Angle-Insensitive Flow Measurement Using Doppler Bandwidth

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Abstract

The ability to measure the velocity of blood flow independent of the orientation of the blood vessel could aid in evaluation of many disease processes, such as coronary lesions. Conventional ultrasonic Doppler techniques require knowledge of the beam-to-flow angle, and the Doppler effect vanishes when this angle is 90^o. By employing a spherically symmetrical range cell and the Doppler bandwidth instead of the Doppler shift, flow measurement of ideal uniform flow that has a blunt velocity profile can be made without knowledge of the orientation of the vessel, even when the angle of orientation is around 90^o. But when the technique is applied to a real flow that has a parabolic velocity profile, the Doppler bandwidth decreases as the beam-to-flow angle increases. Nevertheless, the technique can still be considered for larger blood vessels when the flow velocity in a much smaller range cell is approximately uniform, especially when the velocity profile and the range of beam-to-flow angle can also be estimated. The experimental techniques and results for flow measurements of both the ideal uniform flow and the real flow are presented in this paper.

I. Introduction

Ultrasonic pulsed Doppler flow sensing techniques have been employed to determine blood flow non-invasively [1]. Problems with these techniques, however, are that the transducer angle (the angle between the direction of the blood flow and the line-of-sight of the ultrasonic beam) must be known in order to compute the flow velocity, and that the Doppler effect vanishes when the transducer angle is 90^o. Newhouse *et al.* [2] showed that flow velocity measurement for a transducer angle of 90^o could be obtained using Doppler bandwidth instead of Doppler shift, but the measurement is still a function of the transducer angle. Melton *et al.* [3] showed an angle-insensitive flow measurement technique using bandwidth of the Doppler signal obtained from a range cell that has a spherically symmetrical excitation envelope. The basis of this method is that the amount of transit-time broadening of the Doppler bandwidth [4] as the blood cells moves with a constant velocity across the symmetrical range cell is the same for any flow direction, resulting in angle-insensitive flow measurement.

Experimental results for the angle-insensitive flow-measurement technique proposed by Melton *et al.* are presented in this paper. The measurements were performed on a gelatin flow-mimicking phantom (for mimicking the ideal uniform flow) that contained graphite scatterers, and on real flow in which a parabolic flow profile was established in a flow channel of circular cross section.

II. METHODS

All data were obtained with a twelve-ring annular array transducer (30 mm in diameter and focal length of 65 mm) that was fabricated at Hewlett-Packard Laboratory [5]. All twelve rings were impedance matched and connected in parallel so that they operated as a single element. The width of the beam at the focus was determined by exciting the transducer with a pulser and measuring the reflected signal from a spherical fused-quartz ball (600 μm in diameter) at the end of a fused-quartz capillary. The return signal was monitored as the ball was positioned at different locations across the beam at the focus. The profile was approximately gaussian, and the full-width at half of the maximum (FWHM) of the beam was found to be about 0.78 mm.

The range cell for flow measurement was produced with the annular array by sending bursts of ultrasound waves each with a gaussian excitation envelope. The return signal was sampled at a suitable time delay after the generation of each burst to position the center of the range cell at the focal point of the transducer. By choosing a proper burst length, the dimension of the gaussian envelope along the range direction matched that of the beam profile at the focus to create a spherically symmetrical range cell there.

A. Ideal Uniform Flow

To mimic ideal uniform flow, the transducer was moved with respect to a stationary phantom that contained a large number of scatterers, so that all the scatterers moved at the same velocity relative to the transducer. The phantom was constructed by mixing gelatin (300 Bloom from porcine skin) with graphite particles (sizes less than 44 μm) that served as scatterers. The transducer was moved in small steps by a Klinger [6] motion system, and each time after the transducer was moved, a burst of ultrasound waves was transmitted. After a proper time delay, 128 samples of the RF return signal were recorded centered about the range cell at the focus. The quadrature components of the samples were obtained by a Hilbert transform [7]. If complex sequence x represents the sequence of 128 real samples and complex sequence X represents the fast Fourier transform (FFT) of x, then

$$X = F(x) \tag{1}$$

Where F is the FFT operator. If complex sequence H represents the FFT of the Hilbert transform of x, then

$$H_k = i \operatorname{sgn}(k) X_k \tag{2}$$

Where the subscript k is the frequency index (k is 1 to 63 for the positive harmonics and is -1 to -63 for the negative harmonics), and the DC component X_0 and the Nyquist component X_{64} are set to zero. If h represents the Hilbert transform of x, then

$$h = F^{-1}(H) \tag{3}$$

Where F^{-1} is the inverse FFT operator. The 64'th sample (the indices for x and h run from 0 to 127) of the return signal x_{63} and that of its quadrature component h_{63} were saved in a computer for analysis.

B. Measurement System for Ideal Uniform Flow

The system block diagram is illustrated in Fig. 1. An Hewlett-Packard [8] HP 8175A arbitrary wave generator produces a gaussian envelope for amplitude modulation of a 4.9-MHz sinusoidal transducer excitation signal at the output of an HP 8116A pulse/function generator; the excitation signal is 16 V peak-to-peak at the peak of the qaussian excitation envelope. The return signal is amplified with a Metro Tek [9] MR101 ultrasonic receiver and then digitized with an HP 54112D oscilloscope. Through HPIB, an HP Vectra computer communicates with a Klinger motion controller to control the movement of the transducer along the direction of a stationary vessel phantom, and the computer obtains digitized samples from the oscilloscope for the computation of the quadrature components using Equations (1) - (3). After 1024 pairs (the return signals and their quadrature components) of data are obtained, the power spectrum is computed from a 1024-point complex FFT, and the Doppler bandwidth is determined as the square root of the variance of the spectrum [10]. If complex sequence y represents the 1024 pairs of real (x_{63}) and imaginary (h_{63}) data, complex sequence Y represents the FFT of y, and the real sequence S represents the power spectrum, then

$$Y = F(y) \tag{4}$$

$$S = |Y|^2 \tag{5}$$

The mean frequency $\bar{f}$ is the first moment of the power spectrum S divided by the zero moment. If m is the frequency index running from -511 to 512, and $\overline{m}$ is the mean frequency index, then

$$\overline{m} = \frac{\sum_{m=-511}^{m=512} m S_m}{\sum_{m=-511}^{m=512} S_m} \tag{6}$$

By multiplying $\overline{m}$ with a scale factor that relates to the mimicked pulse repetition frequency (PRF) f_r, the mean frequency $\bar{f}$ is expressed as

$$\bar{f} = \frac{f_r}{1024} \bullet \overline{m} \tag{7}$$

The bandwidth b is the square root of the variance, which is the mean-square deviation referred to $\bar{f}$, *i.e*

$$b = \frac{f_r}{1024} \bullet \sqrt{\frac{\sum_{m=-511}^{m=512} (m - \overline{m})^2 S_m}{\sum_{m=-512}^{m=512} S_m}} \tag{8}$$

Note that the same scale factor is used in Equation (8) to convert the bandwidth expressed in terms of frequency index to a value expressed in terms of frequency. Different flow velocity is mimicked by changing the distance of each step of the transducer movement. For example, a step size of 20 µm mimics a flow velocity of 20 cm/s at a PRF of 10 KHz, or a flow velocity of

10 cm/s at a PRF of 5 KHz, etc.. Although the summations indicated in (6) and (8) are carried out over the entire range of m, the actual computations use only the range of m for which the values of S_m are above the noise floor. In addition, it should be noted that the actual data-collection rate for the uniform flow is not at the mimicked PRF.

C. Velocity Measurement for Ideal Uniform Flow

The mechanical fixture for mounting the transducer permits flow measurement with transducer angles of 45^o, 60^o, 75^o, and 90^o. The relationship between bandwidth and flow velocity was studied with a transducer angle of 60^o, and the step size was increased from 4 μm to 20 μm with an increment of 4 μm to mimic flow velocities from 4 cm/s to 20 cm/s at a PRF of 10 KHz. The FWHM of the gaussian excitation envelope of the range cell, here called the two-way pulse length, was set to 0.6 mm by the arbitrary wave generator.

The Doppler bandwidth as a function of the four transducer angles was studied with a step size of 20 μm to mimic a flow velocity of 12 cm/s at a pulse repetition frequency of 6 KHz. In this case, a velocity of 12 cm/s instead of 20 cm/s was mimicked to facilitate comparison of these flow data with those obtained for the real flow. A set of measurement was obtained for each of three two-way pulse lengths: 0.4, 0.6, and 0.8 mm.

D. Real Flow

When a viscous fluid flows in a channel of circular cross section, the flow velocity increases from zero at the wall to a maximum value at the center of the channel. If the channel is sufficient long and the flow is laminar and steady, the velocity profile becomes parabolic, and the maximum velocity is twice the average velocity of the flow in the channel [11].

A real flow was established in a Doppler flow phantom Model 524 manufactured by ATS Laboratories, Inc. [12]. A Model 700 variable speed flow controller and pumping system from the same company drove a blood mimicking test fluid through a channel (6 mm in diameter) in the flow phantom. Sufficient amount of water was added on top of the phantom surface (the water was confined by special walls) to enable positioning of the range cell at the center of the channel, which was only 25 mm below the phantom surface. The flow velocity recorded for each measurement is the flow velocity at the center of the channel, which is equal to twice the average flow velocity set by the flow controller.

Flow measurement was performed using a PRF of 1 KHz in real time; each time after a burst of ultrasound waves was transmitted, 64 samples of the RF return signal were recorded centered about the range cell at the focus. Limited by the memory size of the digitizing oscilloscope, only 64 samples were acquired for the real flow instead of 128 samples as for the uniform flow. After samples of the return signal for 1024 bursts of ultrasound waves were collected, the raw data stored in the oscilloscope were sent to a computer for processing. Using Equations (1) - (3), the quadrature components of the samples of the return signal were obtained from a 64-point Hilbert transform of the samples. The 34'th sample of the acquired return signal and that of its quadrature component for each of the 1024 bursts were then saved in the computer for analysis.

E. Measurement System for Real Flow

The system block diagram is illustrated in Fig. 2. The basic architecture of this system is similar to that for the uniform flow. In this case, an HP 54512B digitizing oscilloscope has to store all the sample data obtained in real time for each velocity measurement before they are sent to an HP Vectra computer for processing. Also, an additional HP 8116A pulse/function generator is used here to trigger the HP 8175A arbitrary wave generator to initiate bursts of ultrasound waves at a PRF of 1KHz. Because the return signal obtained from the blood mimicking test fluid is much weaker than that obtained from the gelatin flow phantom, an RF power amplifier, ENI [13] Model 325LA, is used to amplify the transducer excitation signal to 130 V peak-to-peak at the peak of the qaussian excitation envelope.

Each flow velocity is estimated based on samples collected from 1024 ultrasonic bursts. After the computer has received the sample data, the quadrature components are computed. Using Equations (4) - (8), the Doppler bandwidth is determined as the square root of the variance of the power spectrum obtained from a 1024-point complex FFT.

F. Velocity Measurement for Real Flow

The mechanical fixture for mounting the transducer was the same as that for the uniform flow; it permits flow measurement with transducer angles of 45^o, 60^o, 75^o, and 90^o. The Doppler bandwidth as a function of flow velocity was determined for each of the four transducer angles by increasing the flow velocity at an increment of 4 cm/s from zero to the maximum value that can be measured with a PRF of 1 KHz and the specific transducer angle without aliasing. Flow velocity measurements were obtained using each of the three two-way pulse lengths: 0.4, 0.6, and 0.8 mm.

Limited by the physical dimensions of the flow phantom used in the real flow measurement, stationary structures in the vicinity of the flow channel reflected ultrasound waves and increased the magnitudes of the DC components in the power spectra. When the transducer angle was 90^o, the mean frequency was zero, so these interferences at a frequency of zero reduced the estimated Doppler bandwidths. In this case, Doppler bandwidths were also estimated with the DC components removed from the power spectra to find out the contributions of the interference to the bandwidth estimations.

III. RESULTS

There are two objectives for the flow measurements. The first objective is to demonstrate a linear relationship between bandwidth and flow velocity, and the second objective is to determine if the flow velocity measurements are insensitive to the transducer angle when the range cell is approximately spherically symmetrical. Experimental results for both the uniform flow and the real flow were obtained. In the data presented below, six data points are plotted for each real-flow velocity measurement to indicate the amount of fluctuation.

A. Bandwidth versus Velocity

For a transducer angle of 60^o and a two-way pulse length of 0.6 mm, the Doppler bandwidth is plotted against the flow velocity for both the uniform flow and the real flow in Figs. 3 and 4, respectively. In both cases, the Doppler bandwidth increases nearly linearly with flow velocity, but each bandwidth obtained from the real flow is smaller than that obtained from the uniform flow at the same flow velocity. Similar linear relationships between bandwidth and flow velocity are also evident for transducer angles of 45^o and 75^o for the real flow, as shown in Figs. 5 and 6, respectively.

Figs. 7 and 8 are plots of bandwidth versus flow velocity for the real flow when the transducer angle is 90^o and the bandwidth estimations are performed with and without the DC components, respectively. In both cases, the estimated Doppler bandwidths increases nearly linearly with flow velocity, but the bandwidths estimated with the DC components are much smaller than those estimated without the DC components.

B. Bandwidth versus Transducer Angle

For an uniform flow of 12 cm/s, plots of Doppler bandwidth versus transducer angle for two-way pulse lengths of 0.4, 0.6, and 0.8 mm are presented in Fig. 9. When the two-way pulse length is 0.6 or 0.8 mm, the bandwidth is insensitive to the transducer angle. However, when the two-way pulse length is 0.4 mm, the bandwidth decreases as the transducer angle increases.

The result for real flow with a flow velocity of 12 cm/s is presented in Fig. 10. Because the bandwidth data obtained for two-way pulse lengths of 0.4, 0.6, and 0.8 mm are almost the same in this case, only the data for a two-way pulse length of 0.6 mm are plotted in this figure. When the transducer angle is 90^o, the magnitudes of the interferences at DC are much larger than those of the Doppler signal, so the bandwidth estimations with the DC components removed are used in this plot to give a more accurate result. The plot in Fig. 10 shows a general trend of decrease in Doppler bandwidth with increase in transducer angle.

IV. DISCUSSION

Each of the Doppler bandwidths obtained for the real flow is smaller than that obtained for the uniform flow at the same recorded flow velocity, because the flow velocity indicated for the real flow is actually the maximum velocity at the center of the flow channel, but the average flow velocity for the scatterers in the range cell is lower. On the other hand, the flow velocity indicated for the uniform flow is equal to the average flow velocity of the scatterers. Taking into account of this difference in the indicated flow velocities, the results shown in Figs. 3 and 4 are in reasonable agreement. The purpose of performing bandwidth measurement on ideal uniform flow is to demonstrate the feasibility of the basic technique and to facilitate comparison with some of the results obtained for the real flow. Because of the time efficiency in collecting data in real time, more data were obtained for the real flow.

In Figs. 9 and 10, the plots of bandwidth versus transducer angle for the uniform flow differ substantially from that for the real flow, which shows bandwidth decreases as the transducer angle increases; the trend is almost independent of the two-way pulse length. As the transducer angle increases, the decrease in Doppler bandwidth is expected for the real flow, because the broadening due to the spread of flow velocity (i.e., the spread of Doppler frequency) in the range cell is less when the transducer angle is larger. However, it is somewhat perplexing that the bandwidth for the real flow is almost independent of the three two-way pulse lengths used for the flow measurement. One explanation is that the change in bandwidth produced by a change in the two-way pulse length is offset by an opposite change in bandwidth caused by a change in flow velocity distribution in the range cell. For example, an increase in the two-way pulse length tends to decrease the bandwidth, but such decrease could be offset by an increase in bandwidth as a result of a larger distribution of flow velocities in the larger range cell created by the longer pulse length.

Although the Doppler bandwidth is sensitive to the transducer angle in the real flow, the change in bandwidth might be small enough for many applications in which the range of transducer angles is limited or can be estimated. For example, as shown in Fig. 10, if the transducer angle is limited to a range of 60^o to 90^o, the bandwidth varies only between 31 Hz to 42 Hz when the flow velocity is 12 cm/s. In certain applications, if the velocity profile is known or can be estimated based on fluid-mechanics considerations, the error in bandwidth determination can be reduced once the range of the transducer angle is estimated. For example, a transducer angle that is close to 90^o can be inferred from a large Doppler bandwidth associated with a small Doppler shift, and the bandwidth obtained can be scaled appropriately by a correction factor stored in a look-up table for the particular flow profile. For this reason, it is feasible to obtain angle-insensitive flow measurement for larger blood vessels, when the flow velocity in a much smaller range cell is approximately uniform.

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Figure 1: System for measuring ideal uniform flow.

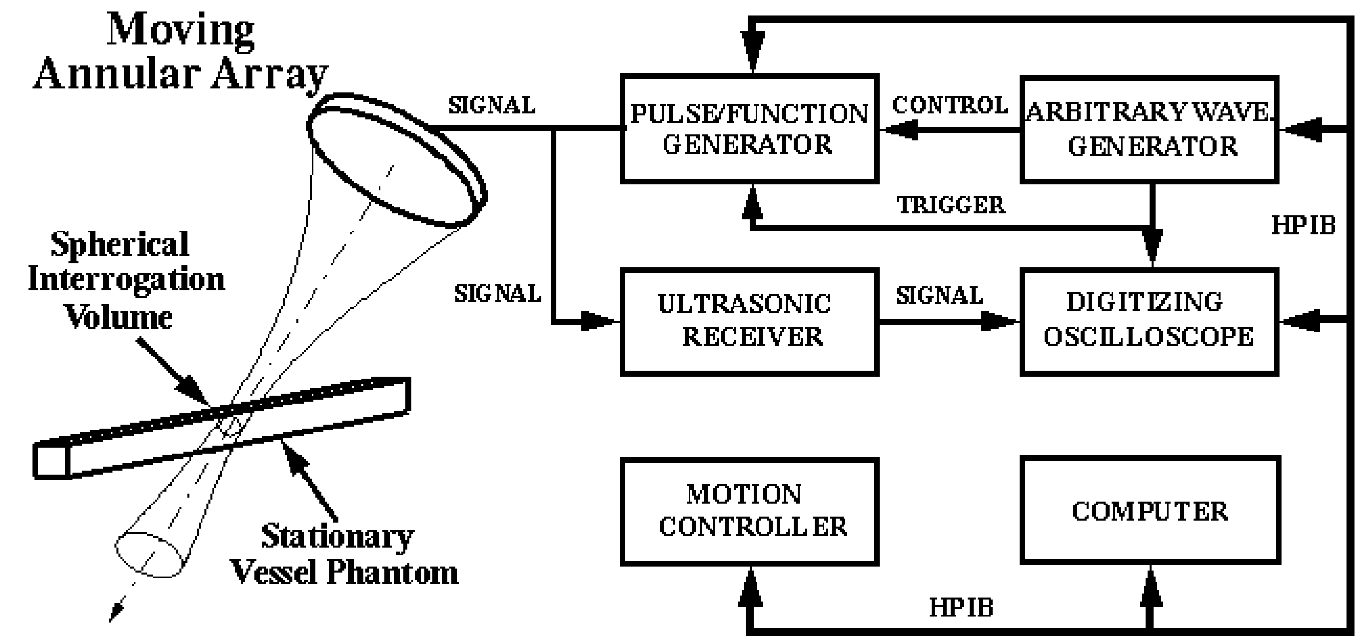

Figure 2: System for measuring real flow.

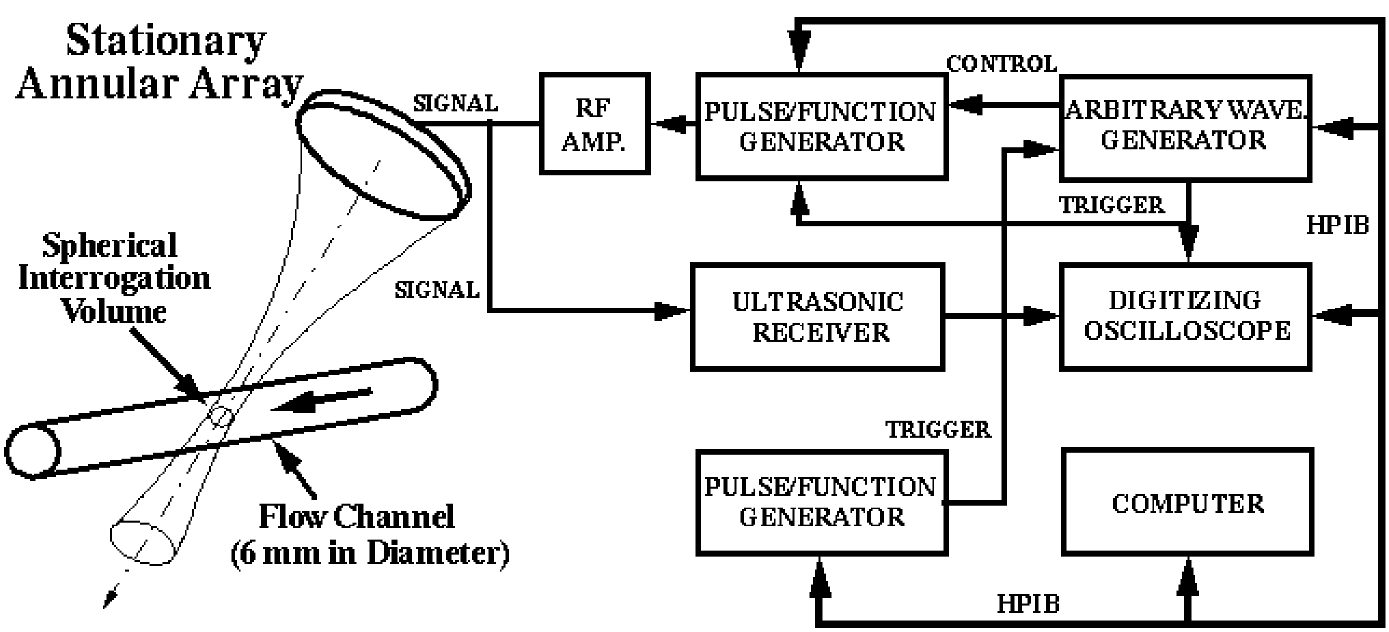

Figure 3: Bandwidth versus velocity for ideal uniform flow. Transducer angle = 60^o; two-way pulse length = 0.6 mm.

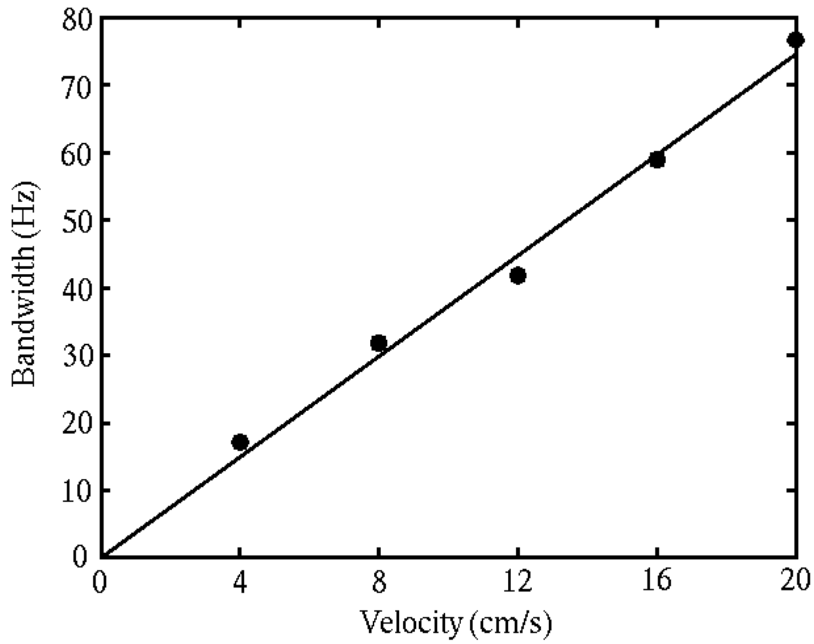

Figure 4: Bandwidth versus velocity for real flow. Transducer angle = 60^o; two-way pulse length = 0.6 mm.

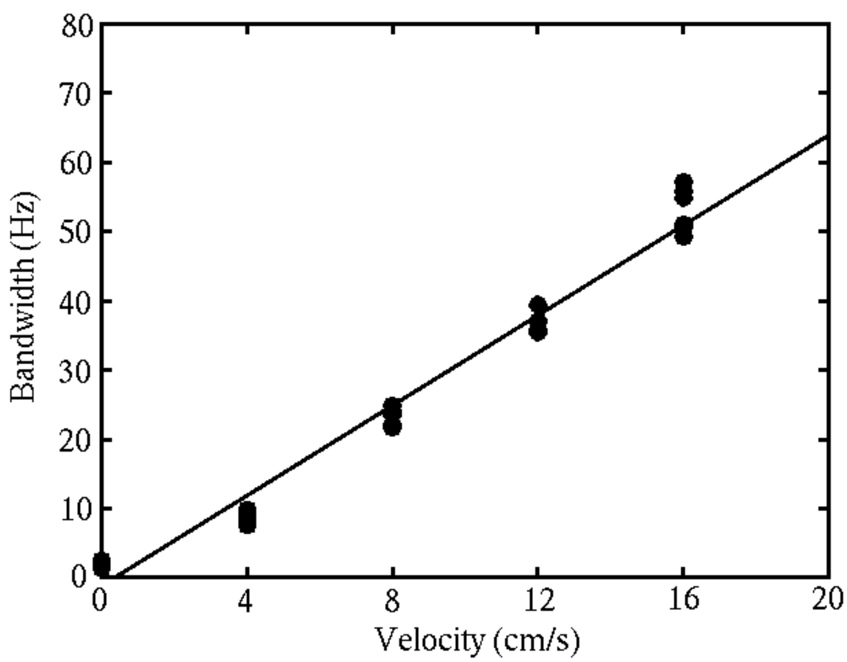

Figure 5: Bandwidth versus velocity for real flow. Transducer angle = 45^{o}; two-way pulse length = 0.6 mm.

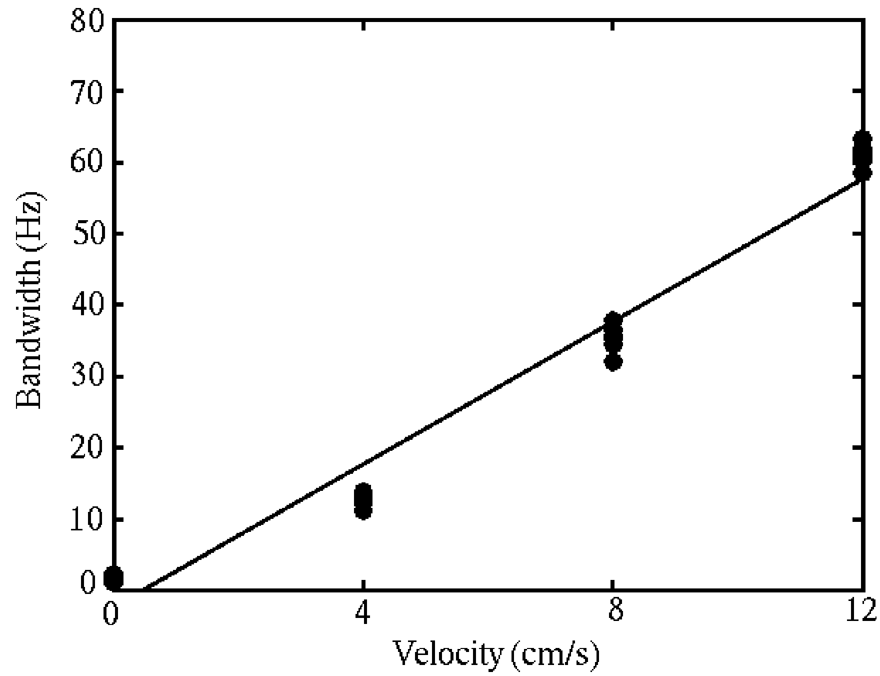

Figure 6: Bandwidth versus velocity for real flow. Transducer angle = 75^{o}; two-way pulse length = 0.6 mm.

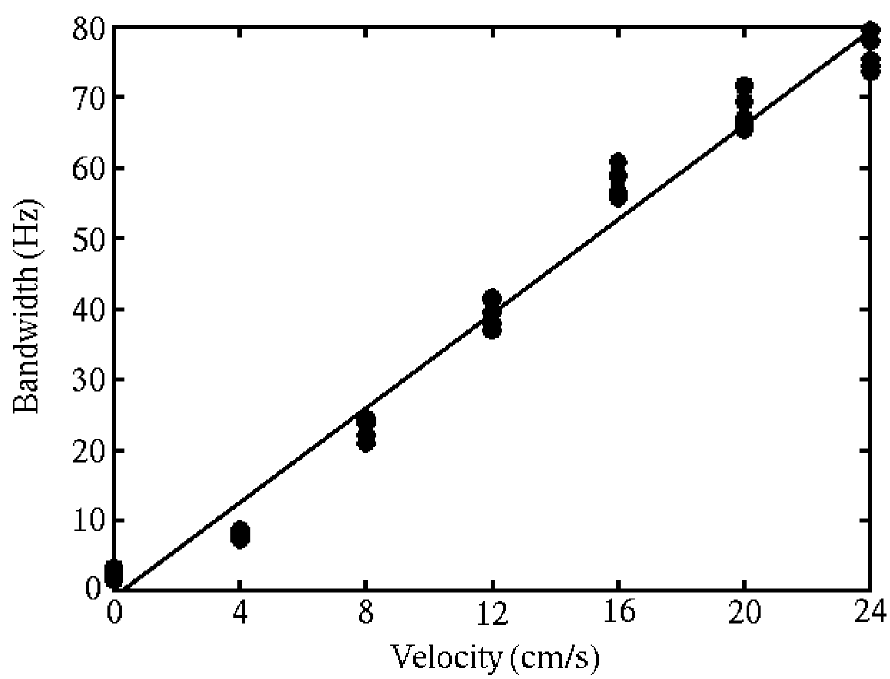

Figure 7: Bandwidth versus velocity for real flow (bandwidths were estimated with DC components in the power spectra). Transducer angle = 90°; two-way pulse length = 0.6 mm.

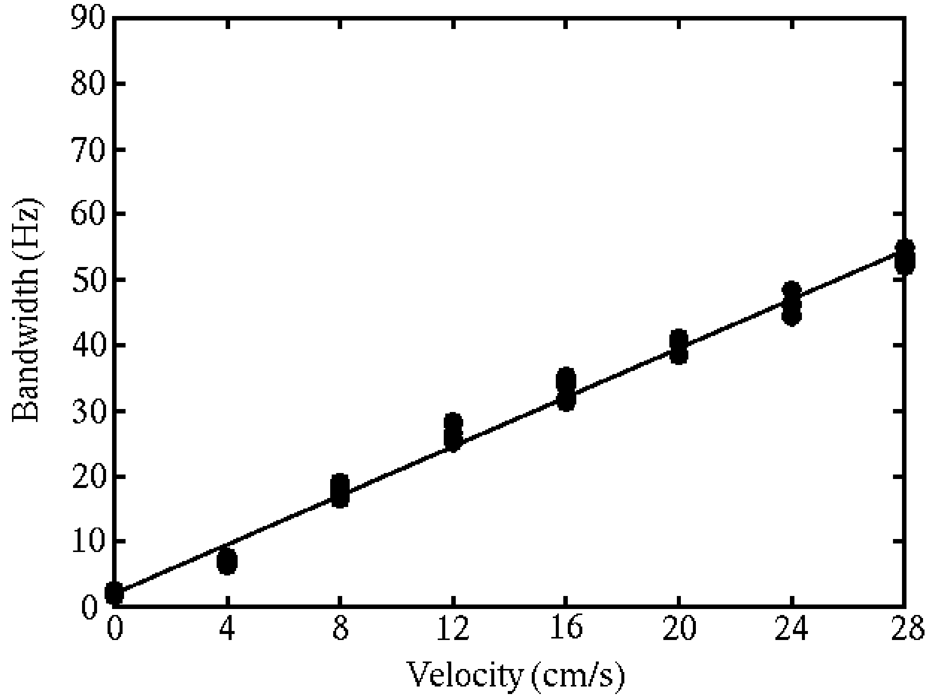

Figure 8: Bandwidth versus velocity for real flow (bandwidths were estimated with DC components removed from the power spectra). Transducer angle = 90°; two-way pulse length = 0.6 mm.

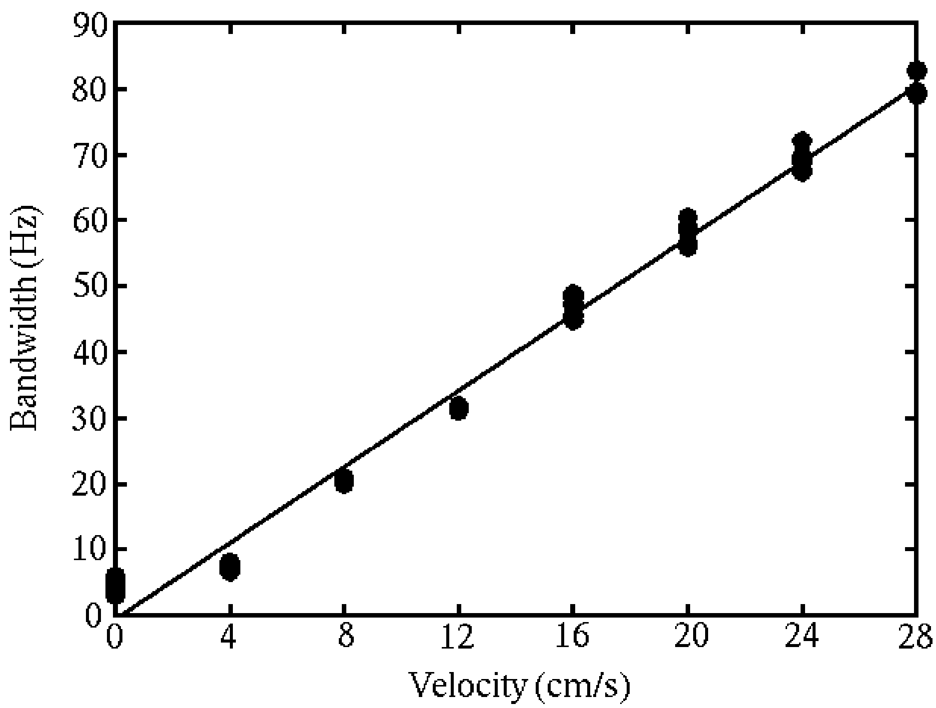

Figure 9: Bandwidth versus transducer angle for ideal uniform flow. Flow velocity = 12 cm/s; two-way pulse lengths = 0.4, 0.6, 0.8 mm.

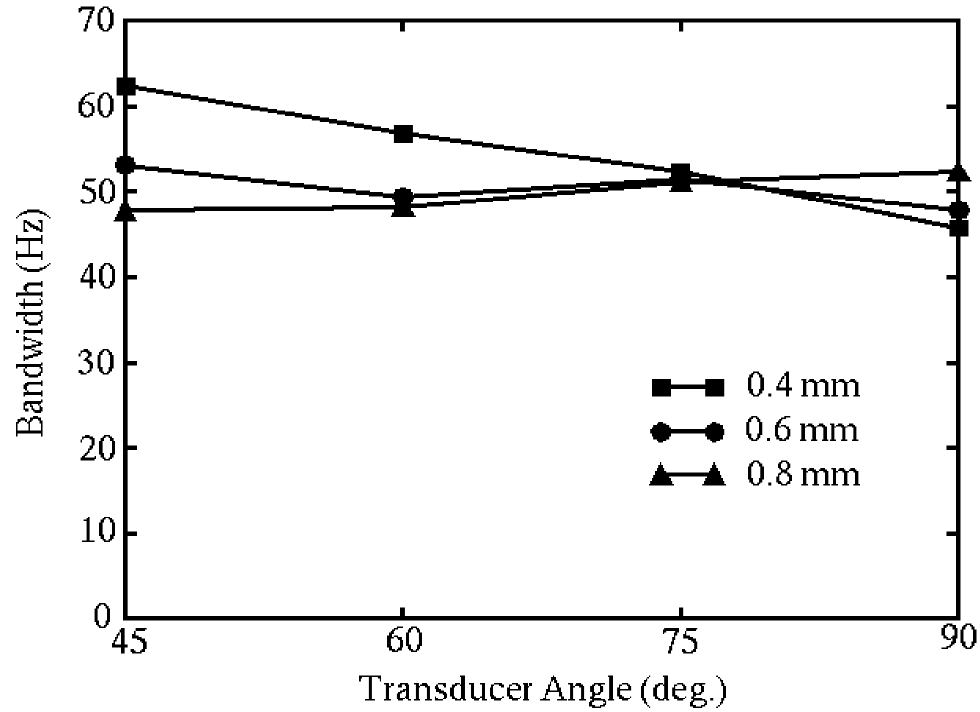

Figure 10: Bandwidth versus transducer angle for real flow. Flow velocity = 12 cm/s; two-way pulse length = 0.6 mm.