J Anesth (1996) 10:133–139

Journal of
Anesthesia

Velocity measurements with a new ultrasonic Doppler method independent of angle of incidence

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Abstract: Blood flow velocity measured by Doppler ultrasound is the relative velocity dependent on the path of the ultrasound beam, which should be influenced by its angle of incidence against the blood flow in the vessel. The angle of incidence generates varying changes in flow velocities that can be measured by the Doppler device. The aim of our study was to develop a new ultrasonic Doppler catheter which could provide a true flow velocity independently of the angle of the ultrasound beam against the flow direction, and to assess the validity of the true flow velocity obtained by a new device using the electromagnetic flowmeter. The newly developed Doppler catheter has a pair of adjoining ultrasonic crystals located on one side of the catheter at right angles to each other. Each Doppler shift, which is detected by two transducers (Δf1, Δf2) that sample the flow velocity at two closely spaced points, is used to compute two velocity measurements (V_1 and V_2); these are the velocities detected by the transducers. The true velocity was calculated using the following equation: $V = ((V_1)^2 + (V_2)^2)^{1/2}$, where V = true velocity. The velocities were calculated by newly developed phase differential techniques. Using a continuous flow model, we compared the flow velocity measured by the new Doppler catheter with that assessed by an electromagnetic flow probe placed into the circuit. At between 0.42 and 4.49 l·min^{-1}, the flow velocity measured by the new Doppler catheter (Doppler velocity) at five sampling depths was compared with the mean velocity calculated from the volumetric flow rate measured by an electromagnetic flowmeter (EMF velocity). The Doppler velocity (y) strongly correlated with the EMF velocity (x) at five sampling depths ($r^2 = 0.99$, respectively). At the maximal velocity sampling depth, the regression equation was y = 1.29x + 2.47 ($r^2 = 0.99$, $P < 0.0001$, $n = 41$, SEE = 0.015). The Doppler velocity also correlated with the volumetric flow rate measured by the electromagnetic flowmeter ($r^2 = 0.99$). The flow velocity measurements using the new Doppler catheter and device we have developed can provide more instantaneous and useful information on hemodynamics.

Key words: Flow velocity, Ultrasound, Doppler catheter

Introduction

Doppler ultrasound instruments are now widely used to measure blood flow velocity in medical practice. Blood flow velocity can easily be measured with pulsed wave and/or continuous wave Doppler ultrasound [1–3]. When an ultrasound beam is intercepted by moving blood cells, the frequency of the back-scattered ultrasound differs from that of the incident beam. If the blood cells are moving toward an ultrasound transducer, the reflected beam has a higher frequency than the transmitted beam. If the orientation of the transducer with respect to the blood vessel remains fixed, the amount of frequency shift produced is proportional to the velocity of blood cells. These facts may be stated mathematically and more precisely by means of the Doppler equation:

$$V = C \cdot \Delta f/(2 \cdot fe \cdot \cos\alpha)$$

where V = velocity of blood flow, C = velocity of sound in tissue (approximately 1540 m·s^{-1}), Δf = measured Doppler frequency shift (i.e., difference between the frequencies of emitted and reflected signals), fe = frequency of the emitted ultrasonic signal, and α = angle of incidence between the direction of blood flow and the direction of the emitted ultrasound beam.

As indicated in the Doppler equation, the flow velocity measured by Doppler ultrasound is the velocity of blood flow relative to the vector of the ultrasound beam (Fig. 1). Thus, the angle between the blood flow and the

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This manuscript was presented in part at the 64th Scientific Sessions of the American Heart Association in New Orleans, USA, November 1992, and at the 40th annual meeting of the Japan Society of Anesthesiology in Morioka, Japan, April 1993.
Received for publication on March 28, 1995; accepted on December 4, 1995

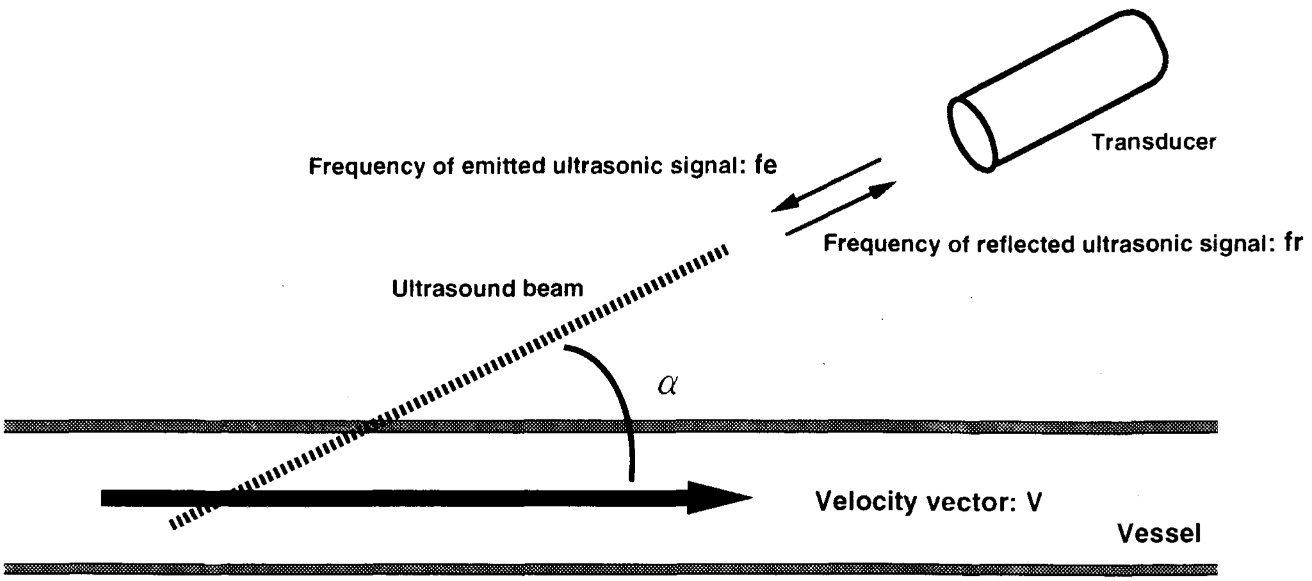

Fig. 1. Schematic diagram of velocity measurement by Doppler ultrasound. The flow vector V and the ultrasound beam create the incidence angle (Doppler angle) α. The flow velocity measured by Doppler ultrasound is proportional to the amount of frequency shift by means of the Doppler equation, and is influenced by the Doppler angle. *V*, blood flow velocity; α, angle of incidence between the direction of blood flow and the direction of the ultrasound beam; *fe*, frequency of emitted ultrasound; *fr*, frequency of reflected ultrasound

incident ultrasonic beam (Doppler angle) influences the flow velocity measurements. For quantitative measurements of blood flow velocity, the Doppler angle must be determined accurately or must be as minimal as possible. Especially when the velocity is measured by intravascular ultrasound, the angle of incidence cannot be determined, and thus it is doubtful whether the velocity measured by Doppler ultrasound is accurate.

Several techniques have been developed that overcome the dependence of the velocity measurement on the Doppler angle [4–10]; however, none of these is suitable for clinical application. We have developed a Doppler catheter incorporating a pair of transducers positioned at a fixed angle to obtain accurate flow velocity measurements independently of the Doppler angle.

Methods

Principle of velocity measurements independent of angle of incidence

To measure blood flow velocity independently of the angle of incidence of the ultrasound beam to the blood flow, it is essential to obtain velocity estimates from at least two different angles of incidence, as illustrated in Fig. 2. The flow vector V and the ultrasound beam 1 create the angle of incidence (Doppler angle α), and the flow vector V and beam 2 create another Doppler angle (α + θ). According to the Doppler equation, the Doppler shift for beam 1 (Δf1) will be given mathematically by the first of the following equations (Eq. 1), while the Doppler shift for beam 2 (Δf2) will be given by the second equation (Eq. 2).

$$\Delta f1 = C^{-1} \cdot 2 \cdot fe \cdot V \cdot \cos\alpha \quad (1)$$
$$\Delta f2 = C^{-1} \cdot 2 \cdot fe \cdot V \cdot \cos(\alpha + \theta) \quad (2)$$

where two ultrasound beams (beam 1 and beam 2) with two different angles of incidence (α, α + θ) generate Doppler shifts Δf1, Δf2, respectively. Note that C = velocity of sound in tissue, fe = frequency of the emitted ultrasonic signal, V = velocity of blood flow, α = the unknown angle of incidence of the emitted ultrasound beam 1, and θ = the known angle between the direction of the emitted ultrasound beam 1 and that of ultrasound beam 2. From these two equations, we can eliminate the unknown angle α, and the true velocity V can be given by the next equation (Eq. 3).

$$V = (C/(2 \cdot fe \cdot \sin\theta))\ ((\Delta f1)^2 - 2 \cdot \Delta f1 \cdot \Delta f2 \cdot \cos\theta + (\Delta f2)^2)^{1/2} \quad (3)$$

The true velocity (V) can be calculated from two Doppler shifts (Δf1, Δf2); however, with this method, two ultrasound beams need to be emitted with different Doppler angles to the same points of blood flow. Although this method seems to be theoretically valid, it is not as easily applicable to clinical use because manufacturing of the Doppler probe and the velocity measurement device is complicated. To improve clinical

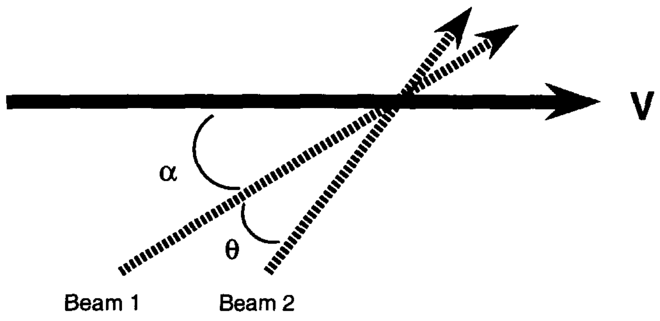

Fig. 2. Schematic diagram of true velocity measurement with Doppler ultrasound independently of angle of incidence. The flow vector *V* and the ultrasound beam 1 create Doppler angle α, and beam 2 creates the Doppler angle (α + θ). According to the Doppler equation, the Doppler shifts for beam 1 (Δf1) and beam 2 (Δf2) will be given mathematically by Eqs. 1 and 2 (see "Methods"). From these two equations, the unknown angle α can be eliminated, and the true velocity V can be given by Eq. 3. V, blood flow velocity; α, unknown angle of incidence of the emitted ultrasonic beam 1, θ, known angle between the direction of beam 1 and that of beam 2

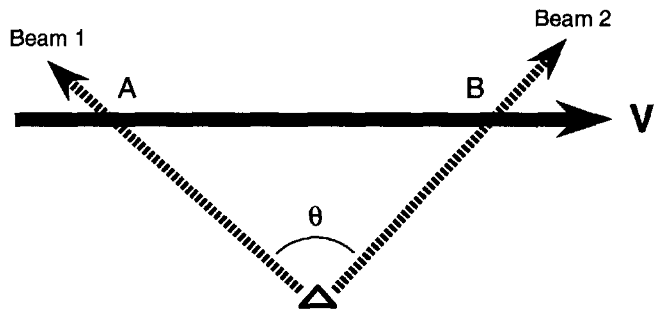

Fig. 3. Schematic diagram of true velocity measurement with the newly developed Doppler catheter independently of angle of incidence. When a flow field is sampled at two closely spaced points A and B, the true velocity can be calculated independently of the angle of incidence using Eq. 3 (see "Methods"). *A*, sampling point of ultrasonic beam 1; *B*, sampling point of ultrasonic beam 2

applicability, we developed a new method which is illustrated in Fig. 3. Since we had some technical difficulties in developing a catheter using the configuration in Fig. 2, we approximated the configuration in Fig. 2 by using another approach shown in Fig. 3, assuming the velocities in positions A and B are equal under ideal conditions. When a flow field is sampled at two closely spaced points A and B, as illustrated in Fig. 3, the true velocity can be calculated independently of the angle of incidence by Eq. 3. We have designed a Doppler probe to implement the methodology described above.

Doppler catheter

Our newly developed Doppler catheter incorporated a pair of adjoining ultrasound transducers located on one side of the catheter (Fig. 4). The catheter was 2.0 mm in diameter. The transducers were located 5 mm proximal to the catheter tip, and were 1 mm in length and 1 mm in width, respectively. The sampling depths were adjustable from 2 mm to 30 mm. The transducers were deliberately positioned at 90° with respect to each other, so that Eq. 3 could be simplified and the true velocity V could be given by the following equation (Eq. 4):

$$V = (C/(2 \cdot fe)) \cdot ((\Delta f1)^2 + (\Delta f2)^2)^{1/2} \qquad (4)$$

The 90° angle was chosen because of the invariability and ease of probe manufacturing and management. The distal transducer emitted the ultrasound beam forward at an angle of 45° to the catheter, and the proximal transducer emitted backward at an angle of 45°, as illustrated in Fig. 3. The true velocity V and Eq. 4 can be expressed by velocities as follows:

$V = ((V_1)^2 + (V_2)^2)^{1/2}$, where V = true velocity, and V_1 and V_2 = velocities detected by transducers 1 and 2.

The velocities detected by transducers 1 and 2 were recorded as shown in Fig. 5.

With our Doppler probe, two ultrasound beams were unable to determine the flow velocities at the same sampling point, but provided sampling at two closely spaced points in the axial flow; thus, errors were probably minimal in the velocity measurements.

Doppler flow velocity measurements

To measure the flow velocity using our new catheter, we connected the Doppler catheter to a velocity measurement device and measured the velocity by multi-range-gated pulsed Doppler ultrasound. A driving signal of 20 MHz and a pulse repetition frequency (PRF) of 40 KHz were then applied to the catheter to measure flow velocities. Peak detectable velocity under the emitted frequency of 20 MHz and a PRF of 40 KHz was 76.5 $cm \cdot s^{-1}$. The sample volume was 1 mm thick and the sampling depth was changeable. The Doppler shifts (Δf1 and Δf2) detected by transducers 1 and 2, respectively, were used to compute two velocity estimates and the true velocity estimate (Eq. 4). The velocities were calculated in real time using newly developed phase differential techniques consisting of modified autocorrelation techniques. Newly developed phase differential techniques do not include analysis of the flow velocity frequency spectrum such as fast Fourier transformation methods, and are superior in rapidity of calculation to conventional methods. In addition, the mean velocity over 5 s was digitally displayed and updated every second.

In vitro experiments

To test the validity of our new catheter, a continuous-flow model was set up with 16-mm-diameter polyvinyl

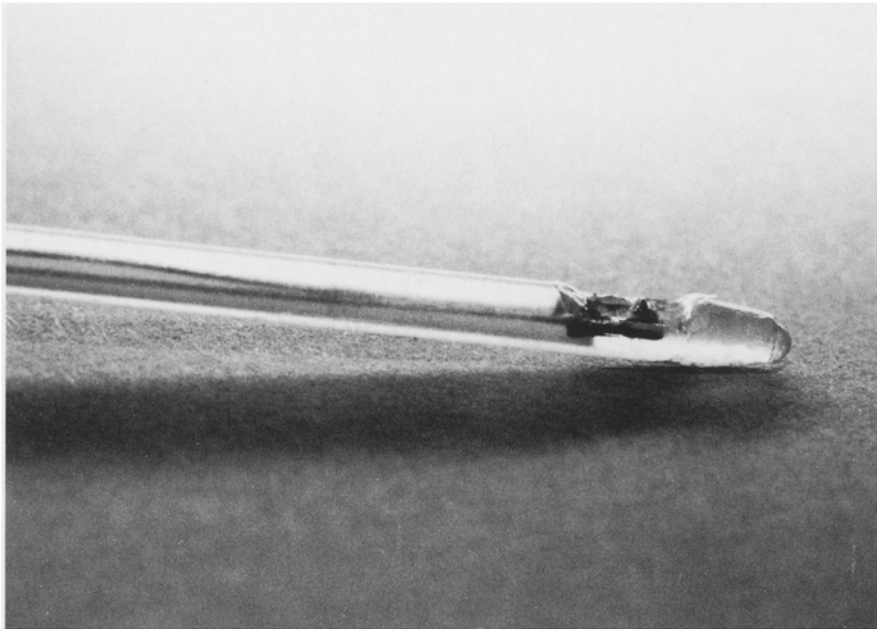

Fig. 4. Our newly developed Doppler probe, which incorporates a pair of adjoining ultrasound transducers (1 × 1 mm) located on one side of the catheter. The transducers are deliberately positioned at 90° with respect to each other

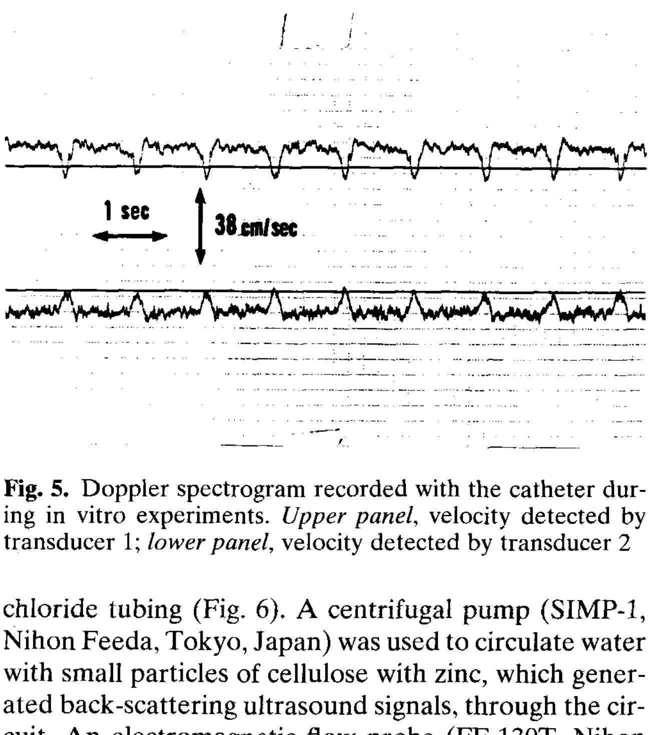

Fig. 5. Doppler spectrogram recorded with the catheter during in vitro experiments. *Upper panel*, velocity detected by transducer 1; *lower panel*, velocity detected by transducer 2

chloride tubing (Fig. 6). A centrifugal pump (SIMP-1, Nihon Feeda, Tokyo, Japan) was used to circulate water with small particles of cellulose with zinc, which generated back-scattering ultrasound signals, through the circuit. An electromagnetic flow probe (FF-130T, Nihon Kohden, Tokyo, Japan) was placed into the circuit and connected to a flowmeter (MFV-3200, Nihon Kohden). The Doppler catheter was then advanced by 10 cm into the circuit through the water bath and connected to the velocity measurement system. The catheter was placed on the bottom wall of the PVC tube at random. The flow rate was varied from 0.42 to 4.49 l·min^{-1} as detected by the electromagnetic flowmeter. The flow velocity was measured by the Doppler catheter (Doppler velocity) at five sampling depths (2.3, 4.6, 6.9, 9.2, and 11.5 mm), and was compared with the mean velocity calculated mathematically from the volumetric flow rate measured by the electromagnetic flowmeter (EMF velocity). The EMF velocity was calculated from the volumetric flow rate Q, measured by the electromagnetic flowmeter according to the following equation:

$$\text{EMF velocity (cm}\cdot\text{s}^{-1}) = Q \cdot 10^5 / (60 \cdot \pi \cdot r^2)$$

where Q (l·min^{-1}) = volumetric flow rate measured by the electromagnetic flowmeter, and r (mm) = radius of the circuit tube.

At the maximal velocity sampling depth, which was of 4.6 mm, the flow velocity was further measured and was compared with the EMF velocity.

Finally, the flow velocity measured with our catheter at the sampling depth of 3.8 mm was compared with the volumetric flow rate measured by an electromagnetic flowmeter (Q EMF).

Statistical analysis

For all flow velocity experiments, analysis of data was performed by standard linear regression with calculation of r^2, slope, intercept, and SEE.

Results

Correlations of the Doppler velocity with the EMF velocity

The Doppler velocities measured at five sampling depths correlated significantly with the EMF velocities (r^2 = 0.99, respectively) (Fig. 7, Table 1). At the sam-

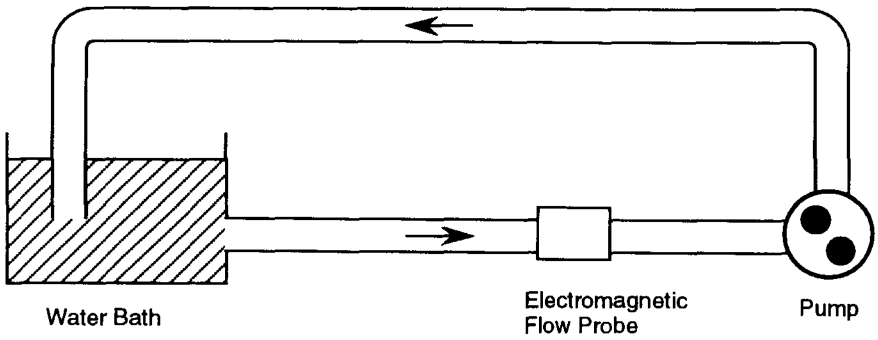

Fig. 6. Schematic diagram of in vitro water bath system

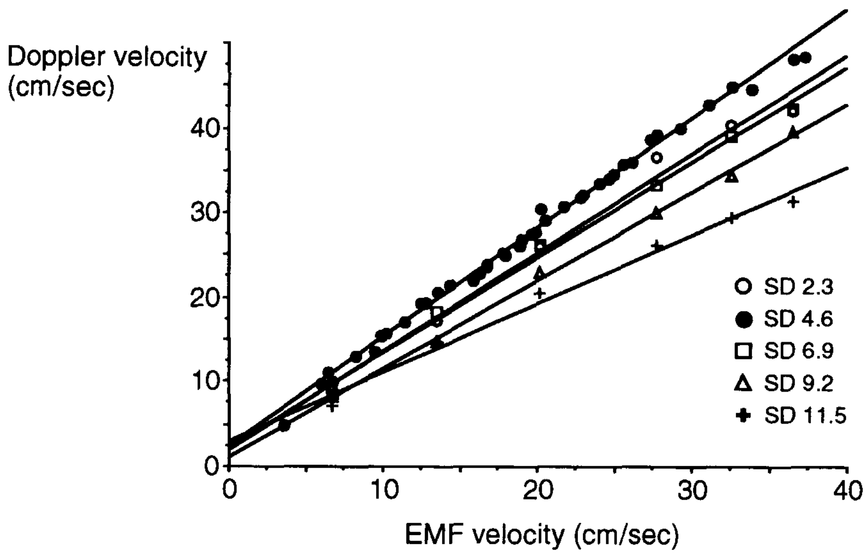

Fig. 7. Scatterplots of correlations between the velocities measured by the Doppler catheter (*Doppler velocity*) at five sampling depths and the velocities calculated from volumetric flow rate measured by electromagnetic flowmeter (*EMF velocity*). *Open circles*, velocity at sampling depth of 2.3 mm; *solid circles*, velocity at sampling depth of 4.6 mm; *squares*, velocity at sampling depth of 6.9 mm; *triangles*, velocity at sampling depth of 9.2 mm; *crosses*, velocity at sampling depth of 11.5 mm

Table 1. Correlations between the velocities measured by the Doppler catheter (Doppler velocity) at five sampling depths and the velocities calculated from volumetric flow rate measured by electromagnetic flowmeter (EMF velocity)

Sampling depth (mm)	r^2	SEE ($cm \cdot s^{-1}$)	Slope	Intercept ($cm \cdot s^{-1}$)
2.3	0.986	0.070	1.17	1.83
4.6	0.995	0.015	1.29	2.47
6.9	0.993	0.046	1.13	2.23
9.2	0.997	0.027	1.05	1.08
11.5	0.987	0.047	0.81	3.00

pling depth of 4.6 mm, the Doppler velocity (y) correlated strongly with the EMF velocity (x), and the regression equation was $y = 1.29x + 2.46$ ($r^2 = 0.99$, $P < 0.0001$, $n = 41$, SEE $= 0.015$).

Correlations of the volumetric flow rate measured by electromagnetic flowmeter with the flow velocity measured by the Doppler catheter

The flow velocity measured by the Doppler catheter (y) at the sampling depth of 3.8 mm strongly correlated with the volumetric flow rate measured by the electromagnetic flowmeter (x) ($y = 8.54x + 4.13$, $r^2 = 0.99$, $P < 0.0001$, $n = 125$, SEE $= 0.065$) (Fig. 8).

Discussion

Doppler ultrasound is widely used to measure blood flow velocity in medical practice [1–3], and has progressed to applications in intravascular ultrasound [11–19]. One of the problems in velocity measurement by Doppler ultrasound is the angle of incidence between the ultrasound beam and the blood flow. The Doppler angle is adjusted as minimally as possible using B-mode imaging and/or Doppler color flow imaging, if possible. Several investigators have reported techniques that overcome the dependence of velocity measurements on the Doppler angle of incidence [4–10]. However, no technique has been developed for clinical use. To measure true blood flow velocity independently of the Doppler angle, it is necessary to obtain velocity estimates from at least two different angles of incidence [7–10,20]. Previous devices for velocity measurement indepen-

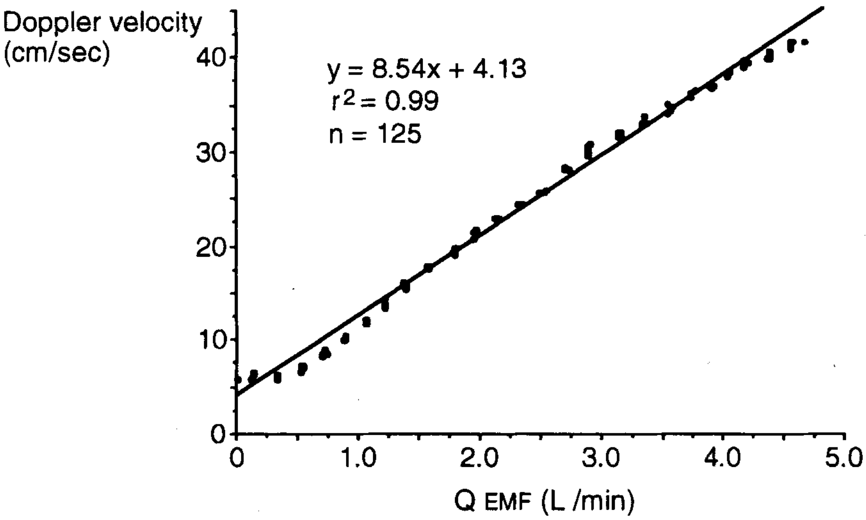

Fig. 8. Scatterplot of correlation between the velocity measured by the Doppler catheter (*Doppler velocity*) at sampling depth of 3.8 mm and the volumetric flow rate measured by electromagnetic flowmeter (*Q EMF*)

dently of the angle of incidence have not been satisfactory for clinical use due to the technical difficulty of emitting the ultrasound to the same sampling point from two directions. We therefore approximated one sampling point to two closely spaced points, assuming that velocities are equal between two points.

Our method can miniaturize the Doppler probe so as to facilitate its handling in clinical use. Our Doppler catheter positioned a pair of transducers at a fixed angle, enabling us to measure true flow velocity accurately, independently of the Doppler angle of incidence. For our catheter, we assumed that the velocities at two sampling points were equal. This assumption is valid when the sampling points are closely spaced to each other, relative to the large diameter of the vessel. Adequate sampling of the axial flow is easy in blood flow with a flat velocity profile, which allows us to minimize the error in velocity measurements.

In our in vitro experiments with circulating water flow, the velocity measured by the Doppler catheter at random angles of incidence closely correlated with the mean velocity calculated from the volumetric flow rate, which was not affected by the angle of incidence. These data suggest that the velocities measured with our catheter at various angles of incidence are almost equal to the true velocity and that our Doppler catheter and the velocity measurement device allow us to measure true flow velocity independently of the angle of incidence of ultrasound beams.

Study limitations

These results verified the validity of the theory of velocity measurement and the proper function of the measurement device in in vitro experiments. In our tests, we assumed that the velocities at two sampling points were equal, and that the sampling points had adequate axial flow. This assumption has been validated by in vitro experiments, but it remains unconfirmed in clinical applications. The sample volume should be positioned proximal to the lumen center to derive accurate flow velocity measurements. Further study is needed regarding clinical use of our Doppler catheter.

Conclusions

We conclude that our new Doppler catheter incorporating a pair of transducers positioned at a fixed angle can enable us to measure true flow velocity independently of the angle of incidence of the ultrasound beam. This measurement by our Doppler catheter can improve the accuracy of velocity measurements, especially for purposes of intravascular ultrasound. Our method can provide new quantitative information on blood flow. An expected clinical benefit of our Doppler catheter is the continuous measurement of blood flow velocity in vessels, e.g., the pulmonary artery. Clinical use of our technique could consist of monitoring hemodynamics by intravascular Doppler ultrasound.

Acknowledgments. We gratefully acknowledge the technical assistance of Kazuhiko Uehara. We are also grateful to Tadashi Tamura for his help in preparation of the manuscript. This study was supported in part by a Grant-in-Aid for Scientific Research from the Ministry of Education, Science and Culture (No. 05557068), Japan.

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