Reconstruction of 3D Flow from Multiple Echo Doppler Views

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Abstract

We present a new method for reconstructing a 3D vector field from multiple 3D Pulsed Wave Doppler echo images. A weakness of Doppler imaging is that only a 1D projection of the true 3D velocity (the component of the velocity parallel to the echo beam direction) is measured. We propose a method to use registered multiple Doppler views to calculate 3D flow vectors using a Least Mean Squares (LMS) minimisation process. Spatial and temporal averaging are used to improve reconstruction accuracy. We investigate the effect on accuracy caused by changes in view angle. We relate our work to clinical practice by using noise values from real data. We report experiments on simulated and phantom data. Simulation data results show that when angles between views are greater than 40° (which can be clinically achieved), 3D flow may be reconstructed with an error of approximately 15% velocity magnitude and 15° vector angle. Phantom data results support these findings.

1 Introduction

Many cardiac abnormalities are characterised by abnormalities of blood flow which can include regurgitation and narrowing of heart valves resulting in abnormal direction and velocity of blood flow patterns. Thus, a 3D characterisation of cardiac blood flow may improve diagnosis of some cardiac diseases. In addition, such information would be useful input to help constrain patient specific cardiac models or to help validate such models.

The recent introduction of 2D matrix array technology allows rapid acquisition of 3D B-Mode and 3D Doppler volumes. Nevertheless, Doppler velocity information is only a 1D projection of the true 3D velocity vector into the echo beam direction. Crossed-beam techniques, which use multiple views to reconstruct the full velocity information, have the potential to overcome this limitation [2]. The first crossed-beam approach [3] used simultaneous acquisition of several 2D Doppler images to compute an instantaneous 3D flow. Xu *et al*. [5] reconstruct 2D flows using more than two views, and improve reconstruction quality using two averaging methods. Arigovindan *et al*. [1] proposed using 2D B-Splines and regularisation to reconstruct smooth 2D flow.

Previous work assumes that the transformation between views is accurately known by calibration. This is generally not true, instead we propose using image registration to calculate these transformations. The accuracy of 3D flow recovery will depend on a number of factors, e.g. noise in Doppler images, registration accuracy, angle between views, velocity range and presence of velocity aliasing. We propose that two major factors are: the noise in Doppler images; and the angle between views (restricted by anatomy). Our experiments use noise calculated from real data and investigate methods to improve the SNR, and we explore the dependence of view angles on reconstruction accuracy. Our aim is to ascertain the feasibility of the reconstruction under clinical conditions of noise and angular limitation.

2 Description of the Method

We initially describe how our method calculates flow vectors from n 3D volumes, by firstly registering the images, and then reconstructing the velocity vector fields. We then describe two clinically compatible methods to improve SNR in input images.

2.1 Image Registration

Echo Doppler images have a reduced Field of View (FOV), which makes registration between views difficult. In order to achieve accurate registrations, we acquire both large FOV B-Mode images and Doppler images at the same probe position which are registered together. We then register the large FOV images acquired from different views using a phase-based registration algorithm [4].

2.2 3D Vector Field Reconstruction

In the general case, we may have more than three echo Doppler datasets. We extend to 3D the method by Xu et al [5] for fusing multiple velocity images based in LMS optimisation.

$$\begin{bmatrix} m_1 & \dots & m_N \end{bmatrix}^\top = \begin{bmatrix} \vec{d}_1 & \dots & \vec{d}_N \end{bmatrix}^\top [\vec{v}] + \begin{bmatrix} g_1 & \dots & g_N \end{bmatrix}^\top \Longrightarrow M = D \cdot \vec{v} + G \quad (1)$$

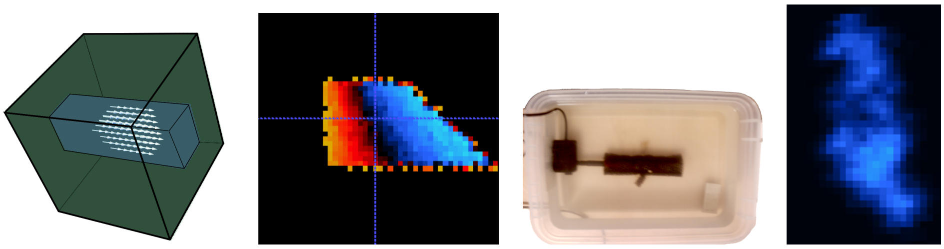
where M is the measured velocity along the beam direction D, $\vec{v}$ is the true velocity vector, N is the number of images and G models the additive Gaussian noise. Applying the LMS algorithm, leads to a linear system:

$$A\vec{v} = b \Longrightarrow \vec{v} = A^{-1}b \quad (2)$$

2.3 Methods to improve SNR

We propose two methods to improve SNR which could fit within a clinical workflow i.e. restricting our input data to available echo views and exploiting the cyclic nature of the cardiac sequences:

1. Temporal Averaging (TA). We propose averaging successive cardiac cycles from one probe position to improve SNR before vector reconstruction. We use an averaging



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

Figure 1: Simulated flow model (a) and resulting Doppler image (b). Flow phantom (c) and acquired Doppler image (d).

method which weights the contribution from frames based on the inverse of the local variance (3×3×3 neighbourhood).

2. Spatial Averaging (SA). We propose acquiring multiple acquisitions from approximately the same view, which are all then input into eq. 2 for vector reconstruction.

3 Experiments

In echocardiography, acoustic windows of the human chest permit acquisition of 3 standard views: apical view (AV), parasternal long view (PLV) and parasternal short view (PSV). These are three independent views in 2D. However, in 3D, both parasternal views are the same except for a 90° probe rotation. Nevertheless, the clinician may change the axis between PLV and PSV views to produce an angle between parasternal views of α_p. The AV axis is approximately orthogonal to both the PLV and PSV axes.

Our experiments are designed to characterise the reconstruction error with respect to the angle α_p. Also, we know that the AV is such that the main flow component through the mitral valve (a typical region of interest for our proposed method) is approximately aligned with the beam direction. Thus, in our experiments we place the AV at a 20° angle to the flow direction and the other views (PLV and PSV) orthogonal to AV and spread at an angle α_p.

3.1 Experiments on Simulated Data

A synthetic phantom was used to measure the impact of noise for different values of α_p on 3D vector accuracy. Our simulated flow model is a rectangular tube with a laminar constant flow (fig 1(a)). In our simulations, we added zero mean Gaussian noise of standard deviation $\sigma = 20\%$ of the real velocity value. This σ value was calculated from real data by comparing corresponding frames from a single view position. We measured σ to be between 10 and 20% of the velocity value, compared to the 9% reported in 2D [1]. The AV was placed at 20° with respect to the flow direction. The angle α_p was given values from 10° to 110° in intervals of 10°. For each configuration, the 3D flow was reconstructed in 4 different ways: 1) using 3 views with no TA (temporal averaging) or SA (spatial averaging); 2) using 3 views and another 3 views at similar but not exactly the same positions and then applying SA; 3) using 3 views with 3 temporal cycles applying TA; and 4) Combining 2) and 3). We report magnitude and angle errors with respect to the theoretical flow, and results show the average of 5 repeat experiments.

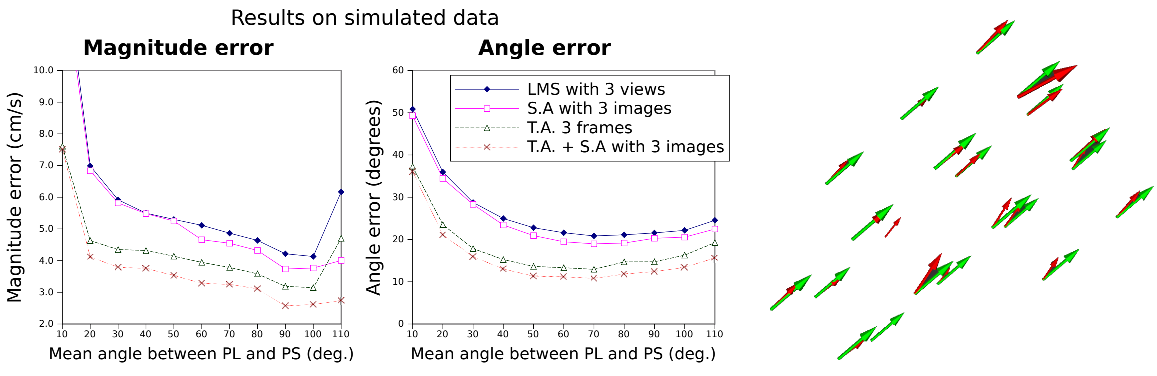
3.2 Experiments on Flow Phantom

Our flow phantom consisted on a reticulated foam pipe lined with a thin layer of latex submerged into a water tank, where an pump injected a constant water flow (fig. 1(c)). We acquired 25 3D+T images, with at least 3 temporal cycles at each position, from approximately uniformly distributed positions which were classified as AV, PLV and PSV. AV images had their beam direction aligned at approximately 20° to the flow direction. The PLVs and PSVs were at an angle of between 70° and 120° to the AV. Images were grouped into sets of three views (one AV, one PLV and one PSV, the angle between the PLV and the PSV ranged from 15.4° to 91.6°). For the SA experiments an additional set of AV, PLV and PSV were used, acquired at approximately the same position as the other three views.

As the true velocity of the flow was not known, flow errors could not be calculated as in 3.1. Instead, a synthetic Doppler image was produced from the reconstructed vector field. This was then compared with independent data acquired at the same probe position. Intensity (i.e. projected velocity value) difference was used as an error measure.

4 Results

Results are presented in fig.s 2 and 3 for the simulated and phantom experiments respectively. From our experiments on simulated data there appears to be three intervals of α_p: 1. $\alpha_p <$ 40°, large errors which rapidly decrease; 2. interval where errors values do not change significantly with respect to angle; 3. $\alpha_p > 90°$, large angles where errors increase again. A significant finding is that clinically obtainable angles lie within interval 2. This finding is also supported by our phantom experiments (fig. 3(a)). In both simulated and real data SA and TA improves the reconstruction (fig. 2(a) and 3(b)). By combining both SA and TA leads to further improvements (almost 50%). Good consistency was observed between vector fields constructed from independent data sets using the phantom data (fig. 3(c)).

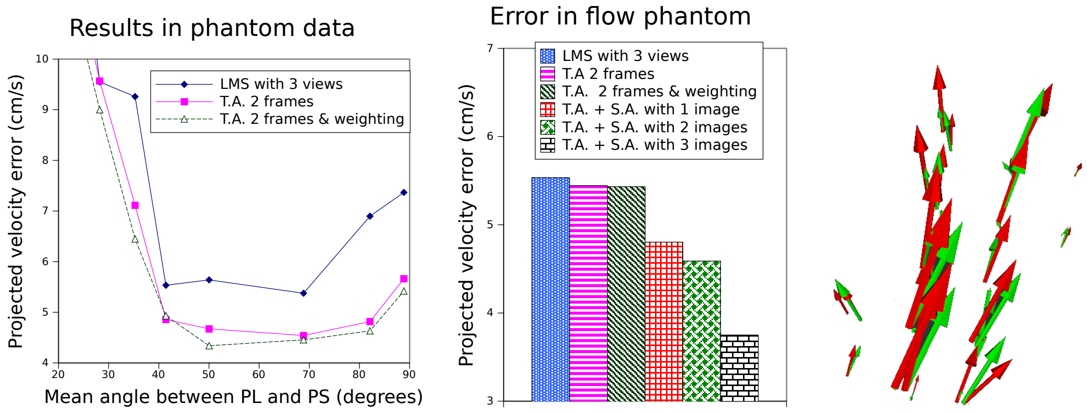


(a) (b)

Figure 2: Magnitude and angle errors for different SNR improvement methods(a). Flow vectors reconstructed from simulated data for $\alpha_p = 70$ (red) and true flow (green) (b)

5 Conclusions and discussion

We have carried out a sensitivity analysis of 3D flow reconstruction with respect to angle between views, α_p. Our results using both simulated and phantom data show that recon-



(a) (b) (c)

Figure 3: Flow phantom results. Improvement of accuracy due to TA on phantom data (a). Performance of TA and SA on phantom data for a given angle of 40 degrees (b). Flow vectors reconstructed from phantom data, with $\alpha_p \approx 90$ (green) and $\alpha_p \approx 70$ (red) (c).

struction error remains approximately constant if the angle between two parasternal views α_p is between 40° and 90°. In discussion with clinicians we believe that it is possible to achieve a value of $\alpha_p > 40°$ in a clinical acquisition. We have used realistic levels of noise in our experiments and investigated the use of two clinically compatible strategies to improve SNR. These were able to improve the reconstruction accuracy by up to 50%.

Flow can only be reconstructed where all the Doppler images intersect. Thus, only small structures can be targeted. However, our interest lies in imaging small, rapidly moving structures, such as valves, which will benefit from the high temporal resolution of echo.

Echo Doppler-based 3D flow reconstruction is a novel technique and needs further improvement. Future work will include validation on clinical data, incorporation of physical knowledge of flow behaviour to the problem and extension to 3D+T flow recovery.

References

[1] M. Arigovindan, M. Suhling, C. Jansen, P. Hunziker, and M. Unser. Full motion and flow field recovery from echo doppler data. *IEEE Trans. Med. Imag.*, 26(1):31, 2007.

[2] B. Dunmire, K. W. Beach, K-H Labs, M. Plett, and D. E. Strandness. Cross-beam vector doppler ultrasound for angle-independent velocity measurements. *Ultrasound in Medicine & Biology*, 26(8):1213–1235, 2000.

[3] M. Fox. Multiple Crossed-Beam ultrasound doppler velocimetry. *IEEE Trans. in Sonics and Ultrasonics*, 25(5):281–286, 1978.

[4] V. Grau, H. Becher, and J. A. Noble. Registration of multiview real-time 3-D echocardiographic sequences. *IEEE Trans. Med. Imag.*, 26:1154–1165, 2007.

[5] S. Xu, H. Ermert, and R. Hammentgen. Phased array pulse doppler tomography. In *IEEE Proc. Ultrasonics Symposium*, pages 1273–1276, 1991.