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The influence of bileaflet prosthetic aortic valve orientation on the blood flow patterns in the ascending aorta

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A B S T R A C T

We investigate three-dimensional pulsatile aortic flow in the ascending aorta with mechanical prosthetic aortic valve implanted at two different orientations under physiological flow conditions using an anatomically accurate aorta. We perform 3D Particle Tracking Velocimetry measurements to assess the phase averaged and fluctuating velocity patterns as well as the shear stresses. A St Jude Medical prosthetic heart valve is implanted in an anatomically accurate silicone model of an aorta obtained from high resolution magnetic resonance imaging of a healthy proband at two different orientations. Our results show that the mechanical prosthetic valve orientation has considerable impact on the local kinetic energy and shear stress distributions but minor effects on the spatially averaged kinetic energy (10%) and shear stresses (15%). We show that the valve orientation plays a distinct role in spatial distribution of wall shear stresses and vortical structures. We show that our results, which show good agreement with the in silico and in vitro studies in the literature, provide full 3D kinetic energy and shear stress information over the entire cardiac cycle for different bileaflet prosthetic valve orientations under physiological flow conditions.

1. Introduction

The interplay with the blood flow and the heart valves, which provide unidirectional blood flow through the heart, is complicated since it involves flow pulsatility and complex, deformable geometries of both the valve and the aorta [1]. Malfunctions of the heart valves can affect the hemodynamics of the whole circulatory system [2,3] and these abnormalities can either be congenital or arise from abnormalities in hemodynamic behavior. Dysfunctional heart valves can usually be replaced with prosthetic heart valves, which, depending on the design and quality of insertion, offer variable hemodynamic performance and may be associated with different levels of energy loss.

The orientation of the mechanical prosthetic valve affects the hemodynamic performance, in particular, by influencing the systolic performance of the valve, considerably changing the coronary flow rates in diastole [4], producing asymmetric regurgitation [5], improving the diagnostic performance of echocardiography [6] and varying the vortex formation [7]. There have been in vitro [8–13], in vivo [4,14] and in silico studies [15,16] on the influence of the valve orientation in the heart on the blood flow patterns. However, there appears to be limited knowledge about how different valve implants perform under physiological flow conditions with respect to the hemodynamical factors, especially with regards to local hemodynamic conditions in proximity of the valve and in the ascending aorta. One reason is that measurements capable of accessing three dimensional space- and time-resolved shear stresses and energy losses have matured only recently to a level where such studies can be performed. Another reason is that both in vitro and in silico studies stated above assumed simplified flow conditions, i.e., rigid model [5,12,15,16], steady state [12], or 2D flow [11,13].

The main aim of the study is to shed a light on the influence of the aortic valve orientation on the overall valve performance including valve kinematics and local shear stress variations. We present here the results of two mechanical prosthetic aortic valve orientation aligned in the direction of (1) non-coronary sinus ("0°" orientation) and (2) sinus left and right sinuses ("90°" orientation) in an anatomically accurate, compliant silicone aortic phantom under physiological flow conditions. The valve orientations are chosen such that the first orientation, i.e., 0°, represents the one recommended for higher left coronary flow [4] and the second orientation, i.e., 90°, is the one optimal associated with lower turbulence level [17]. Three-dimensional particle tracking velocimetry (3D-PTV) is performed to assess the phase averaged and fluctuating velocity fields, kinetic energies and shear stresses.

2. Methods

2.1. 3D PTV measurements

3D-PTV, an optical imaging tool, has been applied to an anatomically accurate silicone replica (Elastrat GmbH, Switzerland)

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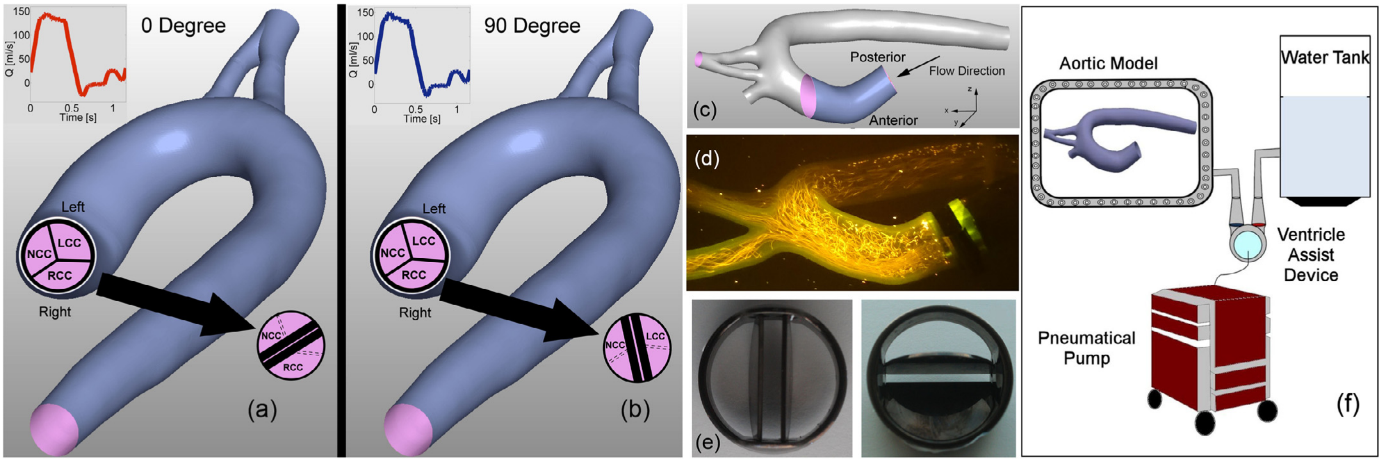

Fig. 1. Geometry of the silicone phantom with mechanical prosthetic aortic heart valves (a) 0° orientation (b) 90° orientation (c) the coordinate system and the flow direction (d) image of tracer particles under laser light (e) St Jude Medical prosthetic heart valves in different orientations and (f) schematic view of the experimental setup including silicone model, pneumetical pump and ventricular assist device.

of the human aorta derived from a high resolution magnetic resonance imaging (MRI) heart scan of a healthy male proband carrying a tricuspid valve. The aorta wall material is a permanently cured but flexible silicone from DOW Chemicals. It is manufactured by brushing the surface of wax positive and curing the layer to the elastic state. The irregularities in the wall thickness were avoided by constantly tumbling the positive wax geometry in a specially designed rotating fixture. The investigation domain for 3D-PTV measurements comprises the ascending aorta. As depicted in Fig. 1a and b, 0° orientation is defined as the heart valve is positioned in the direction of non-coronary sinus whereas 90° orientation is defined as the valve is positioned in the direction of left and right sinuses.

In our measurements, a high-speed camera (Photron SA5, Japan), which allows recording 1.56 s at full resolution of 1024×1024 pixels and 7000 frames per second, with a Nikkon AF Micro Nikkor 60 mm f/2.8 D lens (Japan) was used to capture images during flow. The high-speed camera was synchronized with the heart pump system to trigger the recordings at the beginning of every cardiac pulse. As a light source, a BeamLok 2080 (Spectra Physics) laser, which is a 25 W continuous laser with a wavelength of 514 nm, was used to illuminate the investigation domain. As tracer particles, fluorescent rhodamine particles (Cospheric, USA) with a diameter of 200 µm were used [18]. A mixture of glycerine, water and sodium chloride was used as working fluid, which matches the refractive index of silicone and blood viscosity [18]. During the experiments, a total of 40 heartbeats were recorded to obtain a phase averaged flow field where each heartbeat corresponds to a single pulse with a length of 1.15 s. The material used in this study is different from the native aortic tissue; yet, mechanical properties of the material are in the range of physiological conditions. The silicone model is around 0.05 MPa which agrees with the clinical studies, i.e., elastic modulus of a native aorta varies from 0.06 MPa [19] to 0.1 MPa [20]. Likewise, the volume and diameter variations of the model, which correspond to a relative diameter change of around 5% both cases agrees well with the clinical studies [21].

2.2. Flow field analysis

The Lagrangian velocity information obtained via 3D-PTV was indexed into a $30 \times 20 \times 20$ Eulerian grid system. The Eulerian voxel size was chosen 2 mm. Sixty consecutive time frames of the 8220 frames long heart beat (1.159 s) were binned and the Lagrangian velocity information (u_i) was mapped for each particle (par) in heart beat (cyc) in voxels at spatial positions (x, y, z). The position accuracy of the particles is in the range of 0.15, 0.15, 0.3 mm in x, y and z directions, respectively. The velocity uncertainty of the raw PTV data as obtained from the calibration was 0.033 m/s. A Savitzky–Golay filter was applied for smoothing in time along Lagrangian trajectories, using a cubic polynomial fitted to 21 frames which results in a final voxel accuracy of 4×10^{-4} m/s [18].

The voxel-wise velocity was decomposed into mean and fluctuating components. The phase averaged velocity in a voxel at position (x, y, z) at time point t was obtained as;

$$U_i(x,y,z,t) = \frac{1}{P.N}\sum_{\mathrm{cyc}=0}^{N}\sum_{\mathrm{par}=0}^{P} u_i(x,y,z,t,\mathrm{cyc},\mathrm{par}) \quad (1)$$

where i is the velocity component, N is the number of heart cycles and P is the number of particles in an Eulerian voxel.

Similarly, the square of the fluctuating velocity in a voxel at position (x, y, z) at time point t was obtained (after Gülan et al. [22]):

$$u_i{}'(x,y,z,t) = \frac{1}{P.N}\sum_{\mathrm{cyc}=0}^{N}\sum_{\mathrm{par}=0}^{P} \sqrt{(u_i(x,y,z,t,\mathrm{cyc},\mathrm{par}) - U_i(x,y,z,t))^2} \quad (2)$$

Total kinetic energy is defined as;

$$E(x,y,z,t) = \frac{1}{2}u_i(x,y,z,t)u_i(x,y,z,t) \quad (3)$$

where u_i is the instantaneous velocity.

We can decompose the total kinetic energy into mean kinetic energy (MKE) and turbulent kinetic energy (TKE). The mean kinetic energy (MKE) is defined as;

$$K(x,y,z,t) = \frac{1}{2}\rho U_i(x,y,z,t)U_i(x,y,z,t) \quad (4)$$

where U_i is the phase averaged velocity. Similarly, TKE can be written as;

$$k(x,y,z,t) = \frac{1}{2}\rho u_i{}'(x,y,z,t)u_i{}'(x,y,z,t) \quad (5)$$

where $u_i{}'$ is the fluctuating velocity and summation for repeated indices applies. The MKE and TKE are calculated for each voxel for different phases of the cardiac cycle.