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Particle Imaging Velocimetry Measurements in a Heart Simulator

Knapp, Y.*[1] and Bertrand, E.*[2]

*1 Universite d'Avignon, UMR CSE, INRA-Université Avignon, Avignon, 84000, France.
E-mail: yannick.knapp@univ-avignon.fr
*2 Universite Aix Marseille II, UMR 6594, Equipe Biomecanique Cardiovasculaire, Marseille, 13013, France.

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Abstract : In vitro experiments are often unable to reproduce all the complexities of biological flows observed in vivo. The in vitro models are often rigid, use Newtonian fluids, and/or some ideal geometry tested under ideal physiological parameters. The study presented in this article describes the in vitro assessment of mitral prosthetic heart valves in a setup able to simulate the pulsatile blood flow in a model of the left heart with moving walls. The specific laboratory mockup built for these experiments consists in a Dual Activation Simulator (DAS) that provides a realistic simulation of the atrial and ventricular flow in anatomically shaped silicone models cavities. This mockup, initially designed for ultrasonic velocity measurements took recently advantage of the use of particle image velocimetry. We present here some aspects of flow visualization and phase averaged two-dimensional PIV measurements which can provide new insight in the interaction between the flow dynamics and the heart valves.

Keywords : Cardiovascular Fluid Flows, PIV, Visualization, Valve dynamics.

1. Introduction

Heart valves (Fig. 1a)) play an important role in the control of the proper direction of the blood flow in the human body. In the present article we will focus on the left heart in which the blood is pumped into the systemic arterial circulation therefore generating higher pressures, when compared to the right heart. The major dysfunctions of the valves (called mitral for the atrium-ventricle separation and aortic for the ventricle-aorta separation) are generally caused by degeneration and thickening of the valve leaflets which may have two consequences: a reduction of valve opening (called valve stenosis) thus causing an increase in flow resistance and/or an incomplete coaption of the leaflets when the valve is closed (called valve regurgitation). There are several possible medical or surgical treatments to correct valvular dysfunction. However, when the dysfunction is severe, the only efficient treatment is to repair or replace the valve. There are several different types of prosthetic heart valves; the bioprosthetic valves and the mechanical valves that have 1 (1 disc) or 2 leaflets (2 hemi-discs). Unfortunately, the behavior of these prostheses is not as good as the normal native valves and they may cause other pathological problems such as thrombo-embolism, hemolysis, bleeding and patient-prosthesis mismatch (Fradet et al., 1995). These problems can be reduced by a better knowledge of the heart's blood flow, the valve's dynamics and their interactions, under normal physiological conditions as well as under pathological conditions such as high blood pressure or low

cardiac output.

The flow dynamics downstream artificial heart valves have been described in vitro (see Marassi et al., 2004 for example), in vivo (see Kim et al., 1995, or Botnar et al., 2000, for example) or even by various numerical simulations. However, despite good descriptions of the flow within the left ventricle, most studies focused on valves in the aortic position i.e. in the case of a duct geometry downstream of the valve. The flow observed in the left ventricle shows the predominance of vortical structures, with size and number depending on various parameters such as pressure and flow rate but also valve size, type and orientation. However, little is known about detailed interactions between the flow structures and the opening/closure mechanism of the valves in realistic conditions. Moreover there is a controversy whether pressure gradient and/or intraventricular flow patterns govern the valve closure. As clinical observations are difficult (mainly because of access and opacity) and often restricted to animal or patients in pathological conditions, it was relevant to design a laboratory mockup which enables to simulate pulsatile blood flow with dynamic heart cavities. Such a device was setup in order to simulate realistic physiological atrial and ventricular flows in 1:1 scale silicone models of such cavities of the left heart (Fig. 2).

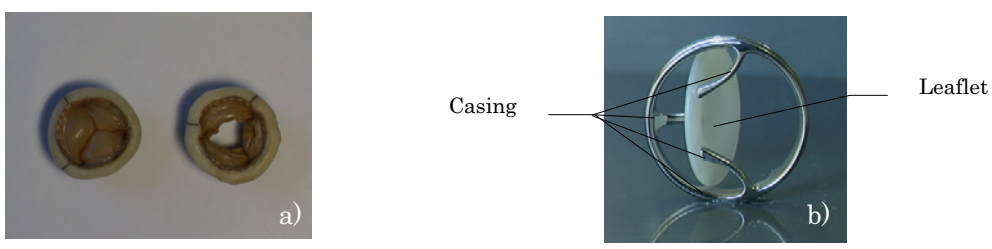

Fig. 1. "Upstream" view of heart valves. a) three leaflet bioprosthetic. Left: closed, Right: open. b) typical mechanical single leaflet valve (open).

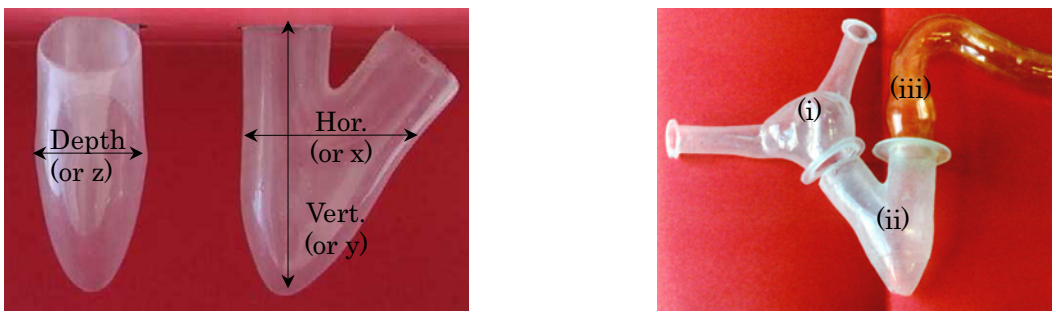

Fig. 2. Silicone shaped models of the heart cavities. Left: ventricular cavity, Right: the atrium (i), ventricle (ii) and aorta (iii) models.

2. Experimental Setup

2.1 The Mockup

The mockup (Figs. 3 and 4) consists of two activation systems, each containing one of the left heart's cavities. These cavities were obtained from molds of original human cardiovascular system. Schematically, an activation system is a closed box in which a fluid is pumped alternatively, resulting in the deformation of the silicon cavity. The blood circuit of the mockup consists in the atrial cavity, the mitral valve, the ventricular cavity, the aortic valve and the aorta completed by a series of resistance and compliance, enabling to take into account the pressure drop and deformation of the body's blood vessels. The flow in this circuit is generated by the out of phase activation of the

computer-controlled gear pumps of the activation systems. Once the proper flow rate shape is chosen, with respect to the desired flow condition, the resistance and compliance are adjusted to obtain the expected flow rate and pressures in the blood circuit.

The fluid used in the simulator is a mixture of water and glycerin. Indeed the blood has a shear thinning non-Newtonian behavior due to the fact that it is a mixture of a fluid and different types of particles. However the viscosity of the blood flow in large arteries shows a Newtonian behavior that is simulated here with the water-glycerin mixture adjusted to obtain a viscosity of 4 cp and a density of 1.13. By using anatomically shaped and dynamic cavities and circulatory fluid with viscosity similar to blood, it is possible to reproduce the anatomic and hemodynamic conditions very similar to those observed in the human circulation. The experiments were performed with ventricle inlet peak Reynolds number of 4350 and inlet Womersley number of 16. Both numbers are similar to the real physiological conditions.

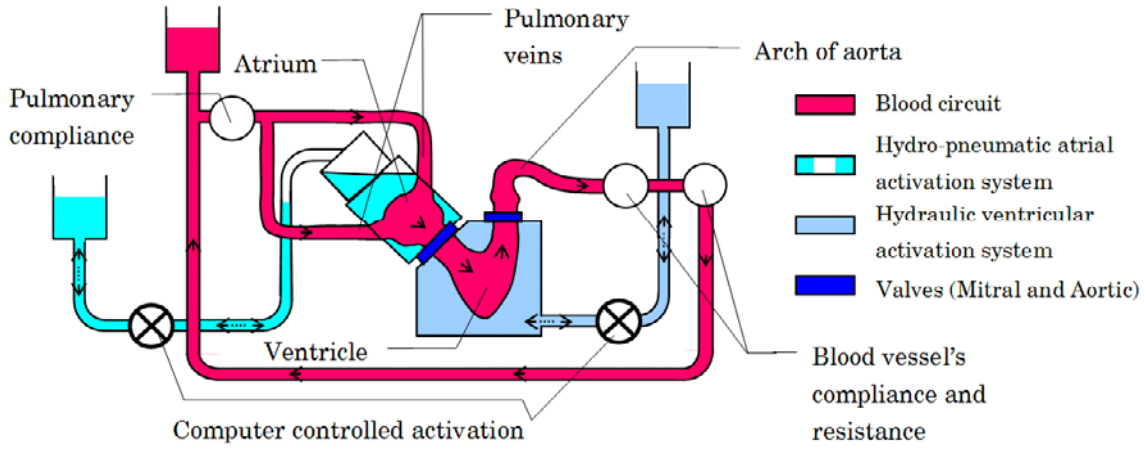

Fig. 3. Double Activated Simulator schematic display.

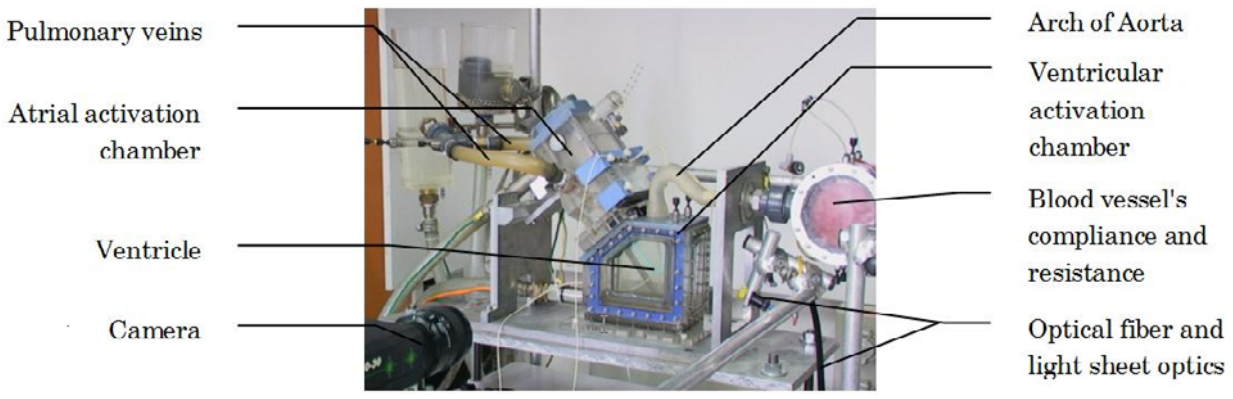

Fig. 4. The Dual Activation System.

Note that special care was taken to limit optical path deformation in the simulator (both for illumination and for image acquisition): perpendicular mounted fine parallel glass walls and very thin silicon walls were used in the mockup.

2.2 Experiments

The present work relates to the flow within the ventricle which is typically a "triangular" cavity with average dimensions of 90 x 60 x 30 mm (vert. x hor. x depth see Fig. 2).

Various studies showed that the flow pattern in the ventricle consists in a series of vortical structures evolving in size and number during the cardiac cycle. The largest vortex is generated by

the fluid's rapid inflow during the filling of the ventricle (Kim et al., 1995, Knapp et al., 2003). The flow visualizations and PIV measurements presented here concentrate on this flow pattern and were therefore performed in the mid section (mid depth) of this cavity. As will be shown, the observation of the vortex's evolution over time can provide some additional insight into the closure mechanism of the valve.

The activation system allows to control and record a set of parameters with a high level of reproducibility (drift less than 2% after 2 hours) and comparisons with clinical observations. These parameters are;

- the flow rate through the mitral and aortic valves measured with a Carolina Medical Model SR670 electromagnetic flow meter, and
- the atrial, ventricular and aortic pressures measured with Millar Instruments Model MPC500 piezoelectric pressure transducers.

The different signals are fed in the computer enabling the generation of control pulses for the synchronization of other measurement devices: PIV in this instance.

The fact that the same fluid is used in the activation system and in the blood circuit allows for a good refractive index matching between the inside and the outside of the silicone cavities. The result of such adjustment can be seen in Fig. 5 were images of the ventricle in the activation chamber are taken in different conditions.

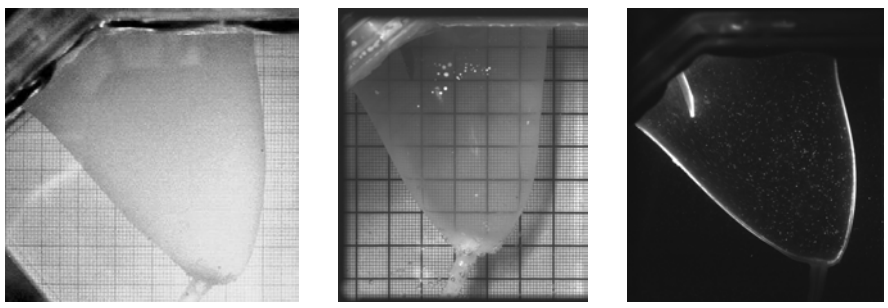

Fig. 5. Images of the silicone ventricle in the activation chamber: Left: both fluids are water-daylight, Center: both fluids are water glycerin mixtures-daylight, Right: both fluids are water glycerin mixtures-laser light sheet.

When considering the choice of a particle, no compromise can be made on both the size and the density of the particles because of the highly variable flow conditions that can be observed in the mockup (from 0 to about 30L/min in terms of instantaneous flow rate). Preliminary tests showed that the best compromise consists in the use of small white polycrystalline polymethacrylate particles in the 40-80 μm range and average density of 1.319 g/cm^3. In order to increase the density matching, the particles were gravity filtered and only the almost neutrally buoyant ones were kept in the experimental setup. The choice of these particles is in good agreement with the theoretical qualification of the particles in terms of settling velocity (less than 0.1 mm/s) and relaxation time (in the range 40 to 150 μs).

The instantaneous velocity fields are determined with the use of an Insight PIV system from TSI Inc. This system, based on a TSI Laserpulse synchronizer, controls the camera, the laser light sheet and the synchronization with the computer. The camera used here is a Kodak 10-30 PIV (able to work in frame stradling mode) equipped with a 60 mm Nikon lens. In order to get a proper flow illumination a Spectra Physics Model 2017 5 Watt Argon Ion laser with 514 nm wavelength is used. The laser beam is driven to the experimental setup through an optical fiber and the synchronizer driving an Isomet Model 1205C acousto-optic modulator controls its intensity. At the output of the fiber the light beam goes through a cylindrical lens resulting in a 1 mm thick 60 degrees opening angle light sheet.

The images are acquired in frame straddling mode (see Raffel et al., 1998 for details) with a 15

Hz frame rate and average of 500 μs between successive images before buffering in pairs in a high performance bi-Pentium PC computer for disk storage and/or processing and display. A proper area of interest of the recorded image pairs is analyzed by a classical FFT based cross-correlation of square 32 pixel windows separated by the fixed distances of 16 pixels in the x and y direction. This simple procedure resulted in a very low number of spurious vectors (less than 1%).

In order to get an adequate scale of the flow a thin millimetric grid was introduced in the activation box filled with the water-glycerin mixture (Fig. 5). Positioning a target directly into the ventricle provided the same scaling factors of about 100 μm/pixel which was constant throughout these experiments.

3. Results

3.1 Experimental Conditions

We present here the results obtained with single leaflet mechanical valves (mitral and aortic). The typical geometry of such valves is given in Fig. 1b).

The model generated a succession of ventricle expansions and contractions that produced normal flow conditions. During the expansion period, called the diastole, the mitral valve opens because of the pressure gradient between ventricle and atrium and the blood inflows the ventricle. A secondary inflow is obtained close to the end of this period, generated by the contraction of the atrial cavity. During the contraction period, called the systole, the mitral valve closes and the aortic valve opens and the blood is ejected into the aorta. As shown by the pressure and flow rate recordings in Fig. 6, the conditions in which the present experiments were performed are typical of the normal physiological conditions observed in a healthy human at rest: 70 bpm, 4.8 L/mn, 100 mmHg Aortic mean pressure.

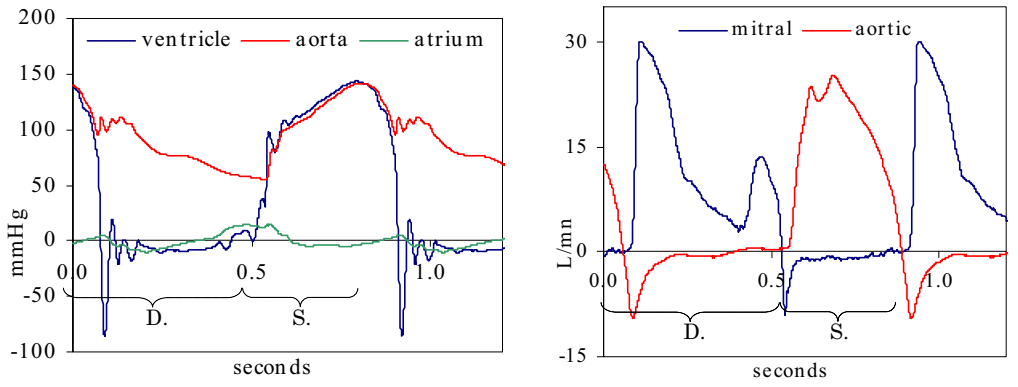

Fig. 6. Pressure (left) and flow rate (right) vs. time recordings: D. diastole, S. systole.

In order to characterize the full cardiac cycle, the acquisitions were performed at 17 time steps over the 850 ms duration of the cardiac cycle in normal condition. For each time step, phase averaged measurements of the velocity field were performed. Preliminary experiments (Mouret et al, 2003) have shown that averages over 50 acquisitions are sufficient to get a converged mean velocity field with less than 2% deviation when compared to an average obtained over 1000 acquisitions.

3.2 Experimental Results

4 instantaneous pictures (the first picture of image pairs) triggered at typical time steps over the complete cardiac cycle are shown on Fig. 7. These pictures underline the difficulty to obtain a good quality image inside the ventricle because of the wall deformation and the partial refractive index matching (time steps 0 to 150 ms and time step 800 ms). However these images may also be useful to

better characterize the flow dynamics, determine the position of the valve (open/closed) and delineate the boundaries of the silicone model (Knapp et al., 2003).

The phase averaged velocity fields presented in Fig. 8 show 3 cases of typical behavior :

- At the very early stages of the cycle (0 to 50 ms) the mitral valve is closed (Fig. 7a)) and the ventricle is "squeezed" reducing drastically the potential area of interest. The flow is here characterized by the final expulsion of fluid in the aorta (end of the previous cycle).

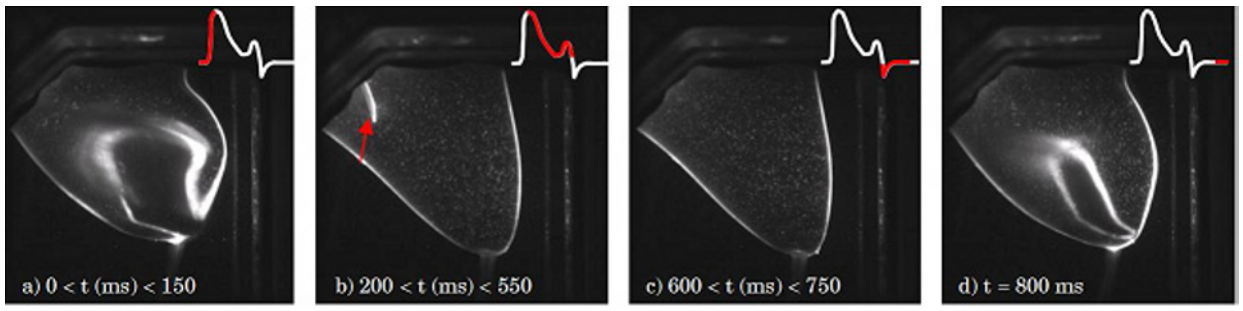

Fig. 7. Typical instantaneous images of the ventricle: a) and d) the silicone ventricle is highlydeformed, b) the silicone ventricle is fully expanded and the valve is open (arrow), c) idem valve closed. Upper right: the mitral flow rate shape with the relative time periods in red.

- The second period (100 to 600 ms) concerns the phase in which the flow rushes into the ventricle through the open mitral valve (Fig. 7b)) under the effect of the ventricle's relaxation. Despite a small pressure variation, the expansion of the ventricle is clearly visible over the different time steps. According to mitral flow rate curve, the flow is initially similar to a jet flow with a strong shear layer close to the lower left wall. This shear layer is formed because of the asymetry of the 2 valvular orifices. The maximum opening angle of the valve is about 70° thus resulting in 2 orifices of different sizes. This burst generates a tridimensional flow in the ventricle and a large vortical structure in the wake of the leaflet. This vortical structure, moves in the ventricle and slowly decreases in intensity due to viscous forces. At approximate time step 500 ms, a second, smaller, inflow takes place generated by the atrial contraction. This second jet re-activates the vortical structure. At this time, since the ventricle is full and totally expanded it is very easy to get the velocity information. The motion of the vortical structure towards the valve and the counter clock-wise rotation of the fluid can contribute to mitral valve closing.

- The last period (600 to 850 ms) concerns the phase in which the mitral valve is closed (Fig. 7c)) and in which the ventricle contracts to expulse the fluid into the aorta. Again at this time, the deformation of the ventricle reduces the visibility inside the ventricle resulting in a reduction of the area of interest. Note that the nature, size and location of the large vortical structure can play an important role in the early stage of this outflow into the aorta ; the vortical structure can indeed contribute to the closure of the mitral valve but the tumbling flow can also obstruct the outflow and thus delay the opening of the aortic valve (Mouret et al., 2003). The regurgitation, or backward flow of blood into the atrium clearly shown as a negative peak in the mitral flow rate curve, can here be limited by the adequate closure of the valve, helped in this process by the existence of the vortical structure.

4. Conclusion

The present article present a heart simulator with transparent moving walls in which the flow structure generated downstream a mitral single leaflet mechanical heart valve is studied. It is clearly shown that a large vortical structure occurs in the left ventricle which was also observed in humans. This flow morphology strongly depends on the valve geometry and can modify the behavior of the valve: the rotating motion of the flow may facilitate mitral closing but on the other hand it

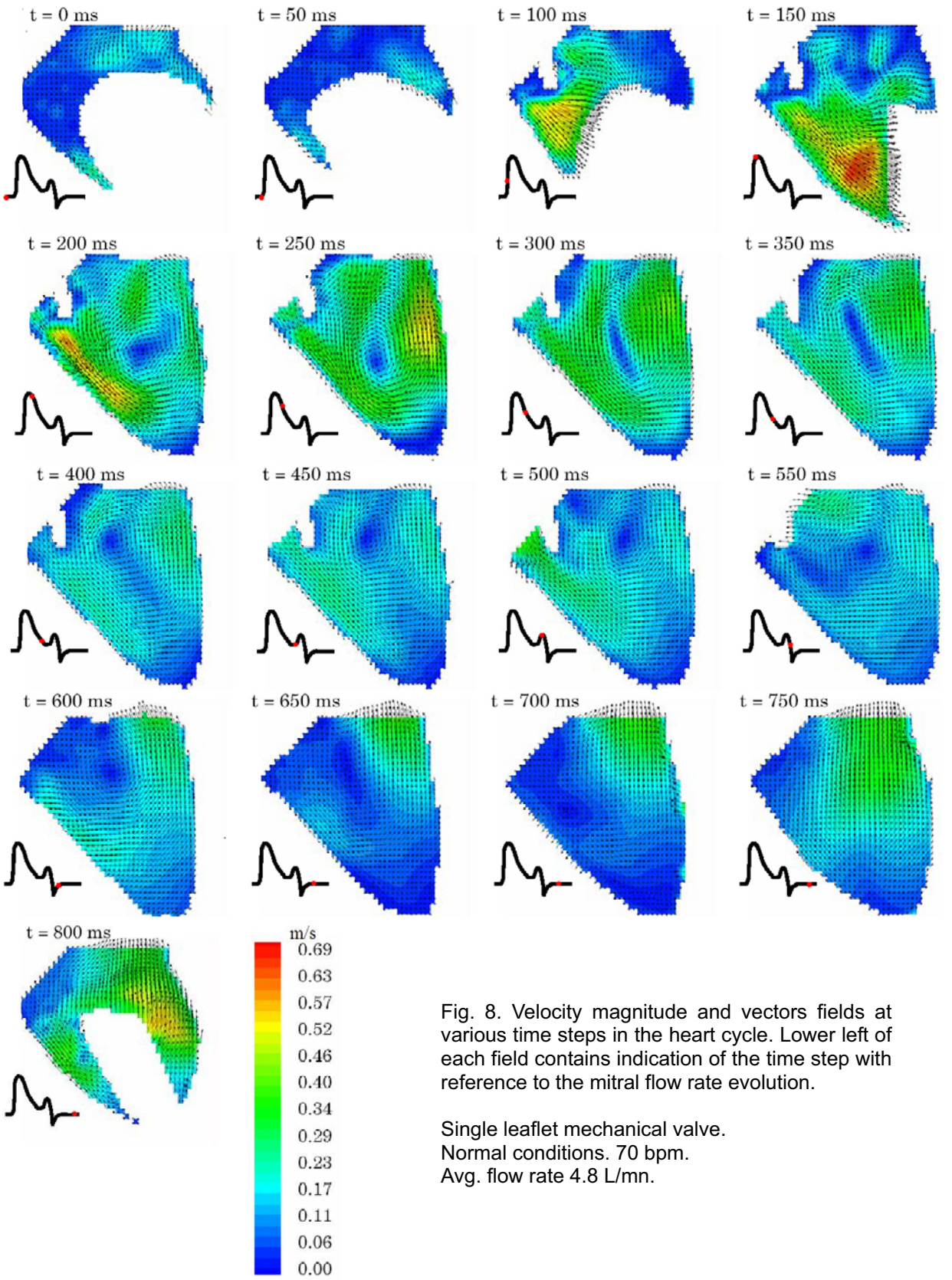

Fig. 8. Velocity magnitude and vectors fields at various time steps in the heart cycle. Lower left of each field contains indication of the time step with reference to the mitral flow rate evolution.

Single leaflet mechanical valve.
Normal conditions. 70 bpm.
Avg. flow rate 4.8 L/mn.

may also interact with left ventricular outflow in early systole. This interaction is of course dependant on the valve geometry: for example bi-leaflet mechanical valves or bioprosthetic valves will not exhibit the same flow structures and will be less sensitive to these structures because of their "shorter" leaflets.

The present work represents therefore only a first step in the application of PIV in the mockup. These macroscopic results are encouraging and the next steps will be to get a closer look into the flow structures close to the valve. The regurgitation is a significant problem for mechanical valves and the accurate description of the flow-structure interaction can contribute to prevent or reduce this problem. This study relays on concentrating acquisitions around the time gate within which the valve closes. Both stereoscopic PIV measurements and High frequency PIV could be useful tools for this study.

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Author Profile

KNAPP Yannick : He received his Ph.D. in Fluid Mechanics in 1997 from Universite de la Mediterranee Aix-Marseille II, Marseille, France. After working as a consultant in fluid-structure interactions for his own company he now works at Universite Avignon – France. His research interests are Quantitative Visualization, PIV and Automated Flow Structure Identification applied to Flow Induced Vibrations.

BERTRAND Eric : He is a research engineer at Universite de la Mediterranee Aix-Marseille II, Marseille, France. He proceeded to this position through the higher education system of the same university starting with an undergraduate degree in mechanical engineering. His current research interests are the design of new experimental setups in the field of biomedical fluid mechanics involving 2 and 3 components PIV measurements.