1996, *The British Journal of Radiology*, 69, 172–177

Accuracy of maximum velocity estimates made using Doppler ultrasound systems

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Abstract

This study was performed in order to provide quantitative data on the estimation of maximum velocity made using modern Doppler ultrasound systems. This is important since the degree of stenosis within arteries is commonly assessed from the maximum velocity. A string phantom was used as the source of Doppler signals. This enables direct comparison between the Doppler estimated maximum velocity and the true filament velocity. Six modern commercial Doppler systems were used. Measurements were made under standard conditions for each probe. In addition a number of factors were varied in turn (beam–filament angle, filament depth, filament velocity and Doppler aperture position). Under standard conditions the maximum velocity was overestimated in all cases (0–29% error). For all measurements maximum velocity errors ranged from −4% to 47%. There was a large intraprobe variation in maximum velocity estimation (mean variation of 25%), and a large interprobe variation (mean variation of 18%). These results indicate that, at present, errors in maximum velocity estimation may be directly translated into significant errors in the estimate of the degree of arterial stenosis made from velocity measurements. As a consequence, some patients may be incorrectly categorized. Consideration should be given to applying angle dependent correction factors to maximum velocity measurements, and to the use of conversion from Doppler frequency shift to velocity using the angle derived from the edge of the Doppler aperture.

Velocity measurements, particularly peak systolic velocities, are one of the most frequently used quantitative measurements in clinical Doppler ultrasound. In the heart, estimates of peak velocity are made from the Doppler signal and used to derive the pressure gradient across the cardiac valve using the modified Bernoulli equation [1]. In arteries, increased velocities occur in the region of a stenosis, and the degree of stenosis may be derived directly from the maximum velocity [2].

In general, there is little testing of the performance of Doppler ultrasound systems, particularly in relation to the errors in maximum velocity estimation. This is probably due to the lack of suitable test objects and test procedures [3]. Measurement of the error in velocity estimation is made by obtaining Doppler waveforms from the test object and comparing the true maximum velocity with the Doppler estimated maximum velocity. This is difficult using a flow phantom as knowledge of the true maximum velocity is obtained indirectly from the measured flow rate and the known cross-sectional area. It is difficult to validate the assumption of parabolic profile which needs to be made [3]. A string phantom is a device in which the moving target is a filament such as nylon, silk or O-ring rubber. Provided the target material is selected appropriately, it may be used to assess the error in Doppler estimated maximum velocity [4, 5]. Estimates of true filament speed are obtained by dividing the measured circumference of the loop of the filament by the time for a fixed point to pass once around the loop.

It is known, for flow in a straight vessel, that maximum velocity is usually overestimated. A small number of studies have used Doppler test objects to demonstrate this [6, 7]. These studies involved use of a string phantom at a time when there were no data on the choice of filament, and both studies used spiral wound filaments, which may possess inappropriate scattering properties [4, 5]. Although these studies correctly concluded that maximum velocity was overestimated, doubt must exist regarding the size of the velocity errors which were measured. Error in velocity estimation is due to intrinsic spectral broadening. The major determinant of intrinsic spectral broadening is associated with the range of velocity vectors which the moving target subtends at the transducer (Figure 1), which will give rise to a range of Doppler frequency shifts. This is called geometrical spectral broadening [8, 9]. Intrinsic spectral broadening also arises from passage of the target through the sample volume, and is called transit time broadening. Recent work has suggested that transit time spectral broadening is small compared with geometrical spectral broadening in clinical practice [10]. The size and variability of the error in maximum velocity estimation is important for clinical Doppler ultrasound studies as the degree of stenosis may be inaccurately assessed. This paper examines the size and variability of error in Doppler estimated maximum velocity for a range of commercial Doppler systems using a string phantom.

Received 19 July 1995 and in revised form 2 October 1995, accepted 19 October 1995.

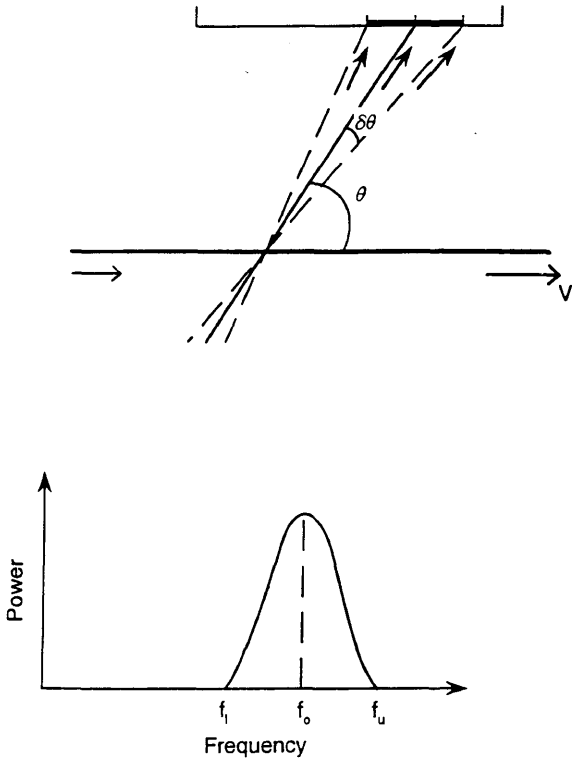

Figure 1. The linear array is shown with the Doppler aperture highlighted. The moving target is a single velocity v. This subtends a range of angles at the Doppler aperture from $\Theta-\delta\Theta$ to $\Theta+\delta\Theta$. The detected Doppler frequency shift is dependent on the cosine of the subtended angle. This leads to a range of Doppler frequency shifts $f_l=f-\delta f$ to $f_u=f+\delta f$.

Methods

String phantom

A commercial string phantom was used (BBS Medical Electronic AB, Sweden). This consists of a filament which passes around three wheels, one of which is connected to the motor. The motor is fully submersible and its speed is controlled by an external control box. Filament speed is indicated on the control box, and the speed may be adjusted manually, with a range of speeds available from -7.5 m s^{-1} to 7.5 m s^{-1}. The string phantom was placed in a tank of dimensions 20 cm × 20 cm × 10 cm (height × length × width) filled with a 9% glycerol solution by volume. The water had been degassed by boiling, and then allowed to cool to room temperature. This solution has an acoustic velocity of 1540 m s^{-1} at a room temperature of 20 °C.

O-ring rubber (RS components, UK) was joined into a loop using cyanoacrylate glue and stretched over the wheels of the phantom.

Accuracy of displayed filament velocity reading

The true filament velocity was checked by dividing the filament circumference by the time for one revolution of the O-ring.

A thin (0.07 mm diameter) nylon thread was knotted at one site on the O-ring rubber with the free ends of the knot cut leaving about 5 mm of trailing nylon thread. String velocity was set to an indicated 100 cm s^{-1}. Doppler waveforms were acquired using a 4 MHz Doptek continuous wave unit with a probe angle of 60°. Passage of the knot produced a high intensity spike on the Doppler waveform which could be clearly seen and which was used to mark the passage of the knot. The trace was frozen and the time T_{30} for the passage of 30 circuits of the knot was estimated using the timing callipers of the Doptek system. The time T_1 for one revolution is $T_{30}/30$. The timing accuracy of the Doptek had been previously tested by injecting into the audio input of the Doptek 4 kHz tone bursts of 1000 ms duration, with each tone burst separated by 1000 ms. The numerical value of accuracy was found to be less than 0.2%.

O-ring rubber is stretched when it is placed over the wheels of the string phantom so this must be taken into account when measuring the circumference. A second nylon thread was knotted on the O-ring 12 cm from the first knot. The distance L_1 between these two points was measured with the O-ring stretched. The O-ring was cut, and the unstretched distance L_2 between the two knots and the total length L_3 of the O-ring measured. The circumference C of the O-ring in its stretched condition on the phantom is then L_3 (L_1/L_2). The true filament velocity is then C/T_1.

Doppler systems

Six colour flow systems and one duplex system used for peripheral vascular and abdominal investigations were tested. These mostly incorporated 7.5 or 10 MHz linear arrays for imaging of the carotid artery (5–6 MHz Doppler), 5 MHz linear arrays for imaging of the lower limb (3.5–4 MHz Doppler), and 3.5 MHz curvilinear or phased array sector probes for imaging of the abdomen (2–2.5 MHz Doppler). The standard convention of referring to transducers by their B-scan imaging frequency has been adopted in this study. Systems tested were representative of most modern machines in use today; and were manufactured by Acuson, ATL, Hitachi, Kretz, Diasonics and Toshiba. These machines will be referred to anonymously in the results section.

Test procedure

The probe was held firmly in place using a clamp and stand. The probe position and the position of the filament of the string phantom were adjusted so that the filament was set at the required position and angle. Filament speed was adjusted using the manual control of the string phantom, the angle correction cursor was aligned along the B-scan image of the filament and a Doppler sonogram acquired. Manual velocity estimation callipers were used to obtain three estimates of the maximum velocity from the sonogram, and the average was taken.

Test conditions

For the velocity measurements to be comparable with those made in a clinical setting, a set of standard conditions was used for each type of probe. The indicated

filament velocity was set to 100 cm s^{-1}, the beam–filament angle to 60°. Filament depth was 25 mm for the 7.5 or 10 MHz probes, 50 mm for the 5 MHz probes, and 100 mm for the 3.5 MHz probes. For the curvilinear probes, the Doppler beam direction was straight down from the transducer face. For the linear array probes, beam-steering for most machines allows the angle between the Doppler beam and the transducer face to be adjusted so that in clinical practice for vessels running parallel to the transducer face, it is easy to set a beam–vessel angle of 60°–70°. Where possible, the Doppler beam direction was adjusted to give the minimum angle between the transducer face and the beam direction. For two machines no Doppler beam-steering was possible and the angle between the transducer face and the Doppler beam was 90°. The Doppler sample gate was placed centrally in the field-of-view, aligned with the B-scan image of the filament.

In addition to the standard conditions the effect on velocity measurement of several factors was investigated in turn for each type of probe. These are listed in Table I.

Reproducibility

Each of the series of measurements made with variation of one parameter, as listed in Table I, included a measurement made at the standard conditions. Therefore, for each probe there were four separate measurements made under identical conditions. These were used to assess reproducibility of maximum velocity measurements.

Results

It was found that the true filament speed was 4–8% higher than the indicated speed on the string phantom (depending on the filament tension associated with differences in unstretched length). For any one filament the degree of overestimation did not change with indicated filament speed. The indicated filament speed for all measurements was therefore multiplied by 1.06 to obtain the true speed.

Table I. Measurement variables

Velocity	50, 100, 150, 200, 250 cm s^{-1}
Angle	40°, 50°, 60°, 70°
Depth	20–150 mm (3.5 MHz probes) 20–100 mm (5 MHz probes) 20–80 mm (7.5–10 MHz probes)
Aperture and sample volume position	All available Doppler aperture angles (*e.g. "L", "C", "R"*) Extreme left, centre, and extreme right sample volume positions at the standard depth

The coefficient of variation of maximum velocity measured under standard conditions for each probe varied between 0% and 3%, with an average value of 0.9%. This, combined with the differences in calibration of 2%, indicates that measured differences in maximum velocity greater than 3–4% are significant.

Figure 2 shows the results displayed for all of the machines included in the survey. In each case the ratio of (Doppler estimated maximum velocity/true filament velocity) has been plotted. For each probe of each machine the results for the standard conditions are shown, along with the maximum and minimum values from the investigation of beam–filament angle etc. In each case maximum velocity is overestimated (Table II). The range of errors in maximum velocity when all probes of all frequencies were considered was high (−4% to 47%). There was a large range of velocity errors for measurements made under standard conditions (0–29% error, mean 18%). Variability in maximum velocity estimation is shown in Table III. Average intraprobe variability over the range of conditions studied is 25%. The average variability between probes of different machines of the same frequency, under standard conditions, is 18%.

Figure 3 shows results for the 7.5 MHz probe from machine J of Figure 2. This shows little or no variation of velocity estimation with filament velocity or Doppler

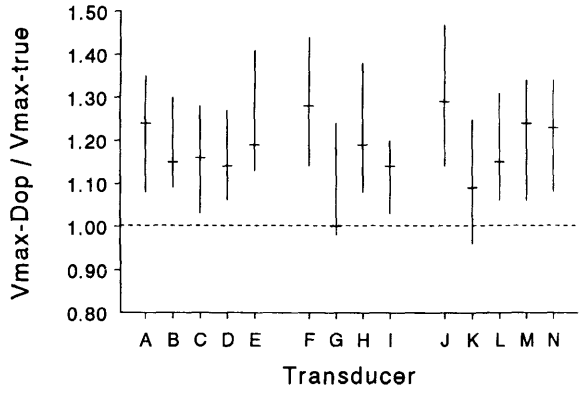

Figure 2. Ratio of estimated to true maximum velocity. For each probe the extreme ends of the line represent the maximum and minimum values found during the series of measurements. The measurements made under standard conditions are indicated by the horizontal mark on the vertical line. Transducers: 3.5 MHz (A to E), 5 MHz (F to I), 7.5 MHz (J to N).

Table II. Maximum velocity errors (%)

	Frequency			
	All probes	3.5 MHz	5 MHz	7.5 MHz
All machines (min, max)	−4 to 47	3 to 35	−2 to 44	−4 to 47
Standard conditions: mean (range)	18 (0–29)	18 (14–24)	15 (0–28)	20 (9–29)

Table III. Variability (%) of maximum velocity errors

	Frequency			
	All probes	3.5 MHz	5 MHz	7.5 MHz
Intramachine[a]: mean (range)	25 (17–30)	23 (19–25)	25 (17–28)	27 (24–30)
Intermachine[b]	18	9	28	18

[a] For each probe: 100[(largest velocity)/(smallest velocity) − 1].
[b] For each group of probes: 100[(largest velocity)/(smallest velocity) − 1].

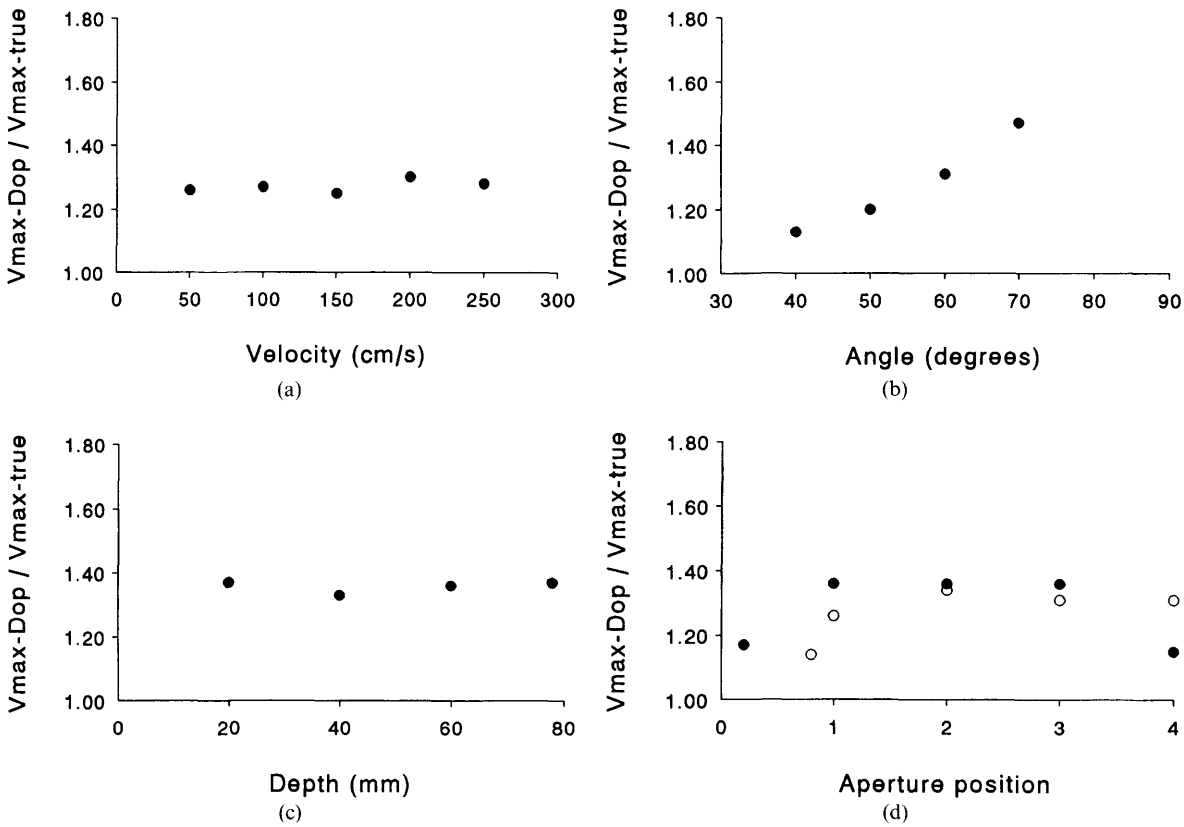

Figure 3. Variation of velocity estimates as a function of various factors for one machine: (a) velocity; (b) angle; (c) depth; (d) Doppler aperture position (○ indicates a Doppler beam angle of 70°, and ● indicates an angle of 90°); note that in this example several more measurements have been made in order to illustrate the variation of estimated maximum velocity for different locations of the Doppler aperture along the linear array (extreme left of the array indicated by "0", and extreme right by "4").

range gate depth. There is some variation of velocity estimation with the Doppler aperture position, and large variation with the beam–filament angle. These are fairly typical results for the machines surveyed.

Discussion

In virtually every case in this study the maximum velocity was overestimated by the Doppler system. This finding is consistent with the known effects of spectral broadening on the Doppler signal.

The variations in velocity estimation between machines, and for each individual machine, are most likely due predominantly to variations in intrinsic spectral broadening. For an individual machine this appears to be most significant for alterations of the beam–filament angle, and alterations of the Doppler beam direction and aperture position. It is common knowledge that the acquisition of Doppler signals for beam–filament angles near to 90° should be avoided. For example, most Doppler systems do not display the sonogram in units of velocity for angles above 70°. This study clearly demonstrates that, in general, there is no critical angle such as 60° below which the error in velocity estimation is acceptably low.

The variation of velocity estimation with Doppler aperture position has been noted previously [7], and is related to the change in size of the Doppler aperture for different positions along the array; as the number of elements of the linear array forming the Doppler aperture will vary. If the Doppler aperture size remained fixed as the depth increased then one would expect the range of angles subtended by the moving filament to be less at larger depths. Hence the degree of intrinsic spectral broadening would be less. For most of the linear array systems investigated there was very little variation of the velocity error with depth. This is consistent with an increase in the size of the Doppler aperture as the depth of the range gate increases.

Table IV shows the conversion between maximum velocity and percent stenosis used by clinical staff in this hospital (probe J of Figure 2). The third column of Table IV shows the range of velocities arising from a $\pm 20\%$ variation in the maximum velocity estimation. It can be seen that there is overlap in the velocity measurements corresponding to the first four stenosis groups (0%, 0–15%, 15–50%, 50–80%). In worst case conditions, it is therefore possible that for an individual machine there may be significant errors in the categorization of the degree of stenosis. Using similar arguments it is clear that similar mis-categorization may occur when different machines are used.

In clinical practice for an individual machine, measurements made on a particular vessel, such as the internal carotid artery, will be associated with approximately similar vessel anatomy and locations for each patient. In this situation it is likely that there will be a smaller range of angles, depths and aperture positions compared with the measurements made in this study. This in turn may be associated with a lower intramachine variation in maximum velocity measurements. The problem of inaccurate velocity estimation may therefore first be noticed after purchase of a second duplex or colour flow system, where it is found that the velocity measurements made on the same patient using the two systems do not agree.

Several approaches could be attempted to correct for velocity estimation errors. The first is to perform an internal audit for each Doppler machine comparing the estimated degree of stenosis with angiography and with clinical follow-up. This would then result in a machine dependent look-up table for conversion of velocity to percentage stenosis. This type of audit is occasionally performed to refine standard tables for use with particular machines. The second approach is to use standard factors to correct velocity estimates for the effect of overestimation due to intrinsic spectral broadening. One factor for each machine could be based on one measurement of velocity error made under standard conditions. This is an approach which may work in practice provided that there is little variation in machine settings such as Doppler aperture position and beam–vessel angle. In practice, it is likely that, for a given vessel, the same Doppler beam direction would be used for all patients. In this case the largest source of error is that associated with the beam–vessel angle. A set of correction factors could be derived for the range of angles used in clinical practice. The estimated velocity values would then be divided by the appropriate correction factor obtained from string phantom experiments. This may be an approach which is worth considering as a practical tool for machines in current use. The third, and possibly the most desirable, approach is for manufacturers to provide conversion from Doppler frequency to velocity using the beam–vessel angle arising from the edge of the Doppler aperture. This approach would probably require the formation of a consensus view among the radiological community, possibly through the professional bodies, that this was a desirable thing for manufacturers to perform.

The recent European Commission *Medical Devices Directive* [11] sets a requirement on manufacturers to indicate limits of accuracy for medical devices with a measuring function and, in future, there is likely to be better information from manufacturers on the errors in all clinical measurements, including velocity measurements.

Table IV. Conversion of maximum velocity to percent diameter stenosis for probe J of Figure 2.

Diameter stenosis (%)	Peak systolic velocity (cm s^{-1})	Peak V (cm s^{-1}) ($\pm 20\%$ variation)
0	<90	<72–108
0–15	<100	<88–132
15–50	<125	<100–150
50–80	>130	>104–156
80–99	>250	>200–300

Conclusion

This study has shown that, in general, for Doppler ultrasound systems, there is overestimation of the maximum velocity, and that there is large variation for measurements made using the same machine, and using different machines. This may lead to variations in the estimate of the degree of stenosis for different machines, and may be a factor which contributes in practice to variations in stenosis estimation for any single machine. Consideration should be given to applying angle dependent correction factors to maximum velocity measurements, and to the use of conversion from Doppler frequency shift to velocity using the angle derived from the edge of the Doppler aperture.

References

1. HOLEN, J, AASLID, R, LANDMARK, K and SIMONSEN, S, Determination of pressure gradient in mitral stenosis with a non-invasive ultrasound Doppler technique, *Acta Med. Scand., 199*, 455–460 (1976).
2. ROBINSON, M L, SACKS, D, PERLMUTTER, G S and MARINELLI, D L, Diagnostic criteria for carotid duplex sonography, *AJR, 151*, 1045–1049 (1988).

3. HOSKINS, P R, SHERRIFF, S B and EVANS, J A (eds), *Testing of Doppler Ultrasound Equipment* (IPSM, York) (1994).
4. HOSKINS, P R, Choice of moving target for a string phantom: I. Measurement of filament backscatter characteristics, *Ultrasound Med. Biol., 20*, 773–780 (1994).
5. HOSKINS, P R, Choice of moving target for a string phantom: II. On the performance testing of Doppler ultrasound systems, *Ultrasound Med. Biol., 20*, 781–789 (1994).
6. DAIGLE, R J, STAVROS, A T and LEES, R M, Overestimation of velocity and frequency values by multielement linear array Dopplers, *J. Vasc. Technol., 14*, 206–213 (1990).
7. HOSKINS, P R, LI, S L and McDICKEN, W N, Velocity estimation using duplex scanners, *Ultrasound Med. Biol., 17*, 195–199 (1991).
8. NEWHOUSE, V L, FURGASON, E S, JOHNSTON, G F and WOLF, D A, The dependence of ultrasound Doppler bandwidth on beam geometry, *IEEE Trans. Sonic Ultrason., SU 27*, 50–59 (1980).
9. CENSOR, D, NEWHOUSE, V L, VANTZ, T and ORTEGA, H V, Theory of ultrasound Doppler spectra velocimetry for arbitrary beam and flow configuration, *IEEE Trans. Biomed. Eng. Comput. BME, 35*, 740–751 (1988).
10. LI, S F, McDICKEN, W N and HOSKINS, P R, The non-equivalence of transit time and geometrical spectral broadening, *IEEE Trans. Ultrason. Ferro. Freq. Cont.* (submitted).
11. EUROPEAN COMMISSION, Medical Devices Directive (93/42/EEC). In *Official Journal of the European Communities, 36, L169/1, clause 10.1* (HMSO, London) (1993).