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Assessment of human left ventricle flow using statistical shape modelling and computational fluid dynamics

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ABSTRACT

Blood flow patterns in the human left ventricle (LV) have shown relation to cardiac health. However, most studies in the literature are limited to a few patients and results are hard to generalize. This study aims to provide a new framework to generate more generalized insights into LV blood flow as a function of changes in anatomy and wall motion. In this framework, we studied the four-dimensional blood flow in LV via computational fluid dynamics (CFD) in conjunction with a statistical shape model (SSM), built from segmented LV shapes of 150 subjects. We validated results in an in-vitro dynamic phantom via time-resolved optical particle image velocimetry (PIV) measurements. This combination of CFD and the SSM may be useful for systematically assessing blood flow patterns in the LV as a function of varying anatomy and has the potential to provide valuable data for diagnosis of LV functionality.

1. Introduction

Complex blood flow dynamics inside the heart, especially inside the left ventricle (LV), are under intensive study for potential early stage biomarkers of cardiovascular health. Of particular interest is visualization of vortex ring formation, pressure distribution, wall shear stress and energy dissipation. Several studies have shown that these indices could provide better understanding of cardiac functionality (Doenst et al., 2009; Elbaz et al., 2017; Eriksson et al., 2011; Gharib et al., 2006; Khalafvand et al., 2014; Kheradvar et al., 2012; Vasudevan et al., 2017). Imaging techniques such as magnetic resonance imaging (MRI), computed tomography (CT) and Doppler echocardiography have been used to visualize blood flow patterns and vortex ring formation in the LV (Ebbers et al., 2002; Elbaz et al., 2017; Eriksson et al., 2011; Gharib et al., 2006; Kilner et al., 2000). In addition, computational methods such as computational fluid dynamics (CFD) and fluid–structure interaction (FSI) in conjunction with MRI and CT data have been widely used to simulate and visualize patient-specific LV flow (Eriksson et al., 2011; Khalafvand et al., 2017; Khalafvand et al., 2014; Krittian et al., 2010; Long et al., 2007; Long et al., 2004; Mihalef et al., 2011; Pedrizzetti and Domenichini, 2005; Saber et al., 2003a,b; Schenkel et al., 2009; Su et al., 2016; Watanabe et al., 2004). The combination of numerical methods and imaging techniques could provide additional data for assessment of blood flow abnormality and so-called predictive medicine.

Blood flow in the LV is directly coupled to LV geometry and wall motion. Remodeling of the LV geometry may alter the blood flow patterns and hence fluid dynamics parameters such as energy dissipation (Pedrizzetti and Domenichini, 2005). Three dimensional (3D) geometric modeling of the cardiac structures is an essential prerequisite for simulations-based blood flow analysis and visualization. Such models can be obtained from patient MRI and CT images. Due to the complexity of image post-processing and computational constraints, most studies are limited to small sample sizes (Mittal et al., 2016). Blood flow analysis in a larger set of data could provide a better understanding of cardiac functionality (Biglino et al., 2016).

Abbreviations: ALE, arbitrary Lagrangian–Eulerian; CFD, computational fluid dynamics; CT, computed tomography; EDV, end-diastolic volume; EF, ejection fraction; ESV, end-systolic volume; FSI, fluid–structure interaction; LV, left ventricle; PIV, particle image velocimetry; MV, mitral valve; SI, sphericity index; SV, stroke volume; TED, time integral of energy dissipation; TRDPIV, time resolved optical particle image velocimetry; TWSS, time integral of wall shear stress; UDF, user-defined function; VFT, vortex formation time.

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Developments in imaging techniques have provided image databases and population-based studies of the heart. Statistical shape models (SSM) have been developed to parameterize the primary modes of variation observed in cardiac shapes (Biglino et al., 2016; Cootes et al., 1995; Metz et al., 2012). SSM provides a method to describe the significant shape variations of the heart geometry over a large population. SSM derived shapes, in conjunction with CFD, can be used to explore the variation of blood flow changes in a population size that would be too cumbersome to analyze on an individual basis. Additionally, analyzing these generalized shapes provides a method to obtain the relation between principal shape variations in a population and their associated hemodynamic effects. This approach may provide more general insights than when directly examining individual cases.

This study aims to quantify the characteristics of blood flow patterns in the LV using a newly developed framework. In this framework, a CFD tool is used on shapes derived from a SSM of the left heart to compute and visualize fluid dynamics characteristics of cardiac blood flow. In this study, the framework is applied on several generated shape instances, derived from the SSM that was built from the shapes of 150 subjects. Also, a shape derived from the model is physically realized by 3D printing and integrated in a dynamic in-vitro setup mimicking the beating heart. Flow in this beating heart phantom was measured using time-resolved optical particle image velocimetry (PIV) and compared with the proposed numerical CFD method.

2. Methodology

2.1. Statistical shape model (SSM)

A previously developed (Metz et al., 2012) 4-D SSM of the LV endocardium, left atrium and aorta of 150 subjects was used to generate 5 characteristic shapes. These five shapes included the mean shape, and mode variations of +3 and −3 standard deviations (SD) along the first and second principal components of shape variation in the population, representing the modes of highest energy and largest shape variation in the dataset (Fig. 1a–c). Each shape model was represented by 3D polygonal surfaces (5170 vertices) at 20 time points per cardiac cycle.

2.2. Computational fluid dynamics modelling

CFD simulation was performed with a developed semi-automatic method. For the 5 characteristic shapes from the SSM geometry reconstructed and separate unstructured grids consisting of tetrahedral cells were generated. The time resolution of the CT-derived shape model data (20 frames per cardiac cycle) was not fine enough for CFD simulation. To ensure a Courant number of less than unity (product of local fluid velocity with the ratio of time step to mesh spacing, $\frac{u\Delta t}{\Delta x} < 1$), intermediate grids were generated for each initial measurement interval between the 20 reconstructed grids from the original CT data. Approximately 2000 grids were generated during a cycle (depending on the case volume) using cubic spline interpolation. In order to solve the fluid flow domain with a finite volume method, the arbitrary Lagrangian-Eulerian (ALE) formulation of the Navier-Stokes equations was used. The integral form of the continuity equation for a volume with surface S is expressed as

$$\frac{\partial}{\partial t}\int_V \rho dV + \int_S \rho(\vec{v} - \vec{v}_b) \cdot \vec{n} dS = 0 \tag{1}$$

where $\vec{v}$ is the velocity vector, $\vec{v}_b$ the velocity on the boundary, $\vec{n}$ the normal vector and ρ the blood density. The momentum equation is

$$\int_V \frac{\partial}{\partial t}(\rho\vec{v})dV + \int_S \rho\vec{v}(\vec{v} - \vec{v}_b) \cdot \vec{n} dS = -\int_S p\mathbf{I} \cdot \vec{n} dS + \int_S \boldsymbol{\tau} \cdot \vec{n} dS \tag{2}$$

where p is the pressure, $\mathbf{I}$ the unit tensor, and $\boldsymbol{\tau}$ the viscous stress tensor. Blood flowing in the heart cavities can be treated as homogeneous Newtonian fluid with a density of 1060 kgm^{-3} and the dynamic viscosity of 0.00316 Pa · s. This is an acceptable assumption for large vessels with inner diameter greater than approximately 0.5 mm and cardiac chambers because of relatively constant apparent viscosity of blood at high shear rates (>100/s) (Kitajima and Yoganathan, 2007). The simulation was performed in a discontinuous time step fashion during one cardiac cycle because for each time step there is a new grid (Khalafvand et al., 2017; Saber et al., 2003a; Schenkel et al., 2009). The velocity boundary conditions at mitral and aortic valve were derived from the change of LV volume (Fig. 1c). To reach a periodic solution, the simulation was repeated for four cycles and second order upwind scheme was employed. The resulting algebraic equation system was solved using the implicit PISO (Pressure Implicit with Splitting of Operators) algorithm. In this study, CFD software ANSYS FLUENT (Version 17.2) was employed to model the blood flow in LV.

The grid dependency analysis was performed to achieve the optimum grid size. Five different grids were generated: 500 K, 1 M, 2 M, 3 M and 4 M cells. An identical simulation was performed for the five model instants from SSM, and several parameters were captured to compare the results. The integral quantity of wall shear stress (WSS) over wall surfaces and the rate of energy dissipation (ED) over volumes during a cycle were calculated for all cases (Fig. 2a). The results show that the difference between 3 M and 4 M cells is less than three percent. The vortex core structure is qualitatively compared in Fig. 2b. There are no visible changes between 3 M and 4 M cells. In the present work, four million cells have been used for more accurate results, though three million cells can also capture the main flow features.

2.3. Time resolved digital PIV

To validate the numerical simulation results against a ground truth flow measurement, an experimental LV flow phantom was designed to be used for laser PIV modality (see Fig. 3). The mean SSM shape geometry at the end of systole was used as an initial geometry in the experimental setup. A compliant, optically and acoustically transparent silicone model of LV was manufactured by painting silicone (HT33 Transparente LT, ZhermackSpA, Italy) onto a mold of the LV mean shape made by 3D printing.

The silicone LV was then encased in a rigid, optically transparent acrylic box and attached to mitral and aortic valve ports fitted with Bjork-Shiley valves, which were connected to an atrial and compliance chamber, respectively. The acrylic box had one open port, which was connected to a piston pump, programmed to reciprocate in a sinusoidal pattern with a frequency of 1 Hz with 80 ml stroke volume. The box had three open surfaces for laser/camera view access for the acquisition of laser PIV recordings. For this study degassed 66% by weight glycerol solution was used as a pumping fluid in the entire phantom. For a global overview on the experimental phantom, a movie is provided as supplementary data 1.

The velocity maps were obtained using a laser-based time-resolved optical particle image velocimetry (TRDPIV) system (LaVision Inc) as described in Table 1. Briefly, the system consists of a 150 W high speed double cavity Nd:YLF pulsed laser (operated at approximately 30 percent of its power) creating a 1 mm thick vertical illumination through the phantom, which was viewed by a high speed CMOS digital camera in planar PIV configuration. The