Experiments in Fluids (2005) 39: 1051–1060
DOI 10.1007/s00348-005-0038-0

RESEARCH ARTICLE

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A new experimental method for the determination of the effective orifice area based on the acoustical source term

Received: 20 December 2004 / Revised: 15 August 2005 / Accepted: 15 August 2005 / Published online: 30 September 2005

Abstract The effective orifice area (EOA) is the most commonly used parameter to assess the severity of aortic valve stenosis as well as the performance of valve substitutes. Particle image velocimetry (PIV) may be used for in vitro estimation of valve EOA. In the present study, we propose a new and simple method based on Howe's developments of Lighthill's aero-acoustic theory. This method is based on an acoustical source term (AST) to estimate the EOA from the transvalvular flow velocity measurements obtained by PIV. The EOAs measured by the AST method downstream of three sharp-edged orifices were in excellent agreement with the EOAs predicted from the potential flow theory used as the reference method in this study. Moreover, the AST method was more accurate than other conventional PIV methods based on streamlines, inflexion point or vorticity to predict the theoretical EOAs. The superiority of the AST method is likely due to the nonlinear form of the AST. There was also an excellent agreement between the EOAs measured by the AST method downstream of the three sharp-edged orifices as well as downstream of a bioprosthetic valve with those obtained by the conventional clinical method based on Doppler-echocardiographic measurements of transvalvular velocity. The results of this study suggest that this new simple PIV method provides an accurate estimation of the aortic valve flow EOA. This new method may thus be used as a reference method to estimate the EOA in experimental investigation of the performance of valve substitutes and to validate Doppler-echocardiographic measurements under various physiologic and pathologic flow conditions.

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1 Introduction

Over the past years, there have been numerous studies on the flow dynamics through orifices in ducts and its various hydraulic applications; an excellent review of these studies can be found in Idelchik's book (2001). As the upstream uniform flow passes through a circular orifice in a duct, it is usually separated in three distinct zones: an axisymmetric jet-like zone, a recirculation zone and a reattachment zone. Due to inertial forces in the jet-like zone, the flow first converges (as far as the vena contracta), leading to a flow area smaller than the orifice area. This area is usually called the effective orifice area (EOA) and is the location of the highest velocity and the lowest static pressure. Downstream of the vena contracta, the flow diverges and reattaches to the wall. The reattachment length is highly dependent on the flow regime and the spreading angle is approximately 20–25° for turbulent steady flows (Landau and Lifshitz 1987; Tkachuk et al. 1993).

In the clinical context, such a flow behavior can be observed in patients with aortic stenosis disease. Aortic stenosis is defined as an incomplete opening of the aortic valve. It is considered as the most common cardiac disease in developed countries after coronary artery disease and arterial hypertension (Braunwald et al. 2001). It is important to estimate the severity of the valvular orifice stenosis because it determines the mechanical overload imposed on the left ventricle and it is a major criterion in the decision process leading to aortic valve replacement surgery. The aortic valve flow EOA is one of the most important parameters to assess the degree of severity of aortic stenosis. It can be measured noninvasively by Doppler-echocardiography with the use of the continuity equation.

However, controversy remains about the validity of the Doppler-echocardiographic method for the estimation of the EOA at low flow rate conditions, which may occur in patients with low cardiac output. For clinical applications, the continuity equation is indeed used under the assumption of a flat velocity profile at the level of the vena contracta. As shown by various authors (Trujillo et al. 1996; Shandas 1996; DeGroff et al. 1998; Mascherbauer et al. 1999) this may not be true for low flow rates ($<2\ \mathrm{l\ min^{-1}}$), where the velocity profile may be more parabolic or semiparabolic leading to an underestimation of the EOA and thus an overestimation of the severity of the valvular stenosis.

Unfortunately, there is no gold standard method for the measurement of valve EOA in vivo, thus precluding the direct validation of the Doppler-echocardiographic method in patients or in animal models. The validation of this method therefore requires the realization of in vitro studies under pulsatile flow conditions, where the EOA is generally measured using particle image velocimetry (PIV) as the reference technique (Mascherbauer et al. 1999; Shandas et al. 2000). However, all previous studies compared the instantaneous EOA measured by PIV at peak ejection with the EOA measured by Doppler-echocardiography, which represents the EOA averaged over the whole systolic ejection period. This is an important limitation given that it has been demonstrated that the EOA may vary markedly in calcified stenosis during ventricular ejection (Arsenault et al. 1998; Bermejo et al. 2000; Beauchesne et al. 2003). Moreover, there is no previously published study comparing the feasibility and performance of the different PIV methods for the measurement of EOA in the context of left ventricular outflow.

Recent advances in hydro-acoustics have shown that the different zones developed downstream of a model of a stenosis are associated with a significant production of acoustic noise (Borisyuk 2002). This is particularly true for the zones corresponding to separated and reattached flow. Within the separation region, the most important sound sources are associated with two groups of large-scale eddies. The first one corresponds to the jet region and the second one to the recirculating flow region near the wall.

The objective of the present study is to propose a reference method to distinguish the different regions within the separated zone downstream of the stenosis based on an identification in terms of acoustic sources, and therefore to obtain an accurate estimation of the aortic valve flow EOA in vitro from PIV measurements under pulsatile flow conditions.

This method is based on the application of the vortex sound theory, introduced by Powell (1964) and used by Howe (2002, 2004). This theory shows that the primary source of sound generation in unsteady low Mach number flows is due to the presence of vorticity:

$$\left(\frac{\mathrm{D}}{\mathrm{D}t}\left(\frac{1}{c^2}\frac{\mathrm{D}}{\mathrm{D}t}\right) - \nabla^2\right)B = \nabla\cdot(\omega\wedge V) \quad (1)$$

where D is the total derivative, c is the speed of sound in the region surrounding the flow region; B is the total enthalpy; ω is the vorticity field and V is the velocity field. This equation has been derived under the following assumptions: (1) the fluid is incompressible and homentropic (uniform specific entropy); (2) only conservative body forces are present; (3) absence of sources of matter; (4) high Reynolds number; (5) the effect of viscosity is neglected. This latest assumption is valid because the viscosity may attenuate the sound once it has been generated and is of no particular interest when studying sound generation mechanisms (Powell 1964; Howe 2002).

The term on the right side $[\nabla\cdot(\omega\wedge V)]$ is considered as the acoustical source term (AST). One can note that this term is included in the Reynolds stress term of the more general Lighthill's acoustical source (Miyauchi et al. 2001).

2 Experimental methods

For the purpose of this study, we used a mock flow circulation model (Fig. 1a, b), previously described and validated (Garitey et al. 1995; Kadem et al. 2002; Knapp and Bertrand 2005). This model is composed of silicone and anatomically shaped models of the left heart cavities and aorta. The areas of the ventricle at the level of the aortic valve (left ventricle outflow tract) and of the aortic root are 7.6 and 4.3 $\mathrm{cm^2}$, respectively. The ventricle is activated by a computer-driven pump that was specifically designed to reproduce physiological flow and pressure waveforms. The fluid is pumped from an open tank through the activation section to a box containing the model of the left ventricle. When the flow is injected into the box it compresses the ventricle. The circulatory fluid contained in the ventricle is then ejected into the ascending aorta and the systemic arterial model (compliance and resistance). When the fluid is extracted from the activation box, the ventricle expands and the circulatory fluid then flows from the left atrium model to the left ventricle. The dilation (diastole) and contraction (systole) of the ventricle lead to the opening of the mitral and aortic valves, respectively. The compliance and resistance of the systemic arterial system can be adjusted to ensure physiological aortic pressure waveform. When a fixed orifice replaces the aortic valve, this has a significant impact on the diastolic phase of the flow waveform (a more important reversal flow) but a minimal impact on the systolic phase. Furthermore, in this configuration, the reversed velocity is low compared to the mitral flow velocity ($\sim$15 vs $\sim$45 $\mathrm{cm\ s^{-1}}$) and the vortex structures that may be present in the left ventricle are probably broken by the ventricular accelerating flow

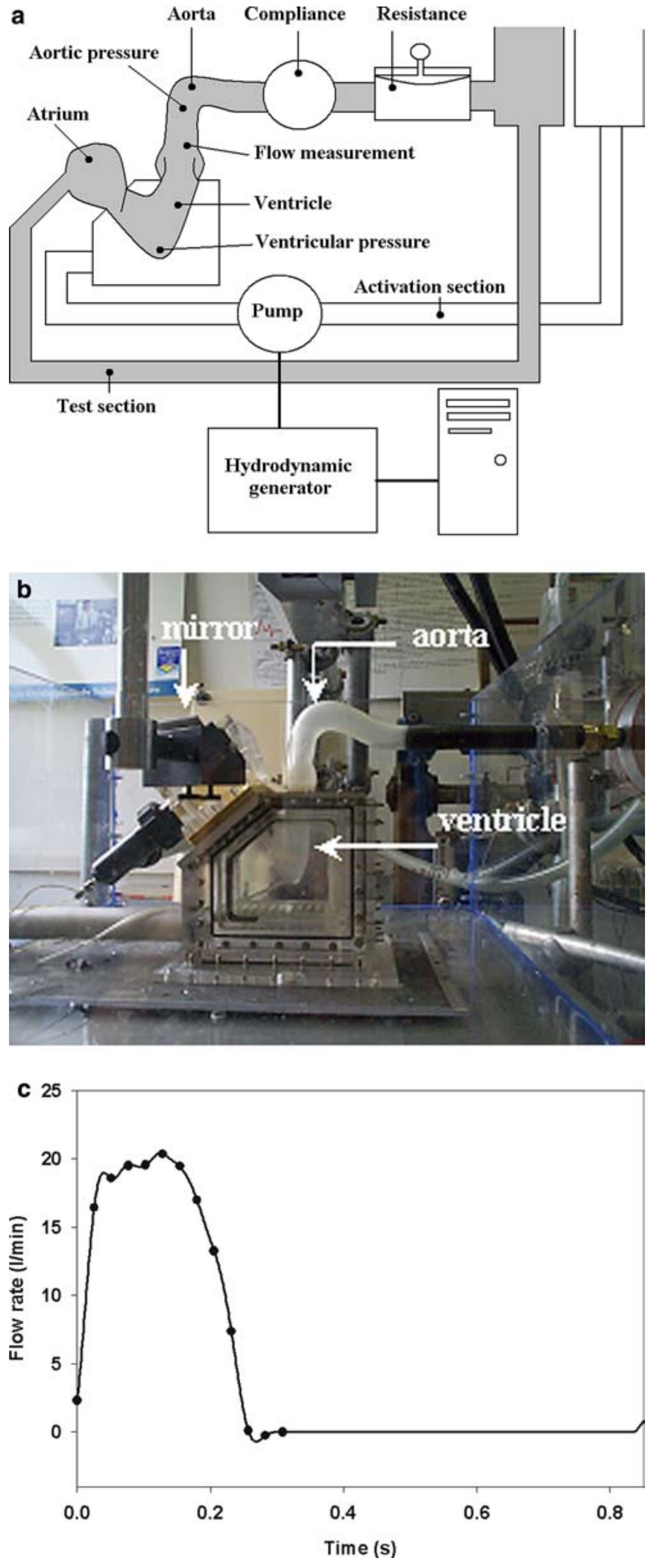

Fig. 1 Schematic representation (**a**) and photography (**b**) of the mock flow circulation model. (**c**) Typical flow measurement downstream of a Medtronic bioprosthetic 25-mm valve, the *dots* represent the instants of PIV measurments

during the systolic phase. Therefore, the presence of a reversed flow should have a minimal effect on the velocity profile at the vena contracta.

The fluid used in the test section is a transparent mixture of water (60%) and glycerol (40%) (viscosity 4×10^{-3} kg m^{-1} s^{-1}) (Nichols et al. 1998). The use of such Newtonian fluid is justified by the fact that blood behaves like a Newtonian fluid at the level of the ascending aorta (the aorta diameter is large compared to that of the red cells and the mean shear rate at the wall is 80 s^{-1}) (Nichols et al. 1998; Zamir et al. 2000; Borisyuk 2002). Pressure measurements were performed using Millar catheters (MPC 500, accuracy 0.5%). The transducers of the proximal and distal catheters were located in the left ventricle and at 7 cm downstream of the aortic valve, respectively (Fig. 1a). Transvalvular flow rate was measured with an ultrasonic flowmeter (Transonic probe 28A31; accuracy 1%) at the level of the ascending aorta (Fig. 1a).

3 Doppler-echocardiographic measurements

Doppler-echocardiographic measurements were performed using an ATL Ultramark 9 with a probe of 2.25 MHz. The probe was located at the level of the aortic arch and the Doppler beam was aligned with the direction of flow to allow measurement of flow velocity in the ascending aorta. In order to avoid aliasing, the continuous-wave Doppler mode was used. This allowed the determination of the temporal repartition of the maximal velocity in the ascending aorta and thus the measurement of the velocity at the location of the vena contracta.

From this measurement, the velocity–time integral (VTI) was obtained, and the EOA was calculated by dividing the stroke volume (SV) (the volume ejected during one beat) measured by the ultrasonic flowmeter by the VTI:

$$\mathrm{EOA} = \frac{\mathrm{SV}}{\mathrm{VTI}} = \frac{\mathrm{SV}}{\int_0^{\mathrm{Ts}} V\mathrm{d}t}$$

where V is the instantaneous velocity measured by echo-Doppler and Ts is the systolic duration. The interexperiment variability of the EOA measurements by Doppler-echocardiography was $\pm 3\%$.

4 PIV measurements

Particle image velocimetry measurements were performed at the level of the ascending aorta using a double-pulsed mini-YAG laser (120 mJ, 15 Hz) and a CCD camera PIVCAM30 (30 Hz) from TSI inc (Shoreview, MN, USA). The seeding particles were small white polycrystalline polymethacrylate particles in the range

40–80 μm and average density of 1.319 g cm^{-3}. The choice of these particles was in good agreement with the theoretical qualification of the particles in terms of settling velocity (less than 0.1 mm s^{-1}) and relaxation time (in the range 40–150 μs). The measurements were performed at 12 instants during the systolic phase (Fig. 1c). The duration between two successive instants was 25 ms. To obtain phase averaged velocity fields, 80 pairs of images were recorded for each instant. We verified that the number of images was sufficient to ensure correct statistical convergence, using the following ratio (Anthoine et al. 2003):

$$R(n, N) = \frac{(1/n)\sum_{i=1}^{n} V_i}{(1/N)\sum_{i=1}^{N} V_i}$$

The velocity was measured at a point located in the shear layer. The fluctuations of $R(n, N)$ did not exceed 4% for $n > 60$ (n is the number of pairs of images, $N = 100$). A standard cross-correlation was performed to determine the displacement of the particles within an interrogation region of 64×64 pixel and a 50% overlapping with a subpixel Gaussian interpolation. In order to increase the spatial resolution, an additional oversampling was used. Since we used the eight-point circulation method for vorticity computation, the data oversampling has a minimal effect on vorticity computation compared to other methods for vorticity computation (Raffel et al. 1998).

This resulted in a spatial resolution of 1×1 mm^2, owing to the constant 56 μm/pixel scale measured in the area of interest. This PIV algorithm is very similar to the one recently used by Balducci et al. (2004) to study the flow past an artificial heart valve.

Since the nature of flow was pulsatile, important temporal velocity gradients may exist. Hence, to obtain adequate velocity profiles at the level of the vena contracta, it was necessary to adjust the duration (Δt) between the two laser pulses depending on the instantaneous flow velocity. The Doppler-echocardiographic signal was thus first analyzed to determine instantaneous flow velocities corresponding to the different instants of PIV measurements. Then, the (Δt) was estimated with respect to the "one quarter rule" (Keane and Adrian 1990; Stitou et al. 2001) and applied to the PIV settings (see Table 1).

Table 1 Characteristics of the flow downstream of the three sharp-edged orifices and the bioprosthetic valve tested in this study

	V_{max} (m s^{-1})	Re_{max}	Δt (μs) mean [range]	St
Bioprosthesis	1.77	4,625	397 [150–1,000]	0.19
Orifice 1.5	4.61	5,383	273 [80–1,000]	0.12
Orifice 1.0	4.80	5,794	220 [80–1,000]	0.069
Orifice 0.5	5.08	5,955	174 [60–1,000]	0.026

V_{max} maximal axial velocity during the systolic phase, Re_{max} maximal Reynolds number, Δt duration between two laser pulses, *St* Strouhal number

5 EOA determination

From the PIV measurements, the velocity field in the ascending aorta was obtained for the 12 instants of the systolic phase. The AST field was then computed over the whole field from the discrete formulation of $(\nabla \cdot (\omega \wedge V))$. The vorticity term was computed using the eight-point circulation method (Raffel et al. 1998) and the spatial derivatives with center difference scheme inside the domain and forward or backward scheme at the domain boundaries. The radial evolution of AST was extracted from each field at the level of the vena contracta. The position of the vena contracta was defined as the section where the velocity was maximal within the velocity field. At each instant of the systolic phase, the diameter of the vena contracta was inferred from this radial evolution as the distance between the two negative peaks of the AST (corresponding to maximal noise production due to vorticity), and the EOA was calculated from this distance assuming a circular shape (Fig. 4). The instantaneous EOA was then averaged over the systolic phase and compared to the EOA predicted from the potential flow theory and to the EOA measured by Doppler-echocardiography. The interexperiment variability of the EOA measurements by AST method was ±4%.

The PIV measurements of the instantaneous EOA at several time points during the systolic phase is an original aspect of the present study since they allow the measurement of the phase averaged EOA, whereas previous studies measured only the instantaneous EOA at peak systole (Mascherbauer et al. 1999; Shandas et al. 2000).

6 Determination of EOA by other PIV methods

The phase averaged EOA over the systolic phase was also determined using three conventional methods that may be used for this purpose:

Streamlines method The streamlines were plotted using the TecPlot software (Tecplot Inc, Bellevue, WA, USA). The location of the vena contracta was first determined by defining the location of the maximal velocity within the flow. For each instant of measurements, the diameter of the vena contracta was defined as the distance between the two streamlines that correspond to the interface between the jet-like zone and the recirculation zone (the limit between the parallel streamlines zone and the zone where the streamlines roll-up). Then EOA was determined assuming a circular shape. A very similar technique was used by DeGroff et al. (1998) using pathlines instead of streamlines since the flow was steady. However, this method is highly operator-dependent.

Method based on the inflexion points on the velocity profile For each instant of measurements, a polynomial fitting was performed on the positive part of the velocity profile at the level of the vena contracta. The polynomial order was automatically determined (the maximal order was chosen arbitrary equal to 8) to minimize the root mean square error between the measured profile and the polynomial profile. The second derivative was then computed using the polynomial fitting to determine the two inflexion points on the velocity profile. The EOA of circular shape was then inferred from the distance between these two inflexion points considered as representative of the diameter.

Method based on the vorticity profile For each instant of measurements, the vorticity distribution within the flow was computed using the eight-point circulation method (Raffel et al. 1998). The vorticity distribution at the level of the vena contracta was extracted and the EOA of circular shape was then inferred from the distance between the two extrema of the vorticity profile considered as representative of the diameter.

7 Experimental conditions

Three sharp-edged orifices with a 5-mm thickness (geometric area: 0.5, 1.0, 1.5 cm^2) and a Medtronic aortic bioprosthetic 25-mm valve were tested in the mock flow model. The stroke volume was fixed at 70 ml, heart rate at 70 bpm and left ventricular ejection time at 300 ms, thus resulting in a cardiac output of 4.9 $\mathrm{l\ min^{-1}}$ and a mean transvalvular flow rate of 233 $\mathrm{ml\ s^{-1}}$. The systemic arterial resistance and compliance were adjusted to maintain systolic and diastolic aortic pressures close to 120 and 80 mmHg, respectively. This experimental setting reproduced the hemodynamic conditions of a normotensive patient with a normal flow rate.

8 Validation of the AST method and comparison with other PIV and Doppler methods

To validate the new method proposed in this study, the phase averaged EOA and the contraction coefficient obtained by the AST method were compared with those obtained from the steady flow theory (potential flow) (Batchelor 1967; Grose et al. 1985; Milne-Thompson 1996). It is indeed well established that the contraction coefficient is close to 0.61 when the following conditions are present: (1) the flow is uniform and steady, (2) the stenotic orifice is a circular sharp-edged orifice with a small aspect ratio (ratio between orifice thickness and the orifice area), (3) the ratio of the geometrical orifice area to the inflow area is small and (4) the Reynolds number is high. In the present study, the aspect ratio ranged from 0.36 to 0.63 and for this range the contraction coefficient is still close to the theoretical value ($\mathrm{Cc} = 0.61$) (the orifice cannot be considered as a small tube). Also, the ratio of the geometrical orifice area to the inflow area ranged from 0.07 to 0.2. It has been demonstrated that the contraction coefficient remains close to 0.61 within this range (Grose et al. 1985). However, one could argue that the transvalvular flow simulated in the present study was unsteady and that this could have limited the utilization of the potential flow theory to validate the AST method. Nonetheless, if we calculate the Strouhal number for sharp-edged orifices using the expression introduced by Clark (1976):

$$\mathrm{St} = 2.87\,\frac{Q_{\max}}{Q_{\mathrm{mean}}}\frac{\mathrm{GOA}^{3/2}}{\mathrm{SV}}$$

where $Q_{\max}$ is the maximal flow rate, Q_{mean} is the mean systolic flow rate, GOA is the geometric orifice area of the stenosis and SV is the stroke volume; it appears that the unsteady effects are relatively small compared to the inertial effects: orifice 1.5 cm^2 ($\mathrm{St} = 0.12$); orifice 1.0 cm^2 ($\mathrm{St} = 0.069$); orifice 0.5 cm^2 ($\mathrm{St} = 0.026$). This suggests that the flow can be considered as quasisteady and that the potential flow theory can be used as a reference method in the present study. The theoretical EOA was thus determined in the three sharp-edged orifices as the product of $\mathrm{GOA} \times 0.61$. The theoretical EOA could not be determined in the bioprosthetic valve because the GOA was not measured. The EOA and contraction coefficient that were obtained by the AST method, the three conventional PIV methods and the Doppler-echocardiographic method were compared to those predicted from the potential flow theory.

Since it has been demonstrated that the Doppler method provides an accurate estimation of the valve EOA at normal flow rates (Mascherbauer et al. 1999; Garcia et al. 2004), we also compared the new PIV method, proposed in this study, to the Doppler method in conditions of normal flow rate.

9 Results

Figure 2a shows typical results of PIV measurements for a sharp-edged orifice of 1.0 cm^2 during systolic outflow (instants 25–150 ms). Figure 2b shows the evolution of the axial velocity profile at the vena contracta during the systolic phase for the same orifice. During the diastolic phase, the velocity is negative within the whole field due to the presence of an open orifice rather than a valve, whereas during the systolic phase, the flow behaves like a circular jet in a confined domain. In fact, at the onset of the systolic phase, a starting structure emerges. The velocity at the center of the field rapidly grows with flow acceleration and the influence of the stenosis is still perceptible, more than five orifice sections downstream of the aortic root. When the flow begins to decelerate (from 150 to 225 ms), the velocity decreases in the whole field and the velocity profile is more spread-out than during the acceleration phase. Then (instants 250–275 ms), due to the presence of a fixed orifice, the

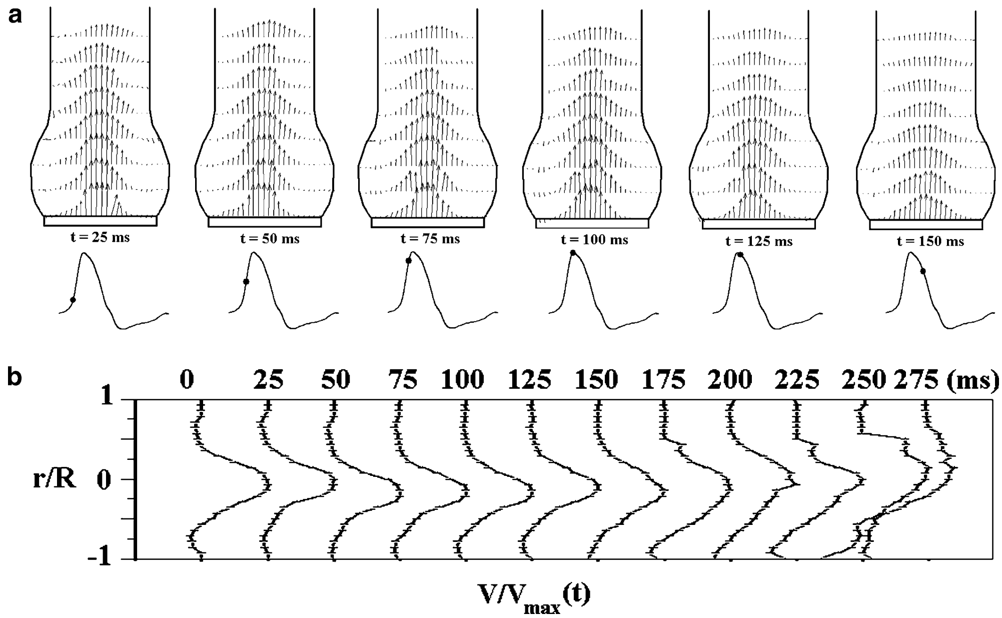

Fig. 2 **a** PIV measurements downstream of a sharp-edged orifice of 1.0 cm^2 (instants 25–150). **b** Evolution of the normalized axial velocity profile at the vena contracta (instants 0–275) (orifice 1.0 cm^2)

reversed flow occupies a more significant region within the field until the onset of the diastolic phase, where the field is dominated by a reversed flow. The maximal axial velocity during the systolic phase was 5.08 m s^{-1} for orifice 0.5 cm^2, 4.80 m s^{-1} for orifice 1.0 cm^2, 4.61 m s^{-1} for orifice 1.5 cm^2, and 1.77 m s^{-1} for the bioprosthetic valve. The peak Reynolds number ranged from 4,625 to 5,955 (maximal velocity and aortic root diameter were used for the calculation of the peak Reynolds number). The frequency parameter (Womersley number) is defined as:

$$\alpha = R\sqrt{\frac{\omega}{\upsilon}}$$

where R is the aortic root radius, ω is the pulsation and υ is the kinematic viscosity, and has a value of 16.6, which is very similar to the one observed in large human arteries (Nichols et al. 1998). Table 1 summarizes the main characteristics of the flow at the level of the vena contracta for the three sharp-edged orifices and the bioprosthetic valve tested in this study.

Figure 3 shows the evolution of the AST fields computed from PIV measurements for an orifice of 1.0 cm^2 (instants 25–150 ms). It seems that most part of the AST is confined in the separation region and the higher sound is produced during the acceleration phase (instants 25–125). Figure 4 shows the radial distribution of the normalized axial velocity and AST at the level of the vena contracta for the same orifice (instant $t = 50$ ms). It can be seen that the AST profile has three extrema: one positive (corresponding to the peak velocity) and two negatives (corresponding approximately to the two inflexion points on the velocity curve). These three extrema define the regions of maximal sound production.

Figure 5 and Table 2 show the comparison between phase averaged EOA measured by PIV using the AST (EOA_{AST}), the EOA measured by Doppler echocardiography ($EOA_{Doppler}$), and the EOA predicted from the theory (i.e., assuming a contraction coefficient of 0.61) (EOA_{Theor}). When compared to EOA_{Theor}, the mean absolute and relative errors were: 0.04 ± 0.02 cm^2 and $10 \pm 8\%$ for EOA_{AST} and 0.03 ± 0.03 cm^2 and $8 \pm 10\%$ for $EOA_{Doppler}$. The mean absolute and relative differences between $EOA_{Doppler}$ and EOA_{PIV} were 0.03 ± 0.02 cm^2 and $3 \pm 2\%$.

During the systolic phase, there was only a minimal variation in the EOA_{AST}: orifice 0.5 cm^2 ($\pm 6\%$); orifice 1.0 cm^2 ($\pm 4\%$); orifice 1.5 cm^2 ($\pm 3\%$) and for bioprosthetic valve ($\pm 3\%$). The variation in valve EOA during systole would likely be more important in aortic valves with stiff leaflets that may occur in patients with calcified aortic valve stenosis (Arsenault et al. 1998; Bermejo et al. 2000; Beauchesne et al. 2003). In this situation, the opening rate of the valve leaflets is indeed lower than in native or prosthetic valves with normal flexible leaflets.

Table 2 shows the results of the phase averaged EOA and the contraction coefficient (only for rigid orifices) obtained by the Doppler-echocardiographic method, the AST method, the three conventional PIV methods, and also the theoretical EOA. The streamlines method generally tends to overestimate the Doppler EOA and the theoretical EOA, especially in the orifices 0.5 and 1.5 cm^2. However in the bioprosthetic valve, this method underestimates the Doppler EOA. As opposed to the streamlines methods, the methods based on the inflexion point and the vorticity profile tend to underestimate the Doppler EOA and the theoretical EOA in all cases, except for the 0.5 cm^2 orifice. It is surprising to see that the conventional PIV methods seem to be less accurate with increasing orifice area. Finally, the AST method slightly overestimates the theoretical EOA and thus the contraction coefficient only in the 0.5 cm^2 orifice, but

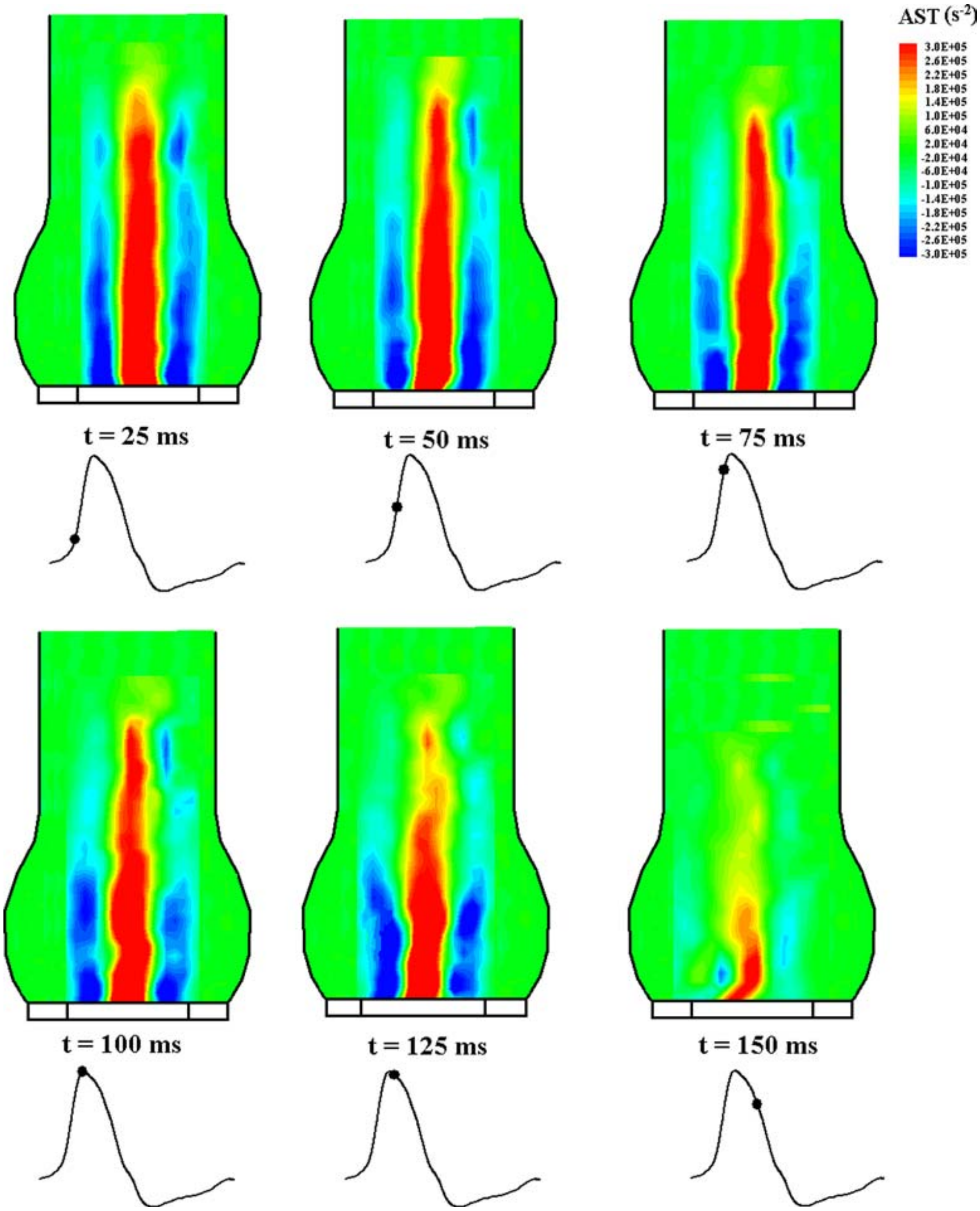

Fig. 3 Acoustical source term (*AST*) field downstream of a sharp-edged orifice of 1.0 cm^2 at instants comprised between 25 and 150 ms

interestingly it is similar to the values obtained by Doppler-echocardiography (Fig. 5, Table 2). On the other hand, in the 1.0 and the 1.5 cm^2 orifices, the EOAs and contraction coefficients determined by the AST method were quite consistent with those predicted from the theory.

10 Discussion

Particle image velocimetry is a very robust technique widely used in experimental fluid mechanics and more recently in the study of the flow downstream of aortic valve substitutes (Shandas et al. 2000; Lim et al. 2001; Brucker et al. 2002; Balducci et al. 2004; Marassi et al. 2004). From PIV measurements, it is possible to compute the AST resulting from the vortex sound theory. This theory allows defining the regions where a sound can be generated within the fluid. In the case of an aortic stenosis, the largest part of the sound generated is concentrated in the region of flow separation downstream of the stenosis. This sound mainly results from the perturbations induced in the flow and in the convected vortices (Fig. 6) (Abdallah 1987; Borisyuk 2002). It is then possible to use an acoustical marker, such as the AST, to separate the structures within the separation region and to accurately define the EOA downstream an orifice.

The results of this study show that there is generally a very good agreement between the EOA measured by the AST method and the EOA predicted from the theory in the sharp-edged orifices. Moreover, there was also a good agreement between the EOA estimated by the AST method and that measured by the Doppler

Table 2 Comparison of the effective orifice area (EOA) and contraction coefficient (Cc) obtained from the theory, the Doppler-echocardiographic method, and the four PIV methods

	Theoretical	Doppler	Acoustical source term	Streamlines	Inflection point	Vorticity
Orifice 0.5						
EOA (cm^2)	0.31	0.37	0.37	0.42	0.32	0.30
Cc	0.61	0.74	0.74	0.84	0.64	0.6
Orifice 1.0						
EOA (cm^2)	0.61	0.63	0.66	0.64	0.35	0.51
Cc	0.61	0.63	0.66	0.64	0.35	0.51
Orifice 1.5						
EOA (cm^2)	0.92	0.91	0.94	1.11	0.44	0.78
Cc	0.61	0.61	0.63	0.74	0.29	0.52
Bioprosthesis						
EOA (cm^2)	–	1.54	1.58	1.37	1.21	0.96
Cc	–	–	–	–	–	–

echocardiographic method at normal flow rate in the orifices as well as in the bioprosthetic valve. However, in very small orifices, the AST method tends to overestimate the EOA and thus the contraction coefficient predicted from the theory. One cannot exclude that this overestimation may be due to a lack of accuracy of the AST method, because of a low spatial resolution, for the measurement of very small EOAs. As a matter of fact, the use of cumulative differentiations reduces the spatial resolution of the original PIV measurements and the relative errors in EOA determination are dependent on $(\Delta r/D)$, where Δr is the spatial resolution in the radial direction and D is the theoretical diameter of the vena contracta. Nonetheless, it should be pointed out that the Doppler-echocardiographic method is consistent with the AST method and also yields to an overestimation of the EOA, when compared with the theoretical value. Hence, it is also quite possible that the AST method indeed provides an accurate measurement of the true EOA and thus of the true contraction coefficient in small orifices. The discrepancy with the theory could then be explained by the fact that for smaller orifices, the pulsatile flow cannot be contracted further. These results are also in agreement with the study of Voelker et al. (1995) where they measured the EOA using an hydraulic model and found slightly higher contraction coefficient in smaller orifices: orifice 1.5 cm^2 (Cc = 0.6); orifice 1.0 cm^2 (Cc = 0.69) and orifice 0.5 cm^2 (Cc = 0.74).

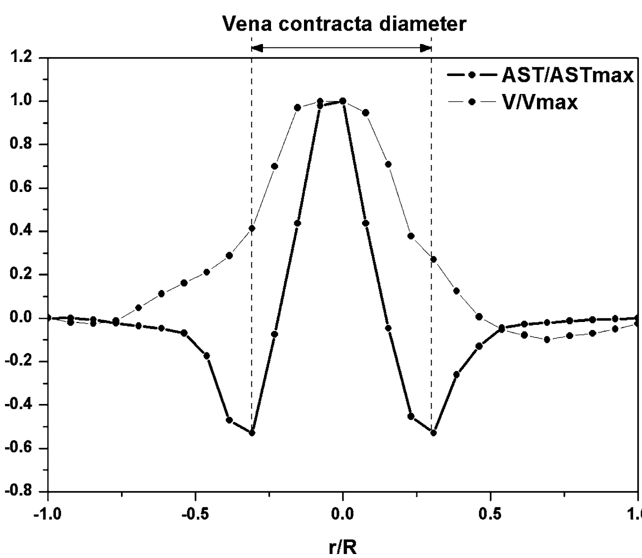

Fig. 4 Radial distribution of the normalized acoustical source term (*AST*) and axial velocity (*V*) at the level of the vena contracta (Orifice 1.0 cm^2; instant $t = 50$). *ASTmax* maximal acoustical source term, *Vmax* maximal axial velocity

The results of the present study suggest that the AST method provides more accurate estimates of the EOA in rigid circular orifices and bioprosthetic valves, compared to other conventional PIV methods. Theoretically, if one assumes an infinitesimally high spatial resolution, conventional methods should also provide an accurate estimation of the EOA. However, experimental techniques such as PIV impose a finite spatial resolution. Hence, a small error in the determination of the diameter of the vena contracta may lead to relatively important errors in the calculation of the EOA. This is particularly true for small orifices. However, in the context of valve prostheses evaluation the orifice is large which tends to decrease the relative errors in EOA determination.

The fact that the new method proposed in this study appears to be equal or better than other PIV-based methods is probably due to the fact that the AST contains nonlinear terms under the form

$$\frac{\partial}{\partial x_i}\left(V_j \frac{\partial V_i}{\partial x_j} - V_j \frac{\partial V_j}{\partial x_i}\right),\ i \neq j$$

which tend to amplify the variations of the vorticity, allowing a better discrimination between poststenosis flow structures. This is a very relevant point, if one considers that in clinical practice, the spatial resolution of the noninvasive techniques for velocity measurements are around 2×2 mm^2 for the echo-PIV technique,

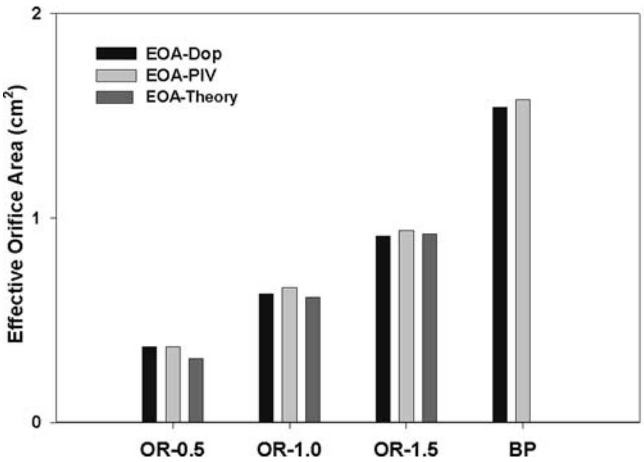

Fig. 5 Comparison between EOA measured by Doppler echocardiography, EOA measured by PIV using the AST and theoretical EOA. *BP* Bioprosthetic valve; *OR-0.5, OR-1.0, OR-1.5* fixed orifices with geometric orifice areas of 0.5, 1.0, and 1.5 cm^2, respectively

recently proposed by Kim et al. (2004), and around $1{\times}1\ mm^2$ for velocity measurements by magnetic resonance imaging (MRI). Therefore, the application of this new method may be useful for in vitro validation of the measurement of aortic valve flow EOA by Doppler-echocardiography or by catheterization in specific pathologic conditions such as in the presence of low flow rate. This method may also be useful to validate and optimize new emerging technologies (e.g., MRI) for the measurement of valve EOA.

However, several future works are still necessary in order to further increase the accuracy of the AST method. The actual spatial resolution of the method induces a filtering effect on the velocity profiles. The response of this filter must be determined under reference flow conditions such as continuous or oscillatory flows in a straight tube with sharp-edged orifice. Also, additional studies based on numerical simulations in pulsatile flow conditions would be useful to further validate the AST method.

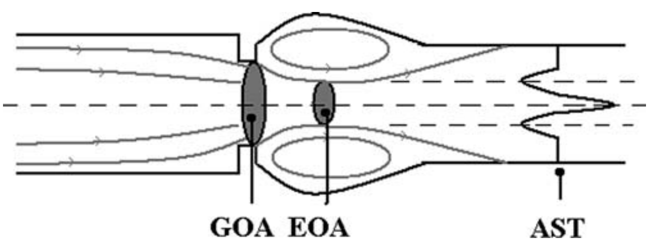

Fig. 6 Schematic representation of the application of the AST to the measurement of the effective orifice area in aortic stenosis. *GOA* geometric orifice area, *EOA* effective orifice area, *AST* acoustical source term

11 Conclusion

In this study, a new method to estimate the valve EOA has been proposed. This method is based on the vortex sound theory and the calculation of the AST from the velocity field obtained by PIV. The results of the study suggest that this new simple PIV method provides a relatively accurate estimation of the aortic valve flow EOA. This new method may thus be useful to validate the Doppler-echocardiographic and catheterization measurements of EOA at low flow rates. It may also be useful to measure the normal reference values of EOA of the different models of prosthetic valve and thus to evaluate and compare their hemodynamic performance.

Acknowledgments The authors thank Claudia Jones, Martine Lauzier and Mickael Savinaud for their technical assistance. This work was supported by a grant of the Canadian Institutes of Health Research (MOP-10929), Ottawa, Ontario, Canada. Dr. Pibarot is the director of the Canada Research Chair in Valvular Heart Diseases, Canadian Institutes of Health Research, Ottawa, Ontario, Canada.

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