31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Real-time Compounding of Three-dimensional Transesophageal Echocardiographic Volumes: the Phantom Study

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Abstract—3D ultrasound has attracted considerable interest in recent years as a low cost, mobile and real-time imaging modality for interventional cardiac applications. However, the low image quality and small field of view have been two major barriers preventing 3D ultrasound from being widely accepted as a solution to the guidance of cardiac interventions. By using the 3D transesophageal echographic (TEE) probe, it is possible to acquire images with better quality compared to the images acquired from traditional transthoracic probe (TTE). However, the 3D TEE volume has even smaller field of view and is insufficient to cover the whole geometry of the heart. Previously, we have developed a technique to compound 3D TTE volumes in real-time. In this study, we extend this technique to compound 3D TEE volumes by using an electromagnetic tracking system. In this pilot study, two different types of phantoms were used to evaluate our technique. The results suggest our method is accurate and efficient. The compounding error is approximately 2.5mm.

I. Introduction

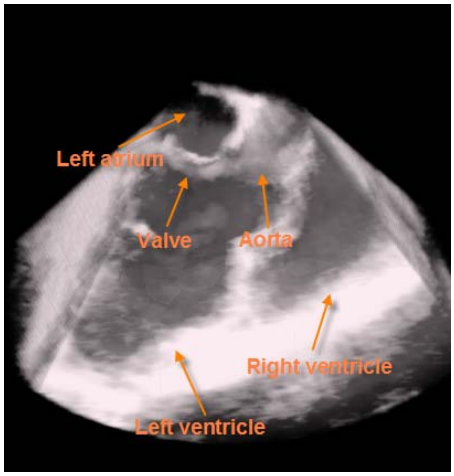

Figure 1. A 3D TEE volume was acquired from a swine model. It shows detailed inter-cardiac structures. The volume visualization was implemented in P-Rex, a software library provided by Philips Healthcare. The image was provided by Hansen Medical (http://www.hansenmedical.com).

Three-dimensional (3D) imaging techniques such as computer tomography (CT) and magnetic resonance imaging (MRI) have been used to guide interventional cardiac applications [1, 2]. As a low-cost, mobile and real-time imaging modality, ultrasound is attracting more and more attention [3, 4] in recent years especially after the introduction of the 3D ultrasound transducer. A 3D ultrasound transducer contains a 2D transducer array which enables a rapid data acquisition and therefore allows for 4D acquistion. This feature is especially useful for cardiac imaging in which the target is constantly moving. However, the 3D ultrasound transducer has a limited field of view and is insufficient to create the whole geometry of the heart. In 4D mode, the field of view could be further reduced because of the limited speed of signal processing. Previous attempts have been able to extend the field of view of ultrasound images by compounding multiple ultrasound images together. Generally, there are two types of approaches for image compounding. Yao at al. [5] reported an image registration based compounding algorithm. The ultrasound volumes with large overlapped regions can be registered to each other. Ma et al. [6] compounded the ultrasound images by using an optical tracking device. Although accurate, Yao's method is computationally expensive and therefore not possible for real-time processing. Compared to Yao's method, the computational burden in Ma's method is low but it requires a clear view of the ultrasound probe to the optical tracking system. In reality, by using a surface probe such as the transthoracic echocardiaography (TTE) probe it is often difficult to acquired clear cardiac images because of the attenuation of the sound wave and the small acoustic window in the thoracic region. Compared to the TTE probe, the transesophageal echography (TEE) can reach deep inside the human body and minimize the distance that the ultrasound beam has to travel. This reduces the attenuation of the ultrasound signal, generating a stronger return signal and ultimately enhances the image and Doppler quality. TEE is a better platform compared to TTE in the identification of aorta, pulmonary artery, the valves of the heart, coronary arteries and etc. (Figure 1). Potentially, it can be a better tool to guide cardiac intervention procedures. In this study, we extend Ma's compounding technique to make it suitable to handle TEE data. Our method involves the use of an electromagnetic (EM) tracking system and a calibration phantom. Our technique is efficient and can be implemented in real-time once the calibration is completed. It was evaluated by using two different types of phantoms. The mean compounding errors are 2.13±0.46mm and

Manuscript received April 7, 2009. This work was co-funded by the Technology Strategy Board's Collaborative Research and Development program, following an open competition (grant 17352), Philips Healthcare (Best, The Netherlands) and the EPSRC (grant EP/D061474/1).

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2.4±0.53mm respectively.

II. METHOD

A. Data acquisition

Live 3D ultrasound was performed on a Philips iE33 3D ultrasound system equipped with a 3D TEE probe (Philips Healthcare, Best, The Netherland). The data acquisition was carried out in Live 3D mode with 10cm depth setting. The ultrasound system was equipped with a software prototype which allows real-time data export via ethernet cable. Approximately 35 volumes were exported per second. Each of the exported volumes covers a region of 10x10x5cm^3. The voxel resolution is 0.7x0.7x0.7mm^3.

During ultrasound data acquisition, the 3D positions and the orientations of the TEE probe were determined by using an EM tracking system (Aurora, Northern Digital, Canada). The accuracy of the EM tracking system has been verified in previous literature [7]. Two EM sensors, which were inclined at 90º to each other, were firmly fixed on the TEE probe, allowing the determination of its position in six degrees of freedom (translations and rotations in X, Y, Z directions). The EM tracking system was connected to a laptop via a RS232 cable. In-house developed software was used to acquire the EM tracking information and the ultrasound data simultaneously.

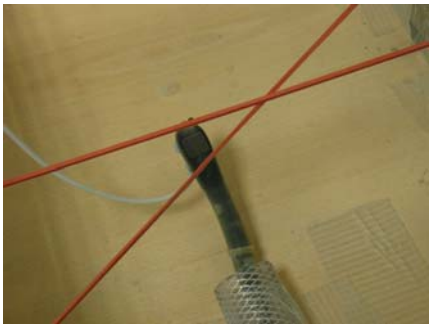


Figure 2. The image shows the cross phantom and the TEE probe. A 6 DOF EM sensor was firmly fixed on the TEE probe.

B. Calibration

The coordinates of the ultrasound images are related to the position of the ultrasound probe. To compound multiple ultrasound images together, it is essential to establish a coordinate system which is irrelevant to the position of the ultrasound probe. Presumably the coordinate system of the EM tracker did not move during the data acquisition, the ultimate target of the calibration is to find the matrix T which transforms the ultrasound image to the EM tracking space. Given the ultrasound probe position T_{probe} in the EM tracking space, matrix T can be defined as:

$$T = T_{probe} \cdot T_{us \to probe} \quad (1)$$

where $T_{us \to probe}$ is the transformation matrix which relates the ultrasound image coordinates to the position of the ultrasound probe. The purpose of the calibration is to calculate $T_{us \to probe}$.

C. Calibration phantom

The calibration procedure is similar to what was described in [6]. The calibration phantom is a 19-litre container filled with water. Two thin strings were placed in the middle of the container, forming a cross (Figure 2). A landmark was placed on one of the strings making the cross asymmetrical and therefore easier for the image registration. Two ultrasound volumes of the cross I_1 and I_2 were acquired from different angles. By using intensity based rigid body image registration algorithm, it is possible to find the transformation matrix $T_{us2 \to 1}$ which aligns image I_2 to I_1. Given P, a set of 3D points distributed evenly in the ultrasound volume, we have

$$T_2 \cdot P - T_1 \cdot T_{us2 \to 1} \cdot P \to 0 \quad (2)$$

T_1 and T_2 are the matrices which transform I_1 and I_2 to the EM tracking space. From equation (1) and (2), we have:

$$(T_{probe2} \cdot T_{us \to probe} - T_{probe1} \cdot T_{us \to probe} \cdot T_{us2 \to 1}) \cdot P \to 0 \quad (3)$$

As mentioned in [6], to optimize the solution to equation (3), we will need to acquire at least one more ultrasound image from a different angle, forming equation (4)

$$T_{probe3} \cdot T_{us \to probe} - T_{probe1} \cdot T_{us \to probe} \cdot T_{us3 \to 1} \cdot P \to 0 \quad (4)$$

A downhill optimizer was used to solve equation (3) and (4), finding the calibration matrix $T_{us \to probe}$.

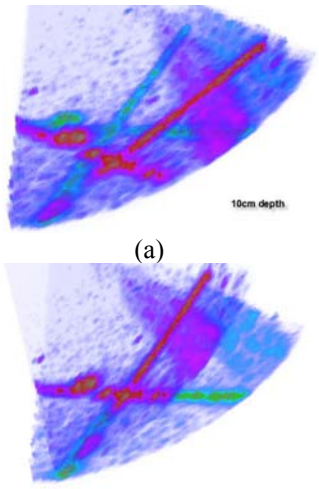
D. Accuracy validation

Figure 3. An ultrasound visible phantom was used to evaluate our compounding technique.

Two phantoms were used to validate the accuracy of our real-time compounding technique. The first phantom is the calibration phantom we used to calculate the calibration matrix. 12 ultrasound volumes were acquired from different angles and three of them were used to calculate the calibration matrix $T_{us \to probe}$. Then we used the matrix to compound the other images. After the calibration, we replaced the cross phantom with an ultrasound visible phantom shown in Figure 3 (Prostate phantom, http://www.CIRS.com). By using the same calibration matrix, we compound multiple ultrasound volumes of the second phantom together to create an extended field of view

ultrasound volume. The compounding errors were evaluated both quantitatively and visually. More detail for the accuracy validation will be discussed in section III.

III. Experimental Results



(a)

(b)

Figure 4 (a). Two ultrasound images of the cross phantom were acquired from different angles; (b). The two ultrasound images were aligned automatically by using an image-based rigid body registration algorithm.

In-house software were developed by using Microsoft Visual Studio 2005 (Microsoft, the USA) and Visualization Tookit (VTK, Kitware, New York, the USA) to acquired data simultaneously from the 3D ultrasound scanner and the EM tracking system. The real-time data export function and corresponding software library were provided by Philips Healthcare. The software library to receive tracking information from the EM tracker was provided by Northern Digital Inc. (http://www.ndigital.com). The rigid-body image registration algorithm was developed previously at King's College London [6]. Figure 4 shows two ultrasound volumes of the cross registered by using the image registration algorithm. Software used in this study was running on a laptop equipped with an Intel Centrino Dual Core processor (1.83MHz) and 2GB RAM.

12 ultrasound volumes of the cross phantom was acquired from different angles. 10cm depth setting was used for the ultrasound data acquisition. Three volumes were used to calculate the calibration matrix $T_{us\rightarrow probe}$. Figure 5 shows the TEE probe positions in the EM tracking space during the data acquisition. The arrows highlighted the positions where the ultrasound volumes were used for the calibration.
The calibration error was calculated by using the following equation.

$$\varepsilon_{cal} = (T_{probe2} \cdot T_{us\rightarrow probe} - T_{probe1} \cdot T_{us\rightarrow probe} \cdot T_{us2\rightarrow 1}) \cdot P$$

In this study, the calibration error was 1.8mm. The compounding error was calculated by using the other 9 volumes. By doing so, each of the 9 volumes was compounded with the volume acquired at probe position 1. A correction matrix was generated manually to correct any visible compounding error. For each of the 9 volumes, the compound error was calculated by:

$$\varepsilon_{cmp} = (T_{correction} T_{probeN} T_{us\rightarrow probe} - T_{probeN} T_{us\rightarrow probe}) \cdot P$$

The mean compounding error is 2.13±0.46mm.

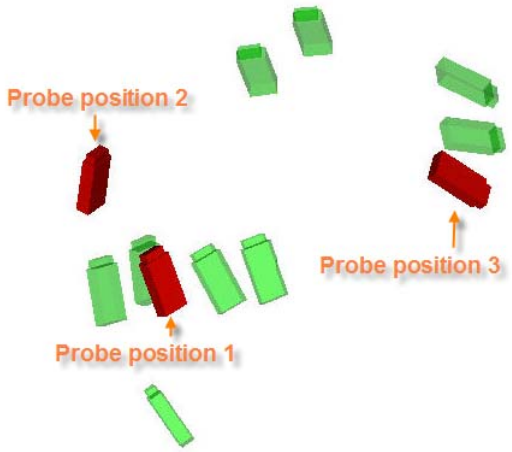


Figure 5. The orientations and the positions of the 3D TEE probe during the data acquisition. The arrows highlighted the positions where the ultrasound volumes were used for the calibration.

Figure 6 shows the ultrasound volume acquired at probe position 9 was compounded with the volume acquired at probe position 1.

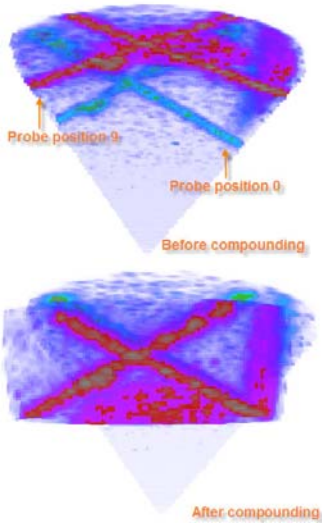


Figure 6. (a) Two ultrasound volumes of the calibration phantom acquired from different angles; (b). By using the calibration matrix, it is possible to compound the two volumes by transferring them into one coordinate system.

After the calibration, we replaced the cross phantom with an ultrasound visible phantom. 5 volumes were acquired from the phantom. By using the calibration matrix, the ultrasound volumes were compounded to create an extended

field of view ultrasound volume. The mean compounding error was calculated to be 2.4mm±0.53 (Figure 7).

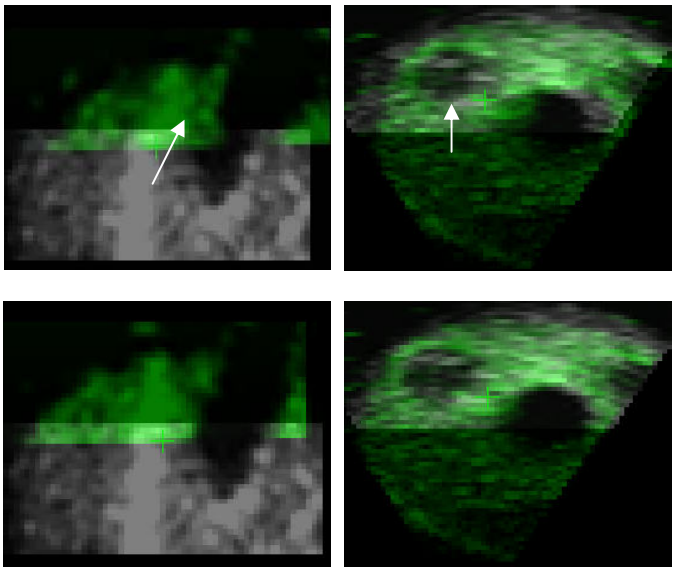


Figure 7. Several ultrasound images were compounded together to form an extended field of view ultrasound volume. The first row shows two examples of the compounding. The arrows highlight the compounding error. The second row shows the compounding errors were corrected manually. The manual correction was used to quantify the compound error.

IV. DISCUSSION AND FUTURE WORKS

3D ultrasound has the potential to be one of the best platforms for interventional cardiac applications. Compared to the other 3D imaging modalities such as CT and MRI, 3D ultrasound is low-cost, mobile and real-time. By using traditional TTE probe, the poor image quality and the limited field of viewer are two major barriers preventing 3D ultrasound from being widely used in interventional cardiac applications. The 3D TEE probe is our solution to the first problem. Compared to the TTE probe, the TEE probe can produce better quality images. However, as the TEE probe is normally closer to the target (normally 12cm depth setting compared to 18~20cm depth setting of the TTE probe), its field of view is even smaller than the TTE probe. In this study, we attempted to track the TEE probe by using a EM tracking system and used the tracking information to compound multiple TEE volumes together, creating an extended field of view ultrasound volume. The preliminary results suggest the EM tracking system is able to accurately track the TEE probe. The calibration and compounding errors are less than 2.5mm. Although the preliminary results seem to be promising, there are issues needed to be further evaluated. Firstly, the EM tracking system is sensitive to the presence of metal. So far, our study was carried out in a meta-free environment. But in reality many of the interventional cardiac applications, such as cardiac electrophysiology (EP) procedures are guided by C-arm X-ray systems. The presence of C-arm X-ray and the other medical devices will affect the accuracy of the EM tracking system. However, solutions to this problem have been proposed [8].

Secondly, the ultimate target of this study is to use our real-time compounding technique in clinical environment. How to integrate the EM tracking system with the current clinical pipeline seamlessly is another issue waiting to be solved. For example, in the case of cardiac EP procedures, how to visualize and overlay the 4D ultrasound volume onto 2D X-Ray images are interesting research topics worth a thorough investigation.

V. CONCLUSION

In this paper, we presented an efficient and accurate method to compound 3D TEE volumes. This method involves the use of an EM tracking system and a calibration technique developed in our previous work. We evaluated the accuracy of our method by using two different types of phantoms. The mean compounding errors for the two phantoms are 2.13mm and 2.4mm respectively. Although the preliminary results seem to be promising, further evaluation must be done before it can be used clinically.

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