-dependent. It is because of the lateral resolution, namely the lateral width of ultrasound beam, point-shaped small objects (like the intersection between nylon wire and ultrasound plane) would express a blurry appearance. Fig. 8(a) shows the various appearances in ultrasound image when nylon wire is at different depth. We can see, more far away from ultrasound focused point, more blurry the appearance is. This phenomenon would influence the accuracy of point recognition algorithm, further be harmful for proposed calibration method. So evaluating how image depth influences the precision and accuracy of calibration matrix deserves concern. Fig. 8(b) shows our experimental facility. We simulate various image depth through adjusting the distance between ultrasound probe and uppermost nylon wire. And then, images are acquired while the probe moves horizontally with the

![Fig. 7. Evaluation for proposed method and N-wire method. Left is precision evaluation. Right is accuracy evaluation.](image)

![Fig. 8. (a) The shape of intersections determined by the ultrasound axial and lateral resolution (b) Schematic diagram for different depth image](image)

| methods     | mean  | standard deviation |
|-------------|-------|--------------------|
| proposed    | CR 0.896 | 0.075              |
|             | FRE 1.022 | 0.085              |
| double-N [14]| CR 0.938 | 0.118              |
|             | FRE 1.140 | 0.134              |

| distance (mm) | precision (mm) | accuracy (mm) |
|---------------|----------------|---------------|
| 30            | 0.822          | 0.954         |
| 50            | 0.741          | 0.936         |
| 70            | 1.053          | 1.251         |
| 90            | 0.956          | 1.156         |
| 110           | 1.178          | 1.400         |
traction of robot arm.

Table. 2 shows how five different image depths influence the precision and accuracy of proposed method. When the distance is 50mm, we can receive a better calibration matrix. That’s because this distance closes to ultrasound focused point (UFP), where the appearance of speckle is close to circle-shaped. Moreover, the precision, especially accuracy would decrease severely when the image depth is far away from UFP. When the distance is 110mm, the accuracy is only 1.400mm, which deserves us more attention during ultrasound guided intervention.

4. DISCUSSION

In this paper, we propose an alternative method for ultrasound probe calibration based on arbitrary wire phantom. This phantom does not depend on special geometric structure (like N-wire phantom). Besides, we provide a novel point recognition algorithm based on point detection and tracking. This algorithm can directly determine the corresponding between points in image space and lines in phantom space, which help us free from precision geometry design. Further, in order to thoroughly evaluate the performance of proposed method, we perform 20 independence trial to verify robustness and 5 trials to explore the response of precision and accuracy under various image depth. All results verify the effectiveness and robustness of proposed method. Besides, we also consider the difference of the sound velocity between phantom and human body and develop a reasonable resolution, which simulate the propagation velocity of ultrasound in fresh body through heating water to 37°C.

However, we just consider a special arbitrary wire phantom - all five wires are not parallel to each other. More detailed geometric constrains worth discussing and exploring. In future, we plan to pay more attention to improve precision and accuracy for deep ultrasound image and more thorough geometrical constrains in arbitrary phantom. We also plan to implement our proposed probe calibration method to clinical ultrasound image guided biopsy and radio-frequency ablation.

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