Pulse-Doppler Ultrasound and Its Clinical Application

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INTRODUCTION

It is unusual to find an ultrasound tomogram which does not contain a section of blood vessel. However, because conventional ultrasonic B-scanning machines are capable of indicating only the shape and route of the vessel, blood flow information, which is available via the Doppler effect, is ignored. This is unfortunate since it is reasonable to suppose that flow parameters could provide clinically useful information relating to the function and viability of an organ. In particular, vascular distortion caused by extrinsic masses or intrinsic abnormalities of the arterial wall will almost certainly affect the blood-velocity waveform in the supply vessels. (A good example of this is a cirrhotic liver where massive distortion of hepatic structure causes an increase in vascular resistance, resulting in a reduction of blood flow through the portal vein.) An atraumatic and preferably transcutaneous blood flow monitoring device ought to find wide clinical application.

This article describes how a pulse-Doppler device has been linked to an ultrasonic B-scan machine so that flow patterns in any ultrasonically accessible vessel can be investigated from the skin surface. Section one is a brief explanation of Doppler principles. Section two explains the different types of Doppler processing which are available, while section three describes the Doppler interface with the B-scan machine. In section four some established and potential clinical uses of the pulse-echo-Doppler hybrid are discussed.

Principles of Doppler Flowmeters

The Doppler principle states that the frequency of the echo reflected from a moving target will be different from the incident frequency. This process is illustrated in Fig. 1. The transmitting transducer emits ultrasound which travels towards the blood vessel. The moving blood cells back-scatter to the receiving transducer ultrasound of
a slightly different, in this case higher, frequency. The frequency difference is detected by comparing the frequencies of the transmitted and received signals. The frequency shift, $\Delta f$, is given by

$$\Delta f = 2f \frac{V\cos \theta}{c}$$

where $f$ is the incident frequency, $V\cos \theta$ is the velocity of the target in the direction of the transducers, and $c$ is the velocity of ultrasound. Furthermore, the power of the Doppler signal relates to the acoustical size and properties of the moving target or, in the case of blood, to the volume flow through the beam. Unfortunately, the variable ultrasonic attenuation of intervening tissues prevents an absolute determination of volume flow.

The continuous-wave device, shown in Fig. 1, suffers from a lack of range resolution since it cannot determine the axial position of the target. This means that the Doppler waveforms would superimpose if there were more than one vessel within the beam or, conversely, that the source of a Doppler signal cannot be localized within the beam. This disadvantage is overcome in the pulse-Doppler flowmeter which combines the fine resolution of a pulse-echo system with the moving target discrimination of a Doppler device. The operating principles of such an instrument are illustrated in Fig. 2. A short pulse of ultrasound is transmitted towards the moving blood and, after a time delay which allows for the pulse to travel to and from the selected range of interest, the scattered echo is sampled for Doppler shifts. This "range-gating" ensures that the Doppler signals originate only from those targets moving within the sample volume. The range of the sample volume is determined by the transmit-sample delay time while its shape is defined by the length of the transmitted pulse and the width of the ultrasonic beam. By transmitting an adequately frequent series of pulses and range-gating the returning echo, it is possible to reconstruct the Doppler difference signal created by movement only within the sample volume.

**Doppler Signal Detection**

The transmitted and received signals must be frequency compared to extract the Doppler difference component. Fig. 3 shows how this comparison is made in a "synchronous detector." This multiplies the input waveforms $A$ and $B$ and then smooths the output to produce $C$, which is highest when $A$ and $B$ are exactly in-phase (peak coincident with peak), and lowest when they are exactly out-of-phase (peak coincident with trough). It can be seen that the output signal oscillates at the difference in frequency between the two inputs. Thus, if $A$ and $B$ represented the
FIG. 2. The pulse-Doppler. The range of the sample volume is determined by the transmit-sample delay time.

FIG. 3. Synchronous detector input-output waveforms.

FIG. 4. Coherent pulse-Doppler waveforms.
transmitted and received (Doppler-shifted) waveforms, then \( C \) would be the audible Doppler difference signal.

The coherent pulse-Doppler (see Fig. 4), operates by comparing (or multiplying) the signal packet \( C \) returning from the sample volume with the transmitted signal, the phase of which is most conveniently stored as the continuously-running local oscillation \( B \). The instantaneous sample voltage \( D \) so produced constitutes one point on the Doppler difference waveform: one sample is joined smoothly to the next to reconstitute the Doppler signal \( E \). It is worth emphasizing that the pulse repetition frequency needs to be significantly higher than the maximum frequency shift in order to allow an accurate reconstruction of the Doppler waveform.

By comparing the received frequency with that transmitted, the coherent Doppler system effectively monitors movement relative to the transducer. This type of processing works well if the sample volume can be positioned accurately and completely inside a blood vessel. However, if the sample volume also happens to contain large, slow-moving targets, such as the vessel wall or surrounding tissue, then low frequency large-amplitude Doppler components (known as clutter) can obliterate the higher frequency but fainter blood signal. This problem is avoided by using non-coherent Doppler detection where, instead of comparing the complete received echo with the transmitted signal, the blood component is allowed to combine with the clutter component. The difference in frequency then corresponds to the velocity of the blood relative to the clutter. Fig. 5 illustrates non-coherent pulse-Doppler detection used on a simple target configuration where the proximal vessel wall is providing the clutter echo. A long pulse of ultrasound \( A \) is transmitted so that the vessel wall echo \( B \) and blood echo \( C \) superimpose ultrasonically. The amplitude of the combined echo \( D \) fluctuates at that Doppler difference frequency which corresponds to the velocity of the blood relative to the wall. The received echo amplitude appears to be constant throughout a single pulse because the blood moves only a small distance during this time period. The contracted time scale on the right-hand side of Fig. 5 shows more clearly how the echo envelope fluctuates from sample to sample as the blood moves relative to the clutter. Each sample represents a single point of the Doppler signal and, as before, the Doppler waveform is reconstructed by smoothly joining one sample to the next. Notice that the echo amplitude would not fluctuate at all if the wall plus blood moved together relative to the transducer; only relative movement within the sample volume is detected. This is the important advantage of non-coherent detection.

**The pulse-echo-Doppler hybrid**

A 2 MHz pulse Doppler system with both coherent and non-coherent processing facilities has been linked to a conventional gray-scale B-scan machine [1]. Fig. 6 shows a schematic diagram of the interface. The investigation procedure first identifies the vessel by B-scan and then interrogates the region of interest by Doppler. Because the same transducer is used for both modes of operation, the sample volume can be maneuvered into position reliably and accurately by suitable adjustment of the beam direction and range-delay time. Furthermore, the interrogation site is identified by using the range delay to mark a bright-up cursor on the B-scan display as shown in Fig. 6. The operator can hear the Doppler signals originating from the sample volume which are broadcast over a loudspeaker as well as being spectral-analyzed.

Preliminary studies [1] have indicated that due to the poor lateral resolution characteristics of ultrasonic transducers, it is not feasible to range-gate across the velocity profile of a deep lying vessel such as the portal vein or renal artery: the
sample volume at this depth encompasses most of the vessel lumen. However, by increasing the length of the transmitted pulse to a point where it bathes the entire vessel cross-section, it is possible to monitor the instantaneous mean flow velocity by computing the average frequency in the Doppler power spectrum. Clutter is reduced by switching to the non-coherent processing mode. If required, the B-scan can be used to measure the calibre of the vessel and correct for \( \theta \), the angle of attack, so that the volume flow can be estimated. This type of investigation was performed on the
portal vein and inferior vena cava in a series of ten normal volunteers. Although the volume flow rates agreed with the commonly accepted values, the accuracy with which this parameter could be estimated was intolerably high at ± 30%. This could have been decreased to ± 20% by using an improved B-scan machine to give a more accurate indication of vessel caliber.

Some clinical uses of the Doppler device

a. Blood-velocity waveform analysis

At this time, Doppler devices do not seem well suited to the accurate estimation of volume flow in deeper lying vessels. They are, however, capable of indicating the velocity to an accuracy of about ± 10% and, furthermore, are extremely sensitive to changes in velocity. It therefore seems that more useful clinical information can be gained by using Doppler devices to their best advantage: that is, as sensitive indicators of velocity change rather than inaccurate computers of volume flow. This type of approach has already been shown to be successful in the demonstration of disease of the peripheral arterial system where the shape of the velocity-time waveform is used to define the vascular properties of an arterial segment [2]. The spatial selectivity of the pulse-echo-Doppler hybrid should now allow the properties of deeper lying vessels to be investigated. For instance, the vascular resistance of an organ can be examined by monitoring the velocity-time waveform of the blood flow in its arterial supply. If the organ exhibits a low resistance to flow then the momentum imparted to the blood during systole is sufficient to cause a continuation in flow through diastole. Alternatively a high resistance leads to a much reduced diastolic flow component: the blood stops when the systolic driving force ceases. This effect is demonstrated in Fig. 7 which shows how the flow through the renal artery to the low impedance kidney compares with the flow through the radial artery to the higher impedance peripheral tissues of the hand.

Development of this principle should allow study of the vascular properties of any organ which has ultrasonically accessible supply vessels, such as the lung, kidney, liver and placenta.

b. Tissue characterization

Non-coherent Doppler detection might prove useful in the qualitative estimation of tissue perfusion. By locating an extensive, spatially-integrating sample volume over the region of interest, it should be possible to investigate generalized movement of blood and thereby differentiate hypervascular tumors from avascular cysts or hyperemic inflammatory masses.

c. Doppler Echocardiography

The spatial selectivity of the pulse-Doppler is valuable when investigating movement in and around the heart. Murmurs have been localized using Doppler by detecting blood turbulence within the heart and great vessels [3]. This technique can provide a useful complement to the more conventional M-mode examination.

d. Fetal breathing detector

The most novel use of the Doppler device is to detect chest wall movements of the normal fetus in utero. A reduction in these movements can provide an early indication of fetal distress [4]. Ultrasonic A-scan machines have been developed which track the position of the thoracic walls in an attempt to monitor the regularity of movement over a period of hours [5]. By adopting a Doppler-type approach, Boyce et al. [6] have discovered that distinctive Doppler sounds originate from the fetal lung during breathing and that fetal breathing can readily be monitored with even a simple continuous-wave Doppler device, identical to a fetal heart detector. Because the lung
provides a diffuse source of echoes, the orientation of the transducer is not as critical as in the A-scan method which relies on strong reflection from the thoracic walls. The Doppler device is, in addition, simpler to use and cheaper to produce, making it more suitable for screening large numbers of people in the antenatal clinic or ward.

SUMMARY

Doppler instruments can provide additional information during an ultrasound examination. When used in conjunction with a B-scan machine, the pulse-Doppler can localize the source of movement to within a few millimeters so that bloodflow patterns in deep-lying vessels can be investigated atraumatically from the skin surface. The clinical use of this facility requires further evaluation.

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