Impact of Finite Orifice Size on Proximal Flow Convergence
Implications for Doppler Quantification of Valvular Regurgitation

Leonardo Rodriguez, Joseph Anconina, Frank A. Flachskampf, Arthur E. Weyman, Robert A. Levine, and James D. Thomas

Analysis of velocity acceleration proximal to a regurgitant valve has been proposed as a method to quantify the regurgitant flow rate ($Q_r$). Previous work has assumed inviscid flow through an infinitesimal orifice, predicting hemispheric isovelocity shells, with calculated flow rate given by $Q_r=2\pi r_o^2 v_o$, where $v_o$ is user-selected velocity of interest and $r_o$ is the distance from that velocity to the orifice. To validate this approach more rigorously and investigate the impact of finite orifice size on the assumption of hemispheric symmetry, numerical and in vitro modeling was used. Finite-difference modeling demonstrated hemispheric shape for contours more than two orifice diameters from the orifice. More proximal than this (where the measured velocity $v_N$ exceeded 3% of the orifice velocity $v_o$), flow was progressively underestimated, with a proportional error $\Delta Q/Q_r$ nearly identical to the ratio of contour velocity to orifice velocity, $v_N/v_o$. For the in vitro investigations, flow rates from 4.3 to 150 cm$^3$/sec through 0.3 and 1.0 cm$^2$ circular orifices were imaged with color Doppler with aliasing velocities from 19 to 36 cm/sec. Overall, the calculated flow (assuming hemispheric symmetry) correlated well with the true flow, $Q_r=0.88Q_o-7.82$ ($r=0.945$, SD=12.2 cm$^3$/sec, $p=0.0001$, $n=48$), but progressively underestimated flow when the $v_N$ approached the orifice velocity $v_o$. Applying a correction factor predicted by the numerical modeling, $\Delta Q$ was improved from $-13.81\pm13.01$ cm$^3$/sec (mean$\pm$SD) to $+1.54\pm5.67$ cm$^3$/sec. These data indicate that flow can be accurately calculated using the hemispheric assumption as $Q_r=2\pi r_o^2 v_o$ when $v_N<v_o$. For larger $v_N$, flow is systematically underestimated, but a more accurate estimate may be obtained by multiplying $Q_r$ by $v_N/(v_N-v_o)$. These observations lend additional support for the clinical use of the proximal acceleration concept and suggest a simple correction factor to make a more accurate estimation of valvular regurgitation. (Circulation Research 1992;70:923–930)

**Key Words** • flow rate calculation • proximal flow convergence • fluid dynamics • finite-difference modeling • in vitro modeling • Doppler echocardiography

Quantification of flow rate through regurgitant valves remains a challenging goal of research and clinical cardiology. No current invasive or noninvasive method fully satisfies the needs of accuracy and applicability. Recently, analysis by color Doppler echocardiography of the velocity convergence zone proximal to a regurgitant orifice has been suggested as a superior method of regurgitant quantification.1–3 Hydrodynamic theory predicts that flow approaches a pointlike orifice in a flat plate as a series of concentric hemispheric shells of decreasing area and increasing velocity. By conservation of mass, flow rate should be calculable as velocity at any of these hemispheric iso-
tachs multiplied by the area of that shell. This is an attractive approach since flow measured in this way appears to be relatively independent of technical factors that alter jet appearance such as gain, transmission power, and transducer frequency.2,4 However, the utility of this method depends critically on the predictable acceleration of blood as it approaches the orifice.

Because regurgitant orifices are not infinitesimally small but rather finite in size, the assumption of hemispheric symmetry for calculating the area of the proximal isovelocity contour may not be appropriate near the orifice where the radius of the isovelocity shell approaches that of the orifice.

The purpose of this study was to validate in a more rigorous manner, using both numerical fluid dynamics analysis and in vitro modeling, the overall concept of proximal acceleration analysis and to examine the influence that orifice size, flow rate, and isotach velocity have on the calculation of flow rate.

**Methods**

**Theoretical Background**

The situation we are considering is that of a round orifice with effective area $A_o$ and radius $r_o$ with blood
flowing through it at velocity \( v_r \) to yield a flow rate \( Q_0 = A v_r \). By using color Doppler flow mapping, it is possible to define the distance \( r_0 \) from the orifice at which blood is moving at a specific velocity \( v_N \) by adjusting the red–blue transition on a color Doppler machine. If the orifice were an infinitesimal point in a flat plate (implying that \( v_r = v \)) and blood were inviscid, then the isovelocity contours would be hemispheric everywhere and the calculated flow, given by \( Q_0 = 2\pi r_0^2 v_N \), would accurately reflect \( Q_0 \) throughout the flow field.

In reality, however, it is evident that blood velocity does not rise infinitely but rather reaches a maximal velocity at the orifice of \( v_N \). Thus, if the aliasing velocity were set to \( v_r \), no red–blue transition would occur, \( r_0 \) would be zero, and the calculated flow would likewise be zero. We term this error in flow rate calculation \( \Delta Q = Q_0 - Q_{\text{true}} \). Thus, when \( v_N \) is set to \( v_r \), we see that all of the flow is missed in the standard formulation of \( 2\pi r_0^2 v_N \), and in this case \( \Delta Q \) would equal \( -Q_0 \). When \( v_N \) is less than \( v_r \), it is clear that this error \( \Delta Q \) is reduced, but it is unclear how rapidly it falls as \( v_N \) is reduced or if it ever becomes zero, implying perfect accuracy of \( Q_0 = 2\pi r_0^2 v_N \). Thus, we were interested in how \( \Delta Q \) varies with orifice area, flow rate, and Nyquist velocity for two reasons: 1) to define where the simple expression \( Q_0 = 2\pi r_0^2 v_N \) is a valid measurement of true flow and 2) to suggest possible modifications to this equation so that the principle of flow convergence might be used in the region near the orifice.

**Numerical Modeling**

To obtain numerical approximations of flow acceleration proximal to a finite orifice, we used a commercially available finite-difference program for computational fluid dynamics (Fluent, Inc., Hanover, N.H.). Calculations were based on a 2,704-node axisymmetric model (2.5-cm radius, 2.5-cm long, 1×1 mm cells), with steady flow through a round orifice surrounded by a planar wall. Three situations were studied: 1) 44 cm\(^3\)/sec flow through a 4-mm orifice (0.126-cm\(^2\) area), 2) 44 cm\(^3\)/sec, and 3) 176 cm\(^3\)/sec flow rates through an 8-mm orifice (0.503-cm\(^2\) area). Figure 1 displays the geometry and boundary conditions for the second of these simulations. Flow was generated by specifying axial (\( v_r \)) and radial (\( v_r \)) velocity components along boundaries A and B to approximate hemispheric symmetry at this distance. These velocity profiles were specified by fifth-order polynomials, the coefficients of which are shown in Table 1. No-slip conditions were specified along the solid wall while the orifice was treated as a simple outlet. Density and viscosity were physiological (1.05 g/cm\(^3\) and 3 centiPoise, respectively). The software solved the Navier-Stokes equations of fluid flow\(^5,6\) throughout the flow domain until acceptable convergence criteria were reached (relative residual error, <3×10\(^{-4}\)) after 500–1,000 iterations. From this converged solution, the velocity along the orifice axis was recorded as a function of distance from the orifice and flow rate was calculated as \( Q_0 = 2\pi r_0^2 v_N \). Because the true flow \( Q_0 \) remained constant, the underestimation of flow \( \Delta Q \) could be readily calculated. The proportional accuracy \( Q_{\text{true}}/Q_0 \) was then analyzed as a function of the ratio of the aliasing velocity to the orifice velocity \( v_N/v_r \) as described in “Statistical Analysis” below.

Because the results of the computer modeling were based on the assumption of perfect imaging machinery, we also studied the practical aspects of this problem in an in vitro model of valvular regurgitation.

**In Vitro Modeling**

We used an in vitro model that has been previously described.\(^7,8\) Briefly, this Plexiglas model consists of two chambers (6×6×57 cm and 6×8×57 cm) divided by a septum with a mount for different orifices. With the distal chamber sealed, air is pumped in, forcing blood into the proximal chamber. On computer command, the

![Figure 1. Graph showing the geometry used for finite-difference simulation. A 2.5×2.5 cm domain, axisymmetric about the x (dashed) axis, was specified with 1×1 mm cells. Velocity profiles along boundaries A and B were specified with fifth-order polynomials, coefficients of which are shown in Table 1. A given profile would be specified as \( v = a_0 x^2 + a_1 x^3 + a_2 x^4 + a_3 x^5 + a_4 x^6 \) using the axial and radial coordinates along the boundaries as shown in the figure.](image-url)

**Table 1. Coefficients Needed to Generate Velocity Profiles Along Boundaries A and B in Figure 1**

<table>
<thead>
<tr>
<th>Coefficients</th>
<th>Boundary A</th>
<th>Boundary B</th>
</tr>
</thead>
<tbody>
<tr>
<td>( a_0 )</td>
<td>1.12×10(^5)</td>
<td>3.96×10(^{-1})</td>
</tr>
<tr>
<td>( a_1 )</td>
<td>1.29×10(^{-2})</td>
<td>8.18×10(^{-2})</td>
</tr>
<tr>
<td>( a_2 )</td>
<td>3.24×10(^{-1})</td>
<td>3.81×10(^{-2})</td>
</tr>
<tr>
<td>( a_3 )</td>
<td>8.39×10(^{-2})</td>
<td>7.68×10(^{-3})</td>
</tr>
<tr>
<td>( a_4 )</td>
<td>8.47×10(^{-3})</td>
<td>3.10×10(^{-2})</td>
</tr>
<tr>
<td>( a_5 )</td>
<td>-3.80×10(^{-3})</td>
<td>7.48×10(^{-3})</td>
</tr>
</tbody>
</table>

For coordinates given in centimeters, this yields velocity in centimeters per second with a flow rate of 44 cm\(^3\)/sec through the orifice. (Coefficients to generate 176 cm\(^3\)/sec were four times as large.) The coefficient for the \( n \)th power of \( r \) (boundary A) or \( x \) (boundary B) is \( a_n \).
air is released, allowing the fluid to flow through the orifice by the force of gravity. We have previously shown that in this model, the pressure gradient follows a predictable parabolic decay, while velocity and flow decay linearly. In this way it is possible to predict the flow rate quite accurately simply by knowing the initial pressure gradient and the time interval from the onset of flow. The absolute flow rate was verified using an electromagnetic flow probe.

Circular orifices of 0.3 and 1 cm$^2$ were used with flow ranging from 4.3 to 150 cm$^3$/sec. A glycerol-water solution with physiological viscosity and density was used with cornstarch added for echo reflectivity.

**Echocardiographic examination.** Doppler flow images were obtained using a Hewlett-Packard 77090 echocardiograph and recorded onto 0.5-in. videotape. A transducer with a carrier frequency of 5 MHz was used.

Color M-mode data were recorded with the M line passing through the center of the orifice in a direction parallel to that of flow. By shifting the color baseline, four different aliasing velocities of 19, 24, 29, and 36 cm/sec were obtained for imaging. Images were analyzed off-line by using the software available in the ultrasound machine.

**Flow rate calculation.** Flow rate $Q_f$ was calculated assuming a hemispheric shape for the isovelocity contours. The proximal isovelocity surface area was calculated as $2\pi r_n^2$, where $r_n$ was the radius measured in the M-mode tracings as the distance from the first color alias to the orifice. This area was multiplied by the aliasing velocity $v_n$, to yield $Q_f$.

**Statistical Analysis**

For the numerical model and the in vitro experimentation, calculated flow rate $Q_f$ was compared with the actual orifice flow rate $Q_o$ by linear regression, and $\Delta Q$ was calculated as $Q_i - Q_o$. The overall determinants of $\Delta Q$ were explored in a multilinear model with $\Delta Q$ as the dependent variable and $Q_i$, $v_n$, and orifice area $A_o$ as independent variables.

It was anticipated that the proportional accuracy $Q_i/Q_f$ would increase with decreasing $v_n/v_n$, since these would represent velocity isochats at greater distances from the orifice. Accordingly, $Q_i/Q_f$ was compared with $v_n/v_n$ by linear regression, and this relation was then used to adjust $Q_i$ to obtain a corrected flow estimation. The agreement between actual flow and calculated and corrected flow was assessed using the method suggested by Bland and Altman, with $\Delta Q$ plotted against $Q_o$. Mean $\Delta Q$ and its standard deviation were used to test whether the correction scheme yielded an agreement with $Q_o$ that was a significant improvement over flow calculations based on the simple assumption of hemispheric symmetry, $2\pi r_n^2/v_n$.

**Results**

**Computer Modeling**

Figure 2 shows the velocity field predicted for idealized inviscid flow converging on a point orifice. The streamlines of flow as they approach the orifice along with the isovelocity contours are shown. These isovelocity contours have a hemispheric shape and converge on the orifice with a velocity proportional to $1/r_n^2$. In contrast, Figure 3 shows data from the finite-difference modeling of viscous flow toward a finite orifice, here with a flow rate of 176 cm$^3$/sec through an 8-mm orifice. Panel A shows the velocity magnitude isochats converging on the orifice. It is evident that these contours are nearly hemispheric in the periphery of the flow field but flatten out significantly in the vicinity of the orifice. Additionally, the velocity falls to zero at the wall. Panel B shows the streamlines of flow along which fluid particles move, approaching the orifice in essentially straight lines except in the immediate vicinity of the orifice. It should be noted that the appearance of isovelocity contours as displayed by color Doppler will differ from those shown in panel A since Doppler displays only one vector component of velocity. For example, panel C shows the axial component of velocity, equivalent to that shown by color Doppler imaging along the central axis through the orifice, while panel D displays the radial velocity component, as would be seen by Doppler imaging from the side of the orifice.

Figure 4 plots velocity against axial distance from the orifice for the four finite orifice simulations (with panel A including the high-velocity region near the orifice and panel B focusing on the low-velocity region away from the orifice). To allow data from the four situations to be shown together, the axial distance is shown relative to orifice diameter and the velocity is expressed relative to the mean orifice velocity. Displayed in this way, the velocity data are essentially identical despite the fourfold variation in flow rate and orifice area for the different simulations. This indicates that the local geometry is the primary determinant of the velocity distribution, with relatively little impact of fluid viscosity. Indeed, repeating the simulation with viscosity set to 100-fold greater than physiological conditions resulted in a velocity distribution varying no more than 12.5% from those shown in Figure 4. Also shown is the velocity distribution predicted for a point orifice (with velocity and distance normalized to unit velocity through a unit diameter orifice; see "Appendix A"). Far from the orifice, the velocity distributions are practically identical, but below about two orifice diameters, they progressively diverge, with the velocity near the finite orifice falling below the $1/r_n^2$ assumption.

**Accuracy of flow estimation.** From the axial velocity distributions of the finite-difference calculations, orifice...
Flow rate was estimated by \( Q = 2\pi r^2 v_w \) and compared with the actual flow rate \( Q_o \). This relation was likewise observed to be geometrically determined: when axial distance was expressed in orifice diameters, a predictable relation between \( Q \) and \( Q_o \) was seen. However, such a relation is of limited clinical use since, in practice, the orifice diameter is unknown. Alternatively, the accuracy of \( Q \) was studied relative to the aliasing velocity as normalized by orifice velocity (\( v_N/v_o \)). This is of potential practical utility since \( v_N \) is specified by the Doppler color baseline shift, while \( v_o \) can be estimated by continuous wave Doppler. Figure 5 demonstrates an almost linear inverse relation between \( Q/Q_o \) and \( v_N/v_o \). When the aliasing velocity is set very low, \( Q \) estimates \( Q_o \) quite well, with proportional underestimation as \( v_N \) approaches the orifice velocity. Another way of viewing these data is that the underestimation of flow by a given \( v_N \) is approximately given by \( v_N/A_o \), where \( A_o \) is the orifice area (see “Appendix B” for this derivation). Clearly then, use of low \( v_N \) should permit more accurate flow estimation. This prediction was dealt with empirically in the in vitro study.

**Flow Model**

Figure 6 (top panel) shows calculated flow assuming hemispheric symmetry (\( Q = 2\pi r^2 v_N \)) plotted against true orifice flow for the two orifices (0.3 and 1 cm²) with various Nyquist velocities. Overall, there was a good linear correlation between them, with \( Q_o = 0.88Q_N - 7.82 \) (\( r = 0.945, \ SD = 12.2 \ \text{cm}^3/\text{sec}, \ p < 0.0001, n = 48 \)). However, it can be seen that for higher Nyquist velocities there was a significant underestimation of true flow; for several low flow states, \( Q_o \) was zero.

The magnitude of this error \( \Delta Q \) was found to be significantly associated with larger aliasing velocities (univariate \( r = -0.406 \)), orifice area (\( r = -0.798 \)), and flow rate (\( r = -0.372 \)). Stepwise multilinear regression selected orifice area, flow rate, and aliasing velocity (in that order and all with \( p < 0.0001 \))) to predict flow error as

\[
\Delta Q = 26.99 - 37.56A_o - 0.897v_N + 0.098Q_o
\]

\( r = 0.929, \ SD = 4.98, \ p < 0.001 \)

Based on the results of computer simulation shown above, we analyzed the proportional accuracy \( Q_o/Q_N \) as

**Figure 3.** Diagrams showing axisymmetric flow toward a finite orifice, with velocity magnitude contours (panel A), streamlines of flow (panel B), and the axial (panel C) and radial (panel D) components of velocity. The domain shown has a radius and axial length of 2.5 cm, with 176 cm³/sec flow through an 8-mm-diameter orifice. Fluid density and viscosity are physiological. Contours in panels A, C, and D correspond (from distal to proximal) to velocities of 1, 2, 3, 5, 7, 10, 20, 30, 50, 70, 100, 200, and 300 cm³/sec. (Panel A has no 1 or 2 cm/sec contours, and panel D has no 300 cm/sec contour.)
Recently, more sophisticated approaches based on the conservation of jet momentum within the receiving chamber have been proposed,\textsuperscript{16,20–22} with promising preliminary results. These methods, however, are computationally intensive, and generally require a free, axisymmetric jet for their implementation.

By contrast, the velocity field proximal to the regurgitant orifice seems “better behaved” than the turbulent downstream jet. Flow here is governed primarily by the physical principle of conservation of mass, which leads to a predictable rise in velocity approaching the orifice. The proximal convergence method has the practical advantage of being independent of Doppler machine parameters and also of other associated regurgitant lesions. However, although the proximal convergence method has been empirically studied in the controlled environment of flow models,\textsuperscript{1,2} its theoretical basis is imperfect, a step necessary for the practical application of this method.

The initial description of the method by Recusani et al.\textsuperscript{1} was based on the assumption that fluid converges toward an orifice in concentric hemispheric shells. Although this is true for the ideal situation of a pointlike orifice, it certainly is not the case for a finite orifice. Utsumiya et al.\textsuperscript{2} for example, found that the assumption of a hemielliptic shape for the isovelocity contours better described the converging flow. The present study offers a theoretical and empirical explanation that helps to reconcile these two different approaches.

**Numerical Modeling**

Computer modeling has the advantage that a myriad of different factors can be studied by manipulating them one at a time. In the present study we used the finite-difference solution of the Navier-Stokes equations to study the impact of finite orifice size on the velocity fields proximal to a regurgitant orifice. Specifically, we examined the velocity profile along the central orifice axis, comparing it with the velocity expected for a point orifice. Our results showed that the velocity distribution far from the orifice (beyond two orifice diameters) is essentially identical for the finite and infinitesimal orifices, and here the assumption of hemispheric symmetry should hold. Closer to the orifice, however, the isovelocity contours flatten out, assuming a more hemielliptic shape, and calculating flow by $2\pi t \nu v_n$ leads to progressive underestimation, the magnitude of which is approximately given by the orifice area multiplied by the aliasing velocity of interest. Equivalently, as shown in Figure 5, the proportional magnitude of this underestimation is a very predictable, and almost linear, function of the ratio of aliasing velocity and orifice velocity. This suggests two conclusions: 1) use of the lowest practical aliasing limit should improve the accuracy of flow calculation, and 2) knowing the Nyquist velocity and orifice velocity (from continuous wave Doppler), one can correct the flow estimation by $Q_{cm} = Q_{c}v_n/(v_n - v_n)$. Note that the $v_n$ used here refers to the mean velocity across the orifice, which will be 20–30% less than the maximal velocity because of vena contracta effects, although the impact of this difference will be relatively minor for most practical $v_n$.
In Vitro Modeling

Because of the assumption of perfect imaging fidelity inherent to the computer models used, we also studied the proximal convergence method in an in vitro flow model using a clinical echocardiographic instrument. Our first observation was that without any correction at all, the assumption of hemispheric symmetry led to flow calculations that were quite good estimates of the true flow (Figures 6 and 7, top panels). Nevertheless, there was a slight but systematic underestimation of true flow, an error that was greater for high aliasing velocities. As predicted from the numerical modeling, the proportional magnitude of this underestimation was a linear function of \( v_n/v_o \), and the absolute discrepancy could be made up by adding \( v_n A_o \) to the calculated flow (Figures 6 and 7, bottom panels).

Practical Implications

The findings of this numerical in vitro study have a number of implications for the practical implementation of the proximal acceleration method in clinical and research cardiology. The first and most important is that for a round orifice in a planar surface, the assumption of hemispheric symmetry leads to an accurate calculation of flow rate when the aliasing velocity is low relative to the orifice velocity. For instance, when the orifice velocity is 5 m/sec (typical for mitral regurgitation) and the aliasing velocity is set to 20 cm/sec, the linear error model suggested here would predict that the true flow would be underestimated by 20/500, or 4%. Reducing the aliasing limit to 10 cm/sec would lower the error to 2%, while also making the aliasing distance easier to measure since it would be farther from the orifice.

Clinical application. Peak regurgitant flow rate may not be the only relevant parameter that can be obtained from analysis of the proximal flow field. Total regurgitant volume could be obtained by integrating calculated flow rates across the several color Doppler images obtained during the regurgitant flow period. Indeed, calculation of flow rate under pulsatile conditions has been shown to be at least as accurate as it is during steady flow. Additionally, an effective regurgitant orifice area (\( A_o \)) could be calculated by dividing the peak flow \( Q_o \) by the peak orifice velocity obtained from continuous wave Doppler. Assuming a relatively constant effective regurgitant orifice area, one could calculate regurgitant stroke volume simply by multiplying \( A_o \) by the time integral of orifice velocity.
distort the local isovelocity shells. We have shown, however, that when aliasing velocities are small relative to the orifice velocity, orifice shape does not greatly impact the accuracy of flow rate calculations. More important is the effect of nonplanar (e.g., funnel-shaped) global geometry surrounding the orifice, effectively truncating the hemispheric isotachs approaching the orifice. In this situation, the constant factor $2\pi$ must be reduced in proportion to the amount of a hemisphere subtended by the global geometry. Additionally, some regurgitant lesions have no quantifiable proximal convergence zone because of the limited resolution of contemporary Doppler flow mapping, implying a very low regurgitant flow rate. In such situations, assessment of the downstream jet by momentum analysis may be applicable. Finally, there may be difficulty precisely locating the regurgitant orifice, with small errors leading to significant errors in flow calculation. We have recently developed an algorithm that uses the full digital velocity map surrounding an orifice to search for the optimal orifice center, an approach that may allow flow rate to be calculated automatically.

**Conclusion**

The proximal convergence method allows for accurate calculation of flow rate; however, there is a consistent underestimation that may be minimized by lowering the aliasing velocity relative to the orifice velocity. For aliasing velocity less than 5% of the orifice velocity, this error appears to be minimal. For aliasing velocities above this, a simple formula is proposed that will correct for this underestimation. Of particular note is that this correction scheme can be implemented knowing only the contour and orifice velocities (the latter by continuous wave Doppler) without reference to specific orifice geometry. These observations should allow the proximal convergence method to be applied with greater confidence in the clinical estimation of valvar regurgitation.

**Appendix A: Normalizing Flow Through an Infinitesimal Orifice to Unit Velocity Flow Through a Unit Diameter Orifice**

The flow through an orifice of diameter $D_o$ is given by $Q_e = \pi D_o^2 v_o / 4$. Conversely, if the flow were converging on a point orifice, velocity and flow would be related by $Q_e = 2\pi r_N^2 v_N$. Equating these two expressions for $Q_e$ yields

$$\pi D_o^2 v_o / 4 = 2\pi r_N^2 v_N$$

Rearranging this so that $v_N / v_o$ is expressed as a function of $r_N / D_o$ yields

$$v_N / v_o = \sqrt{\frac{1}{8(r_N / D_o)^2}}$$

which is the equation plotted in Figure 4 for the point orifice.

**Appendix B: Underestimation of Flow**

As Figure 5 shows, the underestimation of flow with the hemispheric assumption is approximately proportional to $v_N / v_o$:

$$\frac{Q_o - Q_e}{Q_o} = \frac{v_N}{v_o}$$

But $Q_o = A_o v_o$, so we may rewrite $Q_o - Q_e$ as
\[ Q_0 - Q_e = \frac{v_N A_v}{v_o} = A_N v_N \]

Thus, the flow underestimation is equivalent to fluid at velocity \( v_N \) flowing through an orifice of area \( A_N \). Alternatively, we may write the first equation as

\[ 1 - \frac{Q_e}{Q_0} = \frac{v_N}{v_o} \]

or

\[ \frac{Q_e}{Q_0} = 1 - \frac{v_N}{v_o} \]

which is the equation displayed in Figure 5.

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**References**


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