Blood Flow Analysis Using Ultrasound Doppler Technique

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Abstract : *An in vitro system was used for measurement of frequency and amplitude of ultrasonic signals scattered from a moving column of blood analog under physiologic flow conditions. Flow rate and probe orientations (angle, vertical and lateral displacement) were varied in dependently. With the probe centered above the blood vessel model the frequency shift (F) of the scattered signal were fit to the equation:*

$F=C_x$ $+C_\theta\theta+C_v\overset{\triangledown}{}+C_0$ ($10°<\theta<40°$), where θ is the angle (in degrees) to the normal to the flow, V is average velocity of flow (measured indirectly) and C_x, C_θ, C_v and C_0 are constants. Using stepwise regression the constant C_v was found to be insignificant; including the other constants produced a standard error of 31 Hz, about 5-l0% of observed values of F. It was not possible to obtain the diameter of physiologic sized blood vessels using the CW technique because of the low resolution of the system (about 0.5cm). I t was found practical to locate the proper probe position and estimate angle using only variation in observed amplitude to guide positioning. This opens the possibility of accurate transcutaneous blood velocity measurements in animals and humans

Keywords: *Blood velocity, ultrasound, flow rate.*

I.   Introduction

Doppler ultrasound signals have become vital in measuring the disturbances in the instantaneous blood flow velocity profile. In ultrasound flow device, a series of pulses is transmitted to detect movement of blood. Echoes from stationary tissue are the same from pulse to pulse. Echoes from moving scatterers exhibit slight differences in the time for the signal to be returned to the receiver. These differences can be measured as a direct time difference or, more usually, in terms of a phase shift from which the 'Doppler frequency' is obtained. Blood Flow waveform shape and spectrum has proved to be an useful technique in the investigation of many vascular beds. It has the advantage that derived indices are independent of the beam/flow angle. Changes in flow waveform shape have been used to investigate both proximal disease (e.g. in the adult peripheral arterial circulation) and distal changes (in the fetal circulation and uterine arteries). We develop method to analyze the frequency (f) and flow rate (Q) for various ultrasound frequencies and also get vital results in echo amplitude Vs probe angle.

Several methods have developed for analyzing the blood flow velocity profiles produced by ultrasound signals. Some of the methods are Accuracy of continuous wave Doppler measurements of fluid velocity [1], in this paper describe for measuring the average velocity of flow indirectly and also found proper probe position and estimate the angle. Another technique is Influence of Beam Profile and Degree of lnsonation on the CW Doppler Ultrasound Spectrum[2], using this method ultrasound beam profile and the degree to which a vessel is uniformly insonated influences both the spectral density and the estimation of the mean velocity. A computer model that enables these effects to be conveniently studied is described for any axisymmetric flow velocity profile and any probe field response that is symmetric. Specifically, the model provides estimates of the error in the mean velocity compared to the true mean, the spectral density graph, and the total received power. Another method, Theoretical Analysis of the Ultrasonic Doppler Flow meter for Measurements of High Flow Velocities [3], this is based on the calculation of the transit time difference between the ultrasonic waves that are reflected from a moving particle at its various positions. Beam divergence is taken into account, and each path of the ultrasonic wave propagation is approximated by two rectilinear components. In our technique frequency against flow rate is obtained using multiple regression. Subsequent results also obtained with probe angle from 5 to 40 degrees.

Materials and Methods

In the experiments to be described, an -vitro f low model system was used. This system supported continuous flow in vessels of variable known diameter at directly measured physiologic flow rates. The commercial Doppler instrument was connected to a frequency and amplitude detector, which produced voltages proportional to the amp1itude and frequency of the Doppler shifted signal.

Several computer programs were written to model the performance of the ultrasonic flow detection system, and to correct for the frequency and amplitude sensitivities of the frequency measuring equipment. One program was used to correct the frequency meter reading for low amplitude input signals. A second computer

program was used to produce a digital signal having the same characteristics as the signal produced by the ultrasound detector: that is, a band-limited signal with a flat frequency spectrum, but with randomly-varying phase angles. This model enabled comparison of the operation of the flow system with a mathematical analog.

There are basically four kinds of experiments described in this report:
1. Calibration of the velocity detector for varying probe angle, blood vessel diameter, and probe-to-blood vessel distance. This calibration was performed using an indirect approach. The flow was measured and the velocity computed assuming non-turbulent flow. (Measurement of the viscosity of the blood analog and calculation o f Reynolds numbers showed that this assumption was reasonable).
2. Determination of the spatial response of the probe. This measurement was made using a microscope stage to position the ultrasonic transducer accurately and reproducibly with respect to the blood vessel analog.
3. Estimation of vessel diameter using CW ultrasound. Again, the microscope stage was used. In addition, a position transducer was constructed so that plots of probe position versus returned signal amp1itude and/or frequency could be displayed on a storage oscilloscope.
4. Comparison studies of blood analog (evaporated milk and water) to blood and to digital simulation. Blood studies were performed to validate the use of evaporated milk as a blood analog, since it was believed that evaporated milk was much more convenient to use in experimentation

II. Results

Probe calibration. Because the amp1itude of the Doppler shifted signal had significant variation with probe-to-vessel distance, it was necessary to determine the effect of variation of the signal on measured frequencies.

By varying the gain of the detector circuit (Figure I) while flow and probe position were held constant, experimental data (values o f observed frequency "Fobs" and observed "A" were fit to an equation of the form:

$$F_{obs} = F_{lim} - ae^{-bA} \qquad (1)$$

'F_{lim}' represents the frequency that presumably would be measured by a frequency meter of infinite sensitivity. The "a" and "b" are empirically determined constants. Experiments were performed using a constant flow rate while the amp1itde of the input signal to the amp1itude velocity detector was varied. For each flow rate the arbitrary constants "a" and "b" were determined using the transformed equation:

$$\ln (F_{lim} - F_{obs}) = \ln (a) - bA \qquad (2)$$

The determination was carried out by a linear regression using the amp1itude "A" as the independent variable and the logarithm of the frequency difference as the dependent variable. The value o f 'F_{lim}' had to be selected (by trial). It was chosen to maximize the correlation coefficient between the amplitude and the log of the frequency difference. A complete table showing the extrapolated relationships between these variables was produced. When the observed amplitude was greater than **1** volt, the observed frequency was atleast 90% of the limiting frequency. Accordingly, there was no effort made routinely to calculate explicit corrections on the frequency values.

As expected, the relationship of velocity of **flow** to measured frequency shift was linear. Initially, for constant angles of the probe, a least squares fit was obtained in the equation:

In this equation **"F"** is the observed frequency shift, **"Q"** is the volume flow rate, and **"m"** and "b" are constants. One unexpected result in the experiments performed was that the constant "b" was never zero, nor was "b" attributable to experimental error. The value "b" did not appear to depend on tube diameter. Analysis of the situation revealed that the constant took a value of approximately half of the **low** frequency cut-off of the Doppler shift detector("b" was typically about 200 **Hz).** Values for the constants **"m"** and "b" are shown in Table**1.**

Table 1. Values for Flow Constants

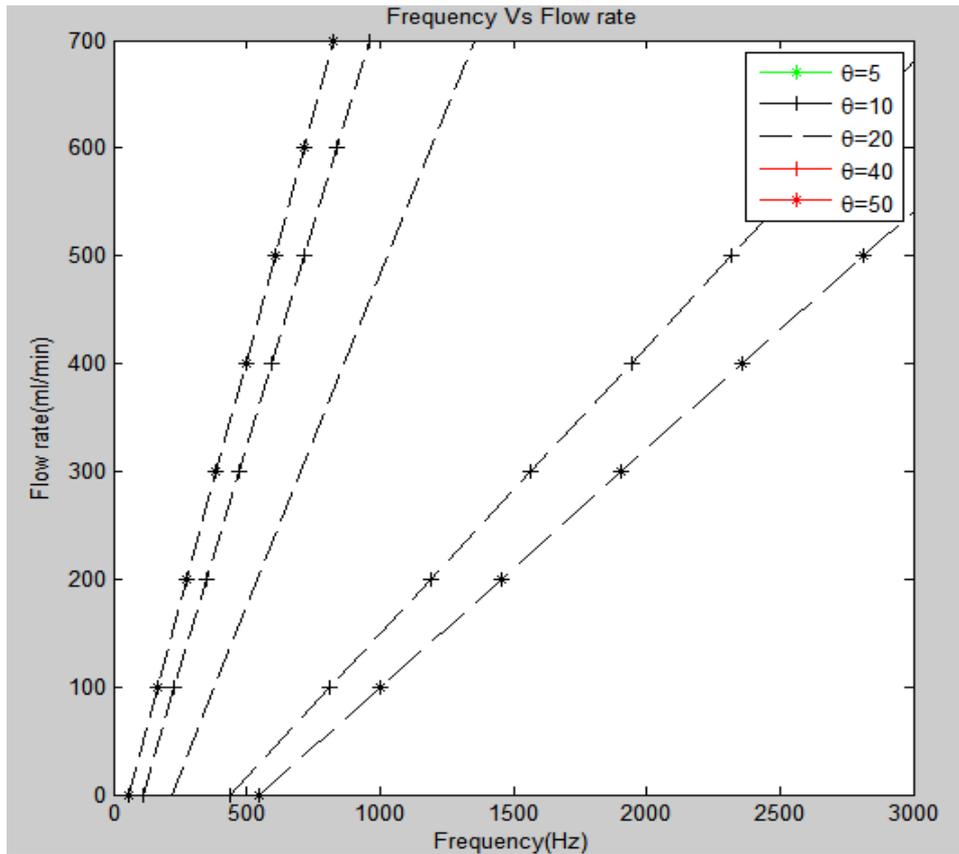| Angle of Probe to Normal (θ) degrees	m Hz/cm³/min	b Hz
5	1.412	54.71
10	0.941	109.43
20	1.618	218
30	2.719	244.4
40	3.766	437
50	4.711	547.05

Fig. 1. flowrate Vs frequency.

Note that they vary with the angle of the probe. Points are plotted in Figure 2**.**

Note that increasing the angle to the normal of the probe increases the slope of regression line. This is an expected result from Doppler theory. The more nearly parallel the probe is to the flow, the greater should be the shifted frequency of the returned signals.

In subsequent experiments the effect of angle in determining the frequency shift of the Doppler signal was included quantitatively using data from experiments with angles from 3˚ to 40˚. The relationship between average flow velocity, angle of probe, and frequency shift was modeled as:

$$F=C_x\theta\nabla+C_\theta\theta+C_v\nabla+C_0 \tag{4}$$

Using mu1tiple regression it was found that the constant "Cv" was not significant but that other constants were. Further, restricting "θ" to the range between 10˚ and **40°,** values f o r the constants were determined to be those listed in Table. Note that the standard error f o r the estimate of frequency shift is only **31** Hz. Thus, for all frequencies observed above **300** Hz, the expected error of estimation using this equation is less than 10%.

Table 2. Constants of equation (4)

$$C_x = 1.034\,\text{Hz.sec/degree.cm}$$

$$C_e = 6.891\,\text{Hz/degree}$$

$$C_o = 125\,\text{Hz}$$

$$\text{Correlation Coefficient}\,(P) = 0.996$$

$$\text{Standard error of estimate} = 31\,\text{Hz}$$

Spatial response of the probe. Measurement of the beam dispersion indicated that the region of maximum sensitivity of the probe lay in its center line approximately 0.6cm from the probe tip. Unfortunately, it was quickly discovered that the region of maximum sensitivity encompassed an area which was about 0.75cm along the axis of the probe and about 0.6cm in diameter, forming a cy1inder around the extended probe axis. Sensitivity tapered off gradually from this region of maximum sensitivity.
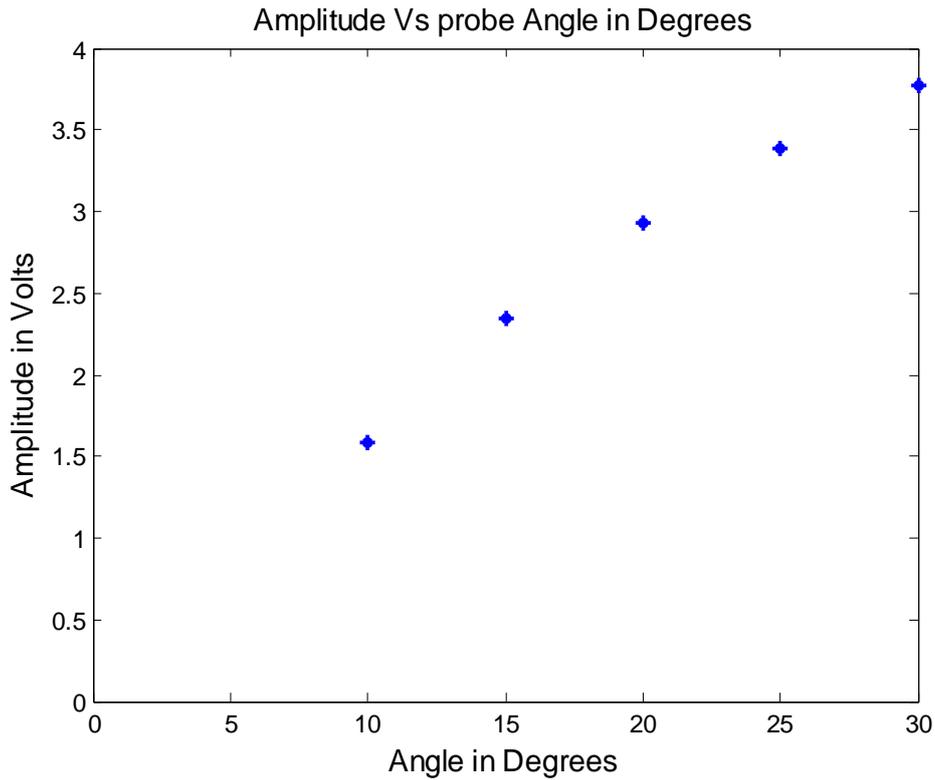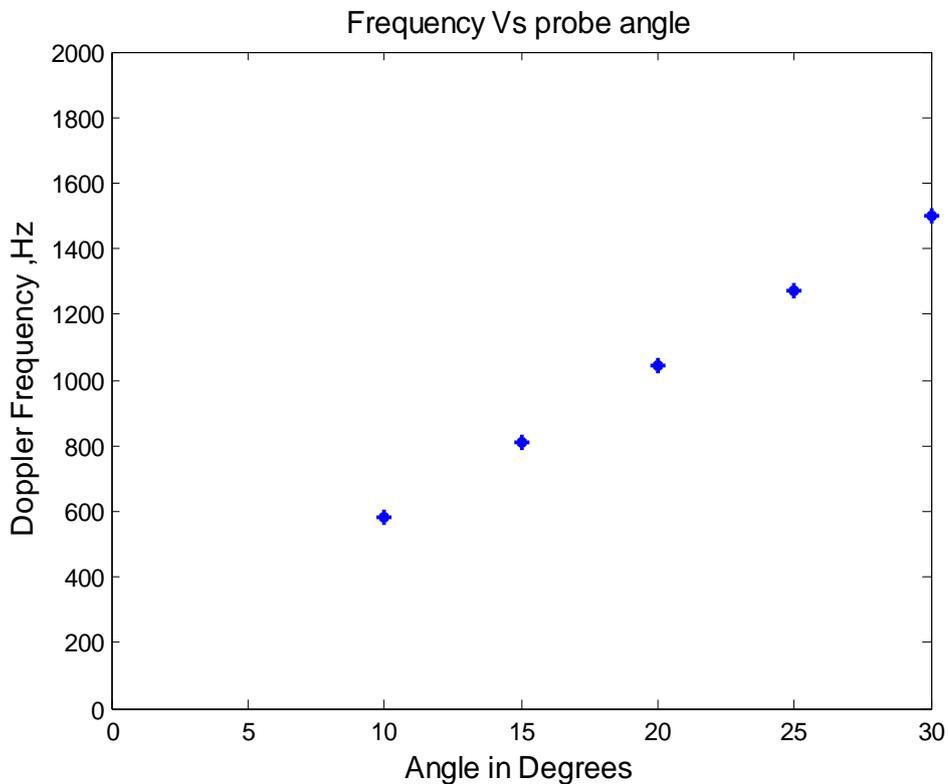
Fig. 2. Amplitude VS Angle in degrees

Fig. 3. Doppler frequency Vs angle in degrees

Brackets represent observed ranges of the amp1itude signal. The relationship of signal amplitude to probe angle is illustrated in Figure 3. Note that the signal diminishes near the angle zero as expected from Doppler theory. If the system were ideal there should be no signal at a zero angle because the fluid is moving perpendicularly to the ultrasonic signal.

Diameter estimation. Since the velocity calibration (Equation 4) contains no constants related to vessel diameter, it might be anticipated that it is independent of diameter. Indeed, for diameters from $1/8^{th}$ to $1/4^{th}$ of an inch this appeared to be the case. Encouraged by these findings, we attempted to measure vessel diameter so that by combining the estimate of velocity with an estimated diameter we could then obtain a figure for blood flow without having to place a probe around a dissected vessel. Initially it appeared that the spatial sensitivity was so diffuse that it would not be possible to determine the size of blood vessel in which velocity was being estimated by simple probe manipulation. This was experimentally confirmed. Neither the velocity signal nor the amplitude signal was affected in any way except by a scale factor when the diameter of the blood vessel analog was changed. Comparison studies. Experiments using blood as the test fluid and digital simulation produced agreement with results predicted by Equations **1** and 3.

III. Discussion

We have shown that it is possible to estimate the velocity of the flow o f blood, or blood analog, with reasonable accuracy using CW ultrasound. Thus far, it has not been possible to estimate the diameter of the vessel using the same technique. However, i f the angle of .the probe to the blood vessel is known, then the possibility of transcutaneous estimation of velocity alone exists. In order to perform such an estimation one would have to construct a probe holder which would enable accurate measurement of the angle o f the probe to the surface of the body or to some other constant reference line. The probe could be manipulated through different angles until the angle of minimum signal was determined. This angle would then be used as the value θ = 0. From this direction relative angle measurements could be made and velocity determined from Equation 4.

The ultimate clinical usefulness of these measurements will depend upon the establishment of standards for blood velocity just as there are now standards for electrocardiography, spirometry, etc. A recent survey of established investigators in ultrasound revealed a high priority on verification of the validity of diagnosis obtained with ultrasonic diagnostic equipment in clinical use. It might be useful to undertake a long-range research project aimed at establishing such standards.

IV. Conclusion

In this experiment results are shown that velocity of the flow of blood using ultrasound Doppler technique. It was also observed alternate detection of frequency and amplitude for various probe angles.

Finally concluded that values for flow constants were derived from linear regression results observed frequency 90% of limiting frequency. So detected frequency accuracy can be increased by other techniques instead of linear regression.

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