Pergamon

Ultrasound in Med. & Biol., Vol. 21, No. 4, pp. 471–480, 1995

0301-5629/95 $9.50 + .00

0301-5629(94)00132-4

●*Original Contribution*

# DOPPLER VELOCITY RATIO MEASUREMENTS EVALUATED IN A PHANTOM MODEL OF MULTIPLE ARTERIAL DISEASE

LOUIS ALLARD, GUY CLOUTIER and LOUIS-GILLES DURAND
Laboratoire de Génie Biomédical, Institut de Recherches Cliniques de Montréal, Montréal, Québec, Canada

(*Received* 19 *April* 1994; *in final form* 19 *August* 1994)

**Abstract—The objective of this study was to evaluate *in vitro* the accuracy of the Doppler velocity ratio (VR) (intrastenotic velocity/prestenotic or poststenotic velocity) under different geometric conditions simulating the presence of multiple stenoses. A steady flow loop model was used to test the influence of the presence of a concentric obstruction of 84% area reduction positioned at a distance of 10, 20 and 30 tube diameters, either proximal or distal to the stenosis under study. The stenosis under evaluation was either concentric or eccentric and had a percentage of area reduction ranging from 20% to 91%. An ultrasound color Doppler system was used to perform both pulsed-wave (PW) Doppler and color-flow velocity measurements. VRs were computed by dividing the maximum velocity of the jet by the velocity at 6 and 10 diameters both proximal and distal to the stenosis under study. A strong correlation was obtained between VR computed using color flow and PW Doppler velocities ($r = 0.99$). Results indicated that using the prestenotic velocity as a reference velocity generally provided a more sensitive VR index to grade arterial stenosis than using the poststenotic velocity. From a curve fit model, the measured percentages of stenosis were calculated from the VR data and compared to the true percentages. The correlation coefficient, $r$, was 0.95. When the proximal and distal stenoses were at 10 diameters of the stenosis investigated, $r$ was 0.91, while it increased to 0.98 when the distance was 20 diameters or more. Although VR is theoretically not influenced by hemodynamic factors, we demonstrated that, in practice, the presence of multiple stenoses reduced its sensitivity. Volumetric flow measurements are suggested to obviate this limitation.**

*Key Words:* **Ultrasonography, Doppler ultrasound, Color Doppler, Pulsed Doppler, Arterial occlusive disease, Peripheral vascular disease, Hemodynamics, Blood flow velocity, *In vitro* model, Velocity ratio.**

## INTRODUCTION

The evaluation of lower limb arterial stenoses in the presence of multiple arterial obstructions remains an important clinical problem. Very few studies have reported observations of the effect of distributed stenoses on the performance of Doppler spectral analysis to grade arterial disease. This observation has recently incited us to perform an exhaustive retrospective study on 642 lower limb arterial segments to see if this factor could explain our classification errors (Allard et al. 1994). During this study, performed with an Ultramark 8 duplex scanner (Advanced Technology Laboratories), criteria based on semiquantitative spectral analysis, similar to those proposed by Jager et al. (1985), were used for grading arterial stenoses. The study showed that duplex scanning had a sensitivity and a specificity of 74% and 96%, respectively, to recognize severe stenoses (50% to 100% diameter reduction). The following technical, methodological and hemodynamic factors were considered as possible sources of the classification errors: (1) no color-flow imaging capability was used to locate the stenoses; (2) limitation of the angiographic exam as the gold standard; (3) technologist subjectivity in the evaluation of the Doppler spectral criteria; and (4) the influence of the presence of stenoses in adjacent segments. Our results demonstrated that the presence of multiple stenoses was an important limitation of duplex scanning since 62% of the errors between duplex scanning and angiography in the detection of severe stenoses involved a segment with at least one 50% to 100% diameter reduction lesion in adjacent segments. These results underlined the importance of performing further fundamental experimentation to provide a better understanding of the effect of multiple arterial disease on ultrasonic Doppler evaluation.

To minimize intrapatient and interpatient differences in peak systolic velocity measurements, a velocity ratio (VR) between two different recording sites

Address correspondence to: Louis Allard, Institut de Recherches Cliniques de Montréal, 110 Avenue des Pins Ouest, Montréal, Québec H2W 1R7, Canada.

(intrastenotic velocity/pre- or poststenotic velocity) was suggested as a diagnostic parameter to grade carotid stenosis (Keagy et al. 1982; Rittgers et al. 1983; Spencer and Reid 1979). Later, the same parameter was applied to femoropopliteal stenoses (Landwehr et al. 1991; Legemate et al. 1991; Leng et al. 1993; Ranke et al. 1992; Whyman et al. 1993). Theoretically, as explained in the next section, this ratio is directly related to changes in cross-sectional area and not to the amount of flow through the stenosis, since the VR is based on the assumption that the amount of blood flowing at one point of the segment is the same as that found at any other point (if there is no arterial side branch). Results obtained using a phantom model of the femoral artery (Whyman et al. 1993) indicated that the variability in measuring VR was generally low and correlated well with the degree of stenosis. In clinical evaluation of lower limb arteries, results showed that the method could distinguish severe from minor stenoses, although a more precise resolution of the degree of obstruction was less reliable (Leng et al. 1993). Although VR seems to be an excellent predictor of the degree of stenosis, further evaluation of this parameter is needed. The objective of this study consisted in evaluating, using an *in vitro* model, the diagnostic power of the VR under different hemodynamic conditions simulating the presence of multiple stenoses.

## THEORY

In steady flow, the quantity of fluid flowing per unit of time through a tube is, by definition, the product of the mean spatial velocity $\bar{v}$ by the cross-sectional area $A$. When fluid flows through an obstruction, the following theoretical relationship is present:

$$Q = \bar{v}_1 A_1 = \bar{v}_2 A_2 \qquad (1)$$

where $\bar{v}_1$ and $A_1$ represent, respectively, the mean spatial velocity and the cross-sectional area proximal (or distal) to the obstruction, and $\bar{v}_2$ and $A_2$ the same quantities measured within the obstruction. From eqn (1), a velocity ratio parameter VR can be defined as follows:

$$\mathrm{VR} = \bar{v}_2/\bar{v}_1 = A_1/\mathrm{A}_2. \qquad (2)$$

When using duplex scanning to evaluate arterial segments, an assumption should be made about the velocity profiles. Both profiles at the two measurement sites should be the same. When this condition is respected, a constant relationship between the maximum velocity and the mean spatial velocity exists so this later can be generally replaced by the Doppler maximum peak systolic velocity, PSV (Legemate et al. 1991; Leng et al. 1993; Ranke et al. 1992). Equation (2) then becomes:

$$\mathrm{VR} = \mathrm{PSV}_2/\mathrm{PSV}_1 = 100/(100 - \%\ \mathrm{stenosis}) \qquad (3)$$

or inversely,

$$\begin{aligned} \%\ \mathrm{stenosis} &= 100(1 - A_2/A_1) \\ &= 100\,(1 - 1/\mathrm{VR}) \qquad (4) \end{aligned}$$

where the percentage of stenosis is expressed in area reduction. VR based on the mean peak systolic velocity, the maximum and mean time-averaged velocities as well as the maximum velocity estimated by color Doppler imaging have also been evaluated *in vitro* (Landwehr et al. 1991) and provided similar results. When the arterial obstruction is located at the origin of the arterial segment, the prestenosic velocity is replaced by the poststenotic velocity recorded distally on the same arterial segment. Although eqn (4) has been used to grade arterial stenoses in *in vitro* models and clinical situations, no study has evaluated its accuracy under controlled hemodynamic conditions simulating the presence of multiple disease.

## MATERIALS AND METHODS

*Steady flow loop model*

All experiments were performed using a closed continuous flow model (Fig. 1). The flow conduit immersed horizontally in a water tank to allow ultrasound transmission was a long and straight silicone tube having an inner diameter of 4.8 mm and a wall thickness of 1.6 mm. The diameter and compliance of the silicone tube allowed good simulation of the lower limb arterial characteristics. A reservoir of 1 L allowed filling of the model with the test fluid. Instead of using gravity driven flow between two reservoirs, the continuous flow rate was controlled directly by the speed of a peristaltic pump (five-head lung–heart peristaltic pump, Sarns Inc.). Damping filters were used in the flow model to eliminate the oscillations produced by the pump. A constant mean flow rate of 100 mL/min was used in each series of measurements, which corresponded to a mean spatial velocity of 9.2 cm/s. This velocity value was chosen to be in the range of those encountered in human femoral arteries (Evans et al. 1989). Assuming the kinematic viscosity $v = 3.8 \times 10^{-6}$ m$^2$/s, the Reynolds number Re = 116 and the inlet length was less than 2 cm (Evans et al. 1989).

An ultrasound Doppler color-flow mapping system (Acuson Model 128XP/10, Mountain View, CA) with a linear array transducer (Model 7384) was used to perform Doppler velocity measurements. For B-mode imaging, the transducer operated at 7 MHz,

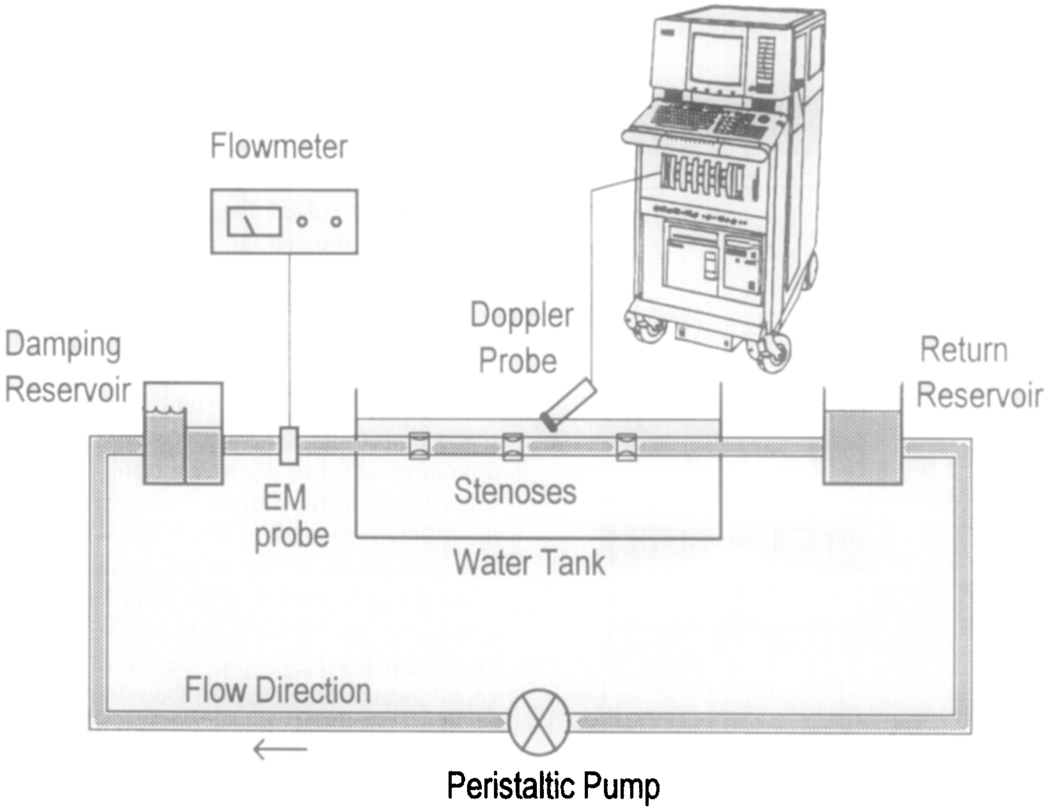


Fig. 1. Experimental block diagram of the steady flow model.

while in the color and pulsed-wave (PW) Doppler modes the frequency was 5 MHz. A cannulating-type flow probe (Model SF616) coupled to an electromagnetic flowmeter (Cliniflow II, Model FM701D, Carolina Medical Electronics) served to monitor, on a Hewlett-Packard digitizing oscilloscope (Model 54503A), the characteristics of the flow at the entrance of the tube. The test fluid was made of a volume of 33% glycerol|66% saline mixture. This mixture is known to have a viscosity similar to that of blood (Law et al. 1989). To obtain similar scattering properties as whole blood, cornstarch particles were suspended in the mixture at a concentration of about 3 g/L (Tamura et al. 1991; Valdes-Cruz et al. 1986). To get a homogeneous solution, a magnetic stirrer constantly mixed the glycerol–saline solution during the experiment.

Few methods have been proposed in the literature to create artificial stenoses. Most of them have the disadvantage of changing permanently the diameter of the tube. A different tube would have to be used with this option for each configuration of stenoses to be tested. To solve this problem, we created a series of Plexiglas strangling devices that temporally compressed the silicone tube to a given shape and diameter simulating concentric and eccentric arterial stenoses. The advantage of this method is that a stenosis can be created and removed quickly. Concentric (20%, 39%, 52%, 75% and 85% area reduction) as well as eccentric (28%, 47%, 59%, 79%, 86% and 91% area reduction) stenoses were realized. The calibration of each strangling device was performed independently of the ultrasound experiments by using a video camera and image processing software. The cross-sectional area of the stenoses created by the strangling devices was characterized by light transmission through a small silicone tube section, while the longitudinal shape was characterized by making a fiberglass cast of the stenoses (Fig. 2).

The strangling devices were positioned at three locations, S1, S2, and S3, along the tube at a distance of about 0.5 m from the inlet ensuring that the flow was parabolic before the first constriction. The distances between each stenosis varied between 5 to 15 cm (10 D to 30 D where D is the inner diameter of the tube). The stenosis at location S2 was the stenosis under study, while those at locations S1 and S3 were used to simulate proximal and distal constrictions (Fig. 3). As summarized in Table 1, the stenosis under evaluation was either concentric or eccentric and had a percentage of area reduction ranging from 20% to 91%, while the proximal and distal stenoses (when present) were a concentric stenosis of 84% area reduction. In all experiments, velocity measurements were respectively recorded: (1) at 6 D and 10 D before and after the

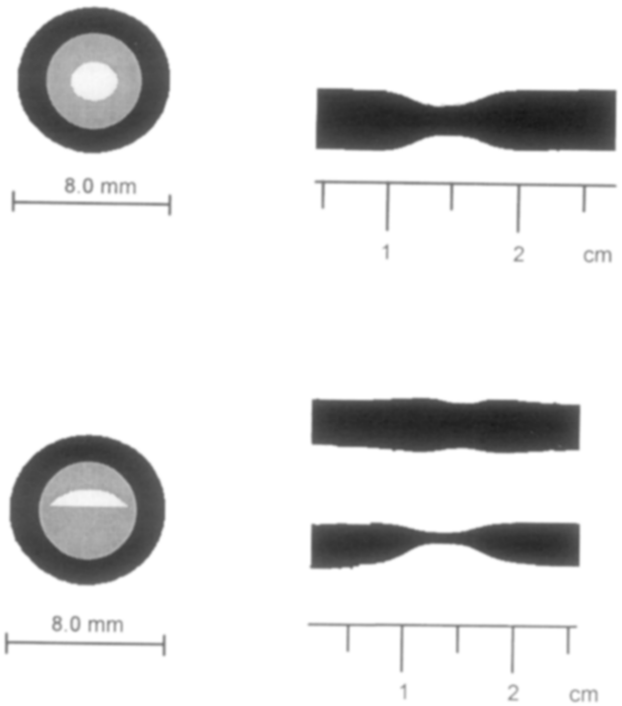


Fig. 2. Cross-sectional and longitudinal views of a 75% concentric (top panel) and 79% eccentric (bottom panel) stenoses in a 8.0-mm outer diameter (4.8-mm inner diameter) silicone tube.

stenosis S2 and (2) within the stenosis S2. The prestenotic velocity ($V_1$) was recorded at the center of the tube (Fig. 4), while the intrastenotic ($V_2$) and the poststenotic ($V_1$) velocities were recorded at the center of the tube only for concentric stenoses. For eccentric stenoses, the flow pattern was determined from the color Doppler flow image. In all cases, velocity measurements using both color Doppler and pulsed Doppler modes were performed as follows: the position where velocities were highest was determined from the color Doppler flow images, and this position was used to locate the pulsed Doppler sample volume within which was estimated the maximum time-averaged velocity. The length of the sample volume in the direction of the axial beam was set to 1.5 mm, which is smaller than the diameter of the tube. For the estimation of color Doppler velocities, the capture and velocity tag functions of the Acuson unit, which displayed in a contrasting color the highest mean velocities during a time interval set to 4 s, were used. The VR parameter was calculated from these measurements according to the following equation: VR = $V_2/V_1$. The transmitting power and gain setting of the Acuson was set to give the best representation of the Doppler signals. All other settings of the Acuson that could influence Doppler measurements (*e.g.*, pre- and postprocessing) were kept constant, including the Doppler angle between the Doppler ultrasound beam and the longitudinal axis of the tube which was set to 45°. This relatively low angle was chosen to reduce the effect of intrinsic spectral broadening (overestimation of the maximum velocity), which is more apparent with a linear array probe (Daigle et al. 1990).

*Data analysis*

The Pearson product moment correlation coefficient, $r$, was computed to measure the association be-

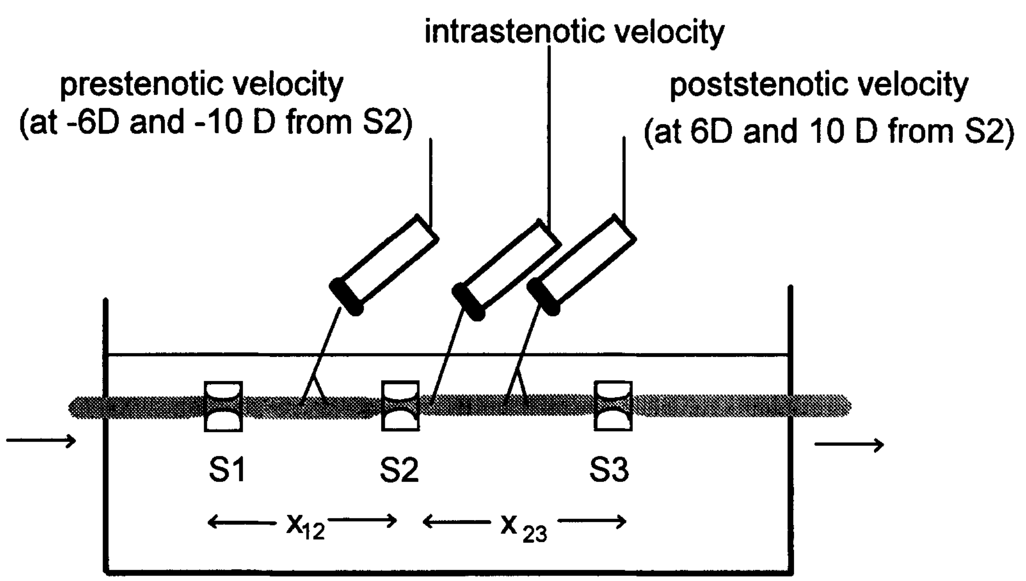


Fig. 3. Recording positions of the Doppler velocity measurements and position of the 84% area reduction concentric proximal (S1) and distal (S3) stenoses. The percentage of area reduction of the stenosis under study (S2) was varied from 20% to 91%. $X_{12}$ and $X_{23}$ represent the distances between the stenoses (10 D, 20 D and 30 D).

Table 1. Patterns of multiple stensoses in percentage of area reduction tested under a continuous flow of 100 mL/min. All distances are expressed in tube diameters.

| Stenosis S1 | Stenosis S2 | Stenosis S3 | Distance between (S1–S2) and (S2–S3) | Distance between S1 and prestenotic site | Distance between prestenotic site and S2 | Distance between poststenotic site and S3 | Distance between S2 and poststenotic site |
|---|---|---|---|---|---|---|---|
| 0% | 20% to 91% | 0% | ∞, ∞ | ∞, ∞ | 6, 10 | ∞, ∞ | 6, 10 |
| 84% | 20% to 91% | 0% | 10, ∞ | 4 | 6 | ∞, ∞ | 6, 10 |
| 84% | 20% to 91% | 0% | 20, ∞ | 10, 14 | 6, 10 | ∞, ∞ | 6, 10 |
| 84% | 20% to 91% | 0% | 30, ∞ | 20, 24 | 6, 10 | ∞, ∞ | 6, 10 |
| 0% | 20% to 91% | 84% | ∞, 10 | ∞, ∞ | 6, 10 | 4 | 6 |
| 0% | 20% to 91% | 84% | ∞, 20 | ∞, ∞ | 6, 10 | 10, 14 | 6, 10 |
| 0% | 20% to 91% | 84% | ∞, 30 | ∞, ∞ | 6, 10 | 20, 24 | 6, 10 |
| 84% | 20% to 91% | 84% | 10, 10 | 4 | 6 | 4 | 6 |

tween the VR obtained using PW Doppler velocity and color-flow imaging. A Student's $t$ test (paired data) was also performed to test the hypothesis that the mean difference between the paired VR data was zero. An analysis of variance (with significance level of 5%) was used to determine whether there were any differences between VR measurements and the percentages of stenosis. The theoretical relationship given by eqn (3) between VR and the known percentage of stenosis (S) was computed and a model of the form VR = $100/(a + b*S)$ was fit to the experimental data. This model was necessary because the experimental VR data did not perfectly match those predicted by eqn (3). The coefficient of determination, $R^2$, was computed to evaluate the goodness of the nonlinear curve fit. From this model, it was then possible to predict the percentage of a given stenosis from the VR. To estimate the accuracy of this prediction, the estimated value of the percentage of stenosis was computed from the experimental VR data and compared to the true value by linear regression analyses. The strength of the association between the two sets of values was measured by the Pearson product moment correlation coefficient. The effects of: (1) using the pre- or post-

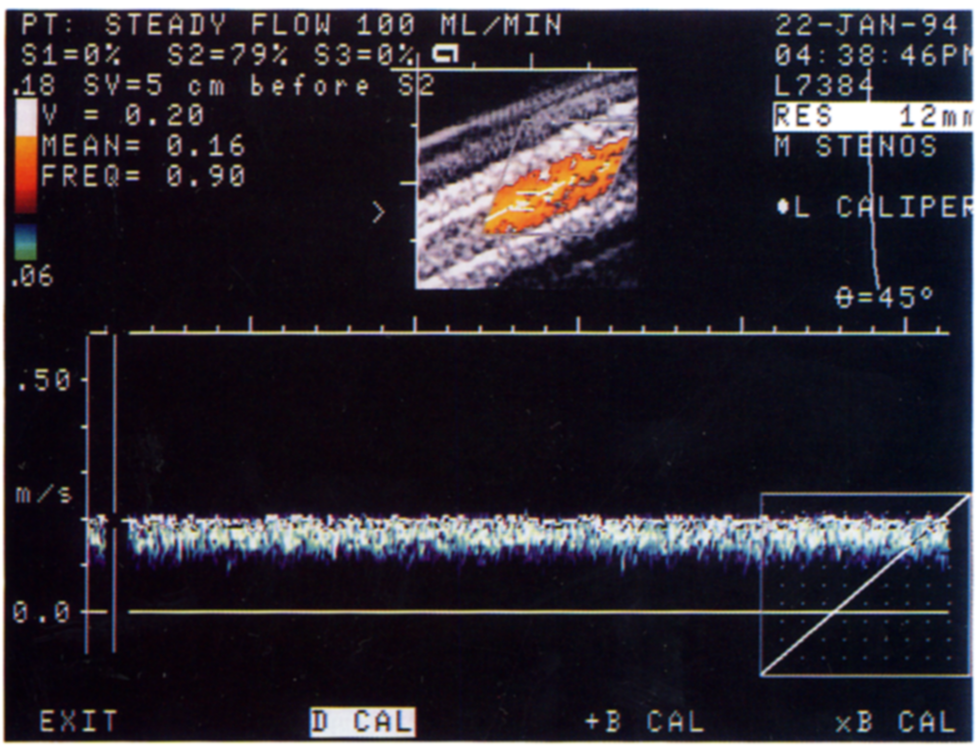


Fig. 4. Example of a Doppler velocity waveform recorded 10 D proximal to a 79% eccentric stenosis. No stenosis S1 or S3 was used in this situation.

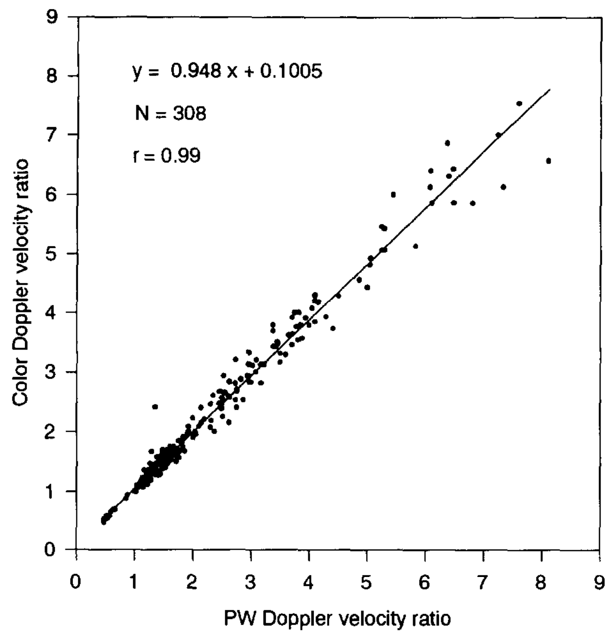


Fig. 5. Relationship between VR computed using PW Doppler and color Doppler velocity measurements for the conditions listed in Table 1.

stenotic velocity; (2) the presence of S1 and its distance to the prestenotic site as well as the presence of S3 and its distance to the poststenotic site and (3) the presence of both S1 and S3 on the discriminant power of VR were also tested.

## RESULTS

Under the conditions listed in Table 1, a total of 308 VR measurements were performed using PW and color Doppler. From all these measurements, the relationship between VR measured using both techniques was established (Fig. 5). The linear regression analysis indicates a high degree of correlation ($y = 0.948x + 0.1005$, $r = 0.99$), while the Student's $t$ test confirms the hypothesis that the mean difference between the paired VR data is not significant ($p > 0.1$). Because both techniques provide the same results, only those obtained using the PW Doppler technique are presented in the next paragraphs.

### *Effect of using pre- or poststenotic velocity*

We observed in some experiments that using the pre- or poststenotic velocity in the computation of VR did not necessarily provide the same result, especially for the assessment of severe stenoses (more than 75% area reduction). This observation was certainly due to one or many of the following factors that provided high velocity values in the poststenotic region: the presence of the jet, the velocity distribution within the sample volume and the presence of turbulent flow. To get the most sensitive VR index for each geometric condition listed in Table 1, the lowest velocity among those recorded before and after S2 was chosen. Based on this criterion, the prestenotic site was usually the best reference because the poststenotic site was chosen only when the stenosis S1 was located at 10 D proximal to S2. As a result, the PW Doppler VRs as a function of the degree of stenosis are summarized in Fig. 6, where the solid line represents the theoretical relationship between VR and the percentage of stenosis given by VR = 100/(100 − S). The experimental data are represented by circles (mean ± standard deviation). The analysis of variance indicated that there was a significant difference ($p < 0.05$) between select stenotic groups (20% to 59%, 75% to 79%, 85% to 86%, 91%). However, the VR for stenoses from 20% to 59% area reduction were not significantly different from one to another ($p > 0.05$). The curve fit model to the experimental data was VR = 100/(134 − 1.29*S) and the goodness of fit as measured by the coefficient of determination $R^2$ was 0.85. From this model, the measured percentage of stenosis was computed and compared to the true one (Fig. 7). Although this model overestimated the degree of stenosis over the range of 20% to 50%, a correlation coefficient of $r = 0.95$ was obtained between the predicted and true percentages of stenoses by linear regression analysis.

### *Effect of presence of S1 or S3*

Measurements involving only the presence of either S1 or S3 were analyzed to study more specifically

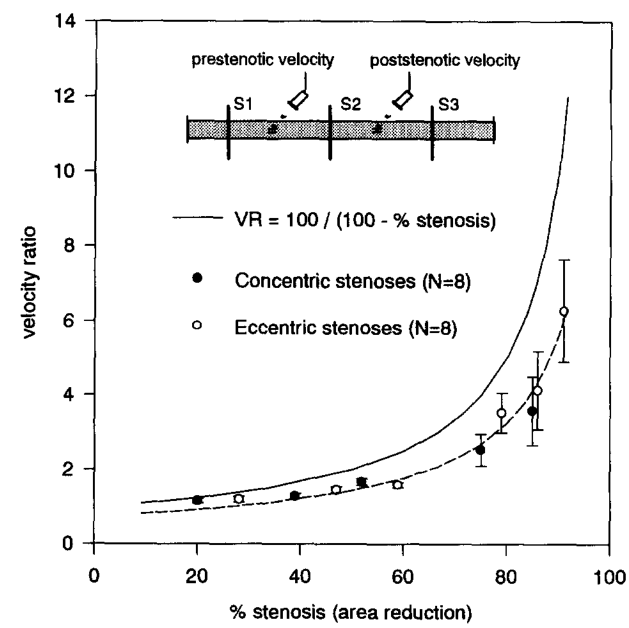


Fig. 6. Theoretical (solid line) and experimental (mean values ± SD) relationship between PW Doppler VR and the percentage of stenosis (S) obtained in steady flow for the eight geometric conditions listed in Table 1. The model VR = 100/(134-1.29*S) was fit to the experimental data (coefficient of determination $R^2$ = 0.85).

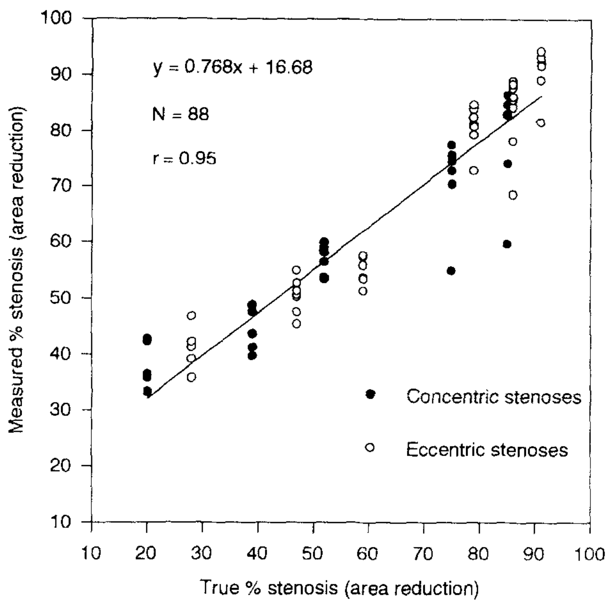


Fig. 7. Relationship between the measured percentage of stenosis calculated from the PW Doppler VR and the true percentage of stenosis.

the individual effect of the presence of each stenosis. Figure 8 shows the relationship between the VR and the percentage of stenosis as a function of the distance between the proximal stenosis S1 and the prestenotic site. The distance between S1 and S2 was 20 D or 30 D when $x(\mathrm{S1}, V_{ref})$ was $\geq$10 D and 10 D when $x(\mathrm{S1}, V_{ref}) = 4$ D. It can be seen in this figure that the VR curve is relatively close to the theoretical one when the distance between S1 and the prestenotic velocity is equal or greater than 10 D. However, for a distance of 4 D, the discriminant power of VR is severely reduced. Figure 9 shows the relationship between the VR and the percentage of stenosis as a function of the distance between the distal stenosis S3 and the poststenotic site. The distance between S2 and S3 was 20 D or 30 D when $x(V_{ref}, \mathrm{S3})$ was $\geq$10 D and 10 D when $x(V_{ref}, \mathrm{S3}) = 4$ D. It can be seen that the distance $x(V_{ref}, \mathrm{S3})$ has no significant effect on VR. Consequently, VR computed from the poststenotic velocity was not affected by the presence of the distal stenosis. However, both curves were less discriminant than the theoretical curve.

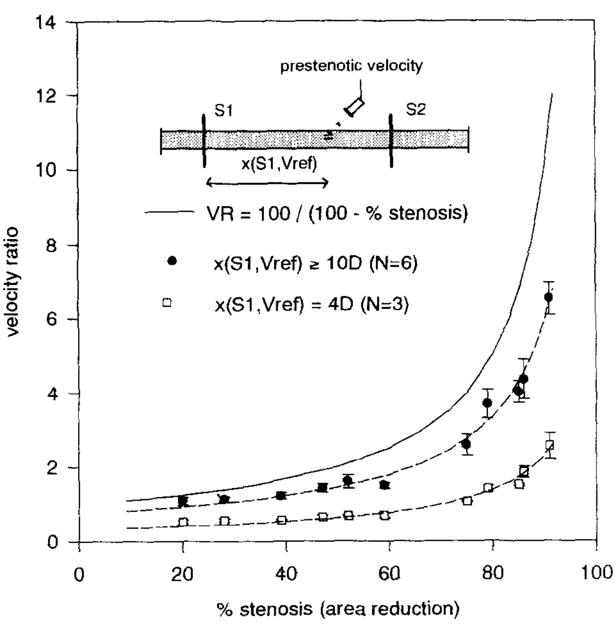


Fig. 8. Relationship between the PW Doppler VR and the percentage of stenosis as a function of the distance between the proximal stenosis S1 and the prestenotic site. The distance between S1 and S2 was 20 D or 30 D when $x(\mathrm{S1}, V_{ref}) \geq 10$ D and 10 D when $x(\mathrm{S1}, V_{ref}) = 4$ D.

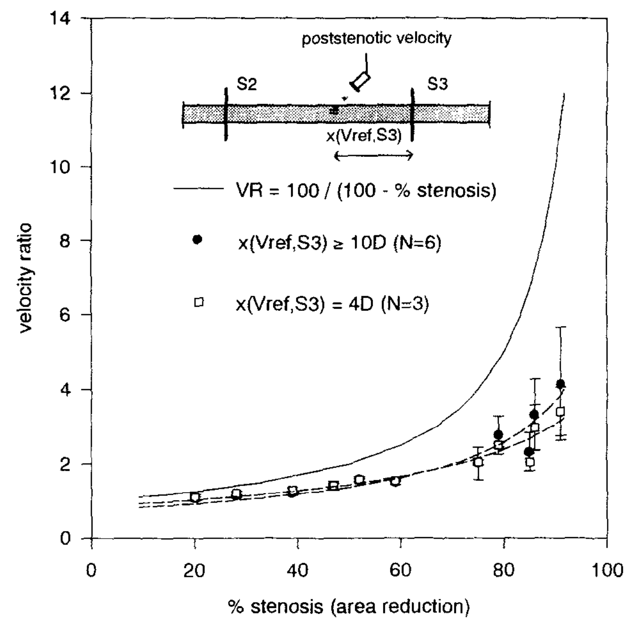


Fig. 9. Relationship between the PW Doppler VR and the percentage of stenosis as a function of the distance between the distal stenosis S3 and the poststenotic site. The distance between S2 and S3 was 20 D or 30 D when $x(V_{ref}, \mathrm{S3}) \geq$ 10 D and 10 D when $x(V_{ref}, \mathrm{S3}) = 4$ D.

*Effect of presence of both S1 and S3*

As a next step, the effect of the presence of two severe stenoses S1 and S3 (84% area reduction) located, respectively, at 10 D proximally and distally to the stenosis under study was investigated. These results were compared to those obtained when the stenosis S1 or S3 was located at 20 D or more of S2. For these situations, the prestenotic velocity was used to compute the VR and results indicated that the VR curve was relatively close to the theoretical one, as illustrated in Fig. 10. The curve fit model to these experimental data was VR = 100/(127 − 1.22*S) and the coeffi-

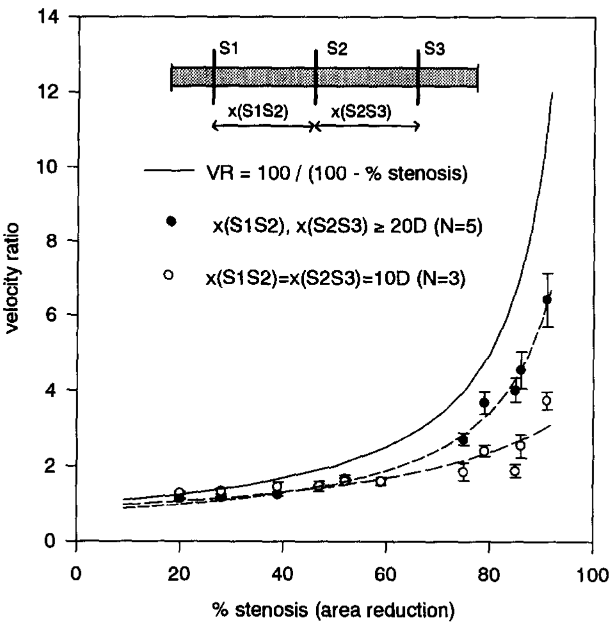


Fig. 10. Relationship between the PW Doppler VR and the percentage of stenosis as a function of the distance between the stenoses S1, S2 and S3.

cient of determination $R^2$ was 0.96. When the distances between S1–S2 and S2–S3 were equal to 10 D (which corresponds to a distance between S1 and the prestenotic velocity of 4 D), the poststenotic velocity was preferred to the prestenotic one in the computation of VR because the latter was influenced by the jet produced by S1. Results indicated that the sensitivity of VR was poor in this situation even if the poststenotic site was used. The curve fit model to the experimental data was VR = 100/(111 − 0.86*S) and the coefficient of determination $R^2$ was 0.76. From these two models, the accuracy of predicting the percentage of stenosis was evaluated by comparing the measured and true percentages of stenosis. Results (Table 2) indicated a correlation coefficient ($r$) of 0.91 and 0.98 when the distance was 10 D and 20 D (or more), respectively.

Although we demonstrated that the presence of severe proximal and distal stenoses close to S2 reduced the sensitivity of the VR (Fig. 10), no change in the intrastenotic velocity was observed. It was found that the prestenotic reference site (Fig. 8) was affected by the presence of a proximal stenosis S1. For instance, a prestenotic velocity increase of 100% was observed when the distance between S1 and the prestenotic site was reduced from 10 D (or more) to 4 D. Although the poststenotic site was not affected by the presence of the distal stenosis S3 (Fig. 9), we observed that the presence of a severe S2 could increase the poststenotic velocity by 100% and, consequently, could reduce the discriminant power of VR.

## DISCUSSION

The design of a Doppler flow phantom that simulates the geometric and flow pattern of human blood vessels is obviously a difficult task. A simplified approach for experimentally studying the effect of multiple stenoses as found in lower limb arteries was chosen. A pulsatile flow model was not used, although it would have been more realistic, because preliminary experiments showed that it was very difficult to reproduce the same waveform shape for different patterns of stenoses. While the mean flow rate was maintained constant, the peak forward and peak reverse velocities changed significantly as the number and the degree of stenosis was increased. Consequently, the comparison of the different results using a pulsatile flow model was not possible. By using a steady flow model with a straight silicone tube and no side branches, more easily interpretable results were obtained.

Our results demonstrated that VRs computed in steady flow using PW Doppler and color Doppler correlated well with each other. *In vitro* investigations performed by Landwehr et al. (1991) using a pulsatile flow model indicated similar results for the evaluation of a single stenosis. This confirms the equivalence of both techniques in determining the degree of stenosis. As predicted by the theory, we observed in Figs. 6 and 7 that the relationship between the VR and the degree of stenosis was independent of the shape (concentric or eccentric) of the stenosis. A similar result was observed by Whyman et al. (1993) in a pulsatile flow

Table 2. Regression analysis between the measured percentage of stenosis (S) and the true percentage of stenosis. From the curve fit model VR = 100/($a$ + $b$*S), and the experimental VR data, the measured percentages of stenosis were calculated.

| Distance (in tube diameters) | | | | |
|---|---|---|---|---|
| S1–S2 | S2–S3 | Velocity reference | Regression line | $r$ |
| 10, 20 and 30 | 10, 20 and 30 | pre- or poststenotic | $y = 0.768x + 16.68$ | 0.95 |
| 10 | 10 | poststenotic | $y = 0.711x + 19.96$ | 0.91 |
| 20 and 30 | 20 and 30 | prestenotic | $y = 0.892x + 7.56$ | 0.98 |

model. Another interesting observation is that the percentage of stenosis calculated from eqn (4) underestimated the true percentage of stenosis, especially for severe constrictions. This underestimation can be explained by the assumption made in the use of eqns (3) and (4) about a predictable (or at least constant) relationship between the maximal velocity within the Doppler sample volume and the mean velocity across the tube. Theoretically, it is known that the maximum velocity at the center of a circular and rigid tube is twice the mean spatial velocity in laminar parabolic flow while, in turbulent flow (flat profile), it is only about 20% higher than the mean spatial velocity (Evans et al. 1989). The velocity profile within a severe stenosis may be flattened compared to that of the prestenotic or poststenotic region. The VR underestimation by a factor of approximately two, shown in Fig. 6 for stenoses greater than 85%, might thus be explained by the variation of the velocity profile as a function of the position along the tube. This explanation is also applicable to the evaluation of VR by color Doppler imaging. For this reason, it was more appropriate to calculate the percentage of stenosis from a curve fit model instead of using eqn (4).

Landwehr et al. (1991) obtained in a pulsatile flow model a very high correlation between the true values of the percentage of stenosis and those obtained from eqn (3), while Whyman et al. (1993) observed an underestimation in similar conditions. In a clinical evaluation of the peripheral arteries using the VR, Ranke et al. (1992) also observed an underestimation of the percentage of stenosis. When using a pulsatile flow model, the flow profile is a function of the flow waveform produced by the pump. In *in vitro* models as well as in *in vivo* conditions, it is difficult to predict a relationship between the mean spatial velocity and the maximum velocity as a function of time. This consideration may explain the different results reported in the literature. With the recent technological improvement in color-flow imaging that allows the determination of the velocity profile and volumetric flow before and within a stenosis, it is anticipated that these conflicting results may be eliminated since VR based on mean spatial velocities will be available.

Based on our previous experience with clinical evaluation of lower limb arteries with duplex scanning (Allard et al. 1991, 1994), it was anticipated that the presence of a severe stenosis in adjacent segments would have a significant effect on the assessment of the disease, even though the distance between the multiple stenoses was relatively important (more than 10 D). From the results of the present study, it is suggested that some classification errors may be explained by the influence of stenoses in adjacent segments. However, we are aware that the conclusions drawn for our study may not be applied directly to clinical situations because some important *in vivo* factors were not included in our flow model. These factors were the flow rate, flow waveform and the peripheral resistance that were all kept constant independently of the degree and number of stenoses involved as well as the absence of any collateral branch and any entrance effect. It is possible that, if these factors were taken into account, the effect of a stenosis may be observed over a more significant distance.

For these reasons, we believe that VR computed from the peak systolic velocities would have not solved all the classification errors reported in Allard et al. (1994). Although the VR is theoretically not influenced by hemodynamic factors such as the velocity profile if volumetric flow is used, some limitations pointed out in the present study may have to be considered. For example, our results have suggested that the presence of more than one stenosis within the same segment may reduce the sensitivity of the VR if the maximum velocity within the tube is used. We also demonstrated the importance of the site of the velocity reference for the computation of VR. This can have important consequences in the clinical evaluation of lower limb arterial segments, since it is not always possible to have a site free of any flow perturbations or accelerations. Volumetric blood flow measurement, which is an emerging and promising application of Doppler ultrasound, could be useful to minimize the influence of these factors.

## SUMMARY

We demonstrated that velocity measurements using PW Doppler and color Doppler provide very similar VR values, although both techniques are based on two different signal processing algorithms. Our results confirmed that the presence of a proximal and/or distal stenosis close to the one under study has an important effect on the VR. It was also demonstrated that the prestenotic site generally provided a more discriminant VR index. However, the best recording site for the velocity reference depends on the distance between the stenoses and the specific degree of stenosis. Finally, we believe that volumetric flow measurements need to be performed to minimize the influence of stenoses close to the one investigated.

*Acknowledgement*—This research was supported by grants from the Medical Research Council of Canada (#MA-11740 and #MA-10030) and the Heart and Stroke Foundations of Canada. The authors acknowledge Mr. Richard Cimon for providing technical assistance and for designing the strangling device used to create the stenoses.

## REFERENCES

Allard, L.; Langlois, Y. E.; Durand, L.-G.; Roederer, G. O.; Beaudoin, M.; Cloutier, G.; Roy, P.; Robillard, P. Computer analysis and pattern recognition of Doppler blood flow spectra for disease classification in the lower limb arteries. Ultrasound Med. Biol. 17:211–223; 1991.

Allard, L.; Cloutier, G.; Durand, L.-G.; Roederer, G. O.; Langlois, Y. E. Limitations of ultrasonic duplex scanning for diagnosing lower limb arterial stenoses in the presence of adjacent segment disease. J. Vasc. Surg. 19:650–657; 1994.

Daigle, R. J.; Stavros, A. T.; Lee, R. M. Overestimation of velocity and frequency values by multielement linear array Dopplers. J. Vasc. Tech. 14:206–213; 1990.

Evans, D. H.; McDicken, W. N.; Skidmore, R.; Woodcock, J. P. Doppler ultrasound, physics, instrumentation, and clinical applications. New York: Wiley; 1989.

Jager, K. A.; Phillips, D. J.; Martin, R. L.; Hanson, C.; Roederer, G. O.; Langlois, Y. E.; Ricketts, H. J.; Strandness, D. E., Jr. Noninvasive mapping of lower limb arterial lesions. Ultrasound Med. Biol. 11:515–521; 1985.

Keagy, B. A.; Pharr, W. F.; Thomas, D.; Bowles, D. E. Evaluation of the peak frequency ratio (PFR) measurement in the detection of internal carotid artery stenosis. J. Clin. Ultrasound 10:109–112; 1982.

Landwehr, P.; Schindler, R.; Heinrich, U.; Dolken, W.; Krahe, T.; Lackner, K. Quantification of vascular stenosis with color Doppler flow imaging: In vitro investigations. Radiology 178:701–704; 1991.

Law, Y. F.; Johnston, K. W.; Routh, H. F.; Cobbold, R. S. C. On the design and evaluation of a steady flow model for Doppler ultrasound studies. Ultrasound Med. Biol. 15:505–516; 1989.

Legemate, D. A.; Teeuwen, C.; Hoeneveld, H.; Ackerstaff, R. G. A.; Eikelboom, B. C. Spectral analysis criteria in Duplex scanning of aortoiliac and femoropopliteal arterial disease. Ultrasound Med. Biol. 17:769–776; 1991.

Leng, G. C.; Whyman, M. R.; Donnan, P. T.; Ruckley, C. V.; Gillespie, I.; Fowkes, G. R.; Allan, P. L. Accuracy and reproducibility of duplex ultrasonography in grading femoropopliteal stenoses. J. Vasc. Surg. 17:510–517; 1993.

Ranke, C.; Creutzig, A.; Alexander, K. Duplex scanning of the peripheral arteries: Correlation of the peak velocity ratio with angiographic diameter reduction. Ultrasound Med. Biol. 18:433–440; 1992.

Rittgers, S. E.; Thornhill, B. M.; Barnes, R. W. Quantitative analysis of carotid artery Doppler spectral waveforms: Diagnostic value of parameters. Ultrasound Med. Biol. 9:255–264; 1983.

Spencer, M. P.; Reid, J. M. Quantification of carotid stenosis with continuous-wave (C-W) Doppler ultrasound. Stroke 10:326–330; 1979.

Tamura, T.; Cobbold, R. S. C.; Johnston, K. W. Quantitative study of steady flow using color Doppler ultrasound. Ultrasound Med. Biol. 17:595–605; 1991.

Valdes-Cruz, L. M.; Yoganathan, A. P.; Tamura, T.; Tomizuka, F.; Woo, Y. R.; Sahn, D. J. Studies in vitro of the relationship between ultrasound and laser Doppler velocimetry and applicability of the simplified Bernoulli relationship. Circulation 73:300–308; 1986.

Whyman, M. R.; Hoskins, P. R.; Leng, G. C.; Allan, P. L.; Donnan, P. T.; Ruckley, C. V.; Fowkes, F. G. R. Accuracy and reproducibility of duplex ultrasound imaging in a phantom model of femoral artery stenosis. J. Vasc. Surg. 17:524–530; 1993.