Three-dimensional blood flow in bifurcations : computational and experimental analyses and clinical applications

Meeting Report

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Abstract
In this report, the issues discussed at a multidisciplinary symposium on blood flow in bifurcations are summarized. Topics addressed are (1) flow analysis in in vitro models, using visualization and laser Doppler anemometer techniques, and numerical models; (2) the influence of (physiological) factors, such as vessel wall distensibility and vessel geometry, on the flow field; (3) the noninvasive assessment of arterial wall properties in humans, and (4) the noninvasive determination of flow patterns in humans, paying special attention to ultrasound techniques and magnetic resonance imaging. It was emphasized that it is of utmost importance to obtain more detailed information, preferably three-dimensional, about flow fields in bifurcations, not only from a diagnostic point of view but also to get more insight into the relation, if any, between flow patterns and atherogenesis. It was agreed that plaque geometry and dynamics should be studied in more detail, especially in relation to plaque fissuring and rupturing. There is a need for the noninvasive assessment of wall shear rate and, hence, to be able to calculate wall shear stress, because these parameters have been shown to be important determinants of endothelial cell function.

Introduction

Several aspects of blood flow distribution in bifurcations were extensively discussed at the Euromech Symposium 286, held in Kerkrade, The Netherlands on October 20-23, 1991. At this meeting, specific attention was paid to issues like flow in rigid in vitro models, using visualization and laser Doppler anemometer (LDA) techniques, flow analysis by means of numerical models, the influence of (physiological) factors such as vessel wall distensibility and geometry on the flow field, the noninvasive determination of arterial wall properties in humans, and the noninvasive assessment of flow patterns in humans with emphasis on ultrasound techniques and magnetic resonance imaging (MRI). Lectures were given by physicists, engineers, mathematicians, biologists and medical doctors.

General (Clinical) Aspects

It was agreed that from a medical point of view it is of outmost importance to obtain more detailed information about flow fields in such complicated geometries as arterial bifurcations. Three-dimensional analysis, in particular, may contribute to a better understanding of the

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flow disturbances induced by minor atherosclerotic lesions (<30% diameter reduction) at sites where the flow pattern is already complicated under normal conditions. It may also shed light on the influence of interindividual geometry variations on the flow field. Yet the relation, if any, between atherogenesis, known to take place in areas of flow separation and low wall shear rate, and flow patterns may benefit from a more detailed description of the flow field.

It has been known for quite some years that arteries loose their elastic behavior with increasing age, a process starting as early as the third decade, and that arteries are less distensible in patients with established and borderline hypertension than in age-matched control subjects. It was shown that in the carotid arteries this loss of distensibility is not homogeneously distributed along the carotid-artery bifurcation. Both at an older age and in borderline hypertensives, the carotid artery bulb is most affected. It was indicated that these inhomogeneities may have consequences for the flow pattern in the carotid artery bulb.

Atherosclerotic lesions without significant intraluminal processes are also associated with inhomogeneities in wall distensibility and locally disturbed flow patterns. This raises the question of cause or effect, and suggests the necessity for detailed flow analysis, preferably three-dimensional, with and without inhomogeneities in arterial wall distensibility under standardized flow conditions. The need to design better models of stress distribution in the arterial wall based upon biological observations was expressed. This makes it necessary to join forces with experts in the field of solid mechanics.

It was considered important to obtain more information about plaque geometry and dynamics, especially in relation to plaque fissuring and rupturing. Moreover, more detailed information about plaque geometry facilitates investigations into the relation between the geometry of the lesion and the type of flow disturbance, as studied in in vitro and numerical models.

It was clear that there is an increasing need for the noninvasive assessment of wall shear rate, and, hence, to be able to estimate wall shear stress, because these parameters have been shown to be important determinants of endothelial cell function and might affect mass transport of atherogenic proteins between plasma and the arterial wall. Accurate determination of wall shear rate requires velocity measurements close to the arterial wall, which is still problematic with the presently available noninvasive methods to measure velocity.

In vitro and Numerical Studies

Blood flow distribution in bifurcations is characterized by several geometrical and fluid dynamic parameters. Moreover, it is characterized by the (assumed) material properties of blood and vessel wall. Finally, boundary conditions like the inlet velocity profile, the shape of the pulse wave, the behavior of the surrounding tissue and the fluid dynamic behavior of the peripheral vascular beds are of great importance. Apart from the distribution of blood velocity and pressure in space and time, derived quantities such as shear stress, wall motion and residence time are also of interest. Depending on the limitations of the model used, more or less accurate information on these quantities can be obtained from in vitro and numerical studies.

Qualitative in vitro studies are often based on flow visualization techniques like dye injection, hydrogen bubble visualization and streaming birefringence in rigid (plexi)glass and deformable (latex or plastic) bifurcation models. These studies are very useful for obtaining an overall picture of the flow phenomena. Quantitative in vitro studies, like LDA and particle image velocimetry (PIV), can give detailed information on local velocity distributions with high spatial and temporal resolution. The data obtained from these studies can be used to analyze the local velocity in time and frequency domain to validate the results of numerical studies.

Numerical studies are based on approximation methods for the solution of the equations of motion (Navier-Stokes equations). The most important methods used at the moment are finite-difference, finite-volume and finite-element methods. With the use of large-scale supercomputers, fully developed three-dimensional time-dependent flow in complex geometries can now be simulated. However, reliable simulation of flow instabilities, flow of viscoelastic fluids and flow in deformable geometries is still problematic.

Geometrical and Fluid Dynamic Parameters

Considering bifurcations with one (straight) main branch in one plane dividing into two downstream (straight) daughter branches, important geometrical parameters are the cross-sectional area ratios of the different branches and the bifurcation angles (i.e., the angles of the daughter branches with the main branch). These geometrical parameters lose part of their significance if curved branches and bifurcations without a plane of symmetry are considered. Moreover, local geometrical factors, e.g., the presence of the carotid artery bulb in the internal carotid artery or the presence of atherosclerotic lesions, can be important for local flow and shear stress distributions. In these situations the area ratios and bifurcation angles can still be used to classify arterial bifurcations at different sites in the cardiovascular circulation.

Blood flow in bifurcation geometries is described by the equations of motion containing stationary inertial (convective), instationary inertial, viscous and pressure forces acting on a fluid particle. The characteristics of flow are determined by the mutual ratios between these forces. By using some representative length D (diameter of the main branch) and some representative velocity U (spatial mean axial velocity in the main branch), the equation of motion can be put into a dimensionless form. The mutual ratio between the convective forces and viscous forces is defined by the Reynolds number as:

$$\mathrm{Re} = \frac{UD}{\upsilon},$$

where υ is the kinematic viscosity of blood. Note that the Reynolds number is fully determined by the localization of the bifurcation in the cardiovascular circulation. It should also be noted that the Reynolds number can easily be defined for Newtonian fluids (υ is constant), but that this is more difficult in the case of non-Newtonian fluid behavior, where the viscosity instantaneously depends on the local shear (generalized Newtonian models) or even on the shear history (viscoelastic models). A second important dimensionless number is the Strouhal number defined by:

$$\mathrm{Sr} = \frac{D}{\tau U},$$

where τ is a representative time constant (period time of one pulse wave). This number represents the ratio of the instationary inertia forces and the viscous forces. For periodic flows, the Strouhal number is often replaced by the Womersley parameter:

$$\alpha = \frac{D}{2}\sqrt{\frac{2\pi}{\upsilon\tau}} = \sqrt{\frac{\pi \mathrm{Sr}\,\mathrm{Re}}{2}}.$$

Note that the Womersley parameter only incorporates the first harmonic of the actual flow pulse. More specific, the flow pulse can be characterized by an acceleration during the first part of the systolic phase of the cardiac cycle, a deceleration during the second part of the systolic phase and an almost steady behavior during the diastolic phase of the cardiac cycle.

Flow in Rigid Bifurcation Models

Two-Dimensional Steady Flow

From steady two-dimensional analyses of the flow in rigid bifurcation models it is known that the flow is characterized by the following. (1) A tendency to boundary layer separation along the nondivider side walls, with extensions into the daughter branches. In this way, regions of relatively low and in some cases even negative axial velocity and shear stress are formed at the nondivider side walls. The axial length of these regions increases for increasing Reynolds numbers and for increasing bifurcation angles. Due to the low velocities, relatively large particle residence times are found in this region. (2) The presence of relatively high axial velocities and high shear stresses at the divider side walls. The high shear region on the divider side of the bifurcation and the low shear region on the nondivider side are separated by a shear layer that becomes more pronounced at higher Reynolds numbers.

Unsteady Flow

In the case of unsteady flow, one can globally say that the flow regions mentioned above are built up during the diastolic phase of flow. During the acceleration phase of flow, the velocity rapidly increases at all sites in the bifurcation, resulting in a less pronounced division of the low- and high-velocity regions. During flow deceleration there is a large difference in the effects of inertial forces in the high- and low-velocity regions. This results in a relatively complicated flow field in which the shear layer mentioned above is rapidly built up. Depending on the geometrical and fluid dynamic parameters mentioned above, both shear layer instability and vortex formation can be found in this phase of the flow cycle. It will be clear that unsteadiness of the flow strongly affects the distribution of wall shear stresses. The observed temporal variations in location and length of the low-shear region induce temporal variations of wall shear stress reaching values that can be significantly different from the mean value.

Three-Dimensional Flow

An additional characteristic of the flow in three-dimensional bifurcation models, as compared to two-dimensional models, is the helical flow pattern in the daughter branches induced by the secondary velocity distribution, i.e., by the velocity components perpendicular to the axial flow. This secondary flow is found to have properties similar to those observed in three-dimensional flow in a bend and is caused by the centrifugal forces acting at the entrance regions of the daughter branches of the bifurcation. Due to these centrifugal forces, fluid from the nondivider side wall of the daughter branches, which has a small or even negative axial velocity component, is transported to the center of the branches. Fluid from the divider side walls, which has a relatively large axial velocity component, is transported along the walls of the branches towards the nondivider walls. The resulting helical flow pattern is observed in both steady and unsteady flow and is found to be more pronounced at larger bifurcation angles. An important dimensionless parameter with report to this secondary flow (Dean vortices) is the Dean number:

$$De = Re\sqrt{a/R},$$

where a denotes the radius of the daughter branch and R is a measure for the curvature radius formed by the main branch and the daughter branch. This dimensionless parameter originates from analysis of the flow in a bend and must be interpreted as the magnitude of the convective and centrifugal forces compared to the magnitude of the viscous forces. It will be clear that this secondary fluid motion strongly influences shear stress and particle residence time in the daughter branches. In particular, regions with relatively low and oscillating wall shear stress are assumed to be preferable sites for the formation of atherosclerotic lesions.

From the observations described above it can be concluded that in the analyses of the flow in arterial bifurcations both incorporation of time-dependent and three-dimensional effects are essential.

The Influence of Physiological Factors

Flow in Deformable Bifurcation Models

The arterial wall is not rigid but consists of nonlinear anisotropic and viscoelastic material. Moreover, the mechanical properties of the arterial wall depend on their anatomical environment. Incorporation of these properties in experimental or numerical models of bifurcations is rather complex. The deformability of the arterial wall not only results in wall motion but also in propagation and reflection of pressure waves. With respect to wave propagation, the arterial bifurcations may be regarded as compact. This means that the reflection and transmission of waves at arterial bifurcations take place at transitions with a length that is small compared to the wave length. Therefore, the wave phenomena can be described using simple wave propagation models. Knowing that the characteristic fluid motion U is small compared to the wave velocity c, it can be shown that the wall motion is mainly determined by the pressure wave. Once the pressure wave distribution is known, either from experimental data or from wave propagation models, the wall motion is determined by the stress-strain relationship. For the arterial wall, the relationship between circumferential stress and circumferential strain is mostly determined from pressure-radius relationships (dynamic compliance) obtained from in situ or in vivo measurements. The wall motion will slightly influence the local velocity distribution and will reduce the areas of low and negative axial velocity on the nondivider side of the daughter branches of the bifurcation. The influence on local wall shear stress distributions is not clear yet and depends on the assumed viscoelastic properties of both the vessel wall and blood.

The Non-Newtonian Behavior of Blood

The viscosity of blood is not a constant but exhibits certain non-Newtonian properties: like (1) shear thinning: the steady viscosity of blood decreases with increasing shear rate; (2) viscoelasticity: the dynamic viscosity of blood exhibits a nonzero elastic component, and (3) thixotropy: the viscosity is time-dependent due to flow-induced breakdown and buildup of nonelastic structures.

In the low-shear regions of the flow in bifurcations especially, the non-Newtonian viscoelastic properties of blood play an important role, and a different flow behavior can be observed as compared to the flow of Newtonian fluids. Moreover, regardless of the difference in flow behavior, a significant influence on the wall shear stresses may be expected as a result of the different stress-strain relationship. Determination of a proper viscoelastic constitutive model for blood, and measurement of the corresponding viscoelastic properties is still a challenge.

Influence of Geometry

Bifurcation geometries used in in vitro and numerical models originate from: mathematically derived geometries (e.g., T-junctions consisting of straight tubes); representative geometries based on a mean geometry obtained from a (large) number of casts, and a single cast.

Although the global flow phenomena will be similar for different bifurcation geometries, research has to be focussed on local flow phenomena and their relationship to localization, size and shape of atherosclerotic lesions. Analyses of flow disturbances induced by minor atherosclerotic lesions, preferable three-dimensional, also remain an issue. Analysis of particle trajectories by experimental and numerical flow visualization techniques is expected to be helpful.

Noninvasive Assessment of Flow Patterns by Doppler Techniques

The methods generally used to determine arterial flow patterns noninvasively in humans are based upon Doppler ultrasound and, more recently, MRI. Considering the size of the arteries of interest (e.g., the proximal part of the internal carotid artery with an internal diameter of the order of 6 mm), any suitable velocity-measuring system should have a high spatial resolution. For ultrasound Doppler systems this is achieved by utilizing a high emission frequency allowing a narrow ultrasound beam. Depth resolution is accomplished (only for pulsed Doppler systems) by emitting short bursts of ultrasound, containing a few (in the order of 4) cycles at the emission frequency, in combination with a short signal analysis window (sample gate). The length of the sample gate is constituted by the effective duration of the impulse response of the low-pass filter following the demodulation of the received signal using two reference signals, both having a frequency equal to the emission frequency but shifted 90 degrees in phase with respect to each other (in quadrature demodulation). This allows, in a later phase of processing, the retrieval of the velocity direction (towards or away from the transducer). The outputs of the demodulator at a predefined delay with respect to emission constitute a sample of the in-quadrature Doppler signal originating from a sample volume (effective gate width in combination with local beam width) at a predefined depth. Subsequent emission-receiving cycles will give the Doppler signal as a function of time. Each cycle of the Doppler signal corresponds to a shift in phase of the scatterers within the sample volume over half a cycle (due to the round-trip effect) of the reference signal, equivalent to a displacement of half a wavelength, where a wavelength equals f_e/c, with c as the velocity of sound in the medium, and f_e as the emission frequency. For an arbitrary velocity v of the scatterers at an angle α with respect to the emitted beam, the above observation transforms to the well-known Doppler equation:

$$f_d = 2f_e \frac{v \cos\alpha}{c},$$

with f_d as the Doppler frequency. The mean Doppler frequency within a time segment can be estimated by evaluating the angle (including its sign indicating the direction of the velocity) of the complex autocorrelation function of the in-quadrature Doppler signal with a time lag of one sample point. It should be noted that this is not equivalent to the mean frequency of the Doppler spectrum since the latter also contains inevitable noise, while the autocorrelation approach tends to remove the noise contribution.

The sample volume is positioned under cursor control using a B-mode image as reference. If the sample volume is located outside the lumen or close to the wall, the Doppler signal originates from slowly moving structures as well. Due to reverberation, even in the center of the lumen, high-amplitude echo signals may be observed. These structures induce high-amplitude Doppler signals obscuring the low-amplitude Doppler signals induced by the moving red blood cells. A high-pass filter applied to the complex Doppler signal prior to frequency estimation should suppress these unwanted signals. The choice of the cutoff frequency then determines the minimum Doppler frequency, i.e., the minimum velocity, that can be measured. A too high cutoff frequency does not permit measurement of low velocities, e.g., close to a vessel wall, while a too low cutoff frequency insufficiently suppresses the unwanted components.

An extension of the single-gate pulsed Doppler system discussed above is the multigate pulsed Doppler system where a number of sample gates are evaluated simultaneously producing a velocity distribution along the ultrasound beam. If the direction of the beam is successively swept across the image the velocity distribution in a plane is obtained. The latter can be displayed color-encoded and superimposed on the B-mode image coded in shades of gray, allowing the visualization of flow phenomena in relation to nearby structures. Thus, the ultrasound B-mode image is very useful for obtaining anatomical information. B-mode imagers can even be adapted to generate a three-dimensional presentation of atherosclerotic lesions.

The transition from multigate pulsed Doppler systems to color-coded Doppler systems, where a large number of sample lines have to be interrogated, requires shortening of the time window used for frequency estimation to a few sample points (typically 8–16 samples of the Doppler signal at each depth). Due to the sequential nature of Doppler systems, the number of flow maps per second is limited by the pulse repetition frequency (PRF) used divided by the number of lines comprising the flow map times the number of sample points within a time window. To obtain smooth flow map transitions the flow map rate should be at least ten. The short time window sets a lower limit for the cutoff frequency of the Doppler high-pass filters. Moreover, it is at the expense of velocity resolution: a shorter time window inevitably leads to a higher variation in the estimated mean frequency. But a velocity map is very helpful to locate quickly those areas with unexpectedly high or disturbed velocities or with a (physiologically) reversed velocity direction. In the second stage of investigation, suspected areas can be interrogated in more detail using the single-gate Doppler modality.

As mentioned above, multigate pulsed Doppler systems allow the noninvasive detection of the velocity distribution along the ultrasound beam. Because the length of the time window is only limited by the dynamic characteristics of the velocity signals, high accuracy can be obtained. However, the Doppler high-pass filter limits the ability to measure low velocities and, therefore, the shear rate close to the vessel wall. The availability of high-speed and cheap mass storage chips allows the capture and temporary storage of the received signals over a large number of subsequent lines. These lines can then be processed off-line with a computer, allowing adaptive settings of processing parameters like the frequency rejection range of the Doppler filter. In this approach, the frequency rejection range is not centered at zero Doppler frequency, but shifts with the Doppler frequency of the nearby vessel wall. This effectively maximizes the frequency range used for the Doppler frequency estimation. With small sample volumes (narrow ultrasound beam, requiring a high emission frequency, in combination with a short burst and a short sample gate), the blood velocity can then be measured rather close to the wall (at a distance of less than 1 mm). The uncertainty in position is mainly caused by the difficulty establishing the exact position of the wall lumen interface at any phase of the cardiac cycle. Preliminary investigations of the common carotid artery of a young, presumably healthy, subject revealed a peak shear rate of 900 sec^{-1}.

Conventional Doppler systems measure only the velocity component along the ultrasound beam; the transverse component is thereby discarded. However, it is possible to modify a (pulsed) Doppler system such that the transverse component can be measured. This approach is based on the width of the Doppler spectrum distributed symmetrically around the zero Doppler frequency (the mean Doppler frequency induced by scatterers moving perpendicularly through the ultrasound beam is zero). The higher the velocity of scatterers, the shorter the observation time, causing a larger Doppler bandwidth (transit time effect). Since the beam is aimed perpendicularly at the artery, the position resolution only depends upon the length of the sample volume. Using short bursts in combination with a short sample gate (requiring a large bandwidth of the Doppler ultrasound system), a high spatial resolution is obtained independent of the local width of the ultrasound beam. Moreover, the velocity measured is absolute, at least for observation angles close to 90 degrees. A minor deviation from the intended angle will show up as a shift in the mean Doppler frequency. Combining this shift with the width of the Doppler spectrum allows for vectorial velocity measurements provided that the spectral broadening to the transit time effects dominates over the spectral width due to the local velocity distribution. The only problem is that the sample window cannot be placed arbitrarily close to a vessel wall since the latter moves in space during a cardiac cycle. However, in vitro experiments have shown that it is feasible to adjust the position of the sample gate in accordance with the movement of a nearby structure.

In pulsed Doppler systems, the displacement of scatterers in between successive emissions is detected using a reference signal with the same frequency as the carrier frequency of the emitted signal. A short ultrasound burst has a wide frequency distribution. Due to the frequency-dependent attenuation (in the order of 1 db/cm/MHz), the higher frequencies will be more attenuated than the lower frequencies, causing a downward shift of the mean frequency, increasing with depth. The assumption that the received signal still has the same mean frequency is essentially incorrect. This has prompted designers to develop a new processing scheme to detect displacement (and velocity). The received signals of subsequent lines within short windows are correlated, where the position of the cross-correlation peak represents the displacement. For this purpose, the received signals are digitized at a high sample frequency (preferably at more than 8-fold the assumed emission frequency). In the simplest approach, only the sign of the received signal is used, simplifying analog-to-digital conversion, intermediate storage and computation of the correlation function. An interpolation scheme is used to estimate the position of the peak between sample points. The contribution of strong reflectors can be suppressed by considering the difference of subsequent lines (echo cancelling).

The result of this processing algorithm is an estimate for the displacement (and thence the velocity) expressed in sample points independently of the effect of depth-dependent attenuation. Moreover, the length of the sample window can be chosen freely independently of the duration of the emitted pulse. For a longer sample window, the estimate for the displacement will be more accurate at the expense of the axial resolution. Of course, sample windows may be partially overlapped to achieve an image with gradual velocity transitions, but this will not change the inherent depth resolution. The use of only the sign of the received signal has a nonlinear effect on the result (high and low amplitude signals do not contribute in an additive way), and makes the accuracy of the result dependent on the phase of the received signal. The difference between both approaches (Doppler velocity imaging and correlation velocity imaging) will be apparent especially for short emitted pulses. Assuming an ultrasound system operating at 5 MHz with a burst length of 2 periods, requiring a signal bandwidth of 2.5 MHz, the downward shift in mean frequency for signals originating from a depth of 4 cm will be 14%, while for a bandwidth of 1 MHz (a burst of 5 periods) the downward shift will be only 3%. The same error (percentage of underestimation) will appear in the estimate for the velocity using the Doppler approach.

Noninvasive Assessment of Arterial-Wall Properties with Doppler Techniques

Actually, a tracking sample volume, as suggested above, is also used with Doppler systems to detect the displacement of arterial walls and, hence, the distension of an artery as a function of time. Using the conventional B-mode, the section of the artery of interest is localized, and then the system is switched to M-mode (observation along a single line). The RF signals received in M-mode (sampled at 4-fold the assumed carrier frequency) are temporarily stored in a buffer memory large enough to hold the raw signals for a registration time of a few seconds. After completion of registration, the first RF line stored is displayed on the screen of the attached computer, and the sample volumes are manually positioned at the wall lumen interfaces. Software Doppler processing of the signals within these sample volumes over subsequent lines reveals the displacement of the wall echo as a function of time. When the observed displacement exceeds a quarter of a period of the RF signal (one sample point) the position of the sample volume is automatically adjusted to track the movement of the wall. The result is a recording of the position of the anterior and posterior walls as a function of time and, taking the difference between both, the distension waveform. Experiments have shown that this method is highly accurate (to the order of a few μm), provided that the sample frequency for data acquisition is close to 4-fold the carrier frequency, and the initial position of the sample volumes is carefully chosen (not covering a transition of two reflections).

Determination of Flow Patterns with MRI

Magnetic resonance (MR) was proposed for measuring velocity over 35 years ago. With the recent advent of whole-body scanners, MR velocimetry (MRV) is now possible for in vivo and in vitro measurement of complex cardiovascular flows. While ultrasound velocity measurement is a mature field, MRV is in its infancy. Typically, two types of velocity measurements are used: (1) phase velocity encoding, where moving particles attain a phase shift along a gradient, and (2) time-of-flight methods, where a band of fluid is magnetized and its location is recorded at a subsequent fixed time. The first method is gaining in popularity because the sequence can be easily incorporated into presently available sequences.

MRV affords several distinct advantages over duplex Doppler ultrasound for velocity measurement. Velocities can be obtained in any direction, resolution is not anatomically dependent since only a magnetic 'window' is required, and a three-dimensional digital data set is acquired directly. The ultimate physical resolution of MRI is on the molecular scale, but typically, body coil resolutions are 0.5 × 0.5 × 0.5 mm. Since this voxel can be oriented along the flow axis, the radial resolution is approximately 0.5 mm. Duplex ultrasound equipment typically has a much greater voxel size because of the large spread of the sound beam in-plane and the necessary Doppler angle.

Several disadvantages should also be noted. MRV measurements are required over minutes instead of milliseconds. While the results can be processed almost in real-time, the velocities actually represent time-averaged velocities and not instantaneous velocities. Pulsatile velocities are obtained by gating the data acquisition over many heart cycles. Secondly, highly turbulent flow through stenoses exhibits a large amount of signal loss, making measurements of stenotic flow highly suspect. Convective acceleration from stenoses also contributes to errors as large as 40%. However, at the present time newer techniques are being explored to address these limitations. Most important, perhaps, is the high cost of the machine and the study. At present an MR imager costs about 10-fold a color Doppler machine, and an MR scan costs approximately 3-fold that of an ultrasound flow study. On the other hand, color flow Doppler and multigate pulsed Doppler only produce the velocity pattern within the observation plane or along the ultrasound beam, respectively.

For normal physiological flow conditions, MRV is accurate and can resolve the secondary and three-dimensional pulsatile velocities typically present in in vivo hemodynamics. Velocity measurements can range from 0.5 cm/s to more than 10 m/s. Measurement of over 2,000 velocity points over the entire cross-section of the aorta can be acquired in minutes. The digital data are easily reconstructed into three-dimensional animated displays. The instrument is noncontacting and a patient study can be performed within thirty minutes. The simultaneously acquired anatomic images have a greater physician acceptability. Ultrasound velocity measurement remains a high standard where the capabilities and limitations have been well documented. MRV, however, holds tremendous promise but remains to be fully characterized.

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