Determination of the Most Appropriate Velocity Threshold for Applying Hemispheric Flow Convergence Equations to Calculate Flow Rate: Selected According to the Transorifice Pressure Gradient

Digital Computer Analysis of the Doppler Color Flow Convergence Region

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Background. While flow convergence methods have been promising for calculating volume flows from color Doppler images, it appears that the velocity threshold used and the transorifice pressure gradient dramatically influence the accuracy of application of the simple hemispheric flow convergence equation for calculation of flow rate. The present in vitro study was performed to determine whether the value of velocity threshold at which the shape of proximal isovelocity surface best fits given shape assumptions with different orifice sizes and flow rates is predictable as a function independent of orifice size from clinically measurable peak velocity or transorifice pressure gradient information.

Methods and Results. In an in vitro model built to facilitate ultrasound imaging, steady flow was driven through circular discrete orifices with diameters of 3.8, 5.5, and 10 mm. Flow rates ranged from 2.88 to 8.28 L/min with corresponding driving pressure gradients from 14 to 263 mm Hg. At each flow rate, Doppler color-encoded M-mode images through the center of the flow convergence region were obtained and transferred into the microcomputer (Macintosh IIci) in their original digital format. Then, the continuous wave Doppler traces of maximal velocity through the orifice were derived for the calculation of driving pressure gradient. Direct numerical spatial velocity measurements were obtained from the digital color encoded M-mode velocities with computer software. For each flow rate, we could calculate flow volume from any number of velocity distance combinations with a number of assumptions and use the results to assess expected flow convergence shape based on a priori knowledge of the progression from oblate hemispheric to hemisphere to prolate hemispheric changes observed previously. Our results showed that for a given ratio of calculated flow rate to actual flow rate (0.7 and 1), the velocity threshold that could be used for the calculation of flow rate with a hemispheric flow convergence equation correlated well with the pressure gradient for a given orifice size, and the differences in velocity threshold that could be used this way among different orifice sizes once they were adjusted for the covariance pressure gradients were not statistically significant (P=.79 for ratio=0.7, and P=.81 for ratio=1).

Conclusions. Our present study provides an orifice size-independent quantitative method that can be used to select the most suitable velocity threshold for applying a simple hemispheric flow convergence equation based on clinically predictable pressure gradients ranging from 40 to 200 mm Hg, and it offers a correction factor that can be applied to the hemispheric flow convergence equation when the pressure gradient is less than 40 mm Hg. (Circulation. 1993;88[part 1]:1699-1708.)

Key Words • flow • echocardiography

Color Doppler mapping of the flow convergence region proximal to a regurgitant or stenotic orifice has recently been proposed as a new method for noninvasive quantitation of flow rate between cardiac chambers.1-8 The theories behind this method state that the flow rate can be obtained by multiplying the proximal isovelocity surface area in an accelerating flow field imaged using color Doppler by the aliasing velocity at that surface. Because of an inability for direct measurement of the proximal isovelocity surface area by color Doppler due to spatial errors, slow sampling rates, and angle dependence of Doppler velocity measurements, simplified methods to

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calculate the proximal isovelocity surface area have been proposed that use the center radius of the flow convergence region in a direction parallel to Doppler sampling by assuming a hemispheric geometric shape. 1 Although both in vitro and animal studies have demonstrated good correlations between actual and calculated flow rate using this assumption, calculated flow rate commonly underestimates or overestimates actual flow rate depending on the different color Doppler aliasing velocities used with different actual flow rates and varying orifice sizes. 3,5,9-11 It is now clear that the geometric shape of proximal isovelocity surface changes with flow rate and orifice size, although a priori knowledge to guide proper choice of these variables is not available in clinical application of these methods. 1,3,9,10,12-17

A recent study from our laboratory 17 has shown that the shape of proximal flow convergence isovelocity region is a function of transorifice pressure gradient and aliasing velocity and that it can assume any shape from flat "pan-shaped" ellipse to hemisphericaloid shapes depending on the value of aliasing velocity (Fig 1A through D) and pressure gradient. Hence, applying the simple hemispheric flow convergence equation to these images will obviously result in different calculated flow rates for different aliasing velocities chosen in the acceleration field even though actual flow rate remains constant. The progression from overestimation to underestimation we observed when using magnetic resonance imaging (MRI) angle-independent velocimetry and calculating hemisphere-based flow rates 14,15 suggested a systematic change that was to some extent velocity related and led to our asking whether continuous wave Doppler velocity could be used to suggest the portion of the zone of acceleration wherein the hemispheric assumption, the simplest to implement, might be applied to calculate flow volume (Fig 2). The present study was designed to allow digital acquisition of centerline acceleration of velocities to allow calculations of the whole range velocities by unwrapping aliasing and to allow tests of the fit of a number of geometric flow convergence models over a range of flows, orifice sizes, and driving velocities in an in vitro model. We had hoped to use these data to look for methods of predicting where within the flow convergence zone and when clinically hemispheric flow convergence geometry assumptions could be used to estimate volume flow rate.
Methods

Experimental Design

For generating a steady acceleration field, steady flow was driven through circular discrete orifices mounted in a dual-chamber model. Planar circular orifices with diameters of 3.8, 5.5, and 10 mm were used. Flow was driven into the top chamber of a circular plastic tube (diameter, 12.4 cm; length, 24 cm) through the orifice mounted in-between the top chamber and bottom chamber, into the bottom chamber, and back out to a recirculating pump. In this model, even for the largest orifice, the rounded walls were infinitely large related to the dimensions of the orifice and flow convergence. A solution of water mixed with 1% by weight cornstarch was used as the fluid medium. Actual flow rate was measured in the model using a rotameter, and flow rates were cross-checked by draining the model into a graduated cylinder and timing the collection with a stopwatch. Flow rate ranged from 2.88 to 8.28 L/min. The model has previously been described.17

Doppler Color Flow Mapping and Image Transfer

Doppler color flow mapping was performed using a Vingmed CFM 750 ultrasound imaging system (Vingmed Sound, A/S, Norway) with a transducer frequency of 5 MHz and aliasing velocity of 50 cm/s. The color Doppler high-pass filter was complex but has a rolloff to minimize flow velocities <8 cm/s (medium). This system is equipped with a digital output port that permits us to transfer two-dimensional and color M-mode Doppler velocity assignments in their original digital format from the digital scan converter in the system directly into a microcomputer before color assignment and without having to convert the signal into analog format. Doppler flow map data are thus transferred as digital velocity assignments.18,19 For scanning, the transducer was placed and fixed on a specially constructed window in the model to image flow going away from the transducer. All other imaging settings,
including gain and depth, were optimized and kept constant throughout the study. Color M-mode flow convergence images were acquired by placing the M-mode cursor line through the center of the two-dimensional imaged flow convergence region aimed towards the orifice.

For every flow state (every discrete combination of orifice size and flow rate), a sequence of 10 to 20 two-dimensional Doppler color flow frames and Doppler color-encoded M-mode images through the center of flow convergence region were acquired on the ultrasound scanner digital cineloop and transferred into the microcomputer (Macintosh IIci) in original digital format for subsequent offline computer analysis. Also at each setting, the continuous wave Doppler traces of maximal velocity through the orifice were obtained and used for the calculation of driving pressure gradient using a simplified Bernoulli assumption.

**Measurements and Data Analysis**

Doppler color-encoded M-mode images through the center of the flow convergence region were analyzed (Fig 1E). The velocity information could be averaged as a function of time on the M-mode tracings because of the steady flow setup used (Fig 1E). Care was taken to always have the bottom of the region of interest at the same position at the orifice and the upper zone of the M-mode line proximal to the beginning of the flow convergence region.

Direct numerical spatial velocity measurements were accomplished using digital values for color pixel intensity obtained with computer software provided by Vingmed (Vingmed Sound, A/S, Norway) with a sampling gate increment of 0.4 mm starting from the orifice position to a point proximal to the beginning of flow convergence region.18,19 Because flow was imaged going away from the transducer and displayed as blue, the velocities measured by computer were represented by negative numbers. If velocity exceeded the Nyquist velocity, the color became red (alias), and the velocities were shown as positive numbers. The velocities measured at the alias region were reassigned and recalculated into true velocities according to the aliasing velocity and the number of times aliasing occurred. Flow rate was calculated with the hemispheric flow equation:

\[ CFR = 2 \cdot \pi \cdot R^2 \cdot V \cdot 60/1000 \]

where CFR is calculated flow rate (L/min), R is radius (cm) from the orifice, and V is velocity (cm/s) at the radius R. Then, the individual velocity thresholds used for the calculation of flow rate were plotted versus radii as well as the ratios of calculated flow rate to actual flow rate. Since a hemispheric flow convergence equation was used to calculate flow rates, the value of the calculated to actual flow ratio contains information about the expected isovelocity surface shape at any given velocity threshold, ie, a value of 1 would be consistent with a hemispheric shape; a value of more than 1 would indicate elongation of shape toward a prolate hemispherial configuration, the larger the ratio, the more asymmetric the shape; and values of less than 1 would indicate flattening toward a pan-shaped ellipse configuration, the lower the ratio, the more flattened the shape (Figs 1 and 2). Because the velocity thresholds available for use in the calculation of flow rate actually represent the serial different velocities on the color bar used in color Doppler imaging system, the resulting curves allow us to determine the threshold velocity at which the shape of the proximal isovelocity surface is best fit for a given shape assumption and what kind of correction factors should be introduced for the calculation of flow rate with the hemispheric flow convergence equation at different flow rates and orifice sizes or pressure gradients. In digital processing of the data transfer for the Vingmed 750 system, any velocity limit could be chosen during data analysis, yielding a whole stream of velocity-distance data. Differences among velocity threshold for a given ratio of calculated to actual flow rate at different transorifice pressure gradients were determined from actual measurements and expressed in absolute numbers.

The pressure gradient for every discrete combination of flow rate and orifice size was calculated using the simplified Bernoulli equation \( \Delta P = 4 \cdot V^2 \), where \( \Delta P \) is pressure gradient (mm Hg) and V is maximal velocity (m/s) at the orifice measured with continuous wave Doppler. Pressure gradients were plotted versus velocity threshold at which the ratios of calculated to actual flow rate were either 0.7 or 1, respectively, to determine whether these velocities (thresholds) were related to the potentially clinically measurable pressure gradients using simple linear regression analysis. The differences among the linear regression lines for different orifice sizes were determined with an analysis of covariance.

**Results**

The direct numerical velocity measures were obtained for the regions containing a single alias. Because of the multialiasing and difficulty of determining precisely how many times aliasing occurred near the orifice, the true velocities were not measurable in these regions close to the orifice.

**Relationship of Velocity to Radius**

Fig 3 shows the relationship of velocities to the radial distance at which they occurred for different flow rates at different orifice sizes and shows how the velocities increase when approaching the orifices. An increase in flow rate resulted in a right and upward shift to these curves. This suggests that for the whole velocity range, the isovelocity surface radius is larger with higher flow rates.

Fig 4 shows the relationship between velocities and radii for two different orifice sizes at two flow rates. Note that the curve has shifted upward and the slope has become steeper in the mid- and high-velocity range when the orifice diameter decreased from 5.5 to 3.8 mm, indicating that velocity acceleration profile of the flow convergence region for different sized orifices changes even though the flow rates are matched.

**Relationship of Ratios of Calculated Flow Rate to Actual Flow Rate to the Velocity Thresholds**

Fig 5 shows the relationship of the ratio of calculated flow rate using a hemispheric flow convergence equation to actual flow rate to the varying velocity thresholds along the center line at different actual flow rates and orifice sizes. It demonstrates that the shape of isovelocity surface is not constant and changes with the velocity threshold used for the calculation of flow rates for a given flow rate.
and orifice size or pressure gradient. As such, the hemispheric flow convergence equation will obviously result in different calculated flow rates for different aliasing velocities with progressive underestimation near the orifice and progressive overestimation far from the orifice even though actual flow rate remains constant.

The position of the curve shifted upward and the slope became flatter with increases in flow rates or pressure gradients for a constant orifice size. This means that for the whole range of ratios of calculated flow rates to the actual flow rates, ie, the given shape of the isovelocity surface and thus the velocity threshold that can be used for the calculation of flow rate using hemispheric flow convergence equation increases with flow rates or pressure gradients for any given orifice size. This relationship was sensitive to variation in transorifice pressure gradient. For example (Fig 5B), increases in transorifice pressure gradient of 15 to 41 mm Hg for the curves resulted in changes in velocity threshold of 7 and 16 cm/s for ratio of 1 and from 4 to 18 cm/s for a ratio of 0.7, suggesting that for the calculation of flow rate, velocity threshold can be adjusted to make the aliasing pattern fit for the given shape of isovelocity surface selected for flow rate or pressure gradient. It also demonstrates that the shape of isovelocity surface for a given velocity threshold changes with the flow rate or pressure gradients for a given orifice size.

From the curves, it was also found that the shape of isovelocity surface was still flattened and calculated flow rates still underestimated actual flow rates even at the low-velocity range (as low as 20 cm/s) for the largest orifices and lowest flow rates (Fig 5D and E) that were corresponding to the pressure gradient of less than 40 mm Hg, and the shape was still prolate hemispherical and calculated flow rates still overestimated actual flow rates even at the high-velocity range (more than 80 cm/s) for the smallest orifice and high flow rates (Fig 5A) that were corresponding to the pressure gradients of more than 200 mm Hg. This suggests that beyond these ranges we could not select isovelocity contour ranges that behaved hemispherically but that different corrections would need to be applied in addition to velocity selection for the calculation of flow rates with the hemispheric flow convergence equation for conditions of both very-low- and very-high-pressure gradients.

**Relationship of Velocity Threshold for a Given Shape of Isovelocity Surface at Different Flow Rates and Orifice Sizes to the Pressure Gradients**

For a given ratio of calculated flow rate to actual flow rate (0.7 and 1 in the present study), the velocity thresholds that were usable for the calculation of flow rate with the hemispheric flow convergence equation were correlated well with the pressure gradient across the orifice for a given orifice size (Figs 6 and 7),
indicating that the velocity threshold used for the calculation of flow rate using the hemispheric flow convergence equation for a given shape of isovelocity surface and a given orifice size was predictable from pressure gradients. All linear regression lines for the calculated to actual flow rate ratios of 0.7 and 1 between the different orifice sizes appeared to have similar slope, and the differences in velocity thresholds among the different orifice sizes adjusted for the covariate pressure gradients were not statistically significantly different (Figs 6A and 7A, $P=.79$ for ratio=0.7 and $P=.81$ for ratio=1). This suggests that for a given ratio of calculated to actual flow rate, i.e., for a given shape of isovelocity surface, the velocity thresholds (selected using pressure gradient as an index parameter) that could be used for the accurate calculation of flow rate by a hemispheric formula over a wide range of pressure gradients was independent of orifice size.
**Discussion**

**Effects of Selected Aliasing Velocities on the Shape of Isovelcity Surface**

The quantitation of flow rates through a regurgitant or stenotic orifice using the proximal flow convergence method is based on the continuity equation. The assumptions in the method predict that the flow rate can be obtained by multiplying proximal isovelcity surface area by color Doppler aliasing velocity. Because the direct measurement of the true proximal isovelcity surface area cannot be made by color Doppler from any plane of information because of angle dependency of Doppler, the simplest usual method to calculate the proximal isovelcity surface area is from the center radius of the flow convergence region by assuming a hemispheric geometry. Several investigators have used this method to calculate the regurgitant flow rates in vitro and in animal studies and obtained good correlation between calculated and actual flow rates. However, calculated flow rates have sometimes either underestimated or overestimated actual flow rates with the hemispheric flow convergence equation in different conditions of flow rate, orifice size, and chosen aliasing velocity, with a general tendency for use of low aliasing velocities in high-flow rate studies and small orifice sizes to overestimate actual flow rate and for high chosen aliasing velocities in low-flow rate studies and with large orifice sizes to underestimate actual flow rate. Our present study demonstrated that the shape of the expected isovelcity surface was not constant but changed with the velocity thresholds for any given flow rate and orifice size (Fig 1A through D) and also changed with the flow rate or pressure gradients for a given velocity threshold. This would explain the overestimation or underestimation of actual flow rates reported previously depending on the different flow rates assessed or the aliasing velocities.
selected for use with a hemispheric flow convergence equation. In the present study, the shape of isovelocity surface was determined by the ratio of calculated to actual flow rate. Our findings are in agreement with previous observations that the shape of isovelocity surface determined by (1) directly viewing the flow convergence region on color Doppler after color angle correction for the lateral dimensions, (2) when flow convergence was imaged by direct optical visualization of particle tracking, or (3) when it was imaged by phase velocity encoded MRI.14 showed that an aliasing region could assume any shape from flat pan-shaped ellipse to hemispheric to prolate hemispherical depending on the value of the velocity threshold selected.21

**Proximal Velocity Profile of Flow Convergence Region**

In the present study, the velocities along the center line of flow convergence region were acquired as estimated flow velocity using computer software with a depth increment of 0.4 mm from the orifice position from images that were directly transferred into the microcomputer from the digital scan converter in the ultrasound imaging system. This digital computer technique has proven to be very accurate for extracting spatial velocity data and has been validated previously.18,19 Our findings are generally consistent with previous reports in which the velocity acceleration profile was obtained by measuring the axial radius of flow convergence region at varying aliasing velocities.22

Since it was very difficult to identify the actual shape of isovelocity surface when planning blind application of hemispheric flow convergence equation due to the dependence of color aliasing pattern on flow rate and orifice size, some investigators22 have analyzed the velocity profile of flow convergence region instead of the calculation of a single isovelocity surface area to try to assess the flow rate by comparing the position of measured velocity profile curves with a nomogram to a previously determined family of velocity-distance profiles of flow convergence regions at different flow rates. Although they developed a nomogram for estimating flow rates independent of orifice size,22 our results indicated that the position of velocity profile curves varied not only with flow rate for a constant orifice size but also with the orifice size even for a constant flow rate, suggesting that the method to analyze the velocity profile to assess flow rate was influenced by orifice size, an unknown in potential in clinical applications such that overestimation and underestimation could occur.

**Correction Procedure for Using the Simple Hemispheric Flow Equation**

As indicated in the present and previous studies,3,12,17,22 the theoretical assumption that the proximal flow convergence region consists of a series of hemispheric surfaces does not fit for the practical situation where the shape of the isovelocity surfaces changes based on the flow rate, orifice size, and aliasing velocity, changing from flattened to hemispherical to hemispheric shapes such that applying the simple hemispheric flow convergence equation to calculate flow rates will obviously overestimate or underestimate actual flow rates (Fig 2). Several investigators9,12,24 have attempted to provide modified methods as enhancements to the simple flow convergence technique. In an in vitro study, Utsunomiya et al3 assumed the shape of isovelocity surface as hemiellipsoid rather than hemispheric and changed the flow convergence equation into one where the area became the surface area of a hemiellipsoid. The surface area was calculated by measuring the long-axis aliasing radius, short-axis aliasing radius, and 90° short-axis radius from two orthogonal scanning planes. They found that although aliasing radius decreased with increasing aliasing velocities, calculated volume flow rate was constant with the percent difference between calculated and actual flow rates ranging from 9.2% to 4.3% for the four aliasing velocity conditions. However, in the clinical setting, the flow convergence method using a hemielliptic assumption to calculate flow rates is limited by the requirement that two orthogonal, a longitudinal, and a 90° short-axis view must be recorded. The short-axis image can almost never be obtained clinically. Another simple and applicable correction method was to measure the axial radius and transverse one of flow convergence region on only one two-dimensional color Doppler scanning plane in the hemispheroidal surface area calculation instead of three radii on two orthogonal scanning planes, and a satisfactory result was reported using this method to estimate the cardiac output in an animal experiment.24 Even though two- and four-chamber orthogonal views are available clinically, it should be kept in mind that because of angle dependence,9 the transverse radius measured by above-mentioned method does not represent the true transverse measurement of the flow convergence region.

Our present in vitro study demonstrated that the most appropriate velocity thresholds for applying the hemispheric flow convergence equation increased markedly and had a linear relationship with pressure gradients ranging from 40 to 200 mm Hg, and this relationship was independent of orifice size (Fig 7). For pressure gradients of 40 to 200 mm Hg, the most suitable velocity thresholds varied from 20 to 80 cm/s, which are within the obtainable limits of the currently used color Doppler system. Those findings are supported by previous qualitative observation of effects of pressure gradient on the proximal flow convergence region using the particle tracking technique where dramatic changes occurred in streamline angle as the pressure gradient varied.17 The present study provided an alternative way of correcting overestimation or underestimation of actual flow rate in view of the uncertainty of determining true color aliasing shape just by allowing selection of the velocity threshold using the pressure gradient yielding an isovelocity surface that fit for the hemispheric assumption to obtain better estimation of actual flow rate. For largest orifices and low-flow rates that were corresponding to pressure gradients of less than 40 mm Hg in the present study, the actual flow rate was still underestimated even using low velocity (velocity threshold) of 10 to 20 cm/s when a simple hemispheric flow convergence equation was used. Under this condition, the flow rate can be calculated with a hemispheric flow convergence equation by selecting the pressure gradient dependent velocity threshold at which the color aliasing pattern is best fit for the shape of isovelocity surface with ratio of calculated flow rate to actual flow rate to be
0.7 and can be corrected by dividing the calculated flow rate by 0.7 (Fig 6).

For the lower pressure gradient applications that account for the low flow rates and large orifice sizes such as mitral stenosis, some investigators have proposed a correction procedure to make the aliasing velocity pattern a better fit for the hemispheric flow convergence equation by decreasing the aliasing velocity as low as 10 to 20 cm/s. Our findings show that for such low aliasing velocities (even quite far from the orifice), surface geometry is still flattened because of slow axial acceleration when gradients are low. In addition, the use of very low velocity thresholds is limited by our findings that the velocity profile curves overlapped in the portion of low velocity range between the varying flow rates. Previous studies also demonstrated that the low aliasing velocity patterns were uncertain and subject to frame-to-frame variability, slow frame rates, and deformation by and fusion with neighboring flow phenomena.

Clinical Implications

The flow convergence method has already been used in patients with native and prosthetic mitral regurgitation,27-29 mitral stenosis,24,26 ventricular septal defect,2,30 and aortic coarctation. It therefore promises to become a useful method for quantifying flow rates. However, there are several problems such as instrument limitations (slow sampling speeds, angle dependence, and poor spatial resolution) and physiological factors that hinder the application of this simple flow hemispheric equation to clinical cases.17 Of the physiological factors, the most important one might be the flow pulsatility and varying hemodynamic conditions in the heart. Both the previous studies and our present one demonstrate that the shape of the isovelocity surface of flow convergence region changes with the flow rate and orifice size and the former will certainly compound application in pulsatile flow. Since the instantaneous regurgitant flow rate and regurgitant orifice size change and therefore the color aliasing pattern also changes for a given aliasing velocity with different hemodynamic conditions within a cardiac cycle, it is easy to understand why the overestimation or underestimation occurs frequently when the simple hemispheric flow convergence equation is used blindly to quantify the flow rate using only one aliasing velocity under different hemodynamic conditions within the whole cardiac cycle. Our present study provides an orifice size-independent relationship between the most suitable velocity threshold for applying the hemispheric flow convergence equation and pressure gradients ranging from 40 to 200 mm Hg that are estimated clinically and analogous to many of the clinical situations, certainly at peak systole, for example. It also offers a method for correcting the underestimation of flow rate calculated with hemispheric flow convergence equation at a low pressure gradient of less than 40 mm Hg by dividing the calculated flow rate by 0.7 (Fig 6B). One way of quantifying the flow rate in potential clinical application using the information provided in the present study is to adjust the aliasing velocity according to the estimated peak pressure gradient to calculate the maximal instantaneous regurgitant flow rate from the color Doppler image showing a maximal flow convergence region. The other way is to calculate the total regurgitant flow volume by integrating multiple instantaneous flow rate over time calculated with a simple hemispheric flow convergence equation from the color M-mode tracing of flow convergence region by adjusting the aliasing velocity to make the color aliasing pattern fit for the hemispheric shape according to the instantaneous pressure gradient obtained from the continuous wave Doppler tracing across the regurgitant orifice. This might be integrated with an automatic computer calculation software method and might provide the most accurate method for quantifying flow rates clinically.

Study Limitations

The present study to demonstrate the relationship between the most appropriate velocity threshold for applying the simple hemispheric flow convergence equation and the pressure gradients is limited to the planar circular orifice. However, this is not always the clinical situation. Although our findings showed this relationship was independent of orifice size, we did not determine the effects of orifice shape (triangle, oval, or rectangular) and the architecture of orifice plane on this relationship. Further studies are needed to determine the influence of orifice shape and architecture on the relationship between velocity threshold and pressure gradients and what kind of correction should be introduced under nonplanar and noncircular orifice conditions.

We did not provide a correction factor for the hemispheric flow convergence equation for hemodynamic condition with pressure gradients of more than 200 mm Hg under which conditions the flow rate calculated with hemispheric flow convergence equation would still overestimate the actual flow rate, even using very high velocity thresholds (more than 80 cm/s). However, hemodynamic condition with pressure gradients of more than 200 mm Hg is rare in clinical situations. Therefore, this limitation should not affect the clinical application of the method provided by the present study for selecting velocity thresholds according to the pressure gradient.

Conclusions

The shape of isovelocity surface changes with velocity threshold for a given orifice size and flow rate or pressure gradient and also changes with pressure gradients for a given velocity threshold. Our present study provides an orifice size-independent quantitative method for selecting the most suitable velocity threshold for applying the simple hemispheric flow convergence equation according to clinically estimated pressure gradients ranging from 40 to 200 mm Hg and offers a correcting factor to the hemispheric flow convergence equation when the pressure gradient is less than 40 mm Hg.

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