Left ventricular flow propagation velocity: insights from a combined hydraulic and numerical model study

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Abstract

Diagnosis of diastolic heart failure remains difficult in clinical practice. Non-invasive assessment of the flow field within the left ventricle (LV) using color M-mode Doppler (CMD) echocardiography provides a potential technique that can differentiate between the normal and diseased heart.

1 Aim

The aim of this study is to evaluate the behaviour of the flow propagation velocity (\(v_p\)), measured using CMD echocardiography for varying settings of physiologic determinants of diastolic function, including filling pressure, compliance, and relaxation (De Mey et al. 1998). First a 2D axi-symmetric numerical model of LV filling has been developed taking into account the interaction between the LV wall and the fluid. The simulation results are also validated with an experimental study. In the hydraulic model, significant interactions between the flow propagation velocity of the early filling wave (\(v_p\)) and LV diastolic variables, including left atrial (LA) filling pressure, the time constant of pressure fall during isovolumic relaxation (\(\tau\)), and LV compliance were found. The aim of this study is to compare the findings in this hydraulic model with numerical simulations in a 2D axi-symmetric LV filling model.
2 Material and Methods

2.1 Numerical Model

Computer models describing cardiac filling are mostly limited to the fluid dynamical process (1D lumped models based on electrical analogy, 2D analytical models or 2D and 3D numerical models) or are focused on the quantification of ventricular wall stresses and deformations (Verdonck et al. 1998, 1999). Peskin (Peskin and McQueen 1989) was the first to model the blood-ventricular wall interaction using the immersed boundary method. The model of Vierendeels et al. used in this study is a 2D axi-symmetric numerical LV filling model, taking into account the interaction between the LV wall and the fluid (Vierendeels 1998).

Intraventricular Blood Model

The fluid domain of the LV is described by an axi-symmetrical model (Figure 1). The unsteady Navier-Stokes equations in a LV geometry with moving walls are solved. The computation starts at the onset of LV relaxation. During the relaxation phase, the fluid is assumed to be quiescent inside the LV. This assumption only holds for a homogeneous relaxation, which is assumed. When the LV pressure drops below the atrial pressure, the mitral valve opens immediately. From this moment on, a mitral velocity pattern is applied at the circular orifice (base) of the LV. After opening of the mitral valve, pressure in the LV is determined by both the relaxation and compliance of the LV wall and the dynamics of the blood inflow. Blood is assumed to behave as a Newtonian fluid with a density of 1050 kg/m³ and a dynamic viscosity of 3.5 mPas.

Ventricular Wall Model

The LV wall is described by a truncated ellipsoid in the zero stress state (Figure 1). At the zero stress state and with blood at rest, the transmural pressure is zero. Away from the zero stress state, the shape of the LV is computed from equilibrium equations for the LV wall. A non-linear extension of the thin shell equations is used (Vierendeels 1998). The parameters involved are Young's modulus $E$, a non-linearity parameter $\alpha$ and LV wall thickness $h$. The mitral valve annulus is kept fixed, while the apical node can freely move in axial direction. This is in contrast with the living case where the mitral annulus is moving and the apical motion is limited.

During LV relaxation, the compliance of the LV is changing. This is modeled by a time dependent Young's modulus:

$$ E = E_{\text{start}} + (E_{\text{stop}} - E_{\text{start}}) \cdot (1 - e^{-\tau}) $$

(1)

where $t$ denotes the time and $\tau$ the time constant of LV relaxation. $E_{\text{start}}$ is the modulus at the onset of relaxation and $E_{\text{stop}}$ is the modulus in the passive state of the heart muscle. Young's modulus $E$ and the non-linearity parameter $\alpha$ are assumed to be constant along the heart wall. They are determined by the fitting of a pressure-volume relationship.
Coupling Procedure
The coupling of the heart wall displacement and the filling dynamics is based on an iterative approach (Vierendeels 1998). For each time step, the procedure alternates between the following steps: 1) calculation of the movement of the boundary, for a known pressure along the boundary, 2) determination of the position of the internal mesh nodes, for a known position of the boundary and 3) calculation of the flow field in a moving domain, which results in an updated velocity pattern and a new approximation of the pressure field to be used in step 1) until convergence for the time step is achieved.

2.2 Hydraulic Model
The left heart model consists of an open rigid cylindrical Perspex atrium (LA, diameter 30 mm) connected to a truncated ellipsoidal left ventricle (LV, zero-pressure equilibrium volume 100ml, base-apex length 75 mm, short axis 50 mm). The LV is composed of latex. In order to prevent inflation at the higher pressure ranges, the ventricle is surrounded by a gauze. The compliance of the LV is proportional to the LV wall thickness. Experiments are performed using a compliant (wall thickness 0.8 mm) and a stiff (wall thickness 3 mm) model LV. To quantify the compliance of the model ventricles, passive pressure-volume relationships are measured by filling the ventricles in steps of 2 ml. Using linear regression the slope of the pressure-volume relationship was determined for the compliant and the stiff ventricle (1.35 and 0.45 ml/mmHg respectively).

In between the two heart chambers a mitral tricuspid porcine xenograft is mounted (diameter 21 mm, McNeilab Inc., Anaheim, CA). A 60%-40% water-glycerin mixture (dynamic viscosity 3.5 mPas) was used as test-fluid. In the experimental set-up, the ventricle is surrounded by a closed rectangular Perspex chamber filled with air (PeCh, 250 mm x 250 mm x 150 mm). An early filling wave is initiated by bringing the ventricle to its end-systolic volume and
pressure, starting from a filled ventricle in equilibrium under a fixed atrial pressure, i.e. for a fixed atrial fluid level. This is done by applying pressure (100 mmHg) in the Perspex chamber around the LV, using a pneumatic system and compressed air. Because of the valve in mitral position, the fluid inside the LV can not freely migrate from the LV into the LA, but it leaks through the bioprosthesis until the LV has reached its end-systolic volume. The actual filling of the LV is initiated by switching an electromagnetic valve of the pneumatic system, releasing the air out of the surrounding chamber through air outlets in the bottom of the surrounding Perspex chamber and allowing the LV to relax. The rate of relaxation can be changed by adapting the area of the air outlets. Filling pressure is adjusted by changing the fluid level in the atrium. The model is instrumented with a Doppler probe (Vingmed) in an apical position.

2.3 Simulated cases

Hydraulic model experiments of LV filling were done for varying hemodynamic settings, including variations in τ (45, 60 and 90 ms), LA pressure (3, 10 and 30 mmHg) and compliance (0.45 and 1.35 ml/mmHg). Each experiment is repeated three times. Analogue to the hydraulic model experiments, in the numerical model 12 simulations were executed (Table 1). 9 simulations were executed in a LV model with compliance 1.35 ml/mmHg, varying τ (45, 60 and 90 ms), and varying LA filling pressure (10, 20 and 30 mmHg). 3 additional simulations were performed in a stiff model LV (compliance 0.45 ml/mmHg) for varying τ (45, 60 and 90 ms) on a fixed LA pressure level (30 mmHg). The passive pressure-volume relationships, measured in the hydraulic model LV with different compliances, are applied to the numerical model and account for differences in compliance. E_start is taken equal to E_stop because the material used in the experiments stays in the passive state. Thus, according to Eq. 1, E remains constant and it appears that differences in τ are not simulated. However, in the hydraulic model, changing τ results in a change in inlet profile. Therefore, because the measured inflow profiles in the hydraulic model experiments are applied to the corresponding numerical simulations, the influence of τ can also be seen in the numerical outcome. The calculated flow patterns are transformed into the format of CMD echocardiograms.
Table 1: Settings of the time constant of LV relaxation ($\tau$), LA pressure (LAP), and LV compliance (C) for LV filling in the Hydraulic model experiment and corresponding numerical simulations.

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<th>$\tau$, ms</th>
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2.4 Data Analysis

CMD images from both the hydraulic experiments and numerical simulations are used for calculation of $v_p$ using the Duval-Garcia method (Duval-Moulin et al. 1997, Garcia et al. 1998). In brief, linear regression analysis is used for fitting a line with the isovelocity boundary of 50% of the maximum velocity, starting at the level of the mitral valve and going three cm into the LV. $v_p$ is defined as the slope of the line in cm/s. Visual comparison between numerical simulations and hydraulic experimental results is enhanced by displaying the numerical CMD simulations in the same format of the CMD images measured in the hydraulic model experiment, i.e. velocities are represented using a red to blue color lookup table.

3 Results

Figure 2 compares CMD images obtained in the hydraulic model experiment and the according numerical model simulations corresponding to cases 1, 7, 3 and 10 of table 1 (respectively corresponding with figure 2 (A), (B), (C) and (D). Overall, a good qualitative agreement is observed.
Figure 2: Calculated CMD images using the numerical model (top) and corresponding measured CMD images in the hydraulic model experiments (bottom) for varying time constant of relaxation ($\tau$), LA pressure (LAP) and compliance ($C$).

The GLM model with dependent variable $v_p,\text{hyd}$ ($v_p$ derived from hydraulic experiments) and covarying factors $\tau$, compliance ($C$), and LA filling pressure (LAP) has a F-value of 82.9 (adjusted $r^2 = 0.82$, $p < 0.001$). A significant ($p < 0.001$) influence of $\tau$, $C$ and LAP is found. All interaction terms are removed as they have no significant influence ($p > 0.05$). The according multiple linear regression equation is (Eq. 2):

$$v_p = 42.67 + 0.30 \cdot \text{LAP} - 0.30 \cdot \tau + 7.86 \cdot C$$

(2)

4 Discussion

In this study numerical simulations corresponding to hydraulic model experiments have been compared.
νₚ in the numerical simulations is consistently lower than νₚ in the hydraulic experiments. Several aspects contribute to this aberration.

In the numerical model, gravity forces are not included in the equations. In contrast, in the hydraulic model, gravity interferes and, e.g., may cause a distortion of the LV geometry, compared to the numerical LV model. Further, in the numerical model, a homogeneous relaxation is assumed. Due to the position of the air outlets at the bottom level of the Perspex chamber that surrounds the hydraulic model LV, a homogeneous relaxation can not be guaranteed in the hydraulic experiments. Also, in the numerical model, inflow velocities are assumed to be orthogonal to the mitral valve plane. In the hydraulic model, as might be the case in vivo, this perfect alignment may not be present. Finally, the more pronounced curve-linear isovelocity lines in the numerical CMD output images, due to higher temporal and velocity resolution in the numerical simulations are also influencing derived νₚ values.

In terms of changes in νₚ as a result of changes in LV diastolic variables, hydraulic model experiments and numerical simulations agree well, especially for changes in τ and LA pressure. Concerning the influence of LV compliance, both models appear to show a deviant behaviour. In the hydraulic model, compliance is positively correlated to νₚ for all levels of τ, i.e. a more compliant LV leads to a higher νₚ. This positive relationship between compliance and νₚ is on one hand in contrast with the Moens-Korteweg equation, stating that wave propagation in a tube is related with the square root of the Young modulus E (i.e. an inverse relationship with compliance of the wall). On the other hand this finding supports the hypothesis that a stiff ventricle, not being an open tube, exerts a stronger resistance on the flow that enters the LV which fits with in vivo observations (Duval-Moulin et al. 1997) and in numero results (Vierendeels, 1998).

5 Conclusion

The interaction between νₚ and LV diastolic variables is studied in a numerical model and compared to corresponding hydraulic model results. Overall a good agreement is found. νₚ increases with increasing LA filling pressure and decreases with increasing τ. A decrease in compliance augments νₚ in the region of the LV base but decreases νₚ due to stagnation of the vortex at the mid-level of the LV.

6 References


