A New Thick-Walled Hydraulic Model of the Left Heart for the Assessment of Blood-Wall Interaction Using Ultrasound

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Abstract

In clinical practice, left ventricular early diastolic performance is most often evaluated non-invasively using (Doppler) echocardiographic techniques. Because the measured blood and myocardial tissue velocities are influenced by various parameters simultaneously, it is not obvious for the clinician to know which parameters are causing a depressed diastolic function. With the help of our newly developed computer controlled hydraulic model of the human left heart, we can individually adjust end-systolic pressure, active relaxation rate, passive myocardial properties and preload. Using this experimental setup, a more fundamental insight in left ventricular isovolumic relaxation and early filling dynamics can be obtained.

1. Introduction

Heart failure is an increasingly important health problem in most developed countries worldwide [1]. The majority of these patients suffer from impaired systolic function. About 30-40% of all patients who have typical symptoms of congestive heart failure, however, have a normal or only slightly reduced ejection fraction. In these patients with a normal systolic function, diastolic heart failure may be the primary cause of congestive heart failure [2]. They experience increased diastolic filling pressures and/or an impaired left ventricular filling due to a delayed isovolumic relaxation or an increased ventricular stiffness.

Doppler echocardiography is the most commonly used non-invasive technique for the assessment of diastolic function as it is safe, fast, relatively cheap and painless. Several time- and/or velocity-related parameters that are derived from blood and myocardial tissue velocities are currently being used. Special attention is payed to the transmitral velocity pattern, the flow propagation velocity and the mitral annulus velocity profile [3].

Diagnosis of diastolic failure and follow-up is rather cumbersome because various interfering physiological factors influence LV relaxation and filling simultaneously. Diastolic performance is directly or indirectly determined by myocardial active relaxation, suction, viscoelastic myocardial properties, systolic performance, compliance, atrial contractility, interaction between the left and right ventricle, and heart rate [4].

The aim of this study was therefore to construct a computer controlled hydraulic model of the left heart, in which critical factors that influence early diastolic performance can be adjusted independently. Using this model, based on a thick-walled phantom ventricle, we can obtain more detailed information about the interaction between fluid dynamics and wall motion during the isovolumic relaxation phase and early left ventricular filling.

2. Methods

2.1. Left ventricular phantom

The left ventricle has been modelled as a thick-walled truncated ellipsoid. Its equilibrium (stress-free) volume is determined by the mould geometry and amounts to 102 ml, which corresponds to a normal human left ventricle. The base-apex length is 85 mm and the short axis diameter is equal to 45 mm. The overall wall thickness is 10 mm, while at the apex, the wall thickness is 5 mm.

Figure 1: Phantom of the left ventricle: top: apex, bottom: base

To mimic mechanical and acoustic myocardial tissue properties, the phantom is made from a Polyvinylalcohol (PVA, Sigma Chemicals, St. Louis, AV. Mol. Wt. 70000-100000) solution mixed with 1% graphite powder to ensure sufficient backscatter of the ultrasonic waves. A unique property of this PVA solution is that its stiffness

increases with the number of freeze-thaw cycles it undergoes during preparation [5]. Two phantoms with a different stiffness were made by using resp. 11% and 13% (mass%) PVA. Both phantoms underwent 3 freeze-thaw cycles (12hrs at -19°C and 12hrs at 22°C). Passive pressure-volume curves are shown in figure 2. Below equilibrium volume, phantom 1 (low compliance) is about 4 times stiffer than phantom 2 (high compliance).

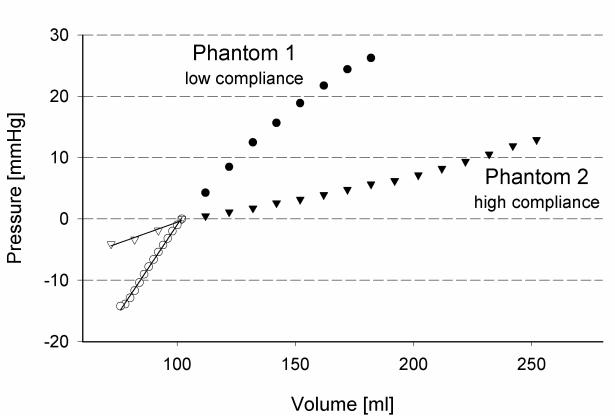

Figure 2: Passive pressure-volume relationships showing that below equilibrium volume (102 ml), the compliance of phantom 1 (1.72 ml/mmHg) is about 4 times lower than in phantom 2 (7.27 ml/mmHg).

2.2. Description of experimental setup

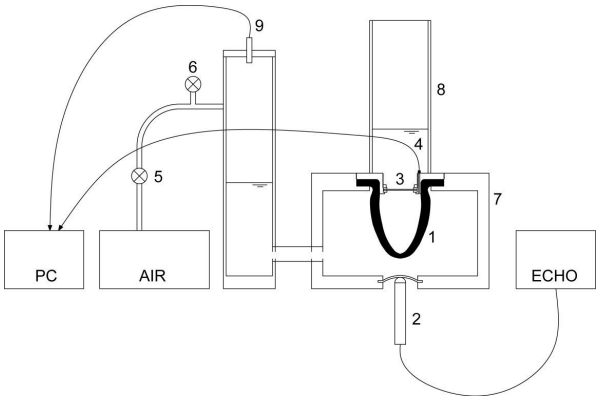

Figure 3: Overview of the experimental setup: (1) LV phantom, (2) Doppler transducer, (3) mitral valve, (4) pressure transducer, (5) proportional valve, (6) valve, (7) plexiglass chamber, (8) preload reservoir and (9) device to measure ventricular volume.

The phantom is mounted in a plexiglass cylindrical chamber containing water (figure 3 (7)). The diastolic phase commences at the end of the systolic phase, defined by its end-systolic pressure (ESP, e.g. 90 mmHg) and its end-systolic volume (ESV, e.g. 70 ml). To obtain these parameter values, we initially calculate the amount of fluid required in the preload reservoir (8) that corresponds to a certain preload level (left atrial pressure, LAP) of e.g. 8 mmHg, corresponding to a volume of 705 ml.

A volume of 705+70 ml is subsequently poured into the ventricle. Intraventricular pressure is then raised to ESP using pressurized air and kept constant by means of a PID control algorithm (partial integration differentiator). Since ventricular pressure is significantly higher than atrial pressure, fluid slowly leaks through the mitral valve leaflets (3) until the predefined preload level of 8 mmHg is reached. At that moment, 2 mechanical valves (5 and 6) are opened, and pressurization ceases. The isovolumic relaxation phase starts now, followed by the early filling phase.

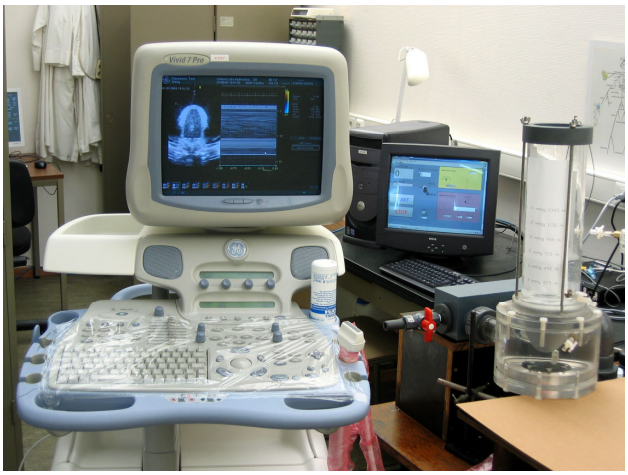

Figure 4: Experimental setup: Ultrasound scanner GE Vivid 7 Pro (left), computer controlled hydraulic model (right).

Pressure and volume registration, opening and closing of the valves is controlled by a virtual instrument panel created in LabView 7 (National Instruments, Austin, TX USA) on a 2.4GHz Intel Pentium IV based Personal Computer.

A mixture of 60% water and 40% glycerine was used as a blood-mimicking test fluid, which has a density of 1060 kg/m³ and a viscosity of 3.5 mPas.

2.3. Measurement protocol

Pressure measurements at the level of the ventricular base have been performed for various ESP (70, 90, 110 and 130 mmHg), LAP (6, 8, 10 and 12 mmHg) and by using 2 ventricles with a different compliance (1.72 ml/mmHg and 7.27 ml/mmHg). Opening degree of the proportional valve 5 is adjusted to 100%, 75% and 50% during relaxation, resulting in 3 different rates of active

relaxation, referred to as 'fast', 'average' and 'slow'. Valve 6, however, is completely opened during relaxation. The peak negative change in ventricular pressure is derived from the pressure curve with a Matlab R13 SP1 code (The Mathworks, Natick, MA, USA). All measurements are repeated 3 times in the same setup.

3. Results

A typical example of a measured pressure time course during ventricular relaxation is seen in figure 5.

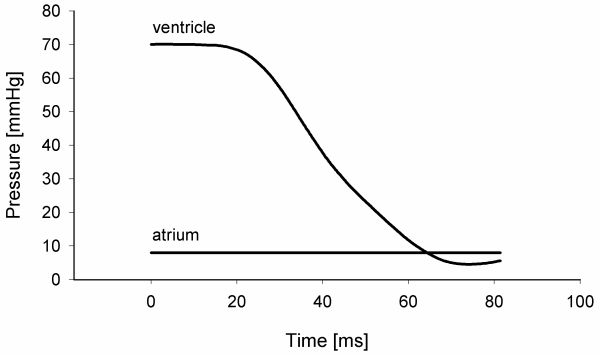

Figure 5: Typical example of a measured pressure time course. Preload (left atrial pressure) equals 8 mmHg and end-systolic pressure is 70 mmHg. Active relaxation was set at the average rate. Pressure decreases due to active relaxation and elastic recoil. Ventricular inflow is initiated when pressure drops below atrial pressure.

Values for peak negative dp/dt ranged between -1.33 and -3.29 mmHg/ms. The coefficient of variation (SD/mean) averaged for every group amounts to 0.007.

Mean values for $(dp/dt)_{min}$ for the average active relaxation rate are presented in figure 6. The effect of varying the active relaxation rate is shown in figure 7. The results show that the pressure decreases faster with increasing end-systolic pressure and active relaxation and with decreasing compliance. Although changing the preload statistically affects peak negative dp/dt as well (ANOVA, $p<0.05$), no general trend could be detected.

This suggests that $(dp/dt)_{min}$ is indeed a measure of ventricular relaxation, which is influenced by (i) the rate of active relaxation and (ii) elastic recoil converted from the stored potential energy at end-systole. The amount of recoil is determined by ventricular stiffness and end-systolic pressure.

Using multivariate linear regression, the peak negative change in pressure can be predicted from the analyzed parameters. Coefficients are shown in table 1. Therelative importance of these parameters is represented by the standardized coefficients. End-systolic pressure has the highest impact on pressure decay rate, followed by ventricular stiffness. Active relaxation rate affects it only moderately, while the effect of preload is almost inconclusive.

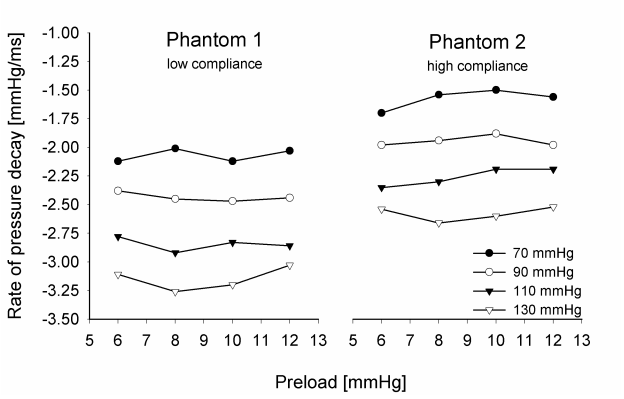

Figure 6: Peak negative change in pressure in a left ventricle with low (left) and high compliance (right). Active relaxation was set at the average rate. Pressure decreases faster with increasing stiffness of the ventricle and with increasing end-systolic pressure.

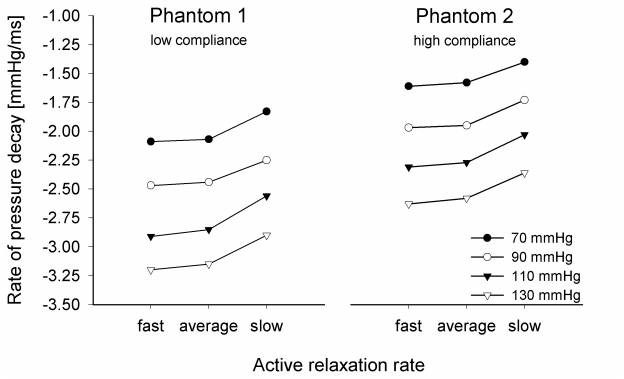

Figure 7: Peak negative change in pressure for different active relaxation rates, averaged for preload. Analagous to figure 6, pressure decreases faster with increasing active relaxation rate and increasing end-systolic pressure.

Table 1: Multivariate linear regression model

Model: $(dp/dt)_{min} = a + b \cdot PL + c \cdot ESP + d \cdot RR + e \cdot C$			
	Coefficients	Stand. Coefficients	Sign.
a	-0.673		<0.001
b	0.008	0.039	<0.001
c	-0.017	-0.794	<0.001
d	-0.529	-0.222	<0.001
e	0.095	0.539	<0.001

PL: preload, ESP: end-systolic pressure, RR: active relaxation rate, C: compliance.

4. Discussion and conclusions

Although knowledge about diastolic function has been increasing vastly with the help of different Doppler based echocardiographic techniques (PW Doppler, Color M-mode, Tissue Doppler), detailed information about the isolated effect of the interacting physiologic parameters is virtually impossible to obtain in vivo.

Our new hydraulic model based on a thick-walled phantom has the advantage that all variables can be adjusted individually without initiating any compensating mechanisms, in contrast with in vivo situations. In this study we have been concentrating on 4 factors that are generally associated with diastolic performance: end-systolic pressure, active relaxation, preload and stiffness. We analyzed the effect of each of the variables on intraventricular pressure during isovolumic relaxation and early filling seperately.

End-systolic pressure was shown to have the largest impact on the peak negative pressure change, when varying it between 70 and 130 mmHg. Although there was a ventricle stiffness ratio of more than 4, the impact was less than that of ESP. The statistically significant influence of preload remains to be clarified. Since there was no trend in this relation, it may be caused by an unknown property of our experimental setup, which could not be unmasked by our measurements. Further experiments with our model may explain this. The result, however, is in accordance with [6]: changing the preload in a physiological range affects $(dp/dt)_{min}$ usually very little (<10%).

It should be emphasized though that our goal was not to make a model that can reflect all peculiarities of the biophysics of a real human ventricle and atrium. Consequently, our computer controlled experimental setup still has a few limitations. We did not account for the myocardial preferential fiber direction which makes a real ventricle heterogeneous and transversely isotropic. Therefore an untwisting motion during diastole could not be simulated.

Furthermore, active relaxation is simulated by releasing the pressure that surrounds the ventricle. In reality though, active relaxation is regulated by oxygen supply for the detachment of actin-myosin cross-bridges. Due to technical reasons, the phantom is required to be attached at its basal side, while in a real ventricle, the atrioventricular plane moves up and down.

Acknowledgements

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