Comparison of Fusion Techniques for 3D+T Echocardiography Acquisitions from Different Acoustic Windows

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Abstract

We propose a new method to combine real-time 3D echocardiography acquisitions from different apical windows. It consists in registering the volumes using only the image data, without external positioning sensors, and combining their intensity values, without requiring a feature extraction method. Registration shows reasonable quantitative results on phantom data and visually acceptable results on in vivo *data. We present two new methods for the fusion process: generalized averaging and multiview deconvolution. The latter, which performed best, improved the myocardium contrast by +49.67% and the signal-to-noise ratio by +4.61 dB. These improvements lead to more accurate quantification of cardiac function.*

1. Introduction

Echocardiography offers many advantages compared to other imaging modalities, such as being non-invasive, real-time, and cost-effective, but suffers at the same time from a limited field of view, shadows, drop-out and a poor signal to noise ratio. Recently introduced into the market, real-time three-dimensional (RT3D) echocardiography systems use a fully sampled matrix array to image volumes in a single acquisition. Each acquisition is completed within four consecutive cardiac cycles, triggered by the electrocardiogram (ECG).

In this paper we present a technique to enhance imaging capabilities of RT3D echocardiography. The goal of the proposed method is to combine volumes from different acoustic windows, to: (i) increase the field of view from individual acquisitions, (ii) complement missing features such as myocardium segments with low contrast, and (iii) attenuate image artifacts such as speckle. Improving these image quality factors can lead to improved pathology diagnosis and more accurate quantification of cardiac function such as left ventricular volume (LVV).

The combination of echocardiographic acquisitions from different acoustic windows has been addressed in the literature for linear arrays. Legget *et al.* [1] used a free-hand 2D system, consisting in a linear probe with a magnetic-positioning sensor. The LV surface is reconstructed using the features extracted from the different views. Ye *et al.* [2] used a rotational 2D probe which generates a sparse 3D reconstructed volume. Features were extracted by a phased-method based algorithm and used to refine the registration given by the position sensor. Features, weighted by the viewing geometry, were combined to estimate the LVV. To our knowledge, the combination of acquisitions from different acoustic windows has not been addressed for RT3D echocardiographic images. The significant overlap of the different volumes enables their registration without the need of an external position probe.

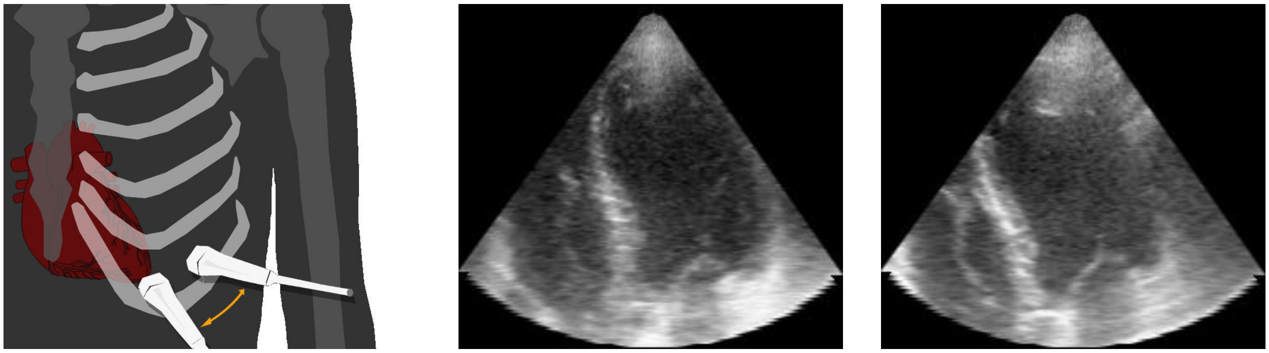
Figure 1 shows an example of two acquisitions from different apical windows. The proposed method consists in registering these different views and fusing their intensity values into a single volume. The proposed registration and fusion methods are described in Section 2 and 3 respectively. Results are reported in Section 4, and conclusions are drawn in Section 5.

2. Registration

The goal of the registration process is to find the geometrical transformation T that makes the two views the most similar, according to a distance metric μ. We define μ as:

$$\mu(v_1, v_2 \circ T) = \frac{1}{n(T)} \sum_{\mathbf{x} \in \mathcal{O}(T)} \rho\left(v_1(\mathbf{x}) - v_2 \circ T(\mathbf{x})\right) \quad (1)$$

where v_1 is the reference volume, v_2 is the moving volume, $\mathbf{x}$ are the voxel coordinates, $\mathcal{O}$ is the overlapping area between the two volumes and n is the number of voxels within $\mathcal{O}$. In order to overcome problems of contrast drop-out and noise, ρ is chosen to be a robust estimator to cope with outliers. For our experiments we used a Huber function [3].



(a) (b) (c)

Figure 1. a) Schema of two different apical window probe positioning, b) slice of RT3D centered apical view, c) slice of RT3D displaced apical view. The goal of our method is to combine views b) and c) to increase the field of view, improve wall contrast and attenuate artifacts.

We restrict T to be a rigid transformation to account for the different probe positions. Since our interest is in evaluating wall movements, we avoid using non-linear registration schemes, e.g. elastic, which could introduce spurious movements not present in the original acquisitions if not accurately regularized.

An initial estimation of T is needed to avoid local minima of μ and to ensure a significant overlap between the two views. To obtain a robust initialization, a LV segmentation is performed at each view. Five points (four in the mitral annulus and one in the apex) are manually selected in each view, and a segmentation mesh is obtained with the commercial software QLab (Philips, Best, The Netherlands). The initial rigid transform is found by minimizing the mean distance between the points of the two generated meshes. Figure 2 shows as example of these meshes.

Once the volumes are initialized, we use a gradient descent algorithm to find the parameters of T which minimize μ. If n is sufficiently large, the gradient of μ can be approximated as:

$$\frac{\partial}{\partial T}\mu(v_1, v_2 \circ T) \approx \frac{1}{n(T)} \sum_{\mathbf{x}\in\mathcal{O}(T)} \rho'\left(v_1(\mathbf{x}) - v_2 \circ T(\mathbf{x})\right) \nabla\left(v_2 \circ T(\mathbf{x})\right) \cdot J(T) \tag{2}$$

where $(\bullet)'$ is the derivate, $\nabla(\bullet)$ is the gradient vector and $J(\bullet)$ is the Jacobian matrix.

Registration of the views acquired at different cardiac cycles rely on the regularity of the heart movement, which showed to be reasonable in most of our experiments. Cases that showed to be irregular within the four cardiac cycles of acquisition or that did not have the same number of frames per cycle were discarded. The registration process was performed for the first frame of the sequence, triggered by the ECG, and used for the whole cardiac cycle.

3. Fusion

Once the images are registered, the fusion process generates a single volume which contains the two (or more) views. Our approach is to fusion the intensities of the volumes, rather than extracted features. We evaluate the maximum (MX) of the two images along with two novel methods, generalized average (GA) and multiview deconvolution (MD). From now on, v_2 denotes the registered volume.

3.1. Generalized Average

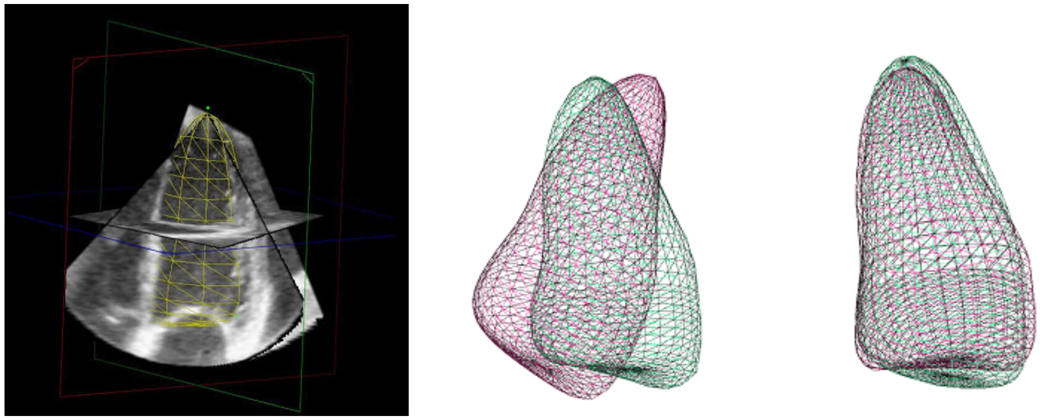
We can basically distinguish between three data fusion scenarios: (i) a feature is present in only one view, (ii) pure speckle patterning in both views and (iii) the same feature is present in both views. To these cases, the appropriate operator should be, respectively: (i) maximum, to keep the feature, (ii) average, to reduce the noise figure and (iii) minimum, to keep the best spatial resolution. Generalized averaging provides a smooth way to switch from minimum to average and to maximum operators in function of data disagreement $\beta(\mathbf{x})$.

$$v_f^{GA}(\mathbf{x}) = \left(\frac{1}{2}\left(v_1(\mathbf{x})^{\beta(\mathbf{x})} + v_2(\mathbf{x})^{\beta(\mathbf{x})}\right)\right)^{1/\beta(\mathbf{x})} \tag{3}$$

The $\beta(\mathbf{x})$ term is computed as:

$$\beta(\mathbf{x}) = \gamma\left(|\bar{v}_1(\mathbf{x}) - \bar{v}_2(\mathbf{x})| - \mathrm{corr}(v_1(\mathbf{x}), v_2(\mathbf{x}))\right) \tag{4}$$

where $\bar{\bullet}$ represents a local mean, $\mathrm{corr}(\bullet, \bullet)$ represents the correlation operator on a neighborhood around $\mathbf{x}$ and γ is



(a) (b) (c)

Figure 2. a) Example of segmented left ventricle, b) meshes at their original location, c) meshes after initial registration

set to accommodate the range of the generalized average operator ($\gamma = 10$ in our experiments).

3.2. Multiview Deconvolution

We proposed in [4] a technique to reconstruct a high resolution volume from anisotropically degraded scans. It consists in solving the regularized inverse problem given the point spread function h_i of each acquisition, as:

$$v_f^{MD}(\mathbf{x}) = \arg\min \Big(\|v_1(\mathbf{x}) - h_1(\mathbf{x}) * v_f(\mathbf{x})\|^2 + \|v_2(\mathbf{x}) - h_2(\mathbf{x}) * v_f(\mathbf{x})\|^2 + \lambda \|\nabla v_f(\mathbf{x})\|^2 \Big) \quad (5)$$

The different h_i are assumed to be Gaussian, with variance linearly depending on the distance to the probe r, $h_i = G(\mathbf{0}, ar(\mathbf{x}))$. The parameter a is estimated using the cross-channel relationship [5] as:

$$a = \arg\min \|h_1 * v_2 - h_2 * v_1\|^2 \quad (6)$$

4. Results

Data was acquired on eight patients with a Philips Sonos 7500 with Live3D echo, using a x3-1 matrix array. An example of registration results are plotted in Figure 3, where can be observed that manual initialization is slightly misaligned at the septum wall region. On the other hand, automatic registration shows reasonable performance. Validation on phantom data (ATS539), acquired with a robotic arm, showed the overall accuracy of the system was $\Delta t = 0.68 \pm 0.18$ mm for translations and $\Delta\alpha = 2.29° \pm 0.82$ for angular displacements. Quantification of registration results on *in vivo* data is on-going.

Figure 4 shows the fusion results with the three commented techniques, maximum, generalized average and multiview deconvolution. To quantitatively evaluate the fusion process, we defined the myocardium contrast increase as $\Delta C = \left(\left| \frac{\bar{M}_f - \bar{B}_f}{\bar{M}_1 - \bar{B}_1} \right| - 1 \right) \cdot 100$, where $\bar{M}$ is the mean value in a manually defined window at the lateral, anterior and posterior myocardium walls, and $\bar{B}$ the mean value in a manually defined window in the LV cavity. The improvement of signal-to-noise ratio is defined as $\Delta SNR = 20 \log \frac{\sigma_1}{\sigma_f}$, where σ is the noise variance in manually defined window in the LV cavity, which is supposed to have a constant value. Table 1 shows the value of these parameters with the proposed fusion techniques.

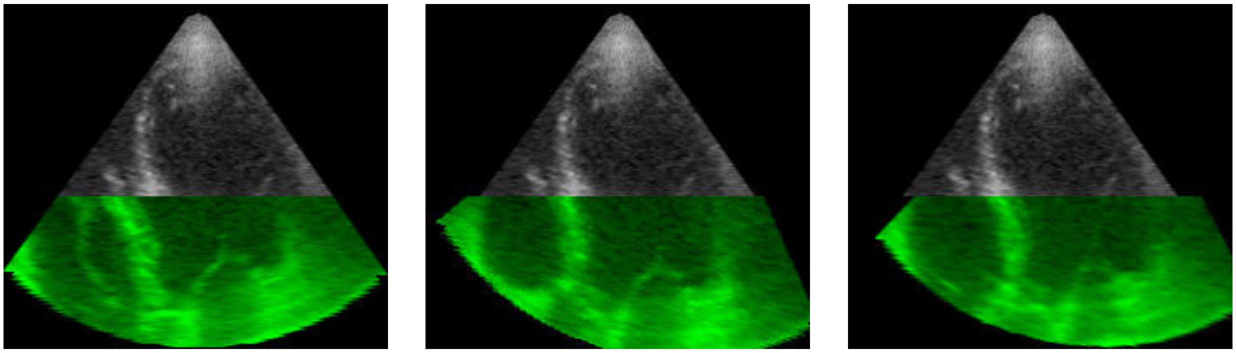
Table 1. Image quality parameters for the different fusion techniques

Fusion	ΔC	ΔSNR
MX	+38.98%	+3.87 dB
GA	+45.32%	+4.74 dB
MD	+49.67%	+4.61 dB

The field of view, measured as $FOV = \left(\frac{n_f}{n_1} - 1 \right) \cdot 100$, was improved by +18% on average for two acquisitions. This value depends uniquely on the position of the probe, thus, it is independent of the fusion technique.

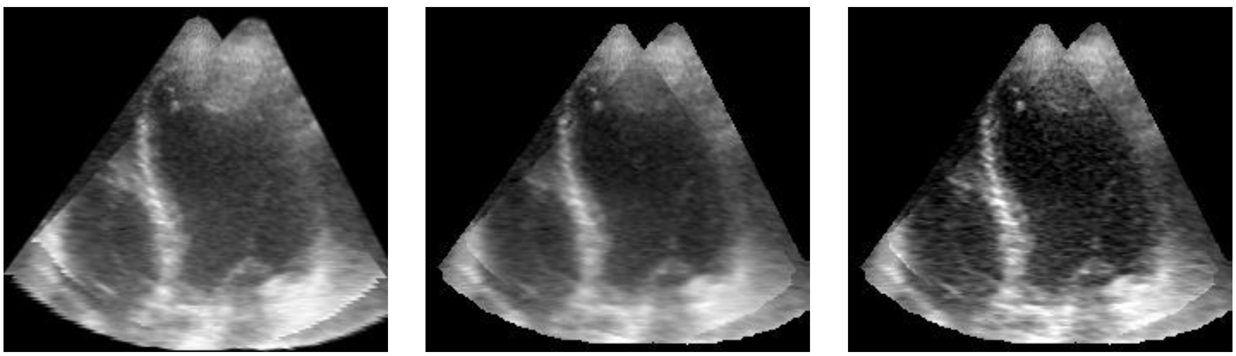
5. Discussion and conclusions

The major benefit of this technique, improving the field of view, shows to be possible without the need of an external position sensor on RT3D echocardiology. All three presented fusion techniques improve the wall contrast of the myocardium, the multiview deconvolution scheme being slightly better. This is permits to see all heart walls within one volume, and enables quantification parameters



(a) (b) (c)

Figure 3. In all images, top (white) is the reference volume v_1. Bottom (green): a) original displaced acquisition, v_2, b) v_2 mesh-based registered, c) v_2 automatically registered.



(a) (b) (c)

Figure 4. Fusion of displaced apical windows. a) Maximum, b) generalized average, c) multiview deconvolution

such as ejection fraction for dilated heart patients. Similarly, the noise figure is diminished by the combination of acquisitions from different points of view, and the generalized average and the multiview deconvolution have comparable performances, slightly better than the maximum operator. These improvements reflect a better tissue delineation which help doctors better asses wall motion asynchrony pathologies. Registration process shows promising results both quantitatively on phantom data and by visual inspection on *in vivo* data. Further work includes the combination of volumes from parasternal and apical windows. It is believed that the multiview deconvolution algorithm will show further improvement due to the bigger disparity of point spread function of the acquisitions.

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