Defining the limitations of measurements from Doppler spectral recordings

Peter A. J. Bascom, PhD, K. Wayne Johnston, MD, FRCS(C), Richard S. C. Cobbold, PhD, and **Matadial Ojha, PhD,** *Toronto, Ontario, Canada*

Purpose: **The purpose of this study was to determine whether Doppler measurements of peak velocity and four other quantitative measures of spectral shape are affected significantly by the site of the Doppler recording in relation to the location of the maximum stenosis.**

Method: **Continuous-wave and pulsed Doppler recordings were made distal to a 70% (area reduction or 45% diameter reduction) asymmetric stenosis in an in vitro flow model under steady and pulsatile flow conditions. Recordings were taken at six different locations proximal and distal to the stenosis. A photochromic dye technique was used to visualize the actual flow field in the model.**

Results: **Distal to the stenosis, the flow visualization results demonstrated a strong radial and axial variation of the velocity field and thus explained why the Doppler measurements of peak frequency and spectral broadening were strongly dependent on the recording site. The peak frequency was maximum within the throat of the stenosis and returned to the prestenotic value five tube diameters distal to the stenosis. Other measurements of spectral broadening and spectral shape varied greatly depending on the location of the recording site in the poststenotic region. Higher order spectral moments such as the coefficient of kurtosis were found to exhibit large temporal variability, which makes them inappropriate as diagnostic indicators.**

Conclusions: **Because of the complex nature of the poststenotic flow field, these results clearly demonstrate that no single Doppler measurement can accurately quantify the severity of a stenosis. Of the Doppler measurements only peak velocity is related to the severity of stenosis. Reproducible peak velocity measurements are obtained only if the Doppler sample volume is positioned at or very near the throat of the stenosis and at an appropriate radial site that may not necessarily be at the center of the vessel. (J Vasc Surg 1996;24:34-45.)**

Doppler ultrasound methods are widely used and have an established role in the noninvasive diagnosis of arterial occlusive disease. Although the B-scan image provides important information on the presence, severity, and architecture of an atherosclerotic plaque, it is Doppler spectral recordings that can provide hemodynamic information that may be crucial for the assessment of the severity of a lesion. With stenoses in straight rigid tubes we have demonstrated that a direct relationship exists between the peak Doppler frequency and the severity of the stenosis[1-3] and that flow disturbances that occur beyond a stenosis can be quantified by Doppler ultrasonography.[4] In a study with the photochromic dye tracer method, Ojha et al.[3] concluded that center-line Doppler measurements can accurately record the peak velocity and thereby determine the severity of a stenosis; however, we also speculated that the assessment of the severity of a stenosis from the extent of spectral broadening from center-line Doppler recordings can have limitations in certain circumstances. Previous studies have relied heavily on the use of steady flow and pulsatile flow models to evaluate the potential applications of the Doppler technique and to improve our understanding of the nature of the Doppler signal[5,6]; however, no studies have reported a direct comparison between Doppler recordings and the flow field visualized by a high-resolution technique.

From the Institute of Biomedical Engineering and the Department of Surgery, University of Toronto.

Presented at the Twenty-second Annual Meeting of the New England Society for Vascular Surgery, Montreal, Quebec, Canada, Sept. 28-29, 1995.

Reprint requests: K. W. Johnston, MD, The Toronto Hospital, 200 Elizabeth St., 5 Eaton Centre, Room 309, Toronto, Ontario M5G 2C4.

0741-5214/96/$5.00 + 0 **24/6/71645**

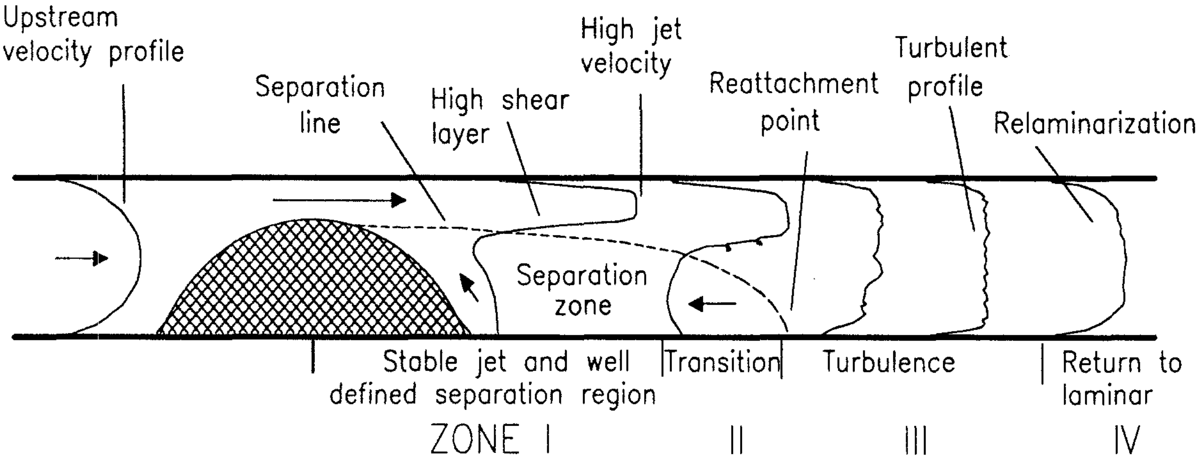


Fig. 1. Illustration of four flow zones and their corresponding flow distribution distal to moderately severe asymmetric stenosis. Length of each zone depends on Reynolds number and precise stenosis geometry.

The aim of this study was to determine whether Doppler measurements of peak velocity and other quantitative measurements of spectral shape are affected significantly by the site of the Doppler recording in relation to the location of the maximum occlusion. This objective was accomplished with a photochromic dye tracer method used to visualize the flow field in the distal region of a 70% asymmetric stenosis in a straight tube. In addition, Doppler recordings were made at different locations proximal and distal to the stenosis of a model of similar geometry and compared with the flow visualization data.

BACKGROUND

At any instance in time the Doppler spectrum is related to the instantaneous velocity distribution of blood flowing through the ultrasound beam. Currently, the diagnostic parameters that are the most useful are those derived from Doppler spectral recordings such as the peak frequency and the subjective grading of the severity of the stenosis from spectral broadening.[7] Unfortunately, our studies have demonstrated that spectral broadening can also arise from factors that are intrinsic to the measurement system and that this intrinsic spectral broadening[8] can have a significant effect on the received pulsed Doppler spectrum when a small sample volume is placed at the center of the vessel.[9] However, in vitro studies have also shown that quantification of pulsed Doppler spectra may still be feasible, if the sample volume is appropriately placed.[10]

Ojha et al.[3,11] have shown that the flow field in the region of a moderately severe axisymmetric stenosis can be qualitatively described by dividing the distal region into four zones as illustrated in Fig. 1. Just beyond the stenosis, a stable jet and a well-defined flow separation zone (zone I) are present. Further downstream, in zone II, the vortexes generated within the shear layer of the jet begin to break down, and the flow exhibits a transition to turbulence. Distal to this region, in zone III, fully turbulent flow develops, and this persists into zone IV, where relaminarization occurs and the velocity profile begins to return to the prestenotic state. The dimensions of each of these zones and their detailed properties will obviously depend on the Reynolds number *(Re)* and on the precise stenosis geometry. Therefore to determine the potential role of Doppler measurements, Doppler recordings were made in a model of a 70% (by area) asymmetric stenosis, and flow visualization experiments were also performed with the photochromic dye technique[12] in a model with similar geometry and flow conditions.

EXPERIMENTAL METHODS

Doppler studies

Flow model. Details of the experimental model were presented previously.[4,6] An asymmetric stenosis with a 70% area reduction was fabricated by heating an acrylic tube (outer diameter, 8 mm, inner diameter, 4.6 mm) to the softening point and then pressing the end of a rod, rounded to a diameter of 9.5 mm, into the tube to a specific depth.[4] After the experiments were completed, the exact shape of the stenosis was determined from cross-sectional slices.

Outdated human red blood cells (RBCs) were washed and resuspended in saline solution to obtain a

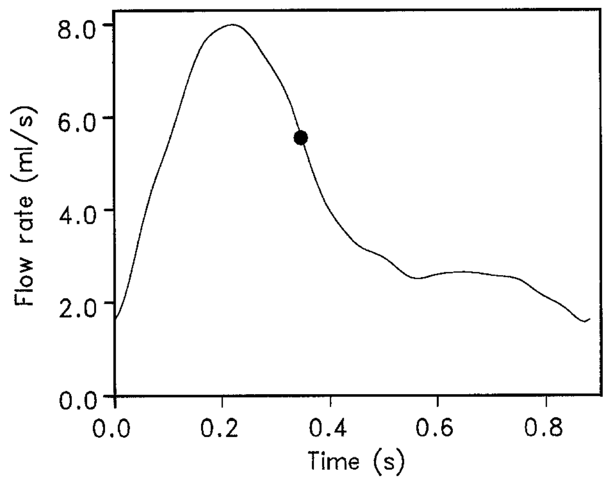


Fig. 2. Pulsatile flow waveform used in Doppler study. Nearly identical flow waveform was used in photochromic studies. Waveform has period of 0.88 sec and fundamental Womersley parameter of 3.8 and is similar to that found in common carotid artery. *Solid dot* represents time in flow cycle when results shown in Fig. 4, *B* were obtained.

hematocrit of 41% (viscosity of 2.6 cP). In addition, for our continuous-wave Doppler steady flow experiments, measurements were also performed with a 4% suspension of gluteraldehyde-fixed RBCs (viscosity of 0.9cP). An important reason for using a 4% hematocrit suspension was that a much higher Reynolds number could be achieved without changing the flow rate. Reynolds numbers of 545 and 1410 were used for the 41% and 4% suspensions, respectively. Air bubbles were eliminated from the flow system, because their presence can profoundly affect the Doppler recordings. To accomplish this elimination all flow connections were carefully made, and the system was run for a minimum of 1 hour before any measurements were taken.

A computer-controlled pulsatile pump was used[5] to generate pulsatile flow waveform similar to that seen in the carotid artery. An electromagnetic flowmeter enabled the instantaneous measurement of the volumetric flow waveform, and a sample of these results are shown in Fig. 2. For these experiments the 41% suspension of RBCs was used.

Doppler recordings. Both continuous-wave and pulsed Doppler recordings were made. A 5 MHz continuous-wave Doppler system (Model D9, Medasonics, Mountain View, Calif.) whose spectral characteristics were thoroughly tested[13] and whose beam width was found to be wide enough to almost uniformly insonate the entire acrylic tube cross-section was used. The Doppler probe was held in place by a clamp at an angle of 50 degrees to the tube axis.

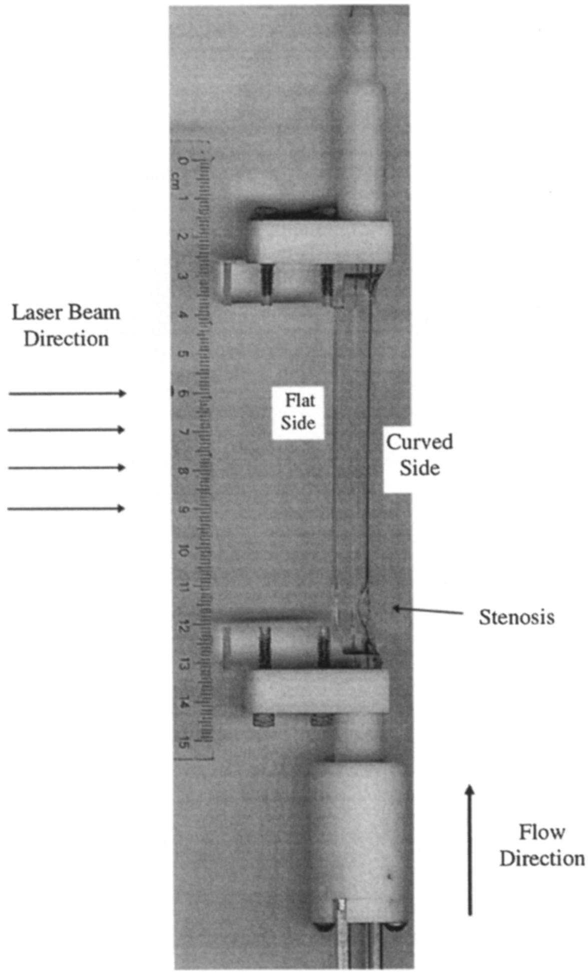
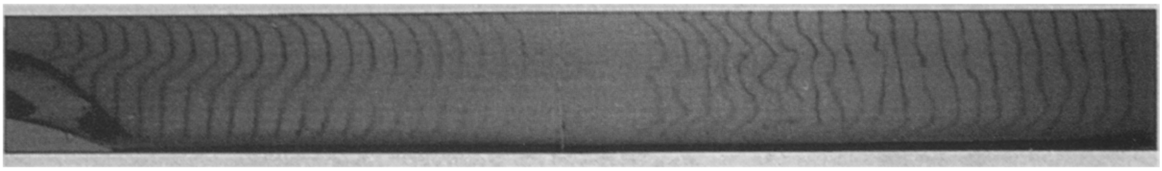


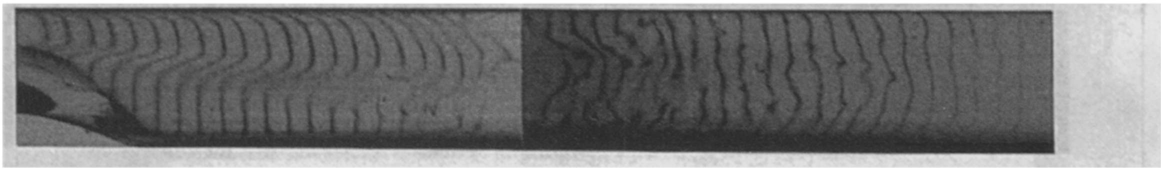
Fig. 3. Photograph of test section used in photochromic experiments. Seventy percent asymmetric stenosis was manufactured out of block of ultraviolet transparent Plexiglas. Entrance length to test section was approximately 90D. To minimize effect of refraction on laser beam, top half of tube was not rounded but maintained flat as shown above. Rounded side was then heated to softening point of Plexiglas, and ⅜-inch steel ball was pressed into tube to depth equivalent (accounting for change in diameter) to that previously used in making stenosis used in Doppler studies.

The distance of the probe surface from the center of the vessel was 2.5 cm, corresponding to the focal region of the transducer. Experiments were also carried out with a pulsed Doppler duplex system (Ultramark IV, ATL Inc., Bothell, Wash.) operating at 5 MHz with a small sample volume (1.5 mm or 28% of the beam length within the tube). The flow-separated Doppler signals were recorded on metal tape with a high-quality audio tape recorder.

Analysis of the Doppler signal. The recorded Doppler signals were digitized off-line with a 12 bit



(a)



(b)

Fig. 4. Photochromic traces distal to 70% asymmetric stenosis for flash delay of 2 msec under **(A)** steady ($Re = 545$) and **(B)** pulsatile flow. Each picture is composite of two photographs as described in text. Time in flow cycle when results in **B** were recorded is indicated in Fig. 2.

A/D converter at a sampling rate of 20 kHz and were analyzed with custom software. For the pulsatile flow ensemble averaging that was synchronized with the flow cycle had to be performed to detect any trend in the recorded data. A modified version of the pulse-foot-seeking algorithm presented by Evans[14] was used to identify the start of each cycle from the maximum frequency envelope.[15] Previous in vitro studies have shown that 20 flow cycles are generally sufficient to reduce amplitude variability in the Doppler spectrogram.[16] Thus to minimize this variability and to establish trends in the recorded data, more than 50 flow cycles were used to calculate ensemble averages at each measurement site.

Quantification of the Doppler spectrum. To reduce the variance (speckle) associated with the Doppler spectra,[17] ensemble averaging was performed. For the steady flow data a single spectrum was calculated every 10 msec, and then 1600 spectra were averaged. For the pulsatile flow data a single spectrum was calculated every 10 msec, and ensemble averaging over multiple flow cycles (>50) was performed.

The maximum Doppler frequency was determined from the averaged Doppler spectrum as the frequency below which 95% of the Doppler power lies,[18] whereas the mean Doppler frequency was calculated from the first moment of the spectrum. In addition, the following four indexes were used to quantify spectral broadening and spectral shape: the spectral broadening index (SBI), the normalized spectral variance (NSV), the coefficient of skewness (CS), and the coefficient of kurtosis (CK). These indexes were calculated with the following equations

$$SBI = 100\left[1 - \frac{f_{\text{mean}}}{f_{\text{max}}}\right] \tag{1}$$

$$NSV = [f_{\text{RMS}}^2 - f_{\text{mean}}^2]/f_{\text{mean}} \tag{2}$$

$$CS = E\{(f - f_{\text{mean}})^3\}/\sigma_s^3 \tag{3}$$

$$CK = E\{(f - f_{\text{mean}})^4\}/\sigma_s^4 \tag{4}$$

where f_{mean}, f_{max}, and f_{RMS} are the mean, maximum, and RMS Doppler frequencies, respectively, σ_s^2 is the variance of the spectrum, and $E\{(f - f_{\text{mean}})^p\}$ is the expected p$^{\text{th}}$ frequency moment of the Doppler spectrum centered around the mean Doppler frequency. The significance of each of these indexes has been addressed in detail by Bascom and Cobbold.[19] In brief, SBI and NSV are two different methods of measuring the spread (band width) associated with a particular spectrum. On the other hand, the two higher order moments, CS and CK, are measures of the skewness and peakedness of a spectrum.

Flow visualization

A photochromic dye technique[12] was used to visualize the precise nature of the flow field in the 5.1 mm diameter* tube distal to a 70% asymmetric

*It should be noted that this is somewhat greater than that used in the Doppler studies. This was necessitated by the physical constraints of the two different measurement techniques.

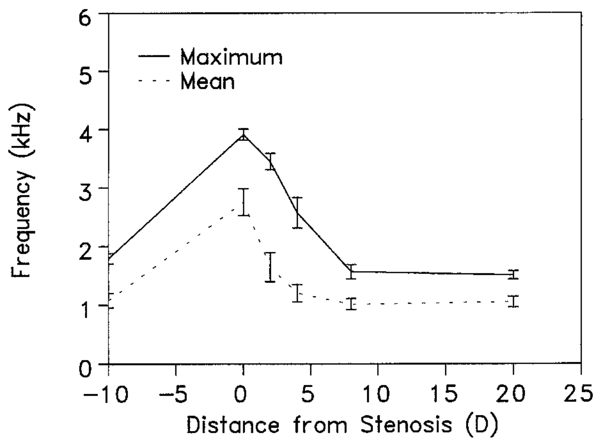


Fig. 5. Maximum and mean Doppler frequencies at various locations relative to stenosis for hematocrit of 4% ($Re = 1410$). *Error bars* represent ±1 SD.

stenosis under both steady and pulsatile flow conditions. This stenosis was constructed so as to ensure that its shape would be similar to the stenosis used in the Doppler studies.[15] The test section used in this study is shown in Fig. 3. The fluid used was a solution of deodorized kerosene and a photochromic dye at a concentration of 100 ppm. At 20° C this solution has a viscosity 1.25 cP and a density of 0.755 gm/cm^3. Both the steady and pulsatile flow was generated by a computer-controlled pump.[20] The Reynolds numbers used were matched to those used in the Doppler studies, as was the Womersley parameter (α) for the pulsatile flow. An entrance length of 90 tube diameters (D) was used to ensure fully developed flow at the stenosis. As a result of hardware limitations only a single photograph could be recorded within a flow cycle. Consequently, visualization of the entire flow cycle required a series of photographs to be taken at various times within different flow cycles. For this study approximately 20 images were recorded at equal intervals throughout the flow cycle.

A pulsed eximer laser (TE460, Lumonics Inc., Xanata, Ontario, Canada) was used[21] in conjunction with a linear lens array to produce traces in the test fluid. The lens array was capable of producing 22 traces spaced 1 mm apart, which meant that at any one instance a 20.4 mm (4D) long region could be visualized. Thus two sets of photographs were taken for every flow condition: one with the edge of the lens array at the throat of the stenosis and the other with the lens array displaced four tube diameters downstream. A 35 mm camera (Nikon Inc., Garden City, N.Y.) in conjunction with a xenon flash lamp was used to photograph the traces 2 msec after the laser was fired. The laser, camera, and flash lamp were controlled with custom software and hardware.

RESULTS AND DISCUSSION

For the sake of clarity the results will be presented and discussed together in this section.

Flow visualization results

The flow field in the region of the 70% (area) asymmetric stenosis was characterized with the photochromic technique. Characteristics of the flow field were not deduced from a single photograph but were based on the comparison of a number of photographs taken at different times. Regions where the displacement profiles showed nonperiodic fluctuations were identified as zones of localized turbulence. An example of the results obtained from steady flow at $Re = 545$ is shown in Fig. 4, *A*. These experimental results were found to be consistent with the flow field depicted in Fig. 1. For pulsatile flow it was observed that turbulence was present only for part of the flow cycle; its onset occurred at around peak systole and then extended throughout deceleration with relaminarization occurring in late diastole (0.6 sec). As shown in Fig. 4, *B* the poststenotic flow field during the turbulent phase of the flow cycle was seen to be composed of the four distinct zones observed under steady flow conditions; however, the length of the turbulent zone was found to reduce during deceleration. Additional details on the flow field distal to this stenosis under a variety of flow conditions are presented by Bascom.[15]

Steady flow Doppler results

Continuous-wave Doppler. The results for the maximum and mean Doppler frequencies shown in Fig. 5 can readily be correlated to changes in the flow field. In this figure the distance from the stenosis is measured in tube diameters from the narrowest part of the stenosis to the intersection of the ultrasound beam with the tube axis. The maximum frequency increased over the region with stenosis and returned to normal within approximately 5 tube diameters downstream from the stenosis. This is in contrast with the flow visualization results that showed that the flow field was still highly disturbed at this location.

The four quantitative indexes of spectral shape were calculated with the forward frequency spectra, and the results are presented in Fig. 6. Both the spectral broadening index and the normalized spectral variance peak at approximately 2 tube diameters downstream from the stenosis in the region where the velocity gradient in the shear layer was the greatest

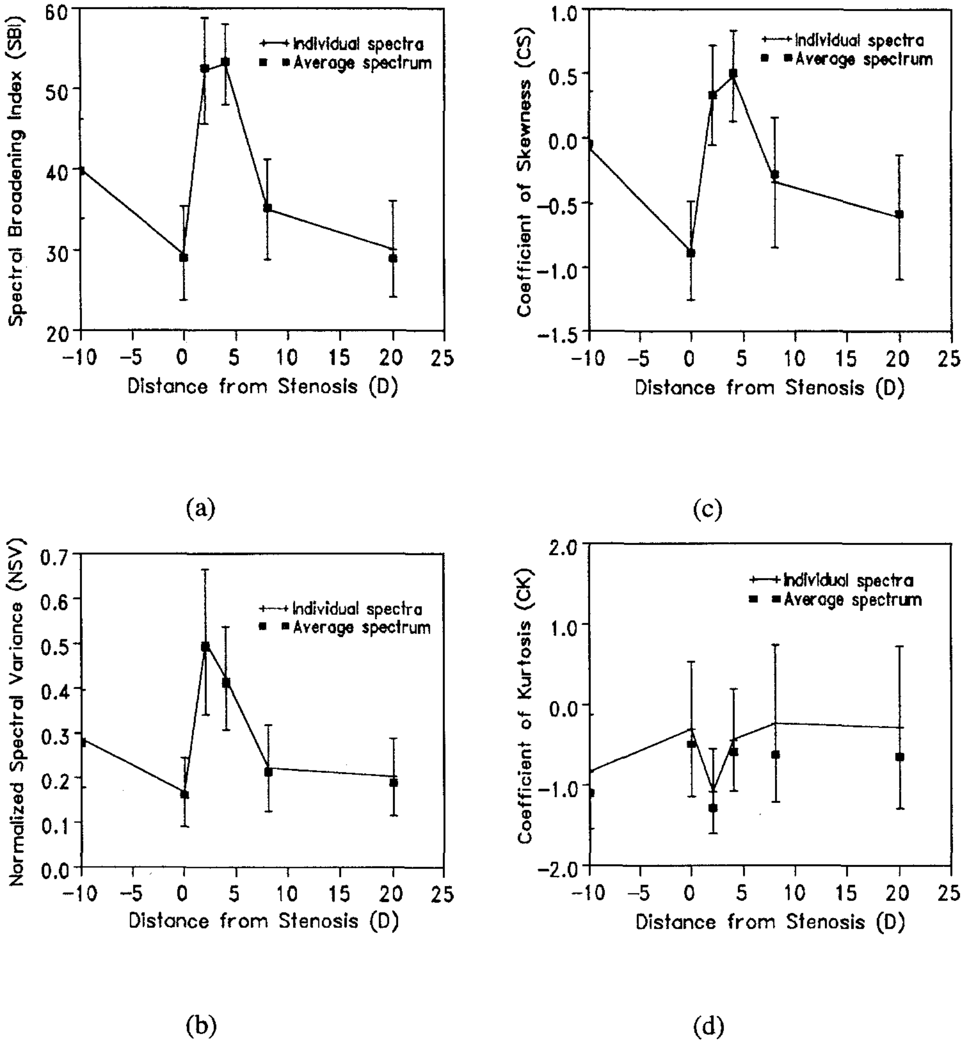
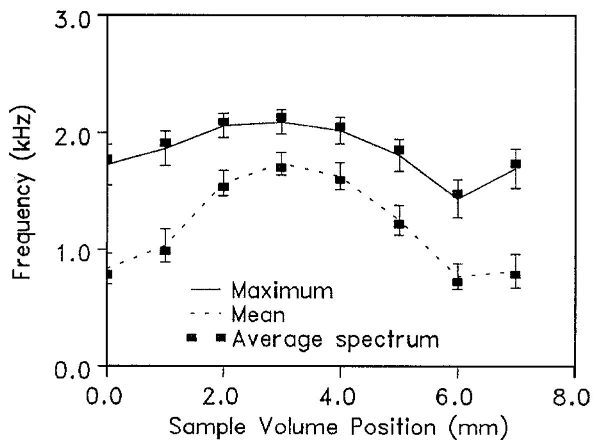


Fig. 6. Variation of indexes with position for both individual spectra and averaged spectra (1600, 10 msec window, hematocrit = 4%, Re = 1410). **A,** Spectral broadening index; **B,** normalized spectral variance; **C,** coefficient of skewness; **D,** coefficient of kurtosis.

rather than where the flow was maximally turbulent. Indeed, this is one of the primary differences between SBI measurements made with continuous-wave Doppler (i.e., with a large sample volume) and those made with pulsed Doppler (i.e., a small volume at the center of the vessel). In the case of pulsed Doppler the maximum value for SBI occurs in regions of maximal turbulence.[22] Instances in which the coefficient of skewness equals zero correspond to spectra that are symmetric, whereas positive and negative values correspond to the skewing of the spectrum towards low and high frequencies, respectively. The results for CS shown in Fig. 6, *C* can now be interpreted in the following manner. Before the stenosis occurred, the spectrum was flat, and CS was therefore zero. At the stenosis the spectrum was skewed towards high frequencies (i.e., plug flow), which resulted in a negative value for CS. Further downstream, flow in the separation region resulted in positive values, indicating skewing of the spectrum towards the low frequency end. Approximately 5D downstream, where the flow was turbulent, the spectra peaked towards the high frequency end, which caused the value of CS to become negative. At 20D CS has still not returned to the prestenosis value, indicating that full relaminarization has not yet occurred. This contrasts the results obtained at the lower Reynolds number, where relaminarization was found to occur at 20D.[4]

The results for the coefficient of kurtosis as a



(a)

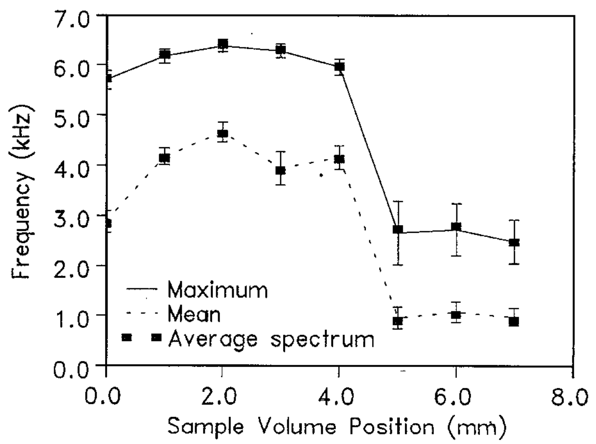


Fig. 7. Maximum and mean Doppler frequencies at various sample volume positions within the tube at **A** 20D and **B** 2D distal to stenosis. This position is defined as the distance of the sample volume from the near wall of tube in beam direction. These results were obtained with a 41% suspension of red blood cells at $Re = 545$.

function of distance from the stenosis are shown in Fig. 6, *D*. If CK is less than 0, then the spectrum is flatter than a normal (Gaussian) distribution curve, whereas the opposite is true if CK is greater than 0. For all the measured spectra CK was less than 0, indicating that all the spectra are less peaked than a Gaussian curve. This was hardly surprising, because insonation was across the entire vessel, thereby encompassing a wide distribution of velocity vectors. In contrast, for small sample volume pulsed Doppler measurements made near the center of a vessel, intrinsic spectral broadening dominates,[9] and the spectrum can be very peaked, resulting in CK being greater than 0.[10] From Fig. 6, *D* it can be seen that CK increased to close to 0 at the stenosis and then dropped to a value of −1.2 at 2D; thereafter CK increased. This figure illustrates the high variance associated with CK, especially in the turbulent zone, which may make this index unsuitable for clinical use.

In calculating these indexes the following points should be noted. When the results from the average and individual spectra were compared, few differences were found. Estimation of CS and CK requires that the third and fourth moments of the Doppler spectrum be evaluated. These high-order moments are very sensitive to noise, especially at high Doppler frequencies. Consequently, these indexes were calculated from spectra in which the power spectral density was set to 0 for $f > f_{max}$.

Finally, it should be noted that these results obtained at a Reynolds number of 1410 (4% hematocrit) are very similar to those obtained at the lower Reynolds number of 545 (41% hematocrit), with the primary difference being a change in the size of the various zones.[15]

Pulsed Doppler. Fig. 7 shows the mean and maximum Doppler frequencies calculated at 2D and 20D distal to the stenosis as a function of the distance of the sample volume away from the near tube wall. The probe was held at an angle of 50 degrees to the tube axis so that the distance across the 5.1 mm tube corresponded to 8 mm. This meant that up to eight positions were obtained when the sample volume was moved in 1 mm increments. The results at 2D differed from those obtained at 20D. This was not a surprise, because the velocity profile just distal to the stenosis was highly asymmetric (Fig. 4) with a high-velocity jet and much lower velocities within the separation region. Furthermore it should be noted that as the sample volume was moved from 4 mm to 5 mm, the mean and maximum Doppler frequencies dropped by 77% and 55%, respectively. This was the result of the sample volume being moved into the separation region. Therefore it is important to realize that the use of a small sample volume can result in a strong spatial dependence in the measured mean and maximum Doppler frequencies.

Quantification of these spectral shapes was performed, and the results are shown in Fig. 8. It is apparent that these results were also dependent on the precise location of the sample volume within the tube. Displacement of the sample volume by as little as 1 mm (e.g., from 4 mm to 5 mm) produced remarkably different values for the various Doppler indexes. Thus in the case of the asymmetric stenosis, large changes in spectral shape occur as the sample volume is moved from the jet flow region through the shear layer into the separation region. This occurrence suggests that reliance on a measurement from a single small sample volume can be misleading when used to

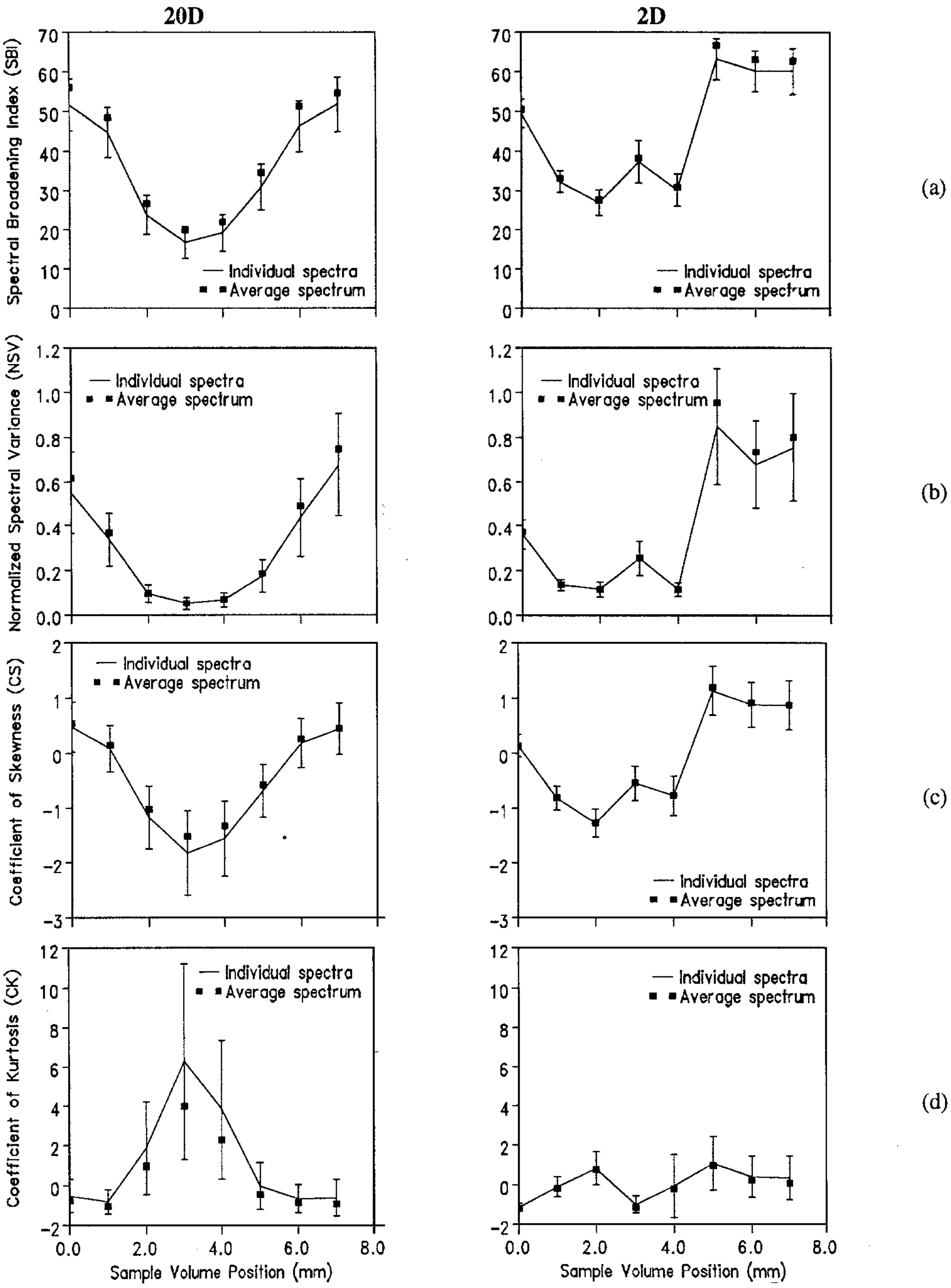


Fig. 8. Variation of indexes with sample volume position for both individual spectra and averaged spectra (1600, 10 msec window) at 20D and 2D distal to stenosis. This position is defined as distance of sample volume from near wall of tube in beam direction. **A,** Spectral broadening index; **B,** normalized spectral variance; **C,** coefficient of skewness; **D** coefficient of kurtosis.

characterize a flow field. If the sample volume is placed inappropriately, the Doppler indexes that are measured may appear quite normal. In the absence of *a priori* knowledge of the flow field being insonated, there is no way of knowing which sample volume location is "appropriate." A comparative examination of the spectral shape at various spatial locations may prove more useful.

Pulsatile flow Doppler results

Fig. 2 shows the flow waveform used in this study. This waveform is similar in shape to that found in the

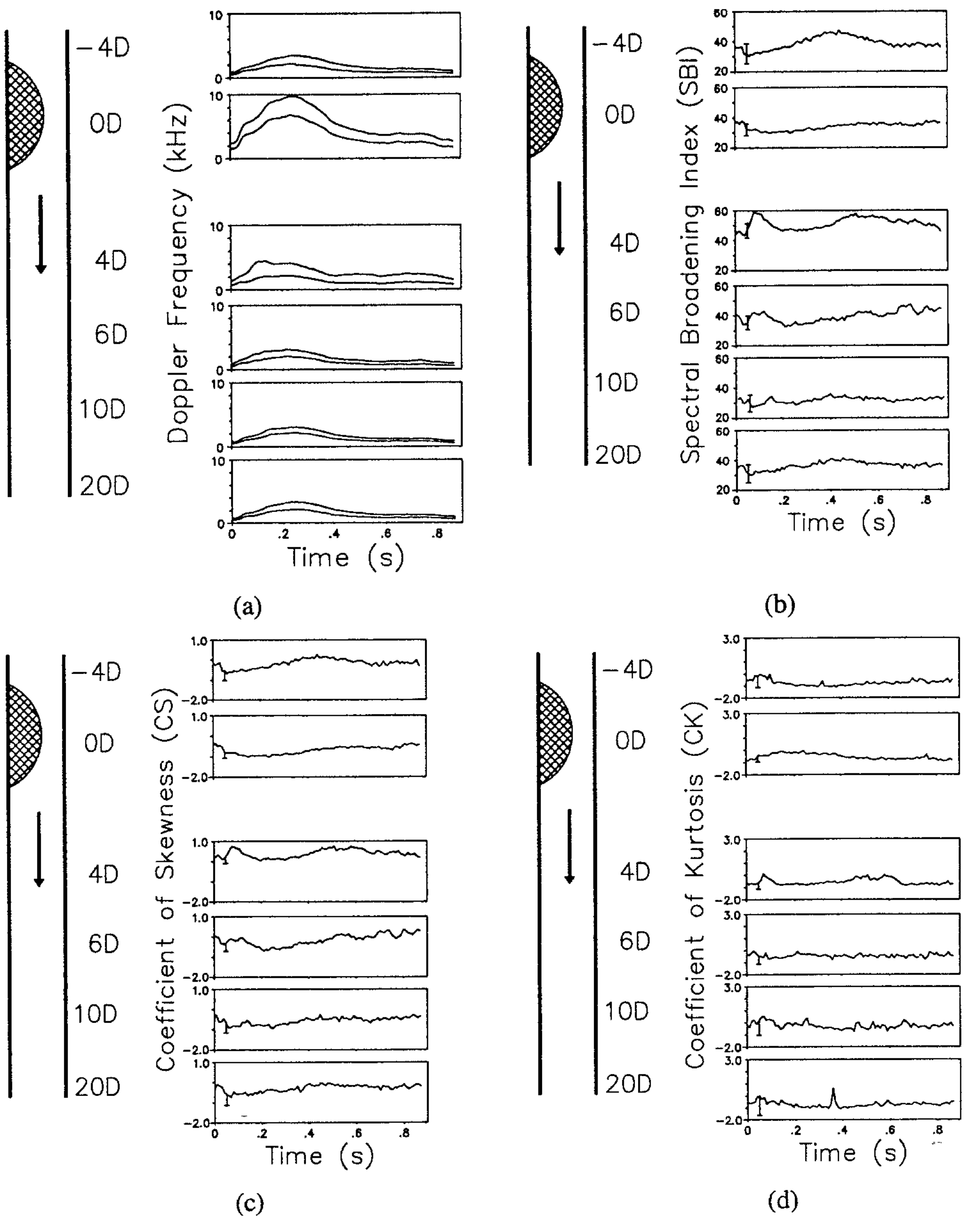


Fig. 9. Ensemble average **(A)** maximum and mean frequency, **(B)** SBI, **(C)** CS, and **(D)** CK envelopes obtained at various distances from stenosis. More than 50 flow cycles were ensemble averaged. These experiments were performed with 41% suspension of human red blood cells.

common carotid artery and has a period of 0.88 sec and a Womersely parameter for the fundamental harmonic of 3.8. Recordings were taken at six different locations relative to the stenosis: –4D, 0D, 4D, 6D, 10D, and 20D. Here again, these numbers represent distances measured from the narrowest part of the stenosis to the intersection of the ultrasound beam with the tube axis. Negative and positive values indicate positions proximal and distal to the stenosis, respectively. The spectra were measured from the forward channel of the Doppler unit, and the ensemble average of more than 50 flow cycles was performed.

The calculated maximum and mean Doppler frequencies are shown in Fig. 9, *A* at the six positions along the tube as a function of time in the flow cycle. In accordance with the flow visualization results and the continuity of flow, the mean and maximum frequencies were found to increase at the stenosis. The ratio of the mean to maximum frequency also increases at this location, indicating that the velocity profile is flatter, which is also in agreement with the photochromic results. At 4D the shape of the maximum frequency envelope has changed, and this change is probably due to the highly disturbed flow and the associated range in Doppler angles. The maximum frequency is still greater than that seen proximal to the stenosis. At 6D the frequency envelopes have effectively returned to their prestenotic value, whereas the photochromic results indicate that the flow is still disturbed in this region.

Three indexes, spectral broadening index, coefficient of skewedness, and coefficient of kurtosis, were used to quantify the forward frequency spectra, and the results are shown in Fig. 9, *B, C,* and *D.* Clearly these indexes exhibit both temporal and spatial variability. The spatial variability associated with SBI and CS in the deceleration phase of the flow cycle (0.3 sec) exhibits trends similar to those observed for the steady flow experiments; that is, their value drops within the throat of the stenosis because the velocity profile flattens, rises to a maximum at around 4D, and then begins to diminish as they return to their prestenotic values. This observation is also consistent with the photochromic results that found that once turbulence was triggered in the deceleration phase of the flow cycle, the flow field was similar in structure to that observed under steady flow. In the acceleration phase of the flow cycle, this trend is not as readily apparent, because the velocity profile before the stenosis is flat, and little or no change occurs in the shape of the velocity profile at the stenosis. Again, this is in keeping with the results obtained from the photochromic flow visualization experiments, where no evidence of turbulence was observed in this segment of the flow cycle. These results suggest that it may be more appropriate to make measurements of spectral width during the deceleration phase of the flow cycle. Similar conclusions can be made for the coefficient of kurtosis, although the inherent variability asociated with this index does make its use difficult.

CONCLUSIONS

It is clear that Doppler signals that are transduced from an artery contain important information on the blood velocity distribution in the region covered by the ultrasound beam. However, the results from this detailed study of the changes in the Doppler spectrum distal to a 70% area stenosis clearly demonstrate that no single Doppler measurement can accurately quantify the severity of a stenosis because of the complex three-dimensional nature of the poststenotic flow field and the fact that the spectral shape was very sensitive to the recording site relative to the position of the stenosis. Of the Doppler measurements, only peak velocity is related to the severity of stenosis. Reproducible peak velocity measurements are obtained only if the Doppler sample volume is positioned at or very near the throat of the stenosis and at an appropriate radial site that may not necessarily be at the center of the vessel. Other measurements from the Doppler spectrum, particularly higher order moments such as CK, showed a large variability even under laminar flow. This variability increased in regions of turbulence and consequently makes such indexes inappropriate as diagnostic indicators. Indeed, it may be unrealistic to expect that consistent reproducible Doppler velocity measurements can be made from such a complex flow field.

Thus relying on a single parameter to adequately represent the Doppler spectrum is likely to lead to erroneous results. In the poststenotic flow field the variability can produce a strong spatial and temporal dependence in the Doppler spectrum. This is particularly true when small sample volumes are used to quantify flow disturbance. Instead, it may be better to generate spatial maps of the indexes like SBI, because this may prove to be a better, more sensitive diagnostic indicator. In this regard color Doppler flow map systems may prove to be extremely useful.

We gratefully acknowledge the financial support of the Canadian Heart and Stroke Foundation and the Medical Research Council of Canada.

REFERENCES

1. Douville Y, Johnston KW, Kassam M, Zeuch P, Cobbold RSC, Jares A. An in vitro model and its application for the study of carotid Doppler spectral broadening. Ultrasound Med Biol 1983;9:347-56.
2. Ojha M, Cobbold RSC, Johnston KW, Hummel RL. Pulsatile flow through constricted tubes: an experimental investigation using photochromic tracer methods. J Fluid Mech 1989;203: 173-97.
3. Ojha M, Johnston KW, Cobbold RSC, Hummel RL. Potential limitations of centerline pulsed Doppler recordings: an in vitro flow visualization study. J Vasc Surg 1989;9:515-20.
4. Bascom PAJ, Cobbold RSC, Routh HF, Johnston KW. On the Doppler signal from a steady flow asymmetrical stenosis model: effects of turbulence. Ultrasound Med Biol 1993;19: 197-210.
5. Law YF, Cobbold RSC, Johnston KW, Bascom PAJ. Computer-controlled pulsatile pump system for physiological flow simulation. Med Biol Eng Comput 1987;25:590-5.
6. Law YF, Johnston KW, Routh HF, Cobbold RSC. On the design and evaluation of steady flow model for Doppler ultrasound studies. Ultrasound Med Biol 1989;15:505-16.
7. Langlois YE, Roederer GO, Chan A, et al. Evaluating carotid artery disease: the concordance between pulsed Doppler/spectrum analysis and angiography. Ultrasound Med Biol 1983;9:51-63.

8. Bascom PAJ, Cobbold RSC, Roelofs BHM. Influence of spectral broadening on continuous wave Doppler ultrasound spectra: a geometric approach. Ultrasound Med Biol 1986;12: 387-95.
9. Bascom PAJ, Cobbold RSC, Law YF, Johnston KW. Problems in the quantitative assessment of pulsed Doppler spectra. IEEE EMBS Conference Proceedings, Philadelphia, Nov. 1-4, 1990, 557-9.
10. Law YF, Bascom PAJ, Johnston KW, Vaitkus P, Cobbold RSC. Experimental study of the effects of pulsed Doppler sample volume size and position on the Doppler spectrum. Ultrasonics 1991;29:404-10.
11. Ojha M, Johnston KW, Cobbold RSC. Evidence of a possible link between poststenotic dilation and wall shear stress. J Vasc Surg 1990;11:127-35.
12. Ojha M, Hummel RL, Cobbold RSC, Johnston KW. Development and evaluation of a high resolution photochromic dye method for pulsatile flow studies. J Phys E Sci Instrum 1988;21:998-1004.
13. Bascom PAJ. An investigation of the factors which affect the received Doppler power over a cardiac cycle. MASc thesis, University of Toronto, 1988.
14. Evans DH. A pulse-foot-seeking algorithm for Doppler ultrasound waveforms. Clin Phs Physiol Meas 1988;9:267-71.
15. Bascom PAJ. Influence of flow conditions on the backscattered Doppler ultrasound signal. PhD thesis, University of Toronto, 1995.
16. Cloutier G, Allard L, Guo Z, Durand LG. The effect of averaging cardiac Doppler spectrograms on the reduction of their amplitude variability. Med Biol Eng Comput 1992;30: 177-86.
17. Mo LYL, Cobbold RSC. Speckle in continuous wave Doppler ultrasound spectra: a simulation study. IEEE Trans Ultrason Ferroelec Freq Contr UFFC 1986;33:747-53.
18. Mo LYL, Yun LCM, Cobbold RSC. Comparison of four digital maximum frequency estimators for Doppler ultrasound. Ultrasound Med Biol 1988;14:355-63.
19. Bascom PAJ, Cobbold RSC. Effect of transducer beam geometry and flow velocity profile on the Doppler power spectrum: a theoretical study. Ultrasound Med Biol 1990;16: 279-95.
20. Holdsworth DW, Rickey DW, Drangova M, Miller DJM, Fenster A. Computer-controlled positive displacement pump for physiological flow simulation. Med Biol Eng Comput 1991;29:565-70.
21. Ojha M. Wall shear stress temporal gradient and anastomotic intimal hyperplasia. Circ Res 1994;74:1227-31.
22. Cambell JD, Hutchison KJ, Karpinski E. Variance of Doppler ultrasound spectral width in the post-stenotic velocity field. Ultrasound Med Biol 1989;15:611-9.

Submitted Oct. 5, 1995; accepted Jan. 3, 1996.

DISCUSSION

Dr. Robert W. Hopkins (Providence, R. I.). I think this is a marvelous study, and we need a lot more of them. I think it does emphasize the inexactness of the measurements we are making, and it really is surprising that they correlate clinically so well with stenoses. What would be theoretically ideal, however, would be measuring not the peak velocity but the true mean velocity. In any nonbranching tube, the mean velocity change is inversely proportional, precisely, to the change in the cross-sectional area; that is, the area times the velocity is equal to the volume flow, which is exactly the same throughout the entire system. So if we could measure true mean velocities, we could measure true area change. Do you think there is any real prospect for that?

Dr. K. Wayne Johnston (Toronto, Ontario, Canada). It is not likely possible. In the article, we present the mean values. Your points are correct; that is, the mean is theoretically the better measurement, because it is more directly related to flow. There are, however, all of the problems associated with the mean that there are with the maximum. First, where should the measurement be made in the large flow field? Second is the fact that you still have to make the mean velocity measurements with a corrected Doppler angle. Third, and most important, is the fact that the mean velocity (compared with the peak velocity) is more likely affected by the presence of signal noise. Any noise that you have in the Doppler signal is going to be averaged along with the true Doppler recording. This is why we prefer to use the peak velocity or the mean velocity.

Dr. Frank Lo Gerfo (Boston, Mass.). I want to compliment you first on your excellent laboratory, which I have visited and I think is one of the only places where there is a complete spectrum of people from the clinicians to the engineers in one place looking at these problems.

I have often wondered why this is not done more often and in particular why the manufacturers of these machines do not actually have a setup where technicians could come and get a feel for what is actually going on in the tube and what they are picking up with their instrument. I think this would be a great way to come to understand the limitations.

Have you looked at other parameters that people frequently use to assess the stenosis, like, for example, the velocity ratio?

I think I would quibble with your use of the term turbulence, which is a chaotic high-energy state. I am not sure at those velocities there is turbulence anywhere. I would agree that the flow is disturbed downstream, but turbulence is a very specific term where there is chaos and there is a lot of flow at right angles to the main flow stream and so forth.

Dr. Johnston. Thank you very much. The comments are very germane. I do not know why the manufacturers do not inform us of the potential errors that are inherent in their machines. Do they not, in fact, understand some of the basic problems? Or are they simply responding to the challenge of balancing the potential innovations that are available to them and the dollars that we are prepared to put towards these instruments?

Velocity ratios will not help us here because we still have the fundamental problem that the velocity measures must be made in exactly the correct position in a very complex flow field. Hence, where exactly do you measure the velocity ratio? If you know precisely where the maximum velocity is located, you can relate it to another velocity upstream and from the velocity ratio determine the severity of the stenosis. However, this study showed great complexity of flow field and the fact that it is very difficult to know clinically where the maximum velocity is.

I agree with you that turbulence is present when flow is random in nature. Downstream from the stenosis, the velocity profiles change from one instance to another, thus indicating the random nature of the flow. Also, if you look very closely at our enlarged pictures, you will see that the traces are blurred, meaning the flow is going in another direction, that is, out of the plane of the photograph.

Dr. William W. Babson (Plymouth, Mass.). I believe that continuous-wave Doppler evaluation of the lower extremities, particularly the veins and the arteries, where you do not depend on a number, a peak velocity, you depend on a pattern of sound to identify problems such as obstructions or stenoses is very useful. I use it. In fact, I feel I am very accurate with my evaluations of in situ saphenous vein grafts in picking up valves, which I then follow-up with either angioscopy or venography or arteriography—in evaluation of peripheral venous disease, in valve closures, and so forth. I think over a period of time you can learn to hear and appreciate what the differences in the signal mean, so I think that you do not need absolute numbers to get a great deal of value out of the continuous wave Doppler.

Dr. Johnston. In general, I agree with your observations in terms of the lower extremity. However, in the literature that assesses the accuracy of carotid duplex examinations, it is a number that is used and not subjective interpretation. We are urging caution in relying on overreliance on the accuracy of numeric measurements. It is hoped that our data will also help explain some of the variation between ultrasound measurements and angiography. Continuous-wave Doppler is an underestimated technique. It avoids one of the major problems that is identified in this study. If you insonate the entire vessel with ultrasonography, you do not have the problem of misaligning the position of pulsed Doppler sample volume. Unlike pulsed Doppler, you do not have to worry if it is in the middle of the artery or just off axis a little, where the signal may be different because of the complex nature of the poststenotic flow field.