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CME

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STATE-OF-THE-ART PAPER

Emerging Trends in CV Flow Visualization

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CME Objective for This Article: At the end of this activity the reader should be able to: 1) understand relevant concepts in cardiac fluid mechanics, including the emergence of rotation in flow and the variables that delineate vortical structures; 2) elaborate on the main methods developed to image and visualize multidirectional cardiovascular flow; 3) develop dedicated imaging protocols for particle imaging velocimetry; and 4) discuss the potential clinical applications and technical challenges of determining multidirectional cardiovascular flow.

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Emerging Trends in CV Flow Visualization

Blood flow patterns are closely linked to the morphology and function of the cardiovascular system. These patterns reflect the exceptional adaptability of the cardiovascular system to maintain normal blood circulation under a wide range of workloads. Accurate retrieval and display of flow-related information remains a challenge because of the processes involved in mapping the flow velocity fields within specific chambers of the heart. We review the potentials and pitfalls of current approaches for blood flow visualization, with an emphasis on acquisition, display, and analysis of multidirectional flow. This document is divided into 3 sections. First, we provide a descriptive outline of the relevant concepts in cardiac fluid mechanics, including the emergence of rotation in flow and the variables that delineate vortical structures. Second, we elaborate on the main methods developed to image and visualize multidirectional cardiovascular flow, which are mainly based on cardiac magnetic resonance, ultrasound Doppler, and contrast particle imaging velocimetry, with recommendations for developing dedicated imaging protocols. Finally, we discuss the potential clinical applications and technical challenges with suggestions for further investigations. (J Am Coll Cardiol Img 2012;5:305–16)

Flow through the heart interacts with the mobile contours of the myocardium, valve, and vessels. Blood flowing through the sequence of compartments is subject to changes in direction and luminal diameter, and as a consequence, flow is multidirectional and vortical with a tendency to curl or spin in the cardiac chambers during various phases of the cardiac cycle (1). The valves, chamber geometry, and wall motion modify the flow patterns to produce a hemodynamic environment comprising normal or pathological adaptation. Therefore, analyzing the spatial and temporal distribution of blood flow in the cardiovascular system may provide diagnostic and prognostic information. However, acquiring and visualizing flow through the volumes of the heart cavity is a complex topic. Doppler echocardiography and cardiac magnetic resonance (CMR), in different ways, have been used to measure the velocities of the flow that is predominantly unidirectional while it passes through a valve, a jet, or an approximately cylindrical vessel segment. Recent technological innovations in imaging modalities and the emergence of flow visualization techniques have provided valuable opportunities for direct in vivo assessment of multidirectional blood flow. Analysis of the flow fields is associated with variables such as pressure differences, shear stresses, and energy dissipation. The display and measurement of such flow variables on a page or 2-dimensional (2D) computer screen can be challenging, and each variable calls for different types of analysis and representation. Therefore, there is a pressing need to guide and advance flow visualization techniques toward the development of consistent measurements and clinical applications (2).

Basic Principles in Cardiovascular Fluid Mechanics

Blood is a corpuscular material that behaves like a fluid and can be deformed throughout its volume. Its movements are powered and constrained by the movements of containing boundaries. Blood's patterns of flow are subject, on the one hand, to the directional momentum of the streams entering and moving through the chambers and, on the other hand, to the slowing, frictional effect of the viscosity both relative to the boundaries as well as between streams, with different velocities sliding against one another within the volume. The liquid motion can be imagined as being composed of numerous flexible layers (laminae) sliding relative to each other (Fig. 1A).

Laminar flow, in which the directions of streamlines remain almost parallel to those of neighboring streamlines or boundaries, occurs in the smaller, more peripheral blood vessels, but it is not typical of flow through large vessels or the chambers of the heart. Laminar flow tends to become unstable, fragmenting and mixing with eddies and counter-eddies when its kinetic energy (which is proportional to the square of its velocity [V]) becomes large relative to its rate of energy dissipation (which is roughly proportional to $\nu V/D$, where D is the diameter of the vessel and ν is the kinematic viscosity). When this ratio, known as the Reynolds number ($Re = VD/\nu$), exceeds a critical value (approximately 2,500 for steady flow in a straight

circular vessel), laminar motion becomes unstable, and the fluid becomes turbulent. Turbulent flow is characterized by chaotic unsteady vortices of many different sizes that increase mixing, friction, and energy dissipation. In turbulence, the rate of energy dissipation increases to approach that of the incoming kinetic energy. Turbulent flows are not common in the healthy circulatory system; rather, flowing blood more commonly develops jets and swirling motions.

The term *vorticity* (typically represented by the Greek letter ω) is a mathematical term that physically corresponds to (twice) the local angular velocity of a fluid particle. Although a vortex is a compact region characterized by the accumulation of vorticity (Fig. 1B), a *shear layer* (or a *vortex layer*) is an elongated layer of vorticity. In general, the *spatial distribution* of vorticity, as well as its development and evolution, can facilitate an understanding of the key features of blood motion in large vessels and heart chambers. Therefore, most fluid dynamic studies focus on the lifespan of vorticity, from its generation to its organization in the form of vortices, and to its dissipation.

Vorticity does not appear spontaneously in a flow field; it develops in the form of a shear layer adjacent to the tissue (boundary layer) as a consequence of the velocity difference between the flow and the solid boundary. Where streamlines separate from the wall, the fluid tends to curl into a vortex or a cascade of vortices (Fig. 1C). Streamline separation and vortex formation are typical of the flow into the heart cavity from a vein or valve or from a narrow to a more dilated vessel segment. Examples include the flow into the carotid sinus and the flow beyond a heart valve or vascular stenosis (Fig. 1D). The presence of such formed vortices affects pressure distribution and shear stresses and is crucial for relating blood motion to pathology (3).

An expanded version of this section, along with assumptions used in numerical modeling, has been included in the online Appendix.

Blood Flow Visualization Strategies

The following section provides an overview of the currently available flow quantification techniques for use in both research and clinical applications. Table 1 lists some of the advantages and disadvantages of each method.

Flow visualization using CMR. Using the velocity-encoded CMR phase-contrast technique (4–6), blood flow velocities can be measured in any direction without using contrast agents. By using bipolar gradients, the signal of the moving blood is in a phase that is proportional to its velocity. Although real-time phase-contrast CMR is possible for 2D measurements, better quality of data is obtained by combining the information from several heartbeats using prospective or retrospective electrocardiogram (ECG) gating. Retrospective ECG gating with time-resolved, through-plane, and 2D phase-contrast CMR is mostly used to obtain peak velocities or volumetric flow to determine the stroke volume. Using ECG and respiratory gating, the complete time-resolved, 3-dimensional (3D), and 3-directional velocity field can be measured over a volume that covers the complete heart or large vessels (7,8). This 3D cine phase-contrast CMR technique is popularly called 4-dimensional flow CMR. The acquisition time depends on several parameters: the region of interest, the location in the body, the spatial and temporal resolution, and the breathing pattern (9,10). Currently, cardiac blood flow can be measured with a spatial resolution of 3 mm^3 and a temporal resolution of 50 ms in approximately 20 min. Blood flow in the aorta can often be measured at a higher resolution and with a shorter acquisition time.

As the complete time-resolved, 3D, and 3-directional velocity field is measured, the blood flow can be analyzed using a number of tools. Three-dimensional streamlines and pathlines (Figs. 2 and 3; Online Video 1) have become popular for intuitively visualizing time-resolved 3D blood flow. The flow structures of interest can be automatically characterized from the 3D cine phase-contrast CMR using a pattern-matching approach, for example (11). Quantitative volume flow measurements can be obtained by retrospectively extracting the 2D planes (12). In laminar flow, the relative pressure field can be computed from the velocity field in the aorta (13) and cardiac chambers (14). Nonlaminar flow can be further characterized by the turbulent kinetic energy, which is a measure of velocity fluctuations (15) that is based on estimating the intravoxel SD of the phase-contrast CMR signal magnitude (16).

METHODOLOGICAL RECOMMENDATIONS. CMR is a uniquely versatile yet complex imaging modality. It is not possible to give concise recommendations on the methods of flow acquisition and post-processing without relating them to the re-

ABBREVIATIONS AND ACRONYMS

CMR = cardiac magnetic resonance

ECG = electrocardiogram

Echo-PIV = echocardiography particle imaging velocimetry

LV = left ventricular

LVOT = left ventricular outflow tract

PIV = particle imaging velocimetry

VFM = vector flow mapping

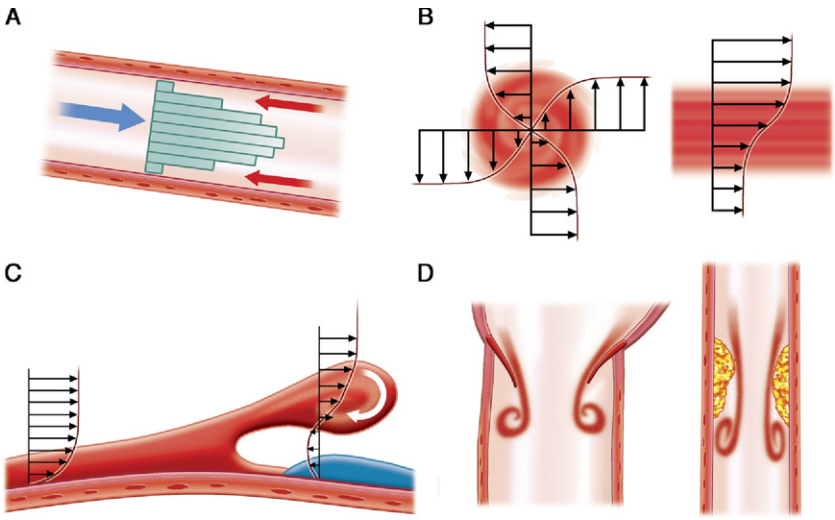

Figure 1. Basic Principles in Fluid Mechanics

(A) Blood motion in a straight vessel under a pressure gradient balances the energy loss due to friction; the fluid velocity is higher in the center and gradually decreases to zero at the wall due to viscous adherence. The shear stresses and the vorticity **(shaded profile)** are larger near the wall. **(B)** A vortex **(left)** is a region of compact vorticity; a shear layer **(right)** is an elongated distribution of vorticity between 2 streams with different velocities. **(C)** The deceleration of the flow produces **(above)** a local thickening of the boundary layer with clockwise vorticity **(red)**. The separated vorticity **(below)** tends to roll up and eventually form a vortex in the main flow. The separating vortex reduces the wall shear stress that may even reverse, which is seen as opposite sign, counter-clockwise vorticity **(blue)** at the wall beneath the vortex. **(D)** Vortex formation occurs when the boundary layer formed for the upstream surface detaches because of a sharp expansion.

search or clinical questions to be addressed. There are trade-offs to be considered among acquisition time, spatial resolution, temporal resolution, signal-to-noise ratio, and the comprehensiveness of the data coverage. Routine breath-hold cine acquisitions using steady-state free precession allow some features of the flow to be visualized in 2D planes. Although not real-time, the local variations in the blood signal can show jet formations, associated para-jet shear, and at least an impression of small-scale flow instabilities within the larger streams. Importantly, cine imaging also depicts the myocardial, valve, and vessel wall dynamics that are inseparable from the flows they deliver and contain.

The methods and applications of such 4-dimensional velocity acquisition (9) are recommended for both displaying the relatively large-scale patterns of multidirectional blood flow that are consistently repeated from beat to beat as well as relating flows in different cardiovascular compartments to one another. However, these methods require relatively long acquisition times of about 8 to 25 min. Each reconstructed movie loop is not in real-time but represents flow that is effectively phase-averaged over hundreds of heart cycles, and, hence, the movie loop does not record any small-scale instabilities or other beat-to-beat variations in the flow. The multidirectional velocity data, although impressively comprehensive, may need to be supplemented by either more selective flow imaging at high temporal and spatial resolutions or computational fluid dynamic simulations to reach conclusions regarding small-scale and time-varying flow features. This solution may be important to improve the calculation of parameters such as fluid energy dissipation, fluid wall shear stresses, and the partitioning versus mixing of adjacent blood volumes or streams.

Flow visualization using echocardiography. Ultrasound flow visualization techniques, with high temporal resolution and relatively low cost, make it suitable for routine clinical use. Recent advances in imaging techniques for mapping flow using echocardiography techniques are discussed here.

COLOR DOPPLER ECHOCARDIOGRAPHY. Color flow was developed in the mid-1980s using multiple range-gated pulsed Doppler. In clinical practice, however, the technique has been used as a qualitative method for demonstrating abnormal flow patterns and superimposed turbulence. Color Doppler measures only axial velocities along the line of each ultrasonic beam. If it were possible to estimate the

Table 1. Flow-Visualization Technologies

		Echocardiography	
	Phase-Encoded MRI	**Echo-PIV**	**Color Doppler**
Resolution and coverage relative to all 3D of space	Good spatial resolution in all 3D or in 2D in shorter acquisition times; unrestricted access	Good spatial resolution in 2D planes, and in multiple planes if required	High spatial resolution in 2D; good resolution in 3D
Coverage relative to all 3-directional components of velocity	All 3 can be acquired, or 2, or 1, through a plane placed in any orientation	Both in-plane components represented but not the through-plane	Only the 1 component directed to or from the transducer is currently measurable clinically
Temporal resolution	Typically 20–50 ms, each phase is calculated from acquisition over many heart cycles	High temporal resolution (4–20 ms), allows assessment of velocity fields even during brief isovolumic contraction and relaxation phases over few heartbeats	Good temporal resolution in 2D (4–20 ms), relatively low in 3D
Scan time	Long (5–20 min) for a dataset covering all 3D and all 3-directional components but seldom real-time	Both scan time and offline analysis can be done over few heartbeats in minutes	Rapid scan times, real-time visualization
Breath-holding	Breath-hold used for short, unidirectional velocity acquisitions, or diaphragm navigation for long acquisitions during free breathing	Imaging relatively easier during breath-hold but also possible during respiration	Not required
Low-velocity accuracy	Low velocities are measurable but less accurate if high velocities also have to be measured	Well-visualized	May be underestimated or affected by noise
High-velocity accuracy	Measurable up to the chosen VENC limit, but only where a stream or jet core is wide enough to include whole voxels	May be underestimated	Well-resolved within aliasing limit
Applications	Flow visualization and measurement of volume flow through all cardiac chambers and large vessels	Flow visualization through all cardiac chambers, aortic flow evaluation may require use of transesophageal echocardiography	
Implanted devices	Metal stents or valve rings cause local artifact; most pacemakers rule out MRI	Flow can be visualized through implanted cardiac devices, and in the presence of pacemakers and defibrillators	

2D = 2-dimensions; 3D = 3-dimensions; Echo-PIV = echocardiography particle imaging velocimetry; MRI = magnetic resonance imaging; VENC = velocity encoding.

radial velocity component, which is directed at right angles relative to the axial velocity, then combining the axial and radial velocities would allow the true flow vector at each site to be calculated. The first method proposed to achieve this goal was crossed-beam ultrasound (17). This technique uses an array of 3 transducers, with 2 acting as transmitters and the third acting as a receiver. All the transducers are focused on the same site but are positioned at slightly different insonating angles. This method has been proven to work both theoretically and in simple phantoms, but it has not been developed for routine clinical applications. More recently, the principle of measuring 2 velocities at the same site to derive the vector has been re-investigated for vascular imaging (18).

An alternative method called echo-dynamography, or vector flow mapping (VFM), has been proposed (19). This technique combines measured axial velocities with estimated radial velocities based on physical principles. The major assumption of the first version of VFM was that all the flow along each radius within an image can be deconstructed into a laminar motion and a vortical component with a zero mean. Subsequently, the transverse velocity is computed assuming that such a vortical component must satisfy the continuity equation (mass conservation) from pixel to pixel, along each radius, and across all the scan lines in the field of the color flow image. Although this method relies on nontrivial assumptions, VFM is often able to delineate the main features of left ventricular (LV) intracavitary flow (Fig. 4).

An analogous approach (20) uses speckle tracking to measure the radial velocity of endocardial wall motion around the left ventricle, achieved by applying the continuity equation in a way such that the resulting radial velocity matches the radial velocity at the wall. This technique involves separate acquisitions of the grey-scale data for the speckle tracking of the wall motion and of the color flow data for estimating the blood velocity. This technique has also been able to discriminate the major features of vortex flow within the left ventricle during the cardiac cycle, but the published images do not show small, low-velocity vortices.

All of these methods currently ignore through-plane flow in the third dimension. Based on

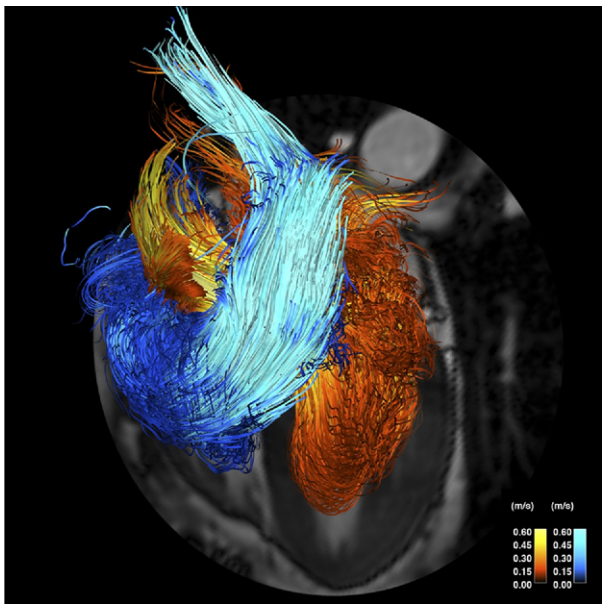

Figure 2. CMR-Derived Pathline Visualization of Cardiac Blood Flow

Pathlines through a flow field track the journey of a particle emitted at a chosen time-frame and incorporate velocity information in color through subsequent phases of the cardiac cycle. By plotting the time-resolved pathlines, the resulting traces reflect the dynamics of the 3-dimensional blood flow over the cardiac cycle. In this figure, the pathlines of the blood flow are traced from the left atrium to the aorta **(red-yellow)** and from the right atrium to the pulmonary arteries **(blue-turquoise)** during a cardiac cycle. CMR = cardiac magnetic resonance.

comparisons with phase-contrast magnetic resonance imaging in apical long-axis planes, the error introduced by this simplification is on the order of 15% for healthy hearts (20). In addition, the accuracy is necessarily dependent on the accuracy of the Doppler color data. Improving the basic color flow with an optimum Nyquist limit to reduce lower velocities from being filtered out and automating the process of de-aliasing the color data will be helpful in future development of these methods (21). Novel techniques for the reconstruction of the blood flow using 2D or 3D Doppler imaging are currently under investigation.

ECHOCARDIOGRAPHY PARTICLE IMAGING VELOCIMETRY. Particle image velocimetry (PIV) is a popular technique to characterize the flow field in fluid dynamics laboratories (22,23). When using PIV, the particle-seeded flow is typically illuminated with a light sheet generated by a pulsed laser. The particle patterns in the field are tracked frame by frame, and the displacement data are converted to velocity using the small time-frame between the illuminating laser pulses. It is important to understand that the PIV technique does not track individual particles; rather, it tracks the patterns produced by groups of particles. Ultrasound beams are used as the imaging source in echocardiography PIV (echo-PIV), which enables visualizing blood flow within an opaque cardiac chamber. The application of ultrasound-based PIV was first reported for imaging kaolin particles in the study of sediment-laden flow (24). After the introduction of ultrasound-based PIV, or echo-PIV as it is known in its current form, as a method for capturing digital B-mode images of contrast agent particles (25), the technique has been successfully explored in both experimental (26) and clinical (27) settings. A careful validation of the technique applied to the carotid flow was recently developed in the laboratory environment and included a clinical feasibility analysis and CMR comparison (28). Echo-PIV provides a tool to assess blood flow directions and streamlines (Fig. 5), map principal flow patterns, and define recirculating regions and vortices with reasonable confidence in a visually reproducible scheme (26).

Many of the limitations of echo-PIV are related to the limitations of current ultrasound technology. For example, the time interval between the 2 frames needs to be short for optimal cross-correlation, which can only be achieved by reducing the sector angle and depth variables. Current applications have mostly been developed for 2D flow tracking and for a sector width size of 45°. Flow can now be measured at a temporal resolution as high as 4 ms and with an effective spatial resolution of about 4 mm. Due to this high temporal resolution, the flow sequences for the phases of the cardiac cycle, including the brief intervals of isovolumetric contraction and relaxation, have been characterized (Online Videos 2 and 3). Although the echo-PIV technique seems to be a promising approach with qualitatively meaningful results, it does have several limitations. An experimental validation study revealed that, depending on the acquisition frame rate, high velocities can be underestimated, and the spatial resolution of current ultrasound imaging limits accurate imaging of the small-scale features of ventricular flow (26,27), which has implications when using echo-PIV for diagnostic purposes.

METHODOLOGICAL RECOMMENDATIONS. During 2D echocardiography, acquiring 3- or 5-chamber views is helpful for visualizing both the filling and the emptying flow patterns. The width of the ultrasound scans, imaging depth, and spatial and temporal settings have to be optimized to achieve the highest possible frame rate with a scan angle

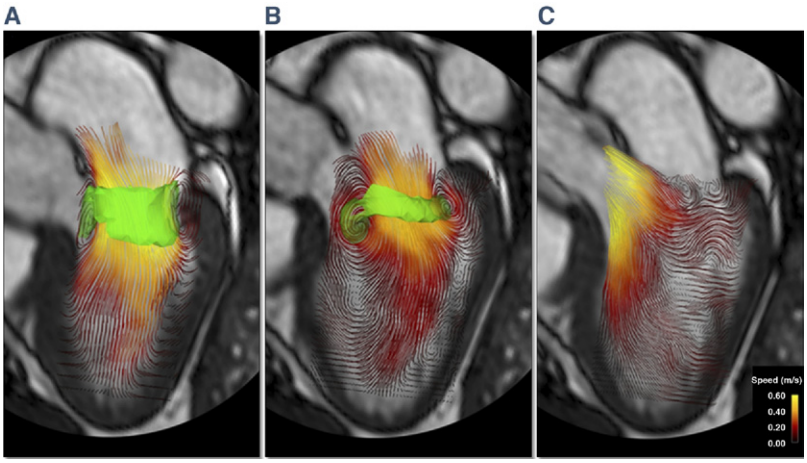

Figure 3. CMR-Derived Short Streamlines

A streamline is a line that is tangent everywhere to the velocity vector field at a given instant in time. In this figure, short streamlines have been computed backward and forward from a 3-chamber plane in the left ventricle of a healthy volunteer during **(A)** early diastole, **(B)** late diastole, and **(C)** ejection phases of the cardiac cycle. An isosurface is shown in **green** that outlines a vortex ring in early and late diastole. CMR = cardiac magnetic resonance. See also Online Video 1.

large enough to ensure that the entire blood pool is visualized. Echo-PIV requires a minimum frame rate of approximately 60 to 100 frames/s; otherwise, the higher velocities are underestimated (as a simple rule, the frame rate must be higher than the heart rate). Much higher frame rates (>150 frames/s) may be required to visualize transient flow patterns during the brief isovolumetric contraction period (Online Video 2). The flow pattern is better visualized with the tissue harmonic modality and by regulating the mechanical index according to the manufacturer's recommendations; as a general rule, a single focal zone at the base of the left ventricle is preferable. The contrast agent must homogeneously fill the cavity for clear visualization of the swirling flow while avoiding both saturation and black areas. To obtain good results, it is preferable to acquire the loop during the dilution phase that follows the peak of bubble concentration using the contrast signal growth and plateau phases to adjust the time gain compensation curve, gain, and focal zone.

In vitro experiments and numerical simulations. Careful replication of the heart chambers and vasculature in vitro has significantly improved our understanding of the physics of blood flow (24,29–33). These models are mainly reconstructed from either anatomical castings or images acquired from CMR, computed tomography scanning, or echocardiography (29). PIV is commonly used to assess cardiovascular fluid dynamics in vitro, particularly for evaluating new imaging modalities (29,34–36), heart valves (32,37), and ventricular assist devices (38,39). One important consideration when using PIV to assess a cardiovascular event is the temporal resolution. The temporal

Figure 4. Flow Velocity Vector Fields Obtained by Using Vector Flow Mapping

Color Doppler data are decomposed into basic and vortex flow components, and flow velocity fields are displayed superimposed on the color Doppler images. The velocity vector at a single point is indicated by the **red dot** to which the **yellow line** is attached. Image courtesy of Tokuhisa Uejima, Cardiovascular Institute, Tokyo, Japan.

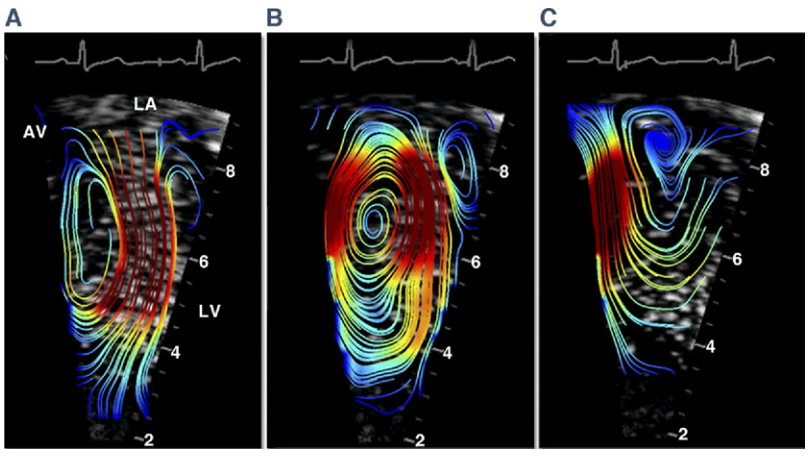

Figure 5. Echo-PIV–Derived Streamlines

Streamlines have been color encoded with kinetic energy derived from echocardiography particle imaging velocimetry (echo-PIV)–derived instantaneous velocity data obtained from a healthy volunteer. The flow is shown along the long-axis view of the left ventricle during the **(A)** early diastole, **(B)** late diastole, and **(C)** ejection phases of the cardiac cycle. See also Online Videos 2 and 3.

resolution of conventional PIV systems that is appropriate for most fluid dynamics applications are about 15 velocity frames/s. These systems are not appropriate for assessing cardiovascular events, and due to their poor time resolution, they cannot capture most flow features in a consistent and reproducible fashion even if they are time-averaged over multiple cycles. Only laser systems with repetition rates around 1,000 Hz, together with correspondingly high-resolution, high-speed cameras, are legitimate for quantitative analyses of unsteady cardiovascular systems in vitro.

Numerical simulations attempt to imitate a dynamic behavior of a system often to predict the sequence of event. Numerical simulations for cardiovascular flow use a combination of computational methods, including medical imaging techniques for determining vascular geometry and techniques for representing the flow domain as a large number (typically millions) of discrete points when evaluating fluid properties. In principle, it is possible to simulate any flow by solving the mathematical laws governing fluid mechanics or by direct numerical simulations. However, it must be stressed that, despite the enormous progress that has been made in computing power, most cardiovascular flow problems remain challenging. Numerical simulations represent a powerful tool that can complement the clinical analyses of cardiovascular disease. These simulations can be used either when in vivo data are not measurable or as a predictive tool for therapeutic options.

Clinical Applications and Research Prioritization

The conventional indexes of cardiac performance usually do not reveal significant changes until there is overt dysfunction, which makes these indexes less effective for the early diagnosis and treatment of cardiac disease. Flow, instead, is immediately affected by changes in cardiac function. Therefore, analyzing the blood flow dynamics opens up new perspectives for our understanding of cardiovascular physiology and for developing very early diagnostic tools (Table 2). The clinical utility of this hypothesis still requires further studies, both to understand the added value over conventional approaches and to develop appropriate flow-based indexes for applications in various cardiac pathologies.

Cardiac function abnormalities. Changes in either the LV shape or contractility will alter the intraventricular flow patterns. In a normal heart, the asymmetric geometry of the left ventricle and the orientation of its valves favor vortex formation and automatic redirection of blood from the LV inflow to the outflow. This efficient arrangement avoids excessive regional wall stress and ensures an optimal LV pump performance. In patients with a dilated left ventricle, the transmitral flow is often directed along the free wall, which gives rise to a well-developed circular flow pattern that turns toward the septum and the outflow tract during diastole. At the end of diastole, blood located close to the left ventricular outflow tract (LVOT) in patients with a dilated left ventricle is directed more perpendicular

Table 2. Clinical Applications of Flow Visualization Techniques

Clinical Condition	Potential Applications
LV systolic function	Paths and kinetic energy changes of blood flowing into the left ventricle for understanding development and progression of dilated/hypertrophic LV remodeling, assessing stagnant flow and risk of thrombus formation, assessment of LV dyssynchrony, optimization of resynchronization therapy and assist devices
LV diastolic function	Transmitral flow patterns and spatial distribution of intraventricular pressure gradients, shear stress, and kinetic energy
Atrial function	Flow features for stratifying risks of left atrial clot formation, efficiency of flow in congenital heart diseases, including Fontan circulation
Valvular diseases	Relationship of regurgitation jet on turbulence and energy dissipation, effects of valve repair and prosthetic replacement surgery on valvular flow direction and LV remodeling
Aorta	Relationship of flow characteristics and shear stress with risks of aortic atherosclerosis, risks of aortic dilation and dissection in Marfan syndrome, retrograde flow from descending aorta and risks of cerebral embolism, optimization of aortic reconstruction surgeries
Pulmonary artery	Characterization of flow features associated with pulmonary artery remodeling in pulmonary hypertension and thrombus formation

LV = left ventricular.

to the LVOT than in the normal left ventricle, in which the blood flow is more directed toward the LVOT (40). Blood flow in the left ventricle has been suggested to contain direct flow routes and recirculating flow patterns, all of which contribute to the efficiency of the pumping function (1,41–44). The presence of a physiological vortex formation process supports such flow routes, which avoids excess turbulence and energy dissipation. Moving further ahead, it can be hypothesized that a proper vortex formation process is associated with the heart's ability to adapt to varying conditions whereas, in its absence, the flow may develop a disordered motion, improper redirection during contraction, and nonphysiological peak pressures. The relationship between abnormal vortex formation and the progression of LV remodeling and symptoms in patients with heart failure remains to be investigated.

Flow visualization may also be useful in understanding pathophysiological mechanisms in LV remodeling. One such mechanism includes, among others, the mechanism of outflow tract obstruction in patients with hypertrophic cardiomyopathy. In the normal left ventricle, the formation of a transmitral vortex helps to maintain the position of the mitral leaflets close to the posterior wall and assists in directing the upcoming systolic stream toward the outflow tract. In hypertrophic cardiomyopathy, however, the anterior displacement of the papillary muscles moves the entire mitral apparatus adjacent to the outflow tract, which reverses the direction of the transmitral vortex and promotes the anterior motion of the mitral leaflets during systole due to the creation of drag forces. This phenomenon requires further investigation (45,46).

Flow visualization may also have application in the assessment of cardiac dyssynchrony. Quantitatively evaluating cardiac efficiency by using intraventricular flow dynamic analysis before and after resynchronization may provide useful indicators of clinical severity and the eventual success of therapy. Moreover, the ability of the technique to discriminate the beat-to-beat changes in the cardiac vorticity and intracavitary pressure gradients could be used to fine-tune biventricular devices.

LV flow visualization can provide a novel and promising method to assess the continuum of systolic and diastolic function. Initiation of LV diastole is related to the release of the elastic energy stored during the previous systole, and this elastic energy generates suction and contributes to normal LV filling (47). Early diastolic suction that exists under physiological conditions may have greater roles during exercise and tachycardia. Therefore, the causal relationship between LV wall recoil, transmitral flow, and intraventricular pressure gradient demonstrates the importance in defining the performance of the left ventricle during diastole. Quantification of diastolic function and intraventricular pressure gradients using echocardiography is typically based on measuring the changes in the ventricular wall, such as the velocity of wall relaxation (48) and the pulsed wave Doppler pattern of the velocity at the mitral tips. There is a need for studies that relate the well-validated parameters of diastolic function (i.e., τ, pressure–volume loops, the stress–strain ratio) to the morphological data of the resulting flow energetics. The formation of transmitral vortices is highly sensitive to the pressure gradient whether due to drop in LV pressure or increases in the left atrial pressure and, therefore, may be a sensitive marker of diastolic function. Conditions that interfere with the normal sequence of regional contraction and relaxation can be ex-

pected to alter early diastolic filling and the vortex formation process. Moreover, a proper characterization of the intraventricular fluid dynamics, energetics, and space-time distributions of pressure and shear stress may allow recognizing early functional changes before they provoke irreversible myocardial dysfunction.

Assessment of atrial function. Rather than being a simple conduit, the atria have also been shown to generate consistent blood flow patterns that are specific to the phase of the cardiac cycle (49). From a fluid dynamics perspective, the atrium and ventricle seem to be tightly linked, as are diastole and systole (40,44). Changes due to disease result in decreased efficiency and elevated turbulence intensity (40,44,50). The function of the right atrium has been explored in healthy volunteers (1,51) and in patients with Fontan circulation (52). The flow across the right side of the heart, in general, is complex, and knowledge of right heart fluid dynamics remains rather limited.

The valvular diseases. Any change to the mitral valve, whether due to disease or surgery, directly affects the LV flow pattern. Some of the problems associated with mitral valve prostheses that may be related to blood flow are platelet activation, thrombus formation, hemolysis, increased myocardial stress leading to LV hypertrophy, and remodeling. Regurgitant flow in mitral prolapse has been shown to result in highly disturbed left atrial flow with both vortices located in proximity to the regurgitant jet as well as elevated values of turbulent intensity in the left atrium during systole, which are related to the regurgitant volume (50). Clinical studies are required to understand the impact of valvular diseases and replacement surgery on unfavorable LV remodeling that has been suggested by numerical simulation (53). Such an understanding would enhance the therapeutic decision-making process and improve surgical procedures, particularly in cases of dysfunctional ventricles.

Controversies still exist regarding the role of the vortices that develop in the sinuses of Valsalva in both optimizing aortic valve closure and allowing coronary blood flow. Several studies of the downstream flow from prosthetic valves and aortic root surgeries have been published in the last 10 years (52,54,55). The impact of changes in valve and prosthesis geometry on intrinsic flow in the aortic sinuses and coronary flow remain to be investigated.

Aortic and vascular flow. Aortic blood flow patterns have been described in healthy volunteers (9,56). Studies of alterations in aortic blood flow patterns, particularly the vortex and swirl behaviors related to age, coronary artery disease, atherosclerosis, congenital abnormalities, surgical interventions, and pulmonary hypertension, have recently been reviewed (9). In addition to its enhanced insight into hemodynamics, 3D cine phase-contrast CMR has demonstrated its advantages in the clinical evaluation of aortic coarctation (57). Using 3D cine phase-contrast CMR, retrograde flow from complex plaques in the descending aorta can explain embolisms to all brain territories and the resulting strokes (58). Larger studies and follow-up imaging of patients with abnormal flow patterns are needed to further evaluate the associations that have been drawn between altered flow in the thoracic aorta and disease outcomes.

Conclusions

The development and validation of novel flow visualization techniques may enable clinical imaging and analysis of multidirectional flow in the cardiovascular system. The identification, verification, and interpretation of flow-related physiology in normal and abnormal states may provide additional/incremental insights into a range of cardiovascular diseases. With advances in imaging, the time is perhaps ripe for further research into the diagnostic and prognostic impact of intracardiac and vascular flow structure.

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Key Words: contrast ■ doppler ■ echocardiography ■ flow ■ magnetic resonance ■ vortex.

▶APPENDIX

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