Deciphering the ventricular intracavitary blood flow by ultrasound imaging

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ABSTRACT

We generalized and improved our technique of two-dimensional intraventricular vector flow mapping (2D-*i*VFM) for a full-volume three-component analysis of the intra-ventricular blood flow (3D-*i*VFM) by triplane color Doppler imaging. We constrained a penalized least-squares minimization problem with fluid dynamics properties, such as mass conservation and free-slip endocardial boundary conditions. The problem was reduced to a linear system by the use of finite differences and Lagrange multipliers. We selected the penalty (smoothness) parameter automatically through the *L*-curve method, which aims to provide a bias-variance trade-off. We tested triplane 3D-*i*VFM on a patient-specific heart-flow CFD (computational fluid dynamics) model. Our preliminary results showed that triplane-derived 3D-*i*VFM can decipher the 3-D vortical flow that forms during left ventricular filling, and thus could offer clinically compatible new insights in diastolic function.

1. INTRODUCTION

1.1 Clinical context

The left intraventricular hemodynamics is characterized by a vortex that is created when blood flow passes through the mitral valve during diastole (filling of the ventricle). This vortex tends to redirect blood to the aorta, which is the natural route. In healthy subjects, it has been claimed that it optimizes the transition from filling to ejection [1]. A change in this flow (diastolic dysfunction) can have a significant impact on the intraventricular vortex [2], [3]. According to the recent literature, it becomes evident that the properties of the ventricular intracavitary blood flow during filling are strongly related to the myocardial relaxation function, which can be non-invasively emphasized by ultrasound color Doppler [4], [5].

To obtain and analyze intracardiac blood flow at the patient's bedside, an imaging modality that is clinically easily accessible and able to provide non-invasive information in real-time is required, which is why ultrasound is the preferred technique [6]. There are two main classes of methods for the analysis of blood flow by ultrasound imaging: 1) speckle tracking [7], such as echo-PIV (particle image velocimetry) [8], and 2) approaches based on color Doppler, such as *i*VFM [9], [10]. Since speckle-tracking methods generally require the injection of microbubbles, techniques derived from color-Doppler are preferable in clinical routine.

1.2 Triplane-echo-based 3D-*i*VFM

Doppler echocardiography is a limited technique for flow imaging because it only gives a scalar velocity field: it returns the velocity projections along the ultrasound beams. In this work, we sought to recover the three vector components of the three-dimensional intracardiac blood flow from Doppler velocities. In other words, when considering the spherical coordinate system linked to the ultrasound transducer, our objective was to estimate a three-component 3-D velocity vector field from the single radial velocity components. To obtain a temporal resolution suitable for the study of vortex formation, a full-volume Doppler ultrasound is not recommended as it provides low frame rates. We thus used triplane color Doppler echocardiography, as represented in Figure 1.

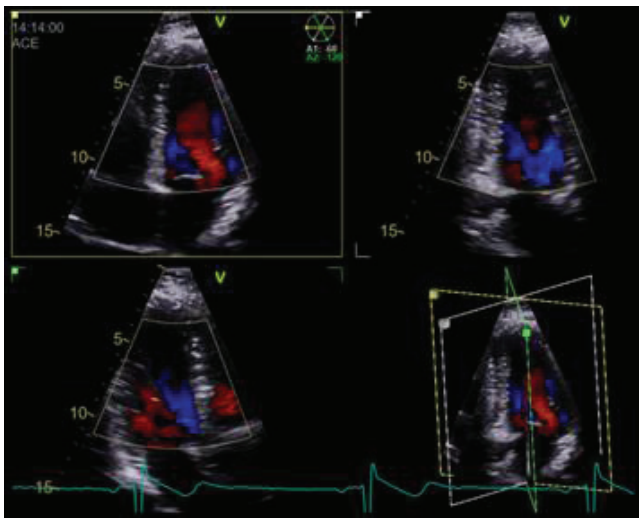


Figure 1: Triplane color Doppler echocardiography of the left ventricle.

In the following, we describe the minimization problem and the method for unsupervised selection of the smoothing parameter. We then introduce the patient-specific CFD (computational fluid dynamics) model, which we used to

simulate triplane Doppler echocardiography and validate the 3D-*i*VFM reconstruction method.

2. METHOD

The objective of 3D-*i*VFM is to recover the three-component intraventricular velocity field from a triplane color Doppler echocardiography after dealiasing of the Doppler velocities [11]. We chose this triplane mode (rather than full-volume color Doppler) since 1) the ultrasound data before scan-conversion are available through the commercially-available software EchoPAC, and 2) the temporal resolution is higher. This clinical ultrasound mode returns data on three long axis apical planes (color Doppler + B mode) separated by an angle of 60°. The data extracted from the triplane mode by the EchoPAC software are given in a spherical coordinate system $\{r, \theta, \varphi\}$. The Doppler velocities, $u_D(r,\theta,\varphi)$, are related to the radial components of the velocity, $v_r(r,\theta,\varphi)$, with a negative sign and additive noise η :

$$v_r(r,\theta,\varphi) = -u_D(r,\theta,\varphi) + \eta(r,\theta,\varphi) \quad (1)$$

To reconstruct the three-dimensional velocity field we need to find the angular and azimuthal components in this spherical coordinate system, $v_\theta(r,\theta,\varphi)$ and $v_\varphi(r,\theta,\varphi)$. For this purpose, we used a cost function with smoothness and physics constraints. The 3D-*i*VFM method is based on a generalized and improved version of the 2D-*i*VFM algorithm introduced by Assi *et al* [12]. In contrast with the previous approach [12], the problem has been rewritten as a constrained least-squares problem, which was solved by the Lagrange multiplier method. The hemodynamic properties used to constrain the minimization problem were based on fluid dynamics: mass conservation for an incompressible fluid, and free-slip boundary conditions on the endocardial inner wall. The problem can be mathematically written as follows:

$$\vec{v}_{VFM} = \underset{\vec{v}}{\operatorname{argmin}} J(\vec{v}) \quad (2)$$

with the cost function given by

$$J(\vec{v}) = \int_\Omega (v_r + u_D)^2 \, \mathrm{d}\Omega \; + \; \alpha \mathcal{L}(\vec{v}) \quad (3)$$

subject to:

$$\begin{cases} \vec{v} \cdot \vec{n}|_{wall} - \vec{v}_{wall} \cdot \vec{n}|_{wall} = 0 \quad \text{on } \partial\Omega, \\ \operatorname{div}(\vec{v}) = 0 \quad \text{on } \Omega. \end{cases} \quad (4)$$

J stands for the cost function to be minimized under the constraints (4). The vector field $\vec{v}_{VFM}$ corresponds to the estimated 3-D intraventricular blood velocities. $\mathcal{L}(\vec{v})$ is a spatial smoothing (penalty) function that uses second-order partial derivatives, with respect to the radial and angular coordinates (r,θ). The parameter α is the smoothing parameter (a scalar). Ω represents the region of interest (the left ventricular cavity) and $\partial\Omega$ its boundary (the endocardial wall). $\vec{n}|_{wall}$ is the vector normal to the wall, and $\vec{v}_{wall}$ is the endocardial wall velocity. The first equality in (4) describes the free-slip boundary conditions. The second equality represents the mass conservation for an incompressible fluid (divergence-free flow).

We applied the Lagrange multiplier method and a finite-difference discretization to convert this constrained least-squares problem into a sparse matrix form. To minimize operator dependency, we selected the smoothing parameter (α) automatically through the L-curve method [12].

3. RESULTS

3.1 *L*-curve

To select an α parameter that provides a good balance between fit to Doppler data and smoothing, we chose the L-curve method [13]. As depicted by Figure 2, the corner of the L-curve can be easily spotted (around -9/-8 in this case), which allowed us to automate the method of selecting the smoothing parameter.

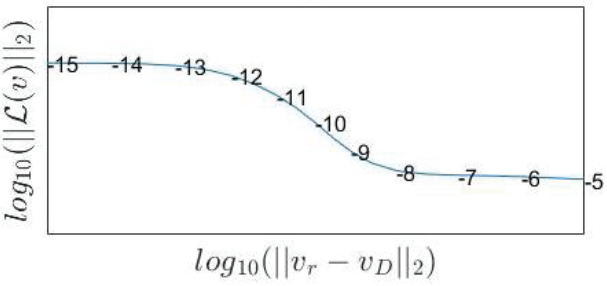


Figure 2: L-curve method applied to our constrained minimization problem. The values on the curve are the $\log_{10}(\alpha)$.

3.2 Velocity reconstruction

Once the smoothing parameter was determined automatically with the L-curve method, we compared the velocities estimated by the 3D-*i*VFM model with the reference velocities of the CFD model. There was a good concordance between the estimated and reference velocities for the radial (v_r) and elevation (v_θ) components (Figure 3, left column). The histograms of the absolute errors were all centered around zero (Figure 3, right column). The histogram for the azimuthal components (v_φ) was stretched out due to the presence of larger errors. It was expected since the spatial resolution in the azimuthal φ-direction is limited (six half-planes = six samples) in triplane color Doppler.

3.3 Intraventricular blood flow streamlines

The 3-D intraventricular vortical flow recovered by 3D-*i*VFM was visible (Figure 4 b) and was globally accurate when compared to the CFD reference (Figure 4 a). Our

results tends to show that the 3D-*i*VFM method enables the reconstruction of the overall vortical pattern in the left ventricular cavity.

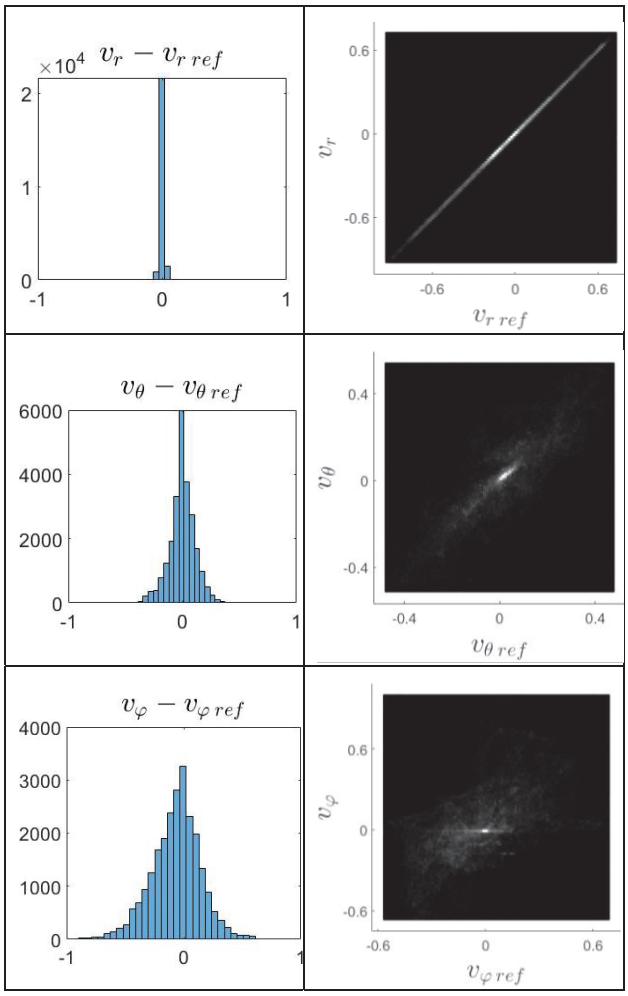


Figure 3: Left column: histograms of the differences between the reconstructed and reference velocities, for each component. Right column: estimated *vs*. ground-truth velocities.

4. DISCUSSION

Our *in silico* study tends to show that triplane-echo-based 3D-*i*VFM can recover the 3-D structure of the intraventricular vortical flow that forms during diastole. The algorithm we have developed is fast and robust, making it clinically compatible. It depends on a single regularization parameter, which can be selected in an unsupervised way using the *L*-curve method. Further *in silico* and *in vivo* investigations will be carried out to ensure that the introduced approach is sufficiently reliable for potential clinical utilization. In particular, we will investigate to which extent the velocity errors in the azimuthal φ-components can have an impact in the determination of a global vortex parameter. In this context, it is likely that access to volume data, rather than triplane data, can reduce the bias of the reconstructed velocities. Whether this difference in bias is significant from a physiological point of view needs to be analyzed.

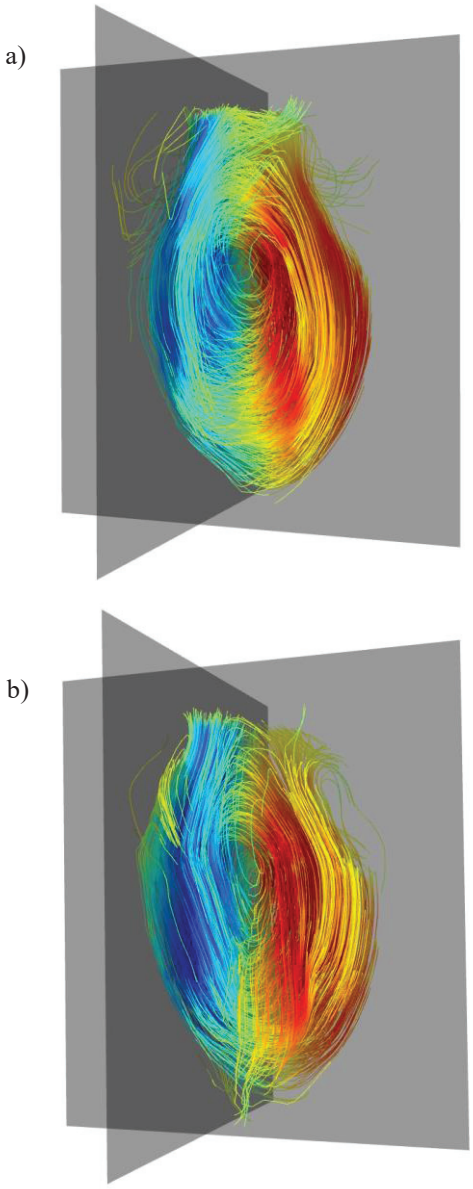


Figure 4: a) Streamlines from the CFD model during left ventricular filling. b) Streamlines from the reconstructed velocities at the same instant.

5. CONCLUSION

Triplane-echo 3D-*i*VFM can quickly decipher the ventricular intracavitary blood flow hemodynamics. It is a 100% automatic method, which makes it clinically compatible. 3D-iVFM could likely offer new echocardiographic insights into left ventricular filling function.

6. ACKNOWLEDGEMENT

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