Echocardiographic Particle Image Velocimetry: A Novel Technique for Quantification of Left Ventricular Blood Vorticity Pattern

Arash Kheradvar, MD, PhD, Helene Houle, RDCS, Gianni Pedrizzetti, PhD, Giovanni Tonti, MD, Todd Belcik, RDCS, Muhammad Ashraf, MD, Jonathan R. Lindner, MD, Morteza Gharib, PhD, and David Sahn, MD, Columbia, South Carolina; Mountain View and Pasadena, California; Trieste and Sulmona, Italy; Portland, Oregon

Background: In this study, the functionality of echocardiographic particle imaging velocimetry (E-PIV) was compared with that of digital particle imaging velocimetry (D-PIV) in an in vitro model. In addition, its capability was assessed in the clinical in vivo setting to obtain the ventricular flow pattern in normal subjects, in patients with dilated cardiomyopathy, and in patients with mechanical and bioprosthetic mitral valves.

Methods: A silicon sac simulating the human left ventricle in combination with prosthetic heart valves, controlled by a pulsed-flow duplicator, was used as the in vitro model. Particle-seeded flow images were acquired (1) using a high-speed camera from the mid plane of the sac, illuminated by a laser sheet for D-PIV, and (2) using a Siemens Sequoia system at a frame rate of 60 Hz for E-PIV. Data analysis was performed with PIVview software for D-PIV and Omega Flow software for E-PIV. E-PIV processing was then applied to contrast echocardiographic image sets obtained during left ventricular cavity opacification with a lipid-shelled microbubble agent to assess spatial patterns of intracavitary flow in the clinical setting.

Results: The velocity vectors obtained using both the E-PIV and the D-PIV methods compared well for the direction of flow. The streamlines were also found to be similar in the data obtained using both methods. However, because of the superior spatial resolution of D-PIV, some smaller scale details were not revealed by E-PIV. The application of E-PIV to the human heart resulted in reproducible flow patterns in echocardiographic images taken within different time frames or by independent examiners.

Conclusions: The E-PIV technique appears to be capable of evaluating the major flow features in the ventricles. However, the bounded spatial resolution of ultrasound imaging limits the small-scale features of ventricular flow to be revealed. (J Am Soc Echocardiogr 2010;23:86-94.)

Keywords: Particle image velocimetry, Echocardiography, Diastolic flow, Transmitial vortex formation

Left ventricular (LV) fluid dynamics represent an elegant flow phenomenon, starting with the transmitral jet reaching a maximal velocity of about 100 cm/s in traveling a distance as short as only a few centimeters. This volume of blood must then turn around to exit the left ventricle via the aortic valve within a fraction of a second while maintaining its initial high velocity. For this to happen properly, nature has chosen a vortex flow (a swirling motion of fluid), which is developed during diastole, to redirect the transmitral flow toward the LV outflow tract. This track of motion guarantees efficient transfer of the fluid momentum and minimizes the ventricular energy dissipation.¹²

Despite improvements in intraventricular flow visualization by means of advanced noninvasive imaging modalities,⁶,⁷ experimental⁵,⁷ and numerical models,⁶ several aspects of the corresponding fluid dynamics phenomena are still unclear. The key reason for the lack of quantitative information concerning cardiac flow is the lack of technology that can reliably map the spatial and temporal details of the flow while being feasible in routine clinical use.

Color Doppler imaging, as a tool to obtain intracardiac flow information, considers only a single component of blood velocity along the transducer scan line and does not provide quantitative information with respect to vortex formation, recirculation areas, and flow stagnation zones. Because of these limitations, color Doppler imaging may not provide adequate information for either the reconstruction of velocity vector fields or the actual direction of streamlines. Several attempts have been made to incorporate quantitative fluid dynamics into echocardiography using particle-tracking algorithms.⁹,¹⁰ This echocardiographic approach has received further attention in the clinical setting, mainly due to its higher image quality and potential for the assessment of intraventricular blood flow. The particle-tracking methods are mostly based on well-known optical imaging techniques of particle image velocimetry (PIV).
previously described by Adrian and further developed by Willert and Gharib to measure two-dimensional and three-dimensional flow velocity fields.

Although the application of PIV to echocardiography seems promising, and results appear to be qualitatively meaningful, the acquired data in clinical settings has yet to satisfy the minimal theoretical requirements. Careful attention is needed to systematically evaluate the accuracy of echocardiographic PIV (E-PIV) for diagnostic purposes. In this study, we compared the results obtained using E-PIV with those simultaneously acquired by digital PIV (D-PIV) from an in vitro model of the left ventricle that reproduced flow conditions similar to the normal human heart. In addition, we used E-PIV to evaluate a number of different clinical subjects to test the reproducibility and the operator interdependency of the technique. The aim of the present study was to assess the reliability, accuracy, and limitations of E-PIV for the quantification of ventricular flow from the postprocessing of ultrasound data. Examples of potential clinical applicability are also presented.

METHODS

Techniques that are based on PIV or, more generally, on optical flow measurements have been proven successful in several applied fields of science, ranging from turbulent fluid dynamics to optical motion recognition. The main challenge in the application of PIV to echocardiography, besides image quality (low signal-to-noise ratio), is the low ratio of spatial resolution to temporal resolution, which prevents the interpretation of the entire range of velocities in the human heart. Good results can principally be obtained using direct PIV when the frame rate is high enough to reduce this limitation. The approach described here introduces a carefully developed hierarchical algorithm that allows for the identification of velocity vector direction despite the deficiency of accurate velocity magnitude. The computed results were then rearranged in terms of fluid dynamics parameters such as vorticity, which is a physical term to properly describe intraventricular flow patterns.

In Vitro Experimental Model

For the purpose of this study, we used a left-heart pulsed-flow simulator system. The system comprised a silicon heart with adult human dimensions suspended in a pressurized Plexiglas (Altuglas International, Philadelphia, PA) container. A 23-mm Carpentier-Edwards PERIMOUNT bioprosthetic heart valve (Edwards Lifesciences, Irvine, CA) and a 21-mm Sorin Biomedica Carbocast mechanical heart valve (Carbomedics, Austin, TX) were placed in the mitral and aortic positions, respectively. The periodic, pulsatile flow in the circulatory system was generated as the response of the ventricular sac to the input waveform provided by the pump. The waveform reproduced a systolic ratio (the fraction of time in a cardiac cycle during which the left ventricle is in systolic phase) of 40%.

Distilled water was used as the circulating fluid, and the system time scale was increased (about 2.5 times) with a heart rate of 30 beats/min to compensate for the viscosity differences between blood and water. This ensured similarity between the model and the human heart at 75 beats/min, with the same Reynolds number and Strouhal number. For optical and ultrasound acquisitions, flow was seeded with neutrally buoyant particles (diameter, 100-120 μm) illuminated with a laser light sheet on a plane containing both mitral and aortic valves. Pairs of optical images, separated by a short time interval, were acquired with a high-resolution monochrome digital camera with a charge-coupled device (30 frames/s; resolution, 768 × 480; TM-9701, PULNiX America, Inc, Sunnyvale, CA) with a rate of 15 pairs/s.

D-PIV

D-PIV is an optical method used to obtain flow characteristic information (eg, velocity and vorticity fields) from the analysis of a pair of images to determine the displacement field of the particles in the field of view (sampling window) by cross-correlating pixels in a subsection of two images. D-PIV acquisitions and processing were performed with intervals of 1, 3, and 5 ms to verify the convergence of results. The time interval between the image pairs is crucial for the accuracy of D-PIV results, because it corresponds to the expected particle displacements. Limitations include too large or too small displacements, which are difficult to detect. Data analysis was performed with PIVview software (PIVTEC GmbH, Goettingen, Germany). Eventually, 3 ms interval was determined as the optimal value for this experiment and was used for further processing. The velocity vector components were obtained on a regular grid of 74 × 61 points, separated by 1.778 mm, covering most of the LV space. Four consecutive beats were processed, and the results for the average heartbeat were used in the analysis. An example of the computed velocity field superimposed on the image is shown in Figure 1 (left).

In Vitro Ultrasound Image Acquisition

An Acuson Sequoia c512 and the 4V1c transducer (Siemens Medical Solutions, Mountain View, CA) were used for the experiment. Ultrasound imaging was performed at a harmonic frequency of H4.25 MHz and a frame rate of 35 Hz. Digital loops of 3 cycles were acquired from the LV system described earlier simultaneously with D-PIV using the same nanoparticles.

The ultrasound transducer was secured externally to the Plexiglas chamber in an apical LV view. Care was taken to have the scan plane coincide with the optically acquired laser sheet; however, this was done visually since no automatic control was in place. Image optimization controlled the artifacts hard interfaces (Plexiglas, valves) and water-silicon-water interfaces (LV chamber). An example of the acquired image is shown in Figure 1 (right).

In Vivo Ultrasound Image Acquisition

Echocardiography with contrast agent imaging was performed on five healthy volunteers by two independent examiners within 2 consecutive days and repeated for one volunteer at a 3-month interval. The Acuson Sequoia c512 and the 4V1c transducer were used to image all volunteers and patients. Imaging was performed at harmonic frequencies of H3.25, H3.75, and H4.25 MHz; frame rates of 60 to 80 Hz; a mechanical index of 0.6 to 0.7; and dynamic ranges of 45 to 55 dB. The acoustic focus was placed at the mitral annular level. Contrast enhancement was produced by intravenous bolus administration of 0.1 to 0.2 mL lipid-shelled octafluoropropane microbubble contrast agent (Definity; Lantheus Medical Imaging, North Billerica, MA), diluted in 10 mL of saline solution. Acoustic captures of three consecutive heart cycles for the apical 3-chamber (A3C), apical 2-chamber (A2C), and apical 4-chamber (A4C) projections were acquired (Video 1, video clip online). During image acquisition, care was taken to maintain a constant homogenous contrast agent concentration in the left ventricle for visualization of swirling flow. In addition, five patients with dilated cardiomyopathy (DCM)
and eight patients with prosthetic heart valves (four bileaflet Carbo-
medics, four biologic Medtronic Hancock; Medtronic, Inc, Minneap-
olis, MN) were imaged. Data were acquired using the same
ultrasound system and imaging acquisition protocol. The patient
study was performed with institutional review board approval, and
all patients provided written consent.

E-PIV

E-PIV data analysis was performed using Omega Flow software pro-
totype version 4.0 (Siemens Medical Solutions). To initialize the pro-
cedure, the examiner was required to trace the contour of the LV
cavity by the interface; this contour was then tracked in other frames
to create a boundary for the region of analysis. The processing results
were initially displayed as velocity vectors overlaid on the contrast-
enhanced B-mode images. The instantaneous vorticity analysis was
reviewed frame by frame. Finally, the steady streaming vorticity field
was evaluated by taking the time-average of the velocity field during
one heartbeat. The steady streaming flow pattern represents the net
dominant flow motion and allowed visualization of the location
and shape of intraventricular vortices and circulatory pattern. The
software performed dedicated PIV processing optimized for echocar-
diography. The technical details of this PIV method and the justifica-
tions are provided in Appendix A.

Fluid Dynamics Analysis

The resulting velocity sequences were analyzed to derive further spa-
tial flow characteristics for better characterization of the flow patterns.
The fundamental quantity that describes the motion of the fluid is rep-
resented by the vorticity defined as the tendency of fluid elements to
spin; more precisely, vorticity can be related to rotation of fluid
elements and the formation of circulatory areas. Quantitative param-
eters of the intraventricular vortex were also extracted on the basis of
the vorticity. The details of the mathematical analyses are provided in
Appendix B.

RESULTS

In Vitro Results

The velocity fields computed by D-PIV are shown in Figure 2 (top
row) at the peak of the filling wave, during inflow deceleration, at
late diastole, and at the beginning of systole, respectively. The corre-
spanding velocity fields obtained by E-PIV are also shown in Figure 2
(bottom row). Despite the differences in spatial resolution, dynamic
range, and velocity accuracy between the two methods, the main
physical features of intraventricular flow could be outlined. The
flow presented an intense transmitral jet that reached the septum,
with two circulatory components on both sides (Figure 2, early dias-
tole); the flow then moved along the septum toward the apex (mid-
diastole), where the apical vortex persisted within the rest of diastole
(late diastole), and finally weakened and moved toward the base at
the beginning of contraction (early systole). The image quality of D-
PIV allowed the evaluation of sharp spatial gradients, whereby sub-
stantial flow changes could be observed within the window of one
or two grid points.

To quantitatively compare D-PIV and E-PIV for local, instantaneous
velocity magnitudes, it was necessary to convert the results of both
methods on the same space-time computational grid. To do this, at
every instant, the D-PIV velocity fields were sampled on the E-PIV
(37 × 29) spatial grid using bilinear interpolation. Prior to this, the
D-PIV velocity fields were preliminarily filtered with a 2 dimensional
Gaussian isotropic filter of 10 mm in width to eliminate the small-scale
fluctuations that were not present in the ultrasound results and to al-
low a punctual comparison. The results presented below were not
considerably affected by changing the filter width from 7 to 10 mm
and to 15 mm, respectively. Sampling time was also different for
both methods (15 and 68 frames/beat for D-PIV and E-PIV, respec-
tively). To accurately compare the two methods, Fourier time decom-
position was performed on both fields, followed by reconstruction at
16 time-instants by 8 Fourier harmonics.

A total of 7331 velocity vector magnitudes (on about 460 spatial
locations at 16 time instants) were compared. The statistical distribu-
tion of the difference between the velocity magnitudes computed by
E-PIV and D-PIV (ie, the error in the absolute velocity) is shown in the
histogram of Figure 3A. The error is reported as a percentage of the
average velocity. The distribution was peaked at zero (mean value,
−2.45%) with a moderate dispersion about the mean value (standard
deivation, 80%). The inset in the same panel shows a polar histogram
of the error in the direction of the computed velocity vectors. The er-
ror of flow direction was centered at zero, with a moderate dispersion
in the range of ±30°, indicating an overall accuracy of estimated ve-
locity direction and the streamlines. The histograms in Figure 3B show
the discrepancy in the magnitude of velocities evaluated only in those
points for which the D-PIV velocity was higher than the median value (3665 samples). The histogram is peaked toward the negative (mean value, $-28\%$) and negatively skewed. The inset shows that the error in direction was reduced.

Further analysis was performed to verify the reproducibility of E-PIV in the controlled laboratory setting. With the apparatus running under the same imaging conditions, four additional sequences (two 3-beats and two 2-beats) were acquired and processed. The flow in all the sequences was found to be equivalent as all four steady streaming fields were about equal. The vortex parameters, such as the sphericity index (SI) and vortex pulsatility coefficient (VPC), differed by <1% (see Appendix B).

In Vivo Results

The prototype Omega Flow software was used to process all ultrasound image sequences. The normal volunteer data were processed by two examiners blinded to each other’s results. The data were also analyzed by the same examiner after a 6-month interval. The results for normal volunteers indicated comparable flow patterns for the same volunteer in the same views on the same day, the subsequent day, and within a 3-month interval. Comparable flow patterns were also similar for the same projections between volunteers. Vortex flow direction for the A4C view showed counterclockwise rotation, while the A3C view showed clockwise rotation for all normal volunteers (Video 2; view video clip online.). Flow patterns were also similar for the same volunteers when analyzed by two different examiners. Examples of comparisons are shown in Figures 4 and 5. The quantitative comparisons for the vortex structure were performed on the basis of (1) the geometric index of Sphericity (SI), which is a dimensionless measure of the mean vortex shape, and describes the ratio of the vortex’s long axis to its short axis, and (2) the dynamic parameter of vortex pulsatility coefficient (VPC), which is a dimensionless measure of vortex unsteadiness as defined further in Appendix B. The average values and the standard deviations of the SI and the VPC for the five normal volunteers in the A3C views were 2.38 ± 0.35 and 0.63 ± 0.21, respectively. The interoperator variability revealed small discrepancies as low as 0.44 ± 0.48 for the SI and 0.04 ± 0.06 for the VPC, respectively. Additionally, the intraoperator variability was 0.04 ± 0.05 for the SI and 0.06 ± 0.15 for the VPC.

Planar displays represent only a partial, 2-dimensional view of the actual 3-dimensional intraventricular flow. Therefore, the results for the A4C, A3C, and A2C views must be consistent with each other to represent the same intraventricular vortical flow structure on different scan planes. This was verified by reconstruction of a three-dimensional pattern from all 3 views. Congruence was found in all the subjects. One example of the reconstructed three-dimensional structure along with the original two-dimensional planes is shown in Figure 6.

The intraventricular flow patterns were distinctively different in patients with artificial valves and in those with DCM (Figure 7; Videos 3 and 4; view video clips online.). The flow pattern in normal left ventricles, as described earlier, revealed a vortex moving in the basal-to-apical direction that initiated during diastole and dissipated during

---

Figure 2 Velocity field, and corresponding vortices, computed by D-PIV (top row) and E-PIV (bottom row). The image orientation is equivalent to the A3C projection. Images were taken at early diastole, mid diastole, late diastole, and early systole. The qualitative features are similar, despite the lower spatial resolution of E-PIV and the corresponding limitations on reporting high velocities.
systole. In patients with DCM, the vortex shape was persistently round, and the parameters for the A3C views were an SI of 1.71 ± 0.15 and a VPC of 0.31 ± 0.08. In all patients with prosthetic mitral valves, the vortex, in either the A4C or the A3C view, showed remarkable dissimilarity in the direction of rotation that was opposite to the normal subjects. An SI of 1.49 ± 0.25 and a VPC of −0.87 ± 0.31 (the negative sign indicates the reversal of the circulatory pattern) were obtained from A3C views.

**DISCUSSION**

The aim of this study was to assess the validity of E-PIV for quantitative assessment of the intraventricular blood flow field. Accurate mapping of intraventricular flow provides novel opportunities to evaluate the role of vortices in ventricular function. Similar to any fluid dynamics phenomenon involving vortices, these flow structures are expected to play fundamental roles affecting the dynamics and the energetics of the left ventricle as a bifunctional pump. The presence of vortical structures that develop along with a strong propulsive jet during normal LV diastole was initially recognized by in vitro visualization of the ventricular flow\(^6,7\) and subsequently confirmed by analyses based on color Doppler mapping,\(^3\) magnetic resonance imaging,\(^4,5\) and computational methods.\(^8\) Vortices reflect the functional status as well as the shape and structures of the left ventricle.\(^17,18\) Shear layers surrounding the transmitral jet lead the flow to “roll up” into vortices, which essentially shortens the pathway of intraventricular flow to LV outflow tract in normal conditions.

Although some initial attempts have already started to sensibly relate intraventricular flow to cardiac function,\(^1,19-21\) most aspects of ventricular flow either in normal or in pathologic conditions are still not clear. The reliability and feasibility of a technology that can quantitatively evaluate the spatiotemporal development of cardiac vortical flow would allow a better understanding of cardiac flow conditions on the basis of tangible clinical observations.

**Quantitative Comparisons In Vitro**

We first compared D-PIV with E-PIV in terms of gross flow pattern and streamlines at various instants shown in Figure 2. In either case, it was observed that the entry jet was directed toward the septum, while the flow was moving along the wall toward the apex, creating a vortex. This vortex that was centered in the apical half of the ventricle finally drifted toward the base as systole began. As mentioned earlier, the spatial resolution of optical acquisition is about five times...
greater than that of echocardiography. As expected, considerable differences were found, particularly for high-velocity magnitudes, due to the lower time resolution of E-PIV (about 30 vs 3 ms). In contrast, the directions of the velocity vectors were well captured by E-PIV. Therefore, streamlines, or lines which are everywhere parallel to the velocity vector, were about equivalent.

The statistical distribution of the differences between the velocity magnitudes computed by E-PIV and D-PIV (Figure 3A) indicates that the velocity vectors obtained by E-PIV were unbiased and consistent with those obtained by D-PIV either in terms of absolute values or in terms of direction. The longer tail in the histogram for negative values, which corresponds to the velocity values that were found higher in D-PIV compared to E-PIV, reflects a tendency of E-PIV to underestimate the higher velocity magnitudes. This can be confirmed by the histogram in Figure 3B, which was constructed entirely from the velocities higher than the median value, peaked toward the negative values and negatively skewed. Nevertheless, the inset shows that the error in direction was further reduced, confirming that the underestimation was accompanied by a persistent accuracy in the assessment of velocity direction. The inability to accurately evaluate high-velocity values corresponds to the E-PIV dynamic range, mainly due to the low frame rate. This is an expected limitation and is further discussed in the “Limitations” section. Nevertheless, E-PIV technique appears capable of proper assessment of flow direction, the streamlines, and the circulatory pattern in a reproducible manner.

Clinical Reproducibility In Vivo

The results obtained either from independent examiners or from the same examiner on different attempts showed comparable flow patterns. There was a slight variability between the initial contours drawn, which led to a slightly different LV geometry. Apart from this, the intraventricular flow pattern and the similarity between the results obtained by the two examiners were evident and confirmed by quantitative parameters (Figure 4). The same flow patterns were found in the same patient on subsequent acquisition days (Figure 5). In all subjects, the flow patterns were found qualitatively similar; however, a larger sample size is required to statistically verify the reproducibility of the technique.

The intraventricular flow quantification on different projections for the same patient also revealed the three-dimensional consistency of the different planar results, as shown in Figure 6. Such a three-dimensional representation of transmitral vortex from A4C, A3C and A2C views would allow a better representation of ventricular flow pattern.

Findings in Patients Versus Normal Subjects

The sensitivity of E-PIV to detect differences in cardiac vorticity patterns in normal versus pathologic conditions was verified. In patients with DCM, the ventricle was found to be more spherical, with reduced radial motion of the LV wall and larger end-systolic volume. Furthermore, the intraventricular vortex, contrary to normal subjects, was neither dissipated nor fully ejected during systole. As a result, the DCM vortex appeared persistent during the entire heartbeat with little pulsatility (one half VPC index). Here again, the vortex possessed a more spherical shape as, confirmed by a reduced SI.

The flow pattern observed in patients with artificial mitral valves was remarkably different compared with the other groups studied. In these patients, the rotational direction of the intraventricular vortex was reversed compared to normal subjects and patients with...
The transmitral jet was also deviated toward the interventricular septum rather than moving parallel toward the lateral wall. The shape of the leading vortex bubble was found to be more circular and symmetric compared to its elliptical and asymmetric shape in normal subjects. We found that this unnatural vortex formation was quite unstable, giving rise to turbulent flow conditions whereby it was fully dissipated within a cardiac cycle and resulting in higher than normal pulsating dynamics. The transmitral jet pattern passing through a prosthetic mechanical mitral valve was fundamentally different than that passing through a bioprosthetic mitral valve due to the presence of side jet(s) in mechanical valves and the geometrical differences in the shape of the orifice. However, dynamics of the

**Figure 6** Results of the steady streaming analysis for A4C, A3C, and A2C acquisitions show the resulting three-dimensional vortex representations arising from the combination of all 3 projections. Color shading in two-dimensional images represents circulatory pattern: blue is clockwise rotation and red is counterclockwise rotation.

**Figure 7** Results of the steady streaming analysis in a normal volunteer (left), a patient with DCM (center), and a patient with a bioprosthetic mitral valve (right). Pictures correspond to the A3C projection. Color shading represents circulatory pattern: blue is clockwise rotation and red is counterclockwise rotation.
dominant vortex upon formation did not show any remarkable qualitative differences between the two valves. One visually noticeable difference was the quantity of small-scale perturbations (turbulence) that was further visible in mechanical versus bioprosthetic valves. These flow structures may be attributed to the geometric complexity of the mechanical valve and potentially to the interaction of the side jet(s) with the mainstream flow. Further studies are in progress to quantitatively study the vortex formation difference between the two types of valves in a statistically significant population database.

Limitations

For comparison of E-PIV and D-PIV, the ultrasound scan plane was not exactly equivalent to the thin laser sheet, because the alignment was indefinite. Because alignment was performed on the basis of visual interpretation, some differences are to be expected. Both methods of E-PIV and D-PIV capture two-dimensional flow field while the studied flow possessed three-dimensional structures. Additionally, the in vitro echocardiographic acquisition was associated with persistent artifacts that prevented accurate calculation near the base and in a small region left to the apex. Overall, the E-PIV approach is subject to the same limitations and requirements of good echocardiographic acquisitions, such as acoustic shadowing from various echocardiographic structures or the problems arising from technically difficult examinations.

Comparison between the D-PIV and E-PIV velocity fields must be performed considering the unavoidable existing instrumental differences. E-PIV is subject to an upper bound of maximum recordable velocities (see Appendix A for technical details), whose values are proportional to the imaging frame rate. The spatial resolution of velocity calculations from echocardiography is much lower (of the order of 20 mm) than that obtained by optical imaging (approximately 3.5 mm).

CONCLUSION

E-PIV is a novel technique for evaluating ventricular fluid dynamics, particularly for assessment of transmitial vortex ring formation. With this new technique, the velocity directions and streamlines can be drawn with reasonable confidence. E-PIV is also capable of evaluating the principal (large-scale) blood flow patterns in the left ventricle. The principal flow features, recirculation regions and vortices, can be detected in a visually reproducible scheme. As a result, this technique may potentially improve the diagnostic capabilities and therapeutic strategies made by clinicians.

The present work describes a technique that allows tracking the vortex formation inside the left ventricle, which can lead to more effective clinical staging of flow conditions that occur in diastolic dysfunction or after surgery. Although the overall velocity values obtained by E-PIV were found to be consistent with those obtained by D-PIV, high-velocity magnitudes are currently underestimated due to the technological limitations in ultrasound imaging frame rates. Accordingly, the bounded spatial resolution of ultrasound imaging limits the small-scale features of ventricular flow being revealed. This method should not be mistaken as a possible competitor to Doppler echocardiography but rather as complementary to it.

Supplementary data

Supplementary data associated with this article can be found in the online version, at doi:10.1016/j.echo.2009.09.007

REFERENCES

Appendix A

In this study, a hierarchical PIV method that iteratively reduces the interrogation window is adopted to handle the long-range correlations due to large velocity values and relatively low frame rate. The interrogation window was reduced iteratively by halving its side from squares $32 \times 32$ down to $8 \times 8$ pixels. To ensure accurate theoretical convergence each level of iteration was performed in two further substeps in which one-half of the value of the first estimate was used to displace the regions of interest (ROIs) for the second attempt. At each step, previously estimated displacement in two consecutive frames would be accounted in either frame to ensure that the evaluation remained centered at the interrogation window. The maximum likelihood between pairs of ROIs would be defined using the transport equation for the brightness field, $B(x, y, t)$:

$$\frac{\partial B}{\partial t} + \nu_x \frac{\partial B}{\partial x} + \nu_y \frac{\partial B}{\partial y} = 0. \quad (A1)$$

This equation would be theoretically zero in perfectly rigid motion. Therefore, the equation $A1$ cannot be satisfied exactly in other situations. Therefore, we seek the velocity field that minimizes the error, in a least squares sense, inside the interrogation region. This definition is practically equivalent to a 2-dimensional cross-correlation maximization but appeared more reliable than other methods in noisy fields, such as echocardiographic images.22

Velocity vectors were evaluated on a regular grid with a 4-pixel or 8-pixel spacing (approximately 1.4 or 2.8 mm, respectively). The computed values of velocity were finally smoothed applying a $3 \times 3$ Gaussian filter with a width given by the grid distance. The dimension of the interrogation ROI together with final smoothing defines the spatial resolution of the E-PIV field, which would be twice the order of the ROI half size (about 32 pixels or 20 mm). This is an intrinsic limitation imposed by ultrasound image quality. These specific processing choices were defined based on an extensive sensitivity study. These processing details resulted in a combination that ensured robust estimation, not significantly influenced by perturbation of the parameters.

The 3-ms time-interval that was used in D-PIV processing is not yet achievable even with high-end ultrasound equipments. This limitation results in an upper bound to the maximum recordable velocity (corresponding to the size of the initial interrogation window, 32 pixels, divided by the time interval between frames). This cutoff velocity is about 30 cm/s at 35 Hz (or 70 cm/s at 80 Hz). However, it could theoretically increase by a larger interrogation window, with the expense of loosing the accuracy. Therefore, given the small variability of the spatial resolution, the cutoff velocity was directly set proportional to the frame rate. Additionally, the low frame rate was particularly critical because of migration of the bubbles out of the scan plane that could have become a significant effect in a large time interval. When particles that cross the scan plane leave a shorter trace, the resulting velocity values correspond to a mixture of the particles that remain on the scan plane and those that leave on a shorter path. These phenomena could potentially lead to significant underestimations of large velocity values. In either case, a statistical correlation remained along the effective direction of motion. Therefore, although the relatively low frame rate represents a limitation for the velocity quantitative estimation, it does not significantly alter the velocity direction. As a result, the estimation of streamlines and the definition of circulatory regions would not be biased by the limitations affecting the high-velocity values.

Appendix B

Vorticity is portrayed as the curl of velocity vector.16 We have evaluated the vorticity, $\omega(x, y, t)$, by spectral accurate numerical derivatives:

$$\omega = \frac{\partial v_x}{\partial y} - \frac{\partial v_y}{\partial x}. \quad (B1)$$

Vorticity ($\omega$) describes the underlying structure of an incompressible flow. A region with compact vorticity constitutes a vortex and a region with an elongated vorticity distribution as a shear layer which is a boundary layer in vicinity of the wall. Alternatively, vorticity is a local quantity (ie, just a single peak of vorticity at a point is sufficient to create a wide circulatory pattern), which explains the difficulty of its evaluation in a system perturbed by noise. A more practical quantity that can represent blood flow pattern is the vorticity stream function, $\psi(x, y, t)$, derived from vorticity field by solving the Poisson equation with homogeneous boundary condition:

$$\nabla^2 \psi = \omega. \quad (B2)$$

The stream function is a smoother version of the vorticity field, which has a clear physical interpretation. The spatial curves at constant values of $\psi$ represent the streamlines related to the vorticity field and thus easily allow the recognition of vortical circulatory regions.

In the present study, we considered a vortex as a region with closed streamlines enclosing the maximal vorticity. The selection of a specific streamline defining the vortex boundary corresponds to selecting a value of the stream function $\psi$. One important description of the periodic flow field is given by the so-called steady streaming flow,23,24 which is the flow obtained by taking the time average over one heartbeat. The steady streaming flow pattern synthesizes the net dominant flow motion from time oscillation and represents a sort of signature for the beating flow.

For developing quantitative measures of the intraventricular circulatory flow (see also Kim et al3), the vortex is automatically delineated as the region enclosed by the vorticity stream function level. The vortex geometry is then measured by the Sphericity Index (SI) defined as the ratio of the length to the width of the vortex, calculated based on the lengths of the principal inertial radii. The dynamical activity of the vortex is evaluated by the correlation, between the steady and pulsatile components of the vorticity within the vortex area. This dimensionless parameter which is called Vortex Pulsatility Correlation (VPC), is normalized with the vortex strength, and represents a measure of vortex dynamics activity. Its value is large when the vortex is coherent and presents lively rhythmic alternation of formation and ejection. It is reduced when the vortex stagnates with minor blood exchanges.