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Clinical Application of an Ultrasonic Duplex System in the Quantitative Measurement of Portal Blood Flow

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Abstract: The ultrasonic duplex system, which is composed of a pulsed Doppler flowmeter and a linear array B-mode scanner, was developed for the purpose of quantitative transcutaneous measurement of blood flow. It was applied here to the study of portal blood flow in a normal series with healthy volunteers as subjects. A preliminary experiment, prepared by setting up a simulation model for blood circulation, demonstrated an excellent correlation between values obtained with the electromagnetic flowmeter (EMF) and the combined ultrasonic B-mode and Doppler system (duplex system). This was confirmed *in vivo* in a study of blood flow in the canine vena cava. The portal blood flow in 88 healthy adult volunteers ranging in age from 18 to 55 years (average 34.3) was studied using this equipment. Blood flow in fasting subjects who were resting was 889 ± 284 ml/min (mean ± SD) or 16.3 ± 5.0 ml/min/kg body weight (mean ± SD).
Indexing Words: Portal blood flow · Ultrasonic duplex system · Pulsed Doppler flowmeter

The ultrasonic Doppler method using a continuous wave was originally devised by Satomura et al.[1] and applied by them in blood flow velocity measurements. Benchimol[2] and Nimura[3] applied the technique clinically in the analysis of cardiac cyclic phenomena. Since then, peripheral arterial blood flow has been widely studied using this ultrasonic Doppler technique.[4,5] However, in all these studies, the technique has been used mainly in qualitative analysis of blood flow near the body surface. The pulsed-wave Doppler method, a technique suitable for transcutaneous quantitative analysis of the blood flow velocity in deep-seated vessels, was developed in 1970.[6,7]

Recently, an ultrasonic duplex system composed of a real-time B-mode scanner and a pulsed Doppler flowmeter has been developed.[8–10] As a preliminary step towards clinical application of this combined B-mode/Doppler system (duplex system), we tested it on a circulation model. This study (1) enabled confirmation of its accuracy and reproducibility, (2) allowed the rheologic characteristics of the blood flow to be studied, and (3) provided information on the most reliable mode of measurement for clinical use.

To test the versatility of the duplex system *in vivo*, we compared its application directly with that of an electromagnetic flowmeter (EMF) on canine inferior vena caval preparations. We then measured the portal blood flow in healthy adults using the method obtained from the experimental study.

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METHODS AND MATERIALS

The Ultrasonic Doppler Duplex System

We used a newly developed combined ultrasonic B-mode/Doppler system (duplex system), the Toshiba SAL-50A/SDL-01A (Toshiba Corp., Tokyo, Japan). The main specifications of this device are

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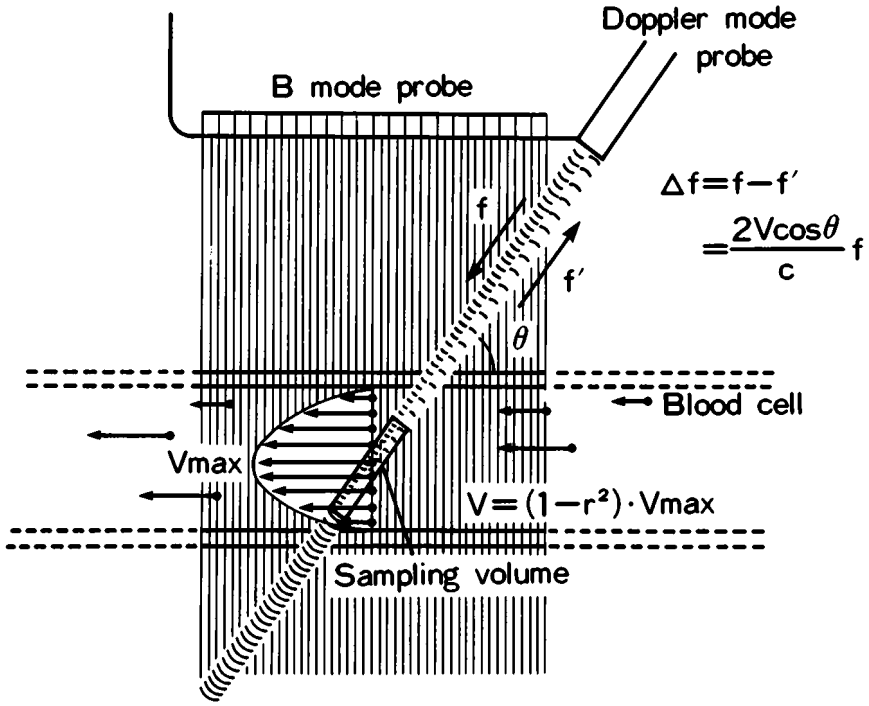

FIGURE 1. Scheme of the duplex system. *f*, frequency of emitted ultrasonic wave; *f'*, frequency of received ultrasonic wave; Δf, Doppler shift frequency; *c*, velocity of ultrasonic beam in body (approximately 1,530 m/s); θ, angle between direction of flow and Doppler beam; *V*, velocity of blood flow (or circulating blood cells); V_{max}, maximum velocity at center of lumen; *v*, velocity of flow at *r*; *r*, distance from center of vessel divided by radius of lumen ($0 < r < 1$).

as follows (Fig. 1). The B-mode electronic linear array scanner had a frequency of 5.0 MHz, a pulse repetition frequency of 3.6 kHz, axial resolution of 1.0 mm, lateral resolution of 1.5 mm, and an effective scanning width of 5.5 cm.

The Doppler mode flowmeter had a frequency of 2.268 MHz, a pulse repetition frequency of 4 or 6 kHz, a focal zone of 7 to 11 cm, at which the width of the beam was about 2 mm (−3 dB). The incidence angle of the Doppler beam was 55° (fixed), and the real-time measurable depth under B-mode monitoring was 3 to 12 cm.

The path of the Doppler beam is depicted by superimposing a schematic broken line on the B-mode display. The selected sample point is also indicated by solid bar-shaped markers on the broken line. The depth of the sample point can be freely adjusted manually using the B-mode display control. The size of the sample volume is selected from 2-, 3-, 5-, or 10-mm settings, and this is also depicted on the B-mode display by the size of the marker.

The Doppler shift frequency signal is displayed alongside the B-mode image of the area under study and is simultaneously recorded on paper using a line scan recorder. The Doppler signal is audible through a built-in loudspeaker.

Circulation Model

To evaluate the correlation between the actual mean blood flow velocity and the velocity determined from the Doppler frequency shift, and especially to define the flow characteristics in a tubular structure like a blood vessel, a simulation model of blood circulation was set up.

As illustrated, bovine blood was circulated in silicon tubing placed in a water tank (Fig. 2). The Doppler shift frequency was recorded while the blood flow velocity was varied.

Actual blood flow was determined by directly measuring the volume of blood accumulating in the gauged container. Measurements were also carried out using silicon tubes with diameters of 7 mm and 11 mm and bovine blood with hematocrits of 40%, 30%, and 20%.

Canine IVC Model

To investigate the capability of the duplex system *in vivo*, the canine inferior vena cava (IVC) was prepared for simultaneous blood flow measurements using the EMF and our duplex system. The electromagnetic flowmeter (MFV-1100 by Nihon Kohden Co., Tokyo) was equipped with a probe

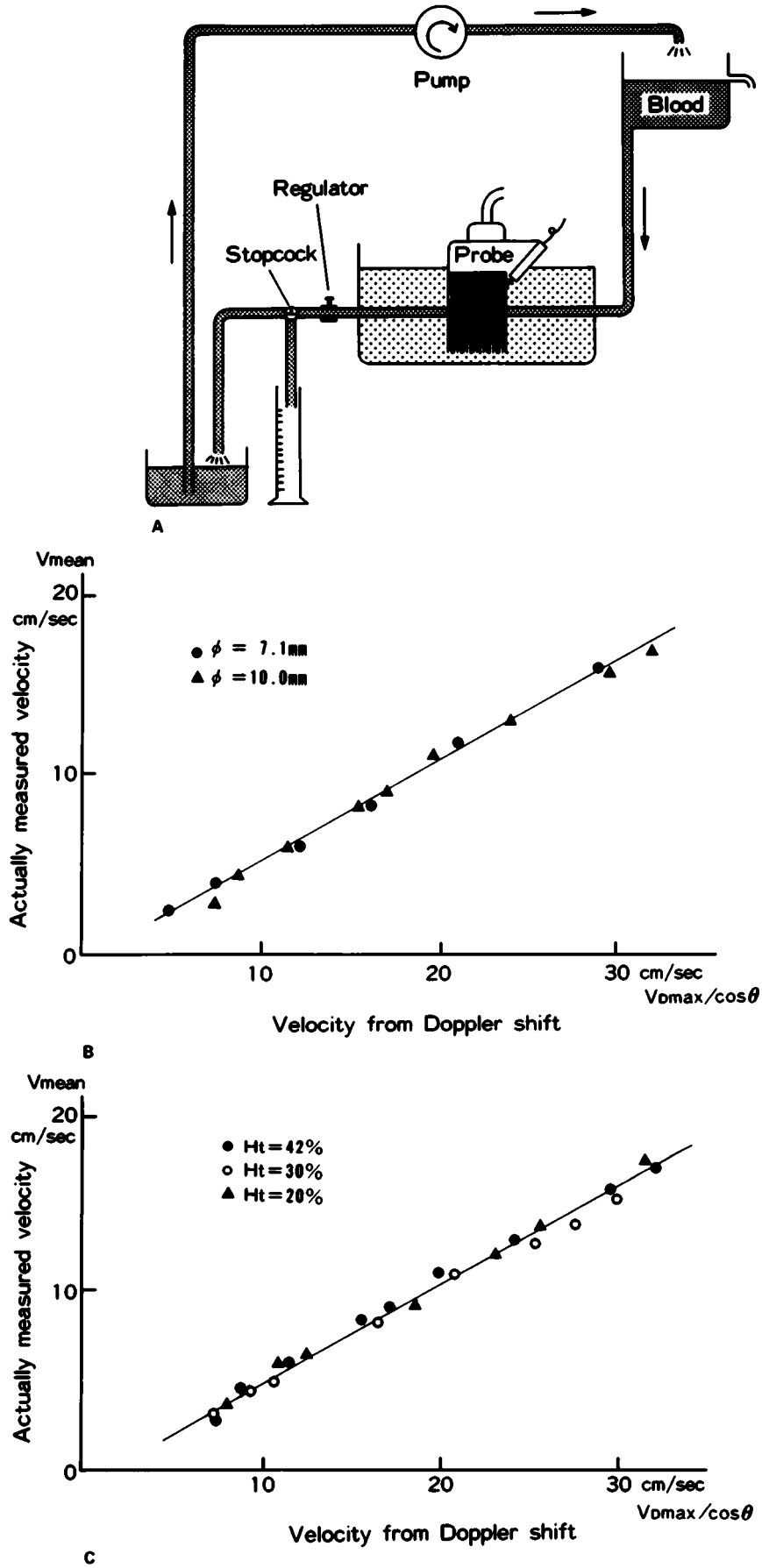

FIGURE 2. Circulation model **(A)**, relationship between actual velocity and velocity determined from Doppler frequency shift **(B)**, and effect of change in hematocrit (Ht) on determining velocity from Doppler frequency shift **(C)**.

having a diameter of 5 mm. The EMF was calibrated before the experiments using a circulation model composed of a silicon tube filled with saline. Two mongrel dogs were anesthetized and maintained on controlled mechanical ventilation. The infrarenal segment of the IVC was exposed through a midline abdominal incision, and the position of the surrounding tissue was adjusted to enable application of the EMF and the duplex system at the same time. A similar experiment was performed on each of the two dogs. Simultaneous blood flow measurements were carried out by placing the probe of the EMF tightly around the IVC just proximal (downstream) to the point at which the probe of the duplex system was aimed. Blood flow in the IVC was varied by placing a vascular tape around the vessel, creating a controlled stenosis.

The equation obtained from the experiments with the circulation model was used to calculate the mean flow velocity.

Portal Blood Flow in Healthy Volunteers

Eighty-eight members of the hospital staff, 44 males and 44 females with a mean age of 34.3 years, were studied. All volunteers were asymptomatic, and their routine physical examinations had consistently been within normal limits. All measurements were carried out in the early morning, 12 h after the subjects' last meal, and with them reclining against a back rest at an angle of 30°.

The portal vein was imaged longitudinally by the B-mode scanner while the Doppler beam penetrated the body from the caudal end of the B-mode probe. The sample point was set at the midpoint between the confluence of the splenic and superior mesenteric veins and the bifurcation of the portal vein at the hepatic hilus (Fig. 3). At this location, the portal vein has few branches, runs straight, and has a cross-sectional image that is more or less round, making it amenable to simple calculation of the cross-sectional area of the vessel. The sample volume of the duplex system was set at 5 mm. To obtain the cross-sectional area of the portal vein, it was scanned at right angles at the sample point, and the image was photographed using Polaroid™ film. Major and minor axes were measured on the photograph, and the formula for the area of an ellipse was used to calculate the cross-sectional area of the vessel. To

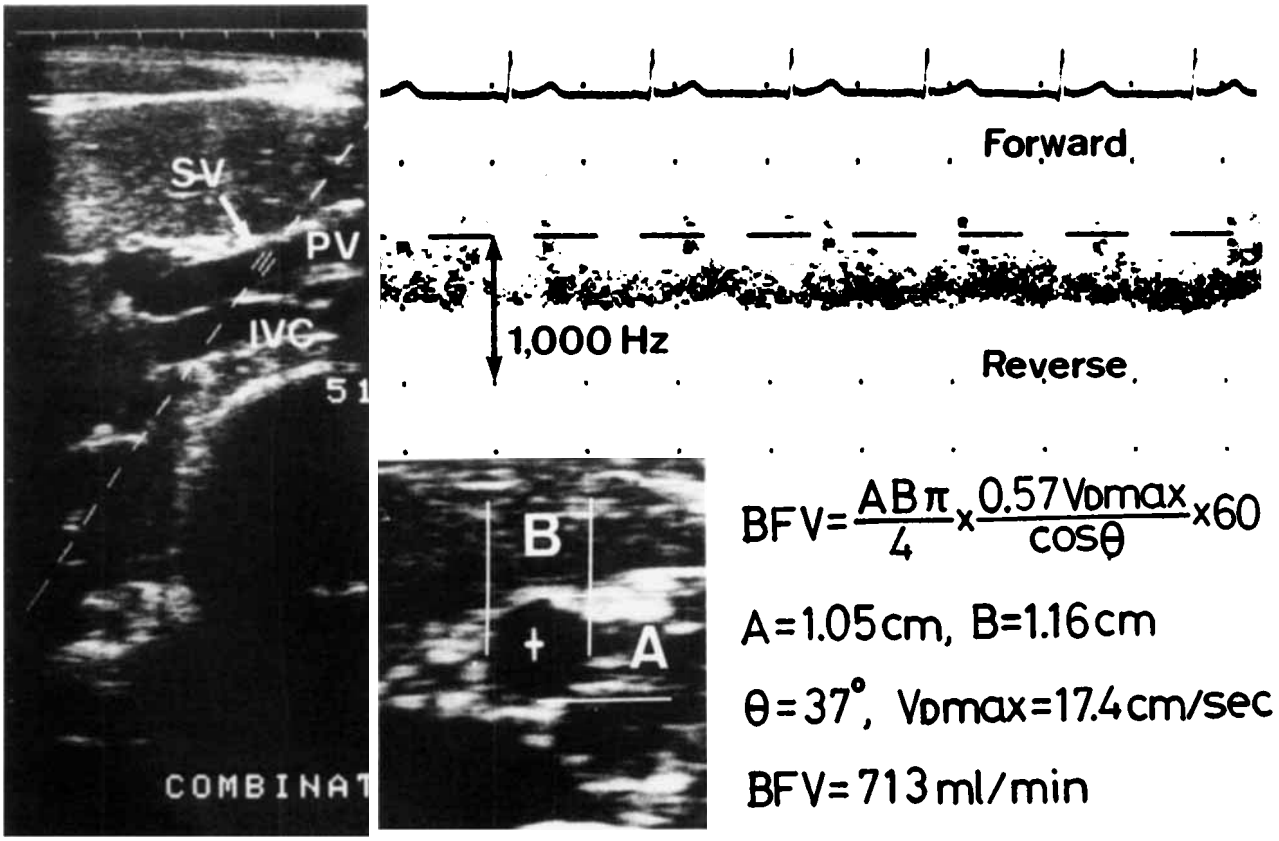

FIGURE 3. B-mode scanning of portal vein (left side), Doppler spectrogram (upper right side), and cross-sectional image of portal vein (lower side) in normal healthy adult. PV, portal vein trunk; IVC, inferior vena cava; SV, sample volume; A, dimension of vertical axis; B, dimension of horizontal axis. Broken line in Doppler spectrogram indicates zero level.

minimize observer bias, the scanning was conducted under decreased levels of gain, dynamic range, and echo enhancement.

When the sample point was adjusted to the center of the lumen of the portal vein, the Doppler signals were recorded on paper.

The mean portal blood flow velocity was calculated using the results obtained from the experiments with the circulation model as follows. Maximum velocity determined from the Doppler spectrogram (V_{Dmax}) was multiplied by 0.57 and divided by cos Θ, where Θ is the angle between the portal vein and the Doppler beam measured on the B-mode display. The coefficient 0.57 is the ratio of the mean velocity and the maximum velocity obtained from the experiments with the circulation model (see results on circulation model).

The blood flow (BF) of the portal vein was calculated from the following equation:

$$BF = \frac{A \cdot B\pi}{4} \times \frac{0.57\ V_{Dmax}}{\cos \Theta} \times 60\ (ml/min)$$

where A is the diameter of the portal vein from top to bottom and B is the diameter from left to right.

In this case, V_{Dmax} is a vector component of the maximum velocity at the central axis in the direction of the ultrasonic beam of the Doppler mode.

Statistical analyses were performed using the Student's *t*-test.

RESULTS

Circulation Model

The maximum velocity obtained from the spectrogram (V_{Dmax}) was compared with the mean blood velocity measured directly (V_{mean}) (Fig. 2). The relationship is expressed in the following equation, where the correlation coefficient (r) is 1.000.

$$V_{mean} = -0.57 + 0.57\ V_{Dmax}/\cos \Theta$$

The velocity range that could be measured was from 2.3 cm/sec to 30.7 cm/sec. When the diameter of the silicon tube and the hematocrit of the circulating blood were changed, the coefficient, 0.57, was not affected.

Canine IVC Model

Figure 4 is a comparison of the results of *in vivo* blood flow measurements using the EMF and the duplex system. A positive correlation with a correlation coefficient of $r = 0.965$ was obtained in the range of 75 ml/min to 500 ml/min ($p < 0.001$).

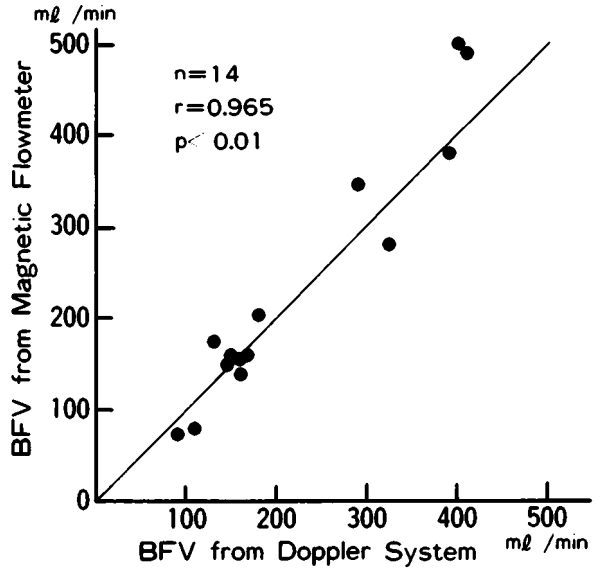

FIGURE 4. Relationship between electromagnetic flowmeter and ultrasonic duplex system. BFV, blood flow.

Portal Blood Flow Measurement in Healthy Volunteers

The results are summarized in Table 1 and Figs. 5–7. The average angle of incidence between the Doppler beam and the portal vein was 39.3 ± 8.0° (mean ± SD, $n = 88$). Mean age, height, body weight, body surface area, and cross-sectional area of the portal vein, along with the measurement of the portal blood flow (mean ± SD), were obtained and analyzed for the male and the female subjects, respectively.

Correlation coefficients were as follows:

Height/blood flow volume: $r = 0.567$ ($p < 0.01$)

Body weight (BW)/blood flow volume: $r = 0.407$ ($p < 0.01$)

Body surface area (BSA)/blood flow volume: $r = 0.437$ ($p < 0.01$)

A weak positive correlation was obtained for each of the above pairs. Significant differences between the sexes ($p < 0.01$) were observed in height, body weight, body surface area, cross-sectional area of the portal vein, and portal blood flow volume. No difference was observed in blood flow velocity. The differences according to sex disappeared when the portal blood flow was expressed either per BW or per BSA.

TABLE 1
Portal Blood Flow Measurement in Healthy Volunteers

	Total (88)	Male (44)	Female (44)
Subjects studied			
Age (years)	34.3 ± 12.3	33.4 ± 11.5	35.1 ± 13.0
Height* (cm)	161.0 ± 9.0	168.0 ± 7.0	154.0 ± 5.0
Body weight* (kg)	55.2 ± 7.9	59.8 ± 7.0	50.5 ± 5.9
Body surface area* (m^2)	1.6 ± 0.2	1.7 ± 0.1	1.5 ± 0.1
Ultrasound measurements			
Velocity of blood flow (cm/s)	15.3 ± 4.0	15.1 ± 4.0	15.6 ± 3.9
Cross-sectional area of portal vein* (cm^2)	0.99 ± 0.28	1.08 ± 0.28	0.90 ± 0.25
Blood flow (BF)* (ml/min)	889 ± 284	965 ± 303	813 ± 239
BF/body weight (ml/min/kg)	16.3 ± 5.0	16.3 ± 5.3	16.3 ± 4.5
BF/body surface (ml/min/m^2)	560 ± 168	573 ± 181	546 ± 154

*Items with significant sex difference ($p < 0.01$). Data expressed as mean ± SD.

DISCUSSION

Ultrasonic B-mode Combined Doppler System

Since the introduction of continuous-wave and, subsequently, pulsed-wave Doppler techniques for blood flow determination, the arterial system has been the primary object of study. However, in measuring blood flow using the pulsed Doppler technique, the intraabdominal venous channels, especially the portal venous system, have considerable advantages over the heart and great vessels because (1) the blood flow is of relatively low velocity, (2) the flow lacks turbulence and is laminar unless pathological conditions are involved, and (3) no phasic changes in velocity can be seen except in the inferior vena cava close to the atrium.

Several disadvantages, however, exist: the vessels are rather small, there are many of them intertwined within a small space, and the straight segment of the vessel is relatively short. These disadvantages require precise localization of the vessels that are to be studied.

The addition, therefore, of a high-resolution real-time B-mode scanner to the pulsed Doppler flowmeter becomes desirable. Furthermore, because

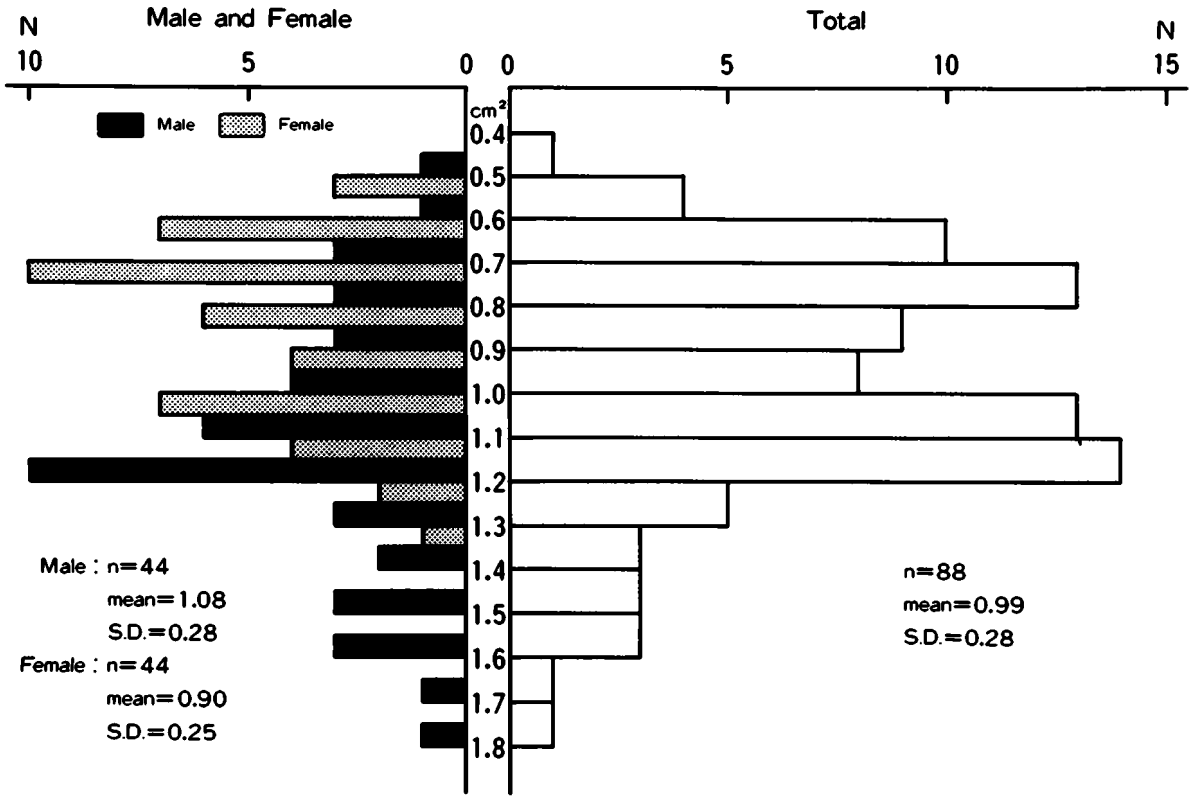

FIGURE 5. Cross-sectional area of portal vein in healthy volunteers. Distribution is wide and bimodal because of sex difference.

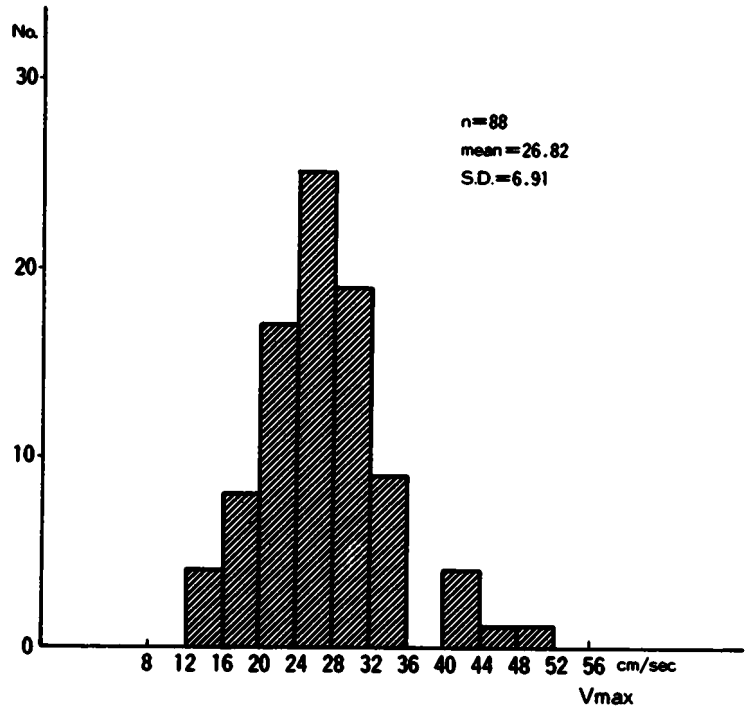

FIGURE 6. Distribution of portal blood velocity (maximum velocity in healthy volunteers).

the majority of the vessels with which we are concerned run parallel to the body surface, the linear array scanner is preferable to a sector scanner.

Thus, the idea of an ultrasonic duplex system was born, in which a linear-array B-mode scanner was combined with a pulsed-wave Doppler flowmeter. With the pulsed-wave technique for the Doppler velocimetry, one can select the Doppler signals that are backscattered from the blood cells from any given portion of the intraabdominal vascular network.

Errors arise mainly from measurement of the cross-sectional area and in guiding the sample volume to the target vessel, especially when measuring smaller vessels such as the portal vein. To reduce the degree of error, a B-mode with high resolution, a wide dynamic range, and an adequate field of view should be used. The duplex system, however, is still not without the technological problems that may arise from the permanent combination of two different probes in a single head (Fig. 1):

1. Deviation of the sample point resulting from differences in the velocities of the sonic waves as they pass at different angles and from different points of origin through tissues of varying densities, such as adipose tissue, muscle layers, and liver tissue.
2. Deviation of the sample point because of refraction.
3. Possibly imperfect surface contact of the Doppler-mode probe, which has to be applied obliquely to the skin.
4. The longer path of the Doppler mode beam because of the oblique transmission, which necessitates a higher ultrasonic beam intensity.

In spite of the need for further refinement in the construction of the system to deal with these problems, the duplex system we used showed a satisfactory capacity for transcutaneous measurement of portal blood flow and will certainly become a preferred method for noninvasive blood flow measurement in the clinical study of hepatic hemodynamics.

Circulation Model and Canine IVC Model

Our experiment of circulating bovine blood in silicon tubing demonstrated a peculiarity in the hemodynamics of the blood flow as well as the accuracy and reproducibility of the duplex system. The relationship between the directly measured velocity (V_{mean}) and the maximum velocity determined from the Doppler spectrum mode ($V_{Dmax}/\cos\Theta$) was expressed by the equation $V_{mean} = 0.57\ V_{Dmax}/\cos\Theta$.

According to the principles of hydrodynamics, the vector profile of the velocity of a laminar flow in a cylindrical tube, which approximates the venous circulation, is parabolic, with the maximum

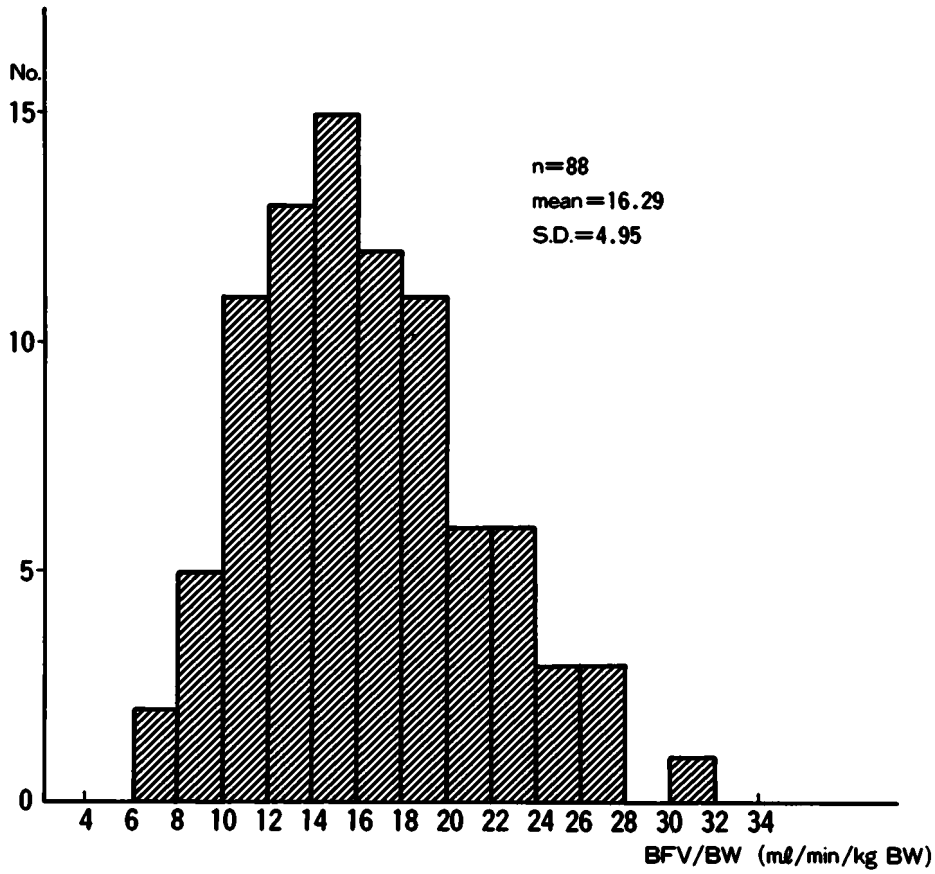

FIGURE 7. Distribution of portal blood flow rate per kilogram body weight in healthy volunteers.

velocity at the center of the lumen (V_{max}) and decreasing velocities toward the vessel wall. This is expressed by the equation, $V = (1-r^2)\, V_{max}$, where V is the velocity of the flow at r, which is the distance from the center of the vessel divided by the radius of the lumen ($0 < r < 1$).[14] The mean velocity (V_{mean}) is obtained by integral calculation of this three-dimensional structure, where

$$V_{mean} = 2 \int_{r=0}^{r=1} (1 - r^2)\, r dr\, V_{max} = \tfrac{1}{2}\, V_{max}$$

The discrepancy between the theoretically derived relationship and that which is directly measured can be explained as follows. The lower density of the red cells and the resultant relatively lower viscosity of the blood in the vicinity of the vessel wall relative to the center of the lumen result in a higher blood flow in the periphery of the lumen than was determined theoretically. That is, blood is not a Newtonian fluid, and the velocity profile cannot be a perfect parabola even under ideal conditions. This leads to:

$$V_{mean} > \tfrac{1}{2}\, V_{max}$$

Regardless of this discrepancy, the maximum velocity determined from the Doppler spectrum display ($V_{Dmax}/\cos\Theta$) and the directly measured mean blood velocity showed the best linear correlation in the velocity range of 2 to 30 cm/s. We therefore decided, in further experiments, to determine the mean blood velocity from V_{Dmax} using the equation $V_{mean} = 0.57\, V_{Dmax}/\cos\Theta$.

Furthermore, although the diameter of the vessel and the hematocrit of the circulating blood were changed, the ratio of V_{mean}/V_{Dmax} remained constant. Therefore, the following reasons for determining maximum velocity and then calculating the mean velocity instead of using mean velocity directly can be cited: (1) the ratio is constant regardless of the caliber of the blood vessel and the hematocrit; (2) it is necessary to cover the entire lumen in order to measure the mean velocity directly,[8] this being difficult because there are many blood vessels in the abdomen that run close to each other; and (3) because it is necessary to make the sample volume smaller than the size of the vessel since, in small vessels, the Doppler signal from the flowing blood cells is relatively small compared with the noise caused by the motion of the vessel wall.

To take the motion of vessels caused by breathing or heartbeat into consideration, a sample volume of approximately one-half the diameter of the vessel was considered to be adequate for the measurement.

Of the methods available for *in vivo* blood flow measurement, EMF, although invasive, has been considered to be the standard against which all other methods are judged. We compared the duplex system with the EMF in *in vivo* measurements and obtained a satisfactory correlation between these two methods.

Portal Blood Flow in Healthy Volunteers

Hepatic blood flow is currently studied using one of the following techniques: measurements of ICG,[15] radiocolloid[16,17] and radiolabeled bile salt uptake by the liver, the ^{85}Kr washout method,[18] and electromagnetic flowmeters.[19,20] According to the results obtained using these techniques, the hepatic blood flow in humans is in the range of 800 to 1,200 ml/min, about two-thirds of which is said to be portal blood flow.[20-23]

However, there are few published references on the successful and precise determination of portal blood flow in humans. Several reports on the measurement in anesthetized dogs, mainly using electromagnetic flowmeters, have been published. According to these reports, the portal blood flow in dogs is in the range of 75 to 90ml/min/100 g of liver.[24-28] Our results using the ultrasonic duplex system showed that the portal blood flow in a healthy adult is 16.3 ml/min/kg of body weight. Assuming that the liver is 2.0% of the body weight,[29-31] the human portal blood flow is 81.5 ml/min/100 g of liver, which is similar to the reported values in dogs. In this study, the values we obtained showed a weak correlation with other physical parameters: body height, weight, and surface area. Statistically, an r value of more than 0.6 is necessary for significance, though the values between 0.4 and 0.57 that we obtained may help confirm the reliability of this method. In general, the ultrasonic duplex system, composed of a pulsed Doppler flowmeter and a real time B-mode scanner, appears to be sufficiently reliable in spite of several potential sources of inaccuracy such as in the measurement of the Doppler frequency-shift spectrum, the incident angle, and the cross-sectional area of the vessel.

In conclusion, the fact that intraabdominal blood flow, particularly the portal blood flow, can be determined noninvasively with this ultrasonic duplex system under physiological conditions is of prime importance to the clinical study of hepatic pathophysiology. With further hardware and software sophistication, this new method will surely become an indispensable tool in the study of gastrointestinal hemodynamics.

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