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A Biventricular Multimodal (MRI/Ultrasound) Cardiac Phantom

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Abstract—**A cardiac phantom can be of crucial importance in the development and validation of ultrasound and cardiac magnetic resonance (MR) imaging and image analysis methods. A biventricular multimodal cardiac phantom has been manufactured in-house that can simulate normal and pathologic hearts with different degrees of infarction. The two-chamber structure can simulate the asymmetric left ventricular motion. Poly Vinyl Alcohol (PVA) is utilized as the basic material since it can simulate the shape, elasticity, and MR and ultrasound properties of the heart. The cardiac shape is simulated using a two-chamber acrylic mold. An additional pathologic heart phantom has been built to simulate aneurysm and infarction. Segmental dyskinesis is modeled based on three inclusions of different shapes and different degrees of elasticity. The cardiac elasticity is adjusted based on freeze-thaw cycles of the PVA cryogel for normal and scarred regions.**

I. Introduction

Heart disease is considered the leading cause of death in the modern world. The major cause of heart disease is vascular occlusion which leads to Ischemic Heart Disease (IHD). It is believed that decreased myocardial perfusion leads to decreased range of displacement of the myocardium and less contractility in the form of hypokinesia, akinesia, and dyskinesia. Other diseases such as cardiomyopathy or myocarditis can present as decreased cardiac motion as well. Clinicians often make use of echocardiography or other imaging modalities for the assessment of heart motion and the underlying myocardial perfusion. However the conventional myocardial motility scoring can be subjective and may suffer from inter and intra observer variability [1]. To solve this problem and as an aid for clinical decision making, registration algorithms are typically applied to the cardiac images. Unfortunately *in-vivo* validation of registration techniques are not straight forward since the ground truth motion field of the cardiac deformation pattern is not known exactly. Older validation techniques such as sonomicrometry or surgical markers are invasive and limited to just one point of the cardiac tissue. Additionally the invasive nature of the surgical implantation changes the local heart structure and properties. One solution to the problem is a controlled experimental phantom setup that can simulate the anatomy and physiology of the heart. Phantoms are easier to handle and can be validated using marker pellets.

To date, there have been a number of studies utilizing ultrasound cardiac phantoms, though typically comprised of simple shapes such as a tube or cylinder. Some examples are gelatin cubic elastography phantom [2] and rotating cylinder-shaped gelatin agar-agar phantom [3]. Claessen et al considered the left ventricle as a thick-walled truncated ellipsoid [4]. Their phantom was designed for apical imaging ignoring the other echocardiography views. Jia et al. [5] introduced another ellipsoidal left ventricular phantom which was able to acquire images from different views. A biventricular phantom limited to ultrasound imaging was recently introduced in [6] that is able to show the interventricular septal motion more accurately. As an exception, a live oxygenated rabbit heart based on Langendorff rabbit heart preparation was used in [7]. However live phantoms have a short life span and do not represent pathologic regions.

Some cardiac MR phantoms have considered applying simple [8], combinational [9] and dilational [10] transformations to a hollow silicon cylinder without fluid. However fluid filled media are crucial for ultrasound imaging and blood simulation. Recently, a multimodal biventricular phantom limited to normal heart was made available commercially (Shelley Medical Imaging Technologies, London, Ontario, Canada).

Cardiac phantoms are mostly based on one or several basic chemical materials such as gelatin agar-agar, silicone, PVA, or agarose. Gelatin-Agar products are not suitable for MRI studies or very dynamic experiments. Silicone products have been utilized as an efficient tissue simulator because of their longevity, stability, variable Young's modulus and the capacity to be marked by scatterers such as glass and plastic microspheres. However, these materials are limited by high attenuation and low speed of sound, reported to be less than 1000 m/s and are not perfect choices for ultrasound experiments [11,12,13,14].

To date, no MRI-compatible phantom has been described in the literature to simulate a two-chamber human heart with multimodal capabilities. In this article, we describe our development of a biventricular cardiac phantom which simulates the cardiac shape and biomechanics and is

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amenable to multimodal imaging. In addition to US, due to the appropriate MR properties of the material and absence of ferromagnetic objects in the set up, the phantom may be imaged with MRI. Additionally, a pathologic phantom representing regions of myocardial scar and aneurysm has been built. The heart phantom is placed inside a mediastinal phantom that has additional structures such as great vessels and esophagus. Therefore transesophageal view can be acquired as well.

II. Methods and Materials

An ideal ultrasound phantom should have attributes such as:

1. Anatomical similarity; i.e., shape of the LV, RV, wall thickness, and 3D orientation in the mediastinum.

2. Physiological/biomechanical similarity such as the elasticity and cardiac dynamics.

3. Pathologic properties such as the ability to model different degrees of scarring from infarctions and hypokinesia of the ventricular wall.

4. Modality properties such as the speed of sound, ultrasound attenuation, and T1/T2 relaxation properties of the myocardium.

5. Easy setup, low cost

In order to build the phantom, a cardiac computerized model considering the size of the left and right ventricles was designed and used to build an acrylic based cardiac mold. Polyvinyl alcohol (PVA) was utilized to mimic the mechanical and acoustical tissue properties of the heart. It has been shown that PVA can simulate the ultrasound attenuation and texture in the heart in addition to T1/ T2 properties in MR studies. PVA is more robust to stress in comparison to the traditional gelatin-based phantoms and can have a wide range of Young's modulus and elasticity based on the chemical reactions [11,12,13]. It is inert and safe even if digested and lasts a very long time without biological decay.

A 10% solution of PVA plus 1% enamel paint was used as the basic material. The enamel paint acts as ultrasound scatterers as well as the color of the heart. The PVA solution was stirred and heated up to 90 deg. C until it became clear. The heating should be in a closed space to decrease water evaporation and changes to the solution concentration. It is important to continue heating until the powder is fully dissolved. This process can vary from 0.5 to 3 hours depending on the PVA manufacturer. Highly hydrolyzed (>99.9%) products dissolve faster. Overheating the solution leads to faster dissolution of the powder but destructs the chemical structure of the molecules. Practically, it is better to discard the final superficial thick layer of the solution in order to decrease inhomogeneity. Subsequently, the solution was gradually cooled down from 80 deg. C to room temperature. Subsequently, it was poured into the cardiac mold and left for 12 hours to extract the bubbles. The manipulations were minimized after this step since any additional manipulation can cause air bubbles. The solution was then gradually exposed to the temperature of –20 deg. C until it froze. The mold and the solution were kept in that temperature for 24 hours. At that time the molecules in the PVA solution were cross linked with each other to make a tougher material called cryogel. Finally, the mold and the frozen gel were gradually exposed to the room temperature to avoid any additional inhomogeneity in the chemical process of the cryogel. At this point, the normal heart phantom has passed one freeze-thaw cycle.

An additional model consisting of the left and right ventricles but with a segmental thin wall in the LV was used to build an additional mold for a pathologically scarred heart. The thinner wall was designed to mimic an aneurysmal dyskinetic motion. Three PVA-based inclusions were separately made as a circle, slab, and cube using nine, six and three freeze-thaw cycles respectively. Each freeze-thaw cycle decreases the elasticity of the heart phantom to mimic scarred myocardium. The attenuation of the PVA and speed of sound increase after each freeze-thaw cycle [11,12,13]. The circular, slab like, and cube like objects were placed in anteromedial, anterolateral, and apical segments of the heart model within the mold. Subsequently, the PVA solution was added to fill the rest of the space in the mold. After one freeze-thaw cycle, the abnormal heart consisted of a background of normal texture with one freeze-thaw cycle plus three infarct-mimicking inclusions having 10, 7 and 4 freeze-thaw cycles. In this set-up, the anterolateral and anteromedial infracted regions surround the aneurysmal part as in the human heart. The mediastinal bed that mimics the great vessels and provides the ability to acquire trans-esophageal images was manufactured with a third mold using a single freeze-thaw cycle.

Since the PVA cryogel properties have been reported in several previous studies [11,12,13], they were not measured in this study . According to previously published studies, the speed of sound in PVA is 1527, 1540, 1545, and 1550 m/s respectively, and ultrasound attenuation is 0.4, 0.52, 0.57, and 0.59 db/cm, respectively for 1, 4, 7 and 10 freeze-thaw cycles. Additionally, T1/T2 is 980/820, 690/605, 540/500, and 520/480 ms for 1, 4, 7, and 10 freeze-thaw cycles [11,12,13].

Finally, two latex balloons were placed in the ventricles to dilate and contract the phantom and a mechanical syringe was used as the actuator. The syringe was attached to the balloons using two long connector tubes so that the piston could be kept away from the bore of the MR scanner. The volume and pressure of each ventricle could be adjusted by using two separate valves controlling the flow to each ventricle. The manual forward and backward motion of the syringe changes the internal pressure of the balloons and therefore the ventricles contract and expand. This structure provides an easy-to-setup, inexpensive approach for dynamic US echocardiography image acquisition. It also supports static MRI image acquisition at different internal balloon pressures. However dynamic MRI image acquisition necessitates gated imaging, requiring an additional trigger signal (mimicking an ECG) which we are currently pursuing.

Pure water is not a good simulator of blood since it is less viscous and lacks the blood particles that act as ultrasound scatterers. In this study, a solution of 50% water and 50% glycerol was used to mimic the blood since

glycerol is able to simulate blood viscosity and ultrasound scattering. The heart phantom was placed in the mediastinal bed and the mediastinum phantom was fixed in a container. The acoustic impedance changes between the container and air can cause artifacts. For ultrasound applications, the container was covered by a polystyrene pad to match the acoustic impedance with the air and decrease back scattering. With the proposed experimental set-up, it is possible to acquire images from any direction including apical and transesophageal views.

III. Results

Figure 1 shows the heart model that was used to build the acrylic cardiac mold. The dimensions were selected close to an average adult human heart. The final cardiac phantom is shown in figure 2(a). Image acquisition was performed using different echocardiographic views on a Philips iE33 workstation. 2D and TDI images were acquired and analyzed for validation studies. MRI images were collected using a 1.5 T Philips Achieva scanner. Six 3D images were acquired at different inflation pressures using 3D T1 weighted FFE, TE/TR 5/25 ms, FA 30^o, slice thickness 3 mm, spatial resolution 0.625×0.625×1.5 mm, 3D FOV 224×224×189 mm, and number of slices=63. An additional volumetric 3D image was acquired using balanced FFE, TE/TR 1/3 ms, FA 60^o, spatial resolution 1.70×1.70×11.35 mm, 3D FOV= 176×176×272 mm, and number of slices=24.

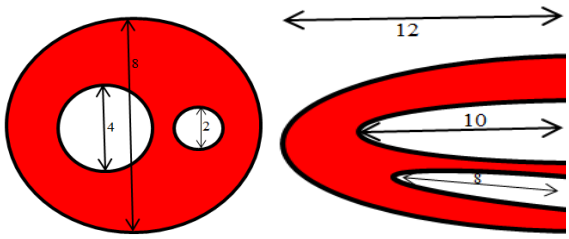
Figure 2(b) shows a two chamber short-axis B-mode image acquired from the phantom. The enamel paint particles are strong scatterers and are visually more pronounced on the B-mode images and can be used as landmarks. Since no marker is perfectly limited to just one pixel on the B-mode images, we use the center of mass of each particle as the landmark.

Short axis and long-axis MRI images are respectively shown in figure 3. The high signal regions in figure 3(a) and the low signal region in figure 3(b) show the regions that simulate the scar. Figure 4 shows the TDI images of the phantom during different phases of the cardiac cycle. Table 1 shows the quantitative analysis of different mid-LV American Heart Association (AHA) segments using TDI data in end systole, mid-systole and end- diastole. We are planning to analyze the phantom motion using different pressure waveforms and compute the regional strain values as an additional validation of the phantom dynamics.

Scope of use: The phantom can be used for the validation of motion detection algorithms. As an example in our work, implementation of the block matching technique on the phantom images shows 0.44 pixel/frame magnitude error and 40.03 degrees/frame angular error compared to the manual landmark tracking. Figure (5) shows the RV and LV volumes (after a small pressure pulse) during different phases of the cardiac cycle measured by manual segmentation. The range of myocardial velocities and cardiac volumes can be adjusted by changing the pressure in the balloons.

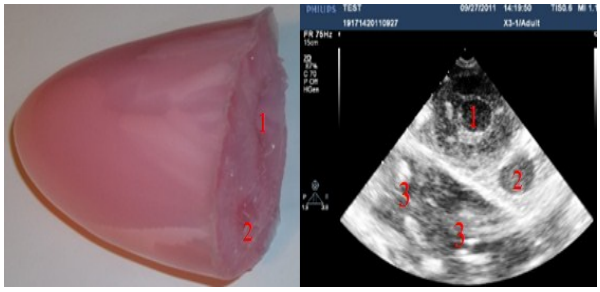
IV. Conclusion

A biventricular PVA based heart phantom was manufactured that mimics the cardiac shape as well as the MRI and ultrasound imaging properties. Pathologic conditions were simulated in an additional phantom simulating aneurysm and scarred regions. The scarred tissue was simulated using stiffer PVC polymers. The two-chamber cardiac structure allows for realistic asymmetric LV motion. All the classical views can be captured in this phantom. Additionally, since the heart phantom is positioned in a mediastinal phantom, it provides the possibility of acquiring transesophageal images. As a future work, we will manufacture a four chamber heart phantom including the valves and analyze the cardiac dynamics based on extensive studies such as tagged MRI images.



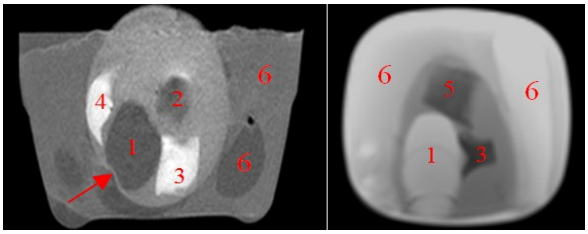
(a) (b)

Figure 1. Biventricular heart model dimensions in cm, (a) Short-axis view, (b) Long-axis view.



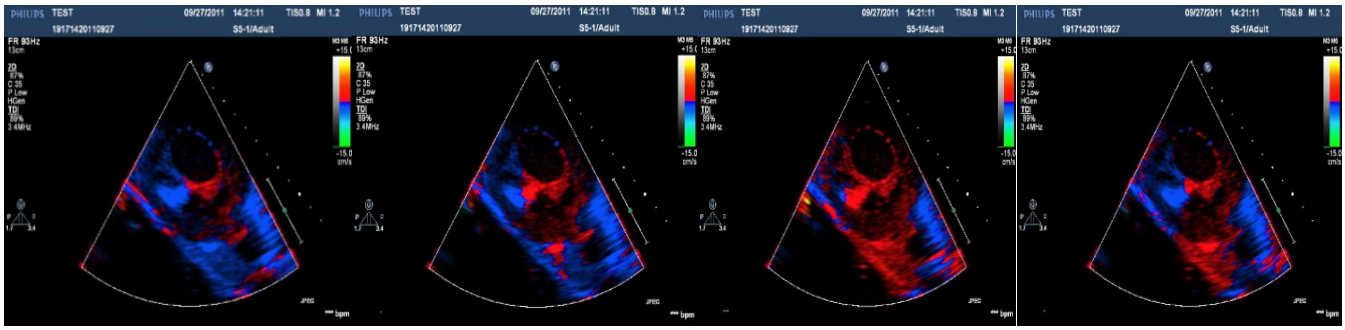
(a) (b)

Figure 2. (a) A picture of the two-chamber phantom (b) Two-chamber B-mode echocardiography image (1: LV, 2: RV, 3: mediastinum and mediastinal structures)



(a) (b)

Figure 3. (a) A static MRI slice of the phantom using T1 weighted FFE, The arrow points to the aneurysm (the thin ventricular wall) (b) A static slice of the phantom using balanced FFE (1: LV, 2: RV, 3: cylindrical inclusion, 4: slab-like inclusion, 5: cube like inclusion, 6: mediastinum and mediastinal structures)



(a) (b) (c) (d)

Figure 4. TDI (Tissue Doppler Imaging) images of the moving phantom, (a and b) systole, (c and d) diastole. Blue indicates the motion away from the transducer and red indicates the motion toward the transducer.

Table 1. Regional TDI motion (cm/s) of the mid ventricular segments in end-diastole, mid-diastole, and end-systole

	Mid antero-septal	Mid anterior	Mid antero-lateral	Mid infero-lateral	Mid inferior	Mid infero-septal
End-systole	-2.1	-0.2	-0.2	0.1	2.4	-2.9
mid-diastole	1.0	0.05	0.07	1.7	2.6	-0.7
End-diastole	0.5	0.4	0.5	2.3	3.2	1.4

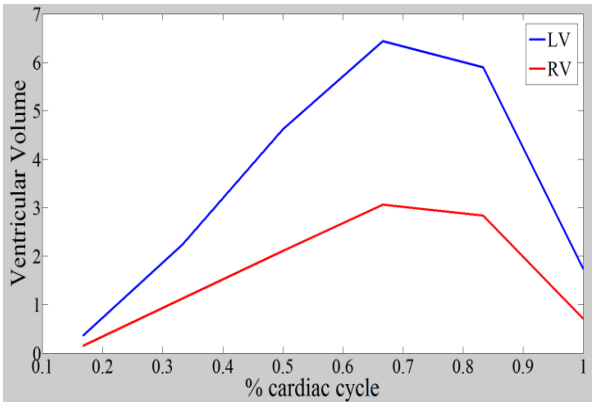


Figure 5. RV and LV volume in cc during different phases of the cardiac cycle

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