JDMS 4:109-120, May/June 1988

Articles

Doppler Physics and Instrumentation

A Review

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The physical principles of continuous-wave and pulsed-wave Doppler as well as the processing methods to detect the direction of flow are reviewed. An overview of Doppler instrumentation, including duplex, zero crossing, range-gated, and color flow imaging detection systems is presented. The technique of spectral analysis is explained with multiple illustrative examples. Aliasing, which arises from the limited sampling rate of pulsed wave instruments, is discussed in detail.

Key words: Doppler instrumentation, spectral analysis, aliasing.

During the last few years, Doppler ultrasound has experienced considerable growth both in the number and diversity of examinations performed. The increased role of Doppler ultrasound is the result of the need to evaluate the vascular system noninvasively as well as the increased sophistication of Doppler instrumentation. A number of different types of Doppler instruments are available. Although each instrument utilizes the Doppler principle to detect motion, the manner in which the Doppler signals are acquired, processed, and displayed distinguish one type of instrument from another. An overview of the various types of Doppler scanners is presented. Emphasis has been placed on the explanation of spectral analysis and aliasing.

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Doppler instruments often quantitate the rate of movement or speed of the moving interfaces within the sound beam. The term "speed" from a physics viewpoint is proper because the magnitude of the movement (a scaler quantity), and not the absolute direction of movement, is of interest. Nevertheless, the term "velocity" has been used traditionally instead of the term "speed" in Doppler ultrasound applications.

DOPPLER EFFECT

The Doppler effect is a physical phenomenon in which an apparent change in the frequency of the sound is observed if there is relative motion between the source of the sound and the receiver of the sound. A sound source produces a series of concentric rings of waves moving away from the sound source. A similar effect can be observed by dropping a stone on the surface of a smooth pond. Concentric ringlets are produced with the most peripheral ring (wavecrest) being the oldest wavecrest and the innermost ring, the newest wavecrest. The medium determines the speed of the waves, and the source determines the frequency of the waves. These combine to uniquely describe the wavelength of the wave (the distance from crest to crest). A receiver sitting still on the water would observe the same number of wavecrests per unit time (frequency) as are being emitted by the source. If, however, the source moves toward the receiver, the receiver observes a greater number of wavecrests per unit time (an increased frequency). The wavecrests are pushed together by the motion of the source. If the source moves away from the receiver, fewer wavecrests per unit time corresponding to a decrease in frequency are observed. The wavecrests are pulled apart by the motion of the source. If the receiver is moving toward a stationary sound source, the receiver will also observe an increased frequency, since the wavecrests are encountered more frequently. Similarily, a decrease in frequency will be observed if the receiver is moving away from the stationary sound source. A change in the observed frequency will also occur if both the sound source and receiver are moving.

The difference between the transmitted frequency and the observed frequency is called Doppler shift. The amount of change in frequency will depend on how rapidly the sound source and/or receiver are moving. That is, an increase in the relative velocity between source and receiver will cause a greater change in the observed frequency. The Doppler shift is also affected by the velocity of sound in the medium and the transmitted frequency.

The Doppler shift (f_D) produced by scanning a moving interface in tissue can be calculated using the following equation:

$$f_D = \frac{2vf}{c} \tag{1}$$

where c is the velocity of sound in tissue, v is the velocity of the interface, and f is the frequency of the transducer. Equation 1 is, in reality, an approximation based on the assumption that the speed of the interfaces for biologic systems are relatively small (0.5 to 200 cm/sec), compared with the velocity of sound in tissue (1,540 m/sec). A specific numeric

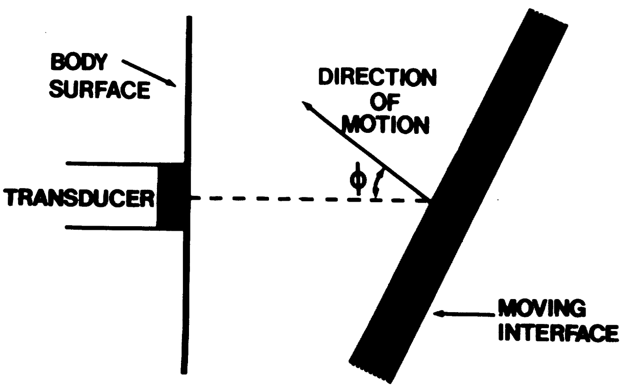

FIG. 1. Motion of interface, which is not parallel to direction of travel of sound beam. The angle ϕ is used in the Doppler shift equation.

example will best illustrate the magnitude of the Doppler shifts observed in ultrasound scanning. Assume that a 5-MHz transducer is aimed at an interface moving toward the transducer with a velocity of 15 cm/sec. The observed frequency is 974 Hz above the original transmitted frequency of 5 MHz. If the interface were moving away from the transducer at 15 cm/sec, then the observed frequency would be 974 Hz below the original transmitted frequency.

Equation 1, however, is only valid if the direction of motion between the sound source and moving reflector is parallel to the direction of sound wave propagation. The equation can be modified for situations in which the sound beam is incident at some angle other than 0° with respect to the motion of the reflecting interface (Fig. 1). This angle is referred to as the Doppler angle. Multiplying equation 1 by the cosine (cos) of the Doppler angle yields the correct equation for the Doppler shift:

$$f_D = \frac{2vf \cos \phi}{c} \tag{2}$$

For the previous example, assume the angle between the transducer and the direction of motion of the interface is 45°. The Doppler shift is now found to be 689 Hz instead of 974 Hz obtained for parallel incidence. The actual determination of this angle may be very difficult. The minimum shift should occur at 90° incidence, because the cos 90° = 0. In practice, the signal never goes to zero, because some portion of the beam is not perpendicular to the motion because of divergence. Using the position corresponding to the minimum shift as a reference point, the incidence angle can be determined by measurement. The maximum shift will occur when the transducer is oriented parallel to the direction of motion, where the cos 0° = 1. Some of the newer Doppler instruments allow the operator to specify the direction of flow on an image and then automatically calculate the Doppler angle.

Doppler units are designed to extract the Doppler shift(s) from the received signals. This change in frequency as illustrated above is in the audible range (between 20 Hz and 20,000 Hz), which enables audioamplifiers with earphones or loudspeakers to be used as output devices. A strip chart recorder may be utilized to generate a hardcopy printout of the Doppler shift(s). The frequency spectrum depicting multiple Doppler shifts could be presented on a video display.[1]

CONTINUOUS-WAVE DOPPLER

Continuous-wave (CW) Doppler units utilize two crystals in the transducer: one to transmit the signal and the other to receive the reflected sound waves. One crystal cannot send and receive at the same time. The transmitter emits a continuous sound wave of constant frequency. The moving interfaces (e.g., heart, vessel wall, red blood cells) acts as "receivers" of the ultrasonic beam. These moving "receivers" will see a change in frequency (increased or decreased), depending upon whether their motion is toward or away from the transducer. Interfaces that are stationary do not experience a change in frequency. Some of the ultrasonic beam will be reflected from the moving interfaces (i.e., the frequency will be shifted up or down, depending upon whether the interfaces are moving toward or away from the sound source) and return toward a second crystal in the transducer, which acts as a receiver (Fig. 2). Even though the receiving transducer is stationary, another change in frequency will occur because the moving interfaces are now acting as a sound source. These two shifts in frequency are responsible for factor 2 in the Doppler shift equation.

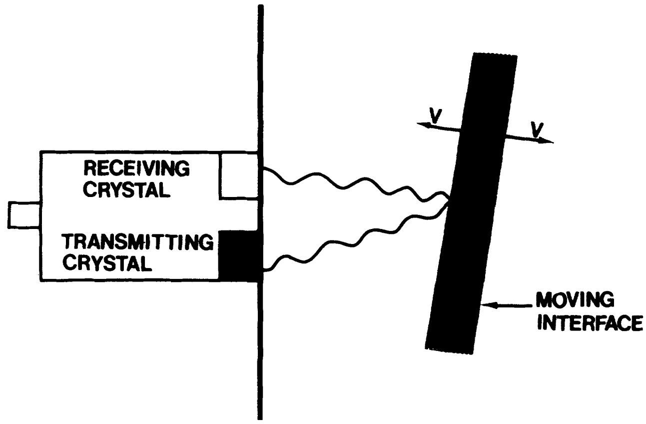

FIG. 2. Continuous-wave Doppler transducer. The transducer has two crystals, one transmitting continuously and the other receiving continuously.

The method used to measure the Doppler shift is based on the principle of wave interference. The reflected wave received from a moving interface varies slightly in frequency, compared with the original transmitted wave, because of the Doppler

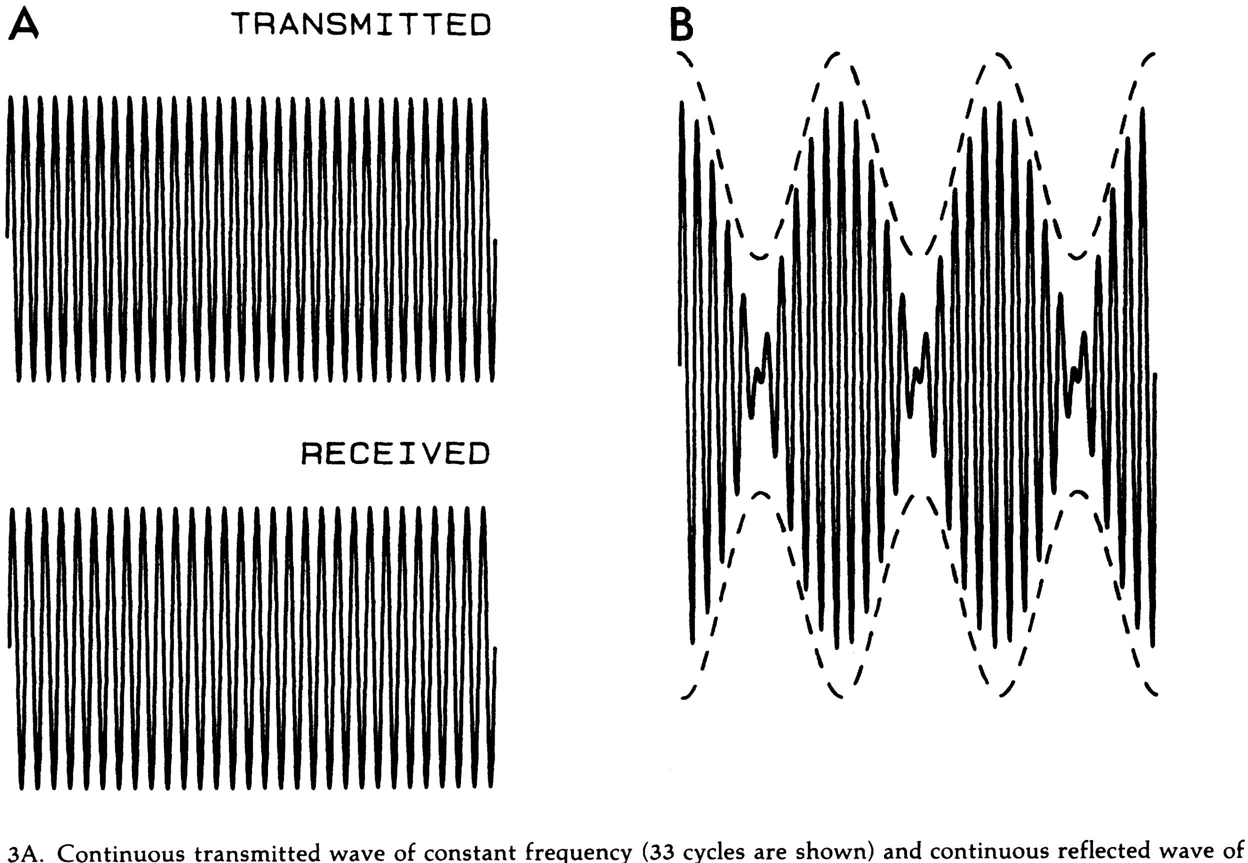

FIG. 3A. Continuous transmitted wave of constant frequency (33 cycles are shown) and continuous reflected wave of constant frequency (30 cycles are shown). B. Addition of the transmitted and received sound waves in (A) results in complex waveform. The beat frequency is illustrated as the outer envelop of this complex waveform.

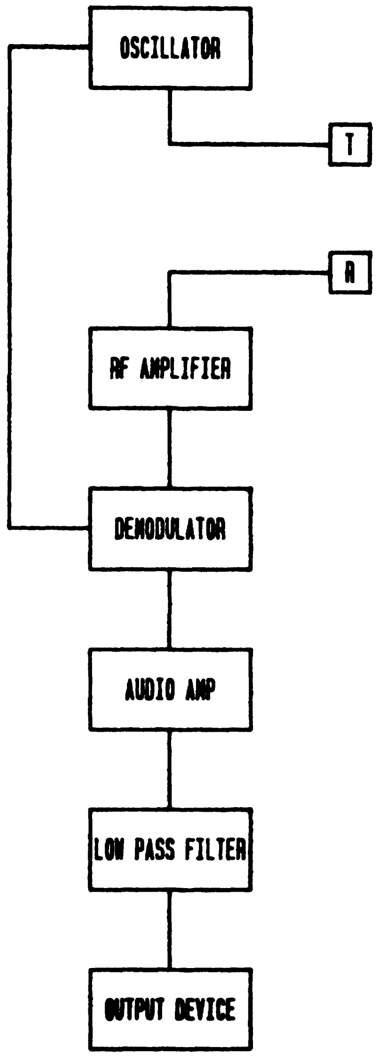

FIG. 4. Block diagram of CW Doppler unit.

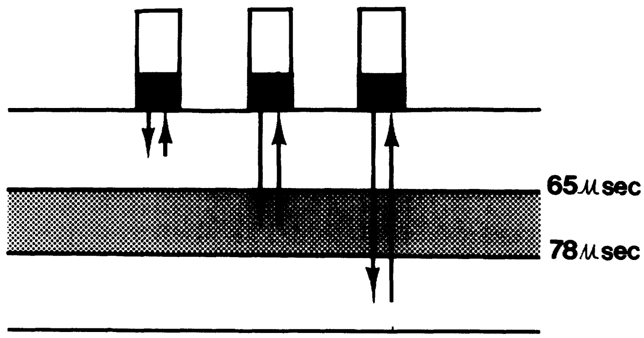

FIG. 5. Gating of the PW Doppler unit for depth discrimination. Sound waves reflected from interfaces in the shaded region are received and processed, whereas signals generated from outside the shaded region are received but are not processed. A change in the gating time will allow sampling of different depths.

shift. Waves of different frequencies algebraically add together, giving a resulting frequency called the "beat" frequency (Fig. 3). The beat frequency corresponds to the Doppler shift. The block diagram in Figure 4 illustrates the steps necessary to generate the Doppler signal. The transmitted frequency is combined with the received frequency, which creates a beat frequency by wave interference. The resultant wave is then demodulated to remove all but the beat frequency. This signal is sent to an audioamplifier, filtered to remove unwanted signals, and then routed to a loudspeaker or earphones for audible "display." Filtering removes the low frequencies from slowly moving structures, such as vessel walls, which are normally not of major interest and could mask other signals. As the speed of flow becomes greater, a higher pitch (greater change in frequency) is heard.

The CW Doppler unit as described here can detect only the speed of movement. The unit must be modified to determine the actual direction of motion (that is, whether movement is toward or away from the transducer), such as in blood flow. The various directional methods are discussed later.

No scanning arm to denote the position of the transducer is necessary because the Doppler shift corresponding to motion along the direction of sight of the transducer is detected. However, the observed Doppler signal can be extremely complex, because it represents the sum of all the Doppler shifts generated by all the moving interfaces within the sound beam. CW Doppler units provide no depth information. Depth discrimination is achieved by pulsing the transmitted signal.

PULSED WAVE DOPPLER

Pulsed wave (PW) Doppler units utilize the echo ranging principle to provide quantitative depth information of the Doppler shift. The transducer is electrically stimulated to produce a short burst of ultrasound, then is silent for a period of time so that one can listen for return echoes before another burst of ultrasound is generated. The received signals are gated for processing, so that only those echoes detected in a narrow time interval after the pulse, corresponding to a specific depth, produce the Doppler signal (Fig. 5). The gate is selectable by the operator, so that the depth of interest can be adjusted.[1,2]

The reflected signals from multiple pulses are mixed with the reference signal (same frequency as the transmitted signal) to produce the beat frequency. The beat pattern is not as well defined as with CW Doppler, because the pulsed echoes are equivalent to sampling the CW signal at discrete intervals (Fig.6). The beat pattern can be more clearly formed if more pulses are used, which would require an increase in the pulse repetition frequency (PRF).

At a minimum, two sample pulses are required per beat cycle in order to define the beat frequency unambiguously. This creates a very important limitation in PW Doppler scanning. The maximum Doppler shift (f_D) that can be detected is related to the PRF:

$$f_D\,(\max) = \frac{PRF}{2} \tag{3}$$

For measurement of reflectors moving with high velocity producing large Doppler shifts, a high PRF is necessary. However, a high PRF will limit the depth that can be sampled, because a certain time is required for collection of the echoes arising from that depth before the next pulse is sent out. The problem becomes more complex when one realizes that the Doppler shift is dependent on the transducer frequency. Nevertheless, the relationship between depth of interest (d), transducer frequency (f), and maximum reflector velocity (V_{max}) can be described in a single equation (Doppler angle assumed to be 0°:

$$V_{max} = \frac{C^2}{8fd} \tag{4}$$

where c is the velocity of sound in tissue. The ramifications from equation 4 are twofold. If the depth of interest is increased, then the maximum reflector velocity that can be measured is decreased. Second, a lower frequency transducer allows greater velocities to be detected. These limitations occur because the motion of the reflector is sampled at discrete intervals and not continuously, as in the CW units. There is no maximum reflector velocity limit for CW units.

If sampling is not adequate for high-frequency Doppler shifts, then artifactual, lower frequency Doppler shifts are displayed. Because the beat frequency is sampled intermittently, the beat frequency must be inferred from the limited data available. If the sampling occurs less than two times during a beat cycle, then the data can be misinterpreted as a lower frequency than it is in reality. As illustrated in Figure 7, imagine that a picture is taken of the amplitude of the beat frequency at various points; then a new waveform is constructed from these images. The actual beat frequency is misinterpreted as a waveform with lower frequency, because sampling occurred only three times over two cycles. This is called aliasing, which is also not present in CW units.

The movie industry provides a visual example of aliasing. In movies of the Old West, a buckboard is often pulled across the prairie by a team of horses. However, the wheels on the buckboard appear to be going backward, which is visually inconsistent with

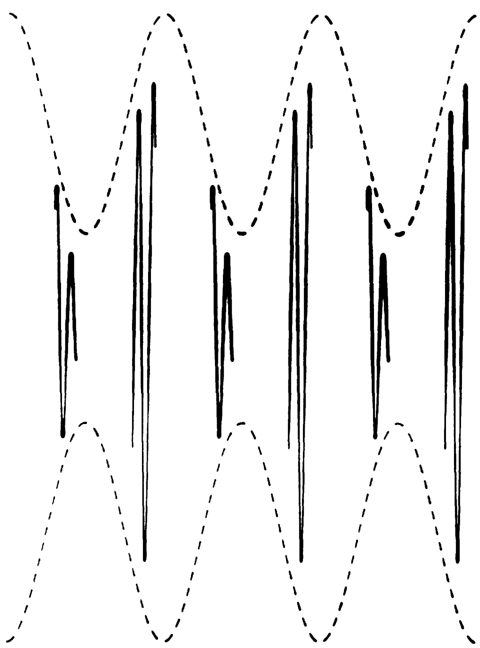

FIG. 6. Addition of pulsed sound waves and the interpreted beat frequency.

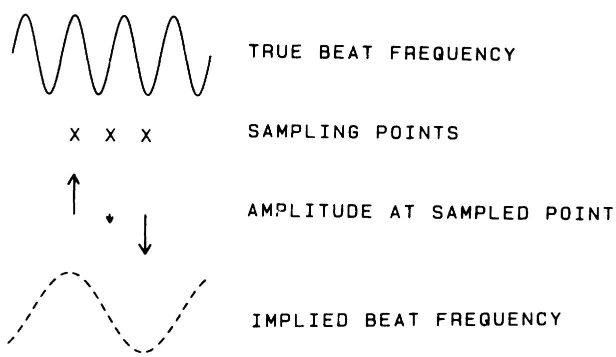

FIG. 7. The true beat frequency is interpreted incorrectly as a lower frequency when the sampling rate is not adequate.

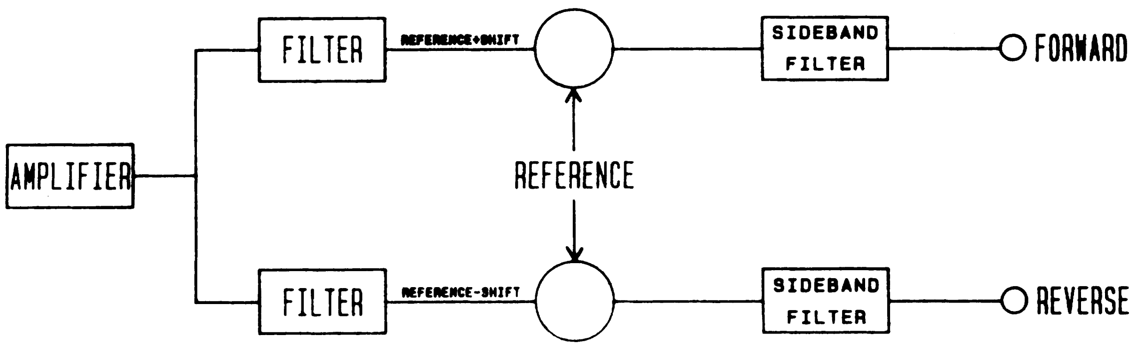

FIG. 8. Block diagram of single-sideband Doppler unit.

the movement of other objects depicted in the scene. In making the movie, a series of stop-action photographs are taken and then shown one after another to give the appearance of motion. However, there is a time delay between frames, which means that the recording system is sampling the motion at discrete intervals. If the motion becomes very rapid, such as the rotating wheel on the buckboard, then the sampling is not adequate to properly represent the motion (the wheel moves a large distance between successive photographs). The solution of this problem is to decrease the time delay between frames (take more photographs) so that the wheel moves a shorter distance between photographs. More frequent sampling allows the recording system to accurately reproduce the motion. Note that in the movie, only objects moving at high velocity are affected by the sampling rate; the motion of slower moving objects is correctly reproduced. In ultrasound the boundary for the correct interpretation of object velocity is given by the maximum Doppler shift, which is dependent on the PRF.

For monitoring flow in vessels the red blood cells act as scattering centers. Because the red blood cells are much smaller than the wavelength of the sound wave, Rayleigh-Tyndall scattering occurs. The intensity of the scattered sound is proportional to the fourth power of the frequency. To produce a strong echo, a high frequency transducer should be used. However, as the frequency is increased the rate of absorption of the sound beam is also increased. These two effects must be balanced in Doppler scanning by matching the transducer frequency with the region of interest (e.g., superficial vessel, 8–10 MHz; carotid, 4–5 MHz; and deep arteries and veins, 2 MHz).

DIRECTIONAL METHODS

The received echo from a moving reflector is shifted in frequency above or below the reference signal, depending on whether the motion is toward or away from the transducer. The process of demodulating the signal indicates there is a shift, but this technique cannot identify whether the shift is positive or negative. Three processing methods, including single-sideband, heterodyne, and quadrature phase detection, have been developed to distinguish between flow toward and flow away from the transducer.[3]

Single-Sideband

The operation of the single-sideband detector system is shown in the block diagram in Figure 8. The received signal is composed of the reflector echoes from stationary structures as well as moving structures. The reflections from the stationary structures are equal in frequency to the transmitted beam, while reflections from the moving structures are offset in frequency.

The signal from the radiofrequency amplifier is split into two components and then filtered. One filter is designed to pass all frequencies above the reference signal (forward motion) and the other is designed to pass all frequencies below the reference signal (reverse motion). The output of each filter is mixed with the reference signal and then filtered by a sideband filter where all components except the shift signal are removed. This results in two separate signals corresponding to the forward and reverse motions, respectively.

Heterodyne

In heterodyne detectors, an offset signal is combined with the reference signal before it is added to the received signal to obtain the Doppler shift. For example, if the offset signal is 5 kHz, then the Doppler shift (f_D) is given by

$$f_D = f_R - (f - 5\ \text{kHz}) \qquad (5)$$

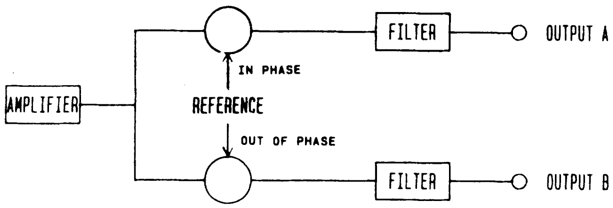

FIG. 9. Quadrature phase detection utilizes mixing the received signal with the reference wave which varies in phase.

where f_R is the frequency of the received signal and f is the transmitted frequency. Note that equation 5 reduces to the usual definition of the Doppler shift (the difference in frequency between the received and transmitted signal) if the offset signal is set to zero. However, the use of the offset signal allows reverse motion (f_R less than f) to be displayed as frequencies below 5 kHz and forward motion (f_R greater than f) to be displayed as frequencies above 5 kHz (usually up to 15 kHz).

Quadrature Phase Detection

Quadrature phase detection is illustrated in Figure 9. The signal from the RF amplifier is split into two components and then each component is mixed with the reference signal (90° out of phase with one another). After filtering, the two output signals are analyzed simultaneously by comparing their relative phase for determination of whether flow is in the forward or reverse direction. However, this detection system does not work properly if forward and reverse flow signals are present at the same time.

COLOR FLOW IMAGING (STATIC MODE)

A two-dimensional image of the flow is constructed by displaying the detected Doppler shifts as the transducer is moved across the scanning area (Fig. 10). Each point in the image corresponds to a particular location of the transducer. The transducer is mounted on a scanning arm (similar to the static B-mode scanner), which enables the position of the transducer to be known. Some devices display the detected Doppler signals as different colors, depending on the magnitude of the frequency shift (color Doppler). Static color flow imaging systems may operate in CW or PW mode.[2]

CW flow imaging systems produce flat-plate images of blood flow across a vessel, but no depth information is possible. Each point in the image is a composite of all the Doppler signals along a single line of sight. Several lines of sight are combined to form a two-dimensional mapping of the flow.

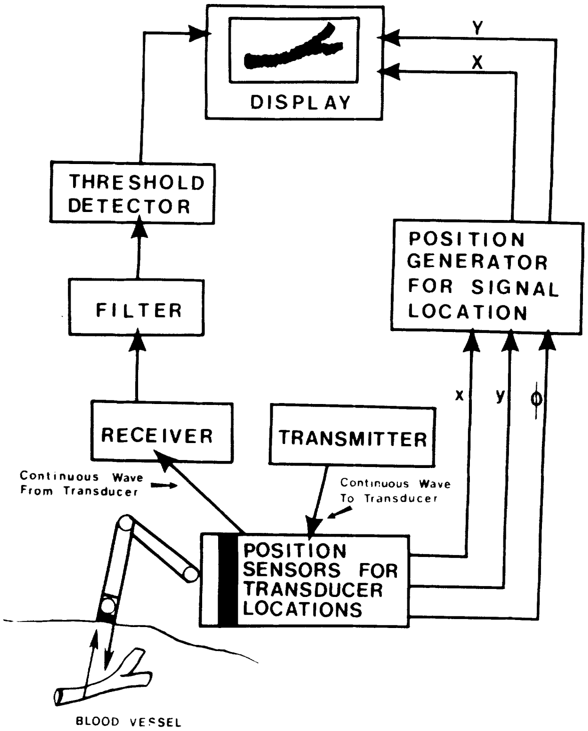

FIG. 10. CW flow imager with scanning arm to denote position of the transducer.

PW flow imaging systems (sometimes called range-gated Doppler) provide depth information and consequently allow flow to be visualized in three dimensions. The depth range along the line of sight is divided into many sections (usually about 30) of 1 mm each. Multiple serial gates are employed for selecting the Doppler signals according to depth. These units can also detect the direction of flow.

DUPLEX SCANNERS

Duplex Doppler units combine real-time imaging with CW or PW Doppler detection. Duplex scanners were developed to overcome two major disadvantages of static color flow imaging: 1) the neighboring stationary structures are not displayed and 2) the scanning time for two dimensional flow mapping is long. Visualization of the physical size and shape of plaque with real-time scanning is an important aid in the diagnosis of vascular disease. The display of anatomical structures allows the operator

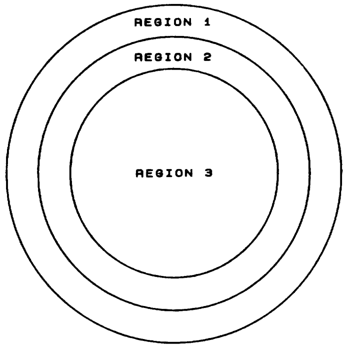

FIG. 11. Vessel lumen divided into three regions of RBC's moving with different velocities.

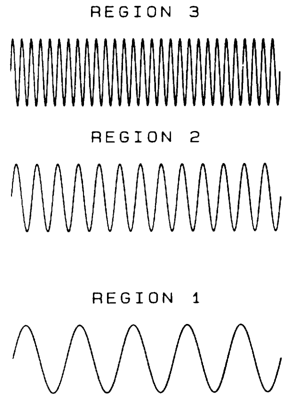

FIG. 12. Beat frequencies for regions 1, 2 and 3. Note that the beat frequency increases with increased velocity of RBC's.

to select the line of sight for CW Doppler or the sample volume for PW Doppler.

The flow patterns in the localized area of interest are displayed as a function of time. The red blood cells across the vessel are moving at different velocities, resulting in a Doppler signal that is a complex combination of all the frequency shifts. The process of determining the individual frequency shifts from this complex Doppler signal is called spectral analysis.

SPECTRAL ANALYSIS

The analysis of the complex Doppler signal is usually accomplished with a mathematical algorithm called the Fast Fourier Transform (FFT). However, other methods, such as the parallel filter band and time compression, have been used to quantitate the individual frequency components in the Doppler signal. Parallel filter band and time compression will be discussed later. Several examples will be used to illustrate the principle of spectral analysis.

A cross-sectional view of a vessel lumen is shown in Figure 11. The regions 1,2 and 3 designate regions where the red blood cells (RBC's) are moving at different velocities ($V_3 > V_2 > V_1$) through the vessel. For simplicity, the same total number of RBC's are assumed to pass through each of the regions and to flow at a continuous rate (nonpulsatile). If the ultrasound beam could be made physically very small so that each of these regions could be sampled individually, then a characteristic Doppler shift would be obtained for each region (Fig. 12). A total of three different frequency shifts would be observed (f_1, f_2, f_3). The frequency shift is largest for region 3, because RBC's in this region are moving at the greatest velocity. Since an equal number of RBC's are present in each region, the amplitude of the individual frequency shifts are the same. This is represented by the heights of the waveforms in Figure 12, which are identical.

If all three regions were sampled by the ultrasound beam at the same time, a very complex Doppler signal would be observed as shown in Figure 13. This complex Doppler signal is an algebraic sum of the three waveforms in Figure 12. This detected signal must be simplified in order to associate groups of RBC's with individual frequency shifts and thus, with rates of movement. Spectral analysis separates the complex signal into the individual frequency components and determines the relative importance of each frequency. That is, the waveform in Figure 13, which is the detected signal, is converted into the various individual frequency shifts shown in Figure 12.

An alternative method to display the spectral analysis is in the form of a power spectrum in which the magnitude of individual frequency components is plotted versus frequency (Fig. 14). The magnitude is determined by the amplitude of the waveform corresponding to the respective shift frequency. The magnitude represents the relative importance of each frequency (the number of RBC's moving at the velocity given by the frequency shift). Suppose that the number of RBC's moving through region 1 is doubled. The amplitude of the beat frequency corresponding to this region will also double and result in a change in the complex Doppler signal (Fig. 15). However, the frequency of the Doppler shift for region 1 remains the same, since the velocity of the RBC's has not changed. The spectral analysis presented by the power spectrum in Figure 16

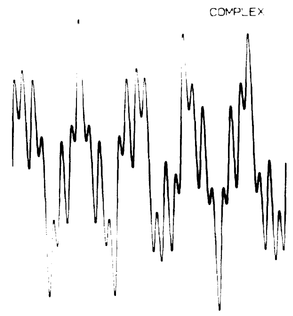

FIG. 13. Complex Doppler signal which is the sum of the waves in Figure 12.

shows the increased importance of the lower frequency by the increased height of the peak corresponding to this frequency.

The relationship between the power spectrum display and the velocity of the RBC's can be further illustrated by the following example. Suppose all the RBC's throughout the vessel are moving slowly at a constant velocity. Then the power spectrum will display a single peak at low frequency (Fig. 17A). If the velocity of the RBC's is now increased to a high rate, then the power spectrum will once again show a single peak but at higher frequency (Fig. 17B).

In practice, blood flow does not occur in discrete velocities, but rather at a wide range of velocities. A velocity profile ranging from zero near the vessel wall to a maximum in the central portion of the vessel would result in a continuous power spectrum as shown in Figure 18.

In vessels the flow pattern is not constant with time, but rather, pressure changes give rise to pulsatile flow. Consequently, it is best to display the power spectrum as a function of time. Three variables (frequency, magnitude, and time) must be included in this display. The magnitude in the

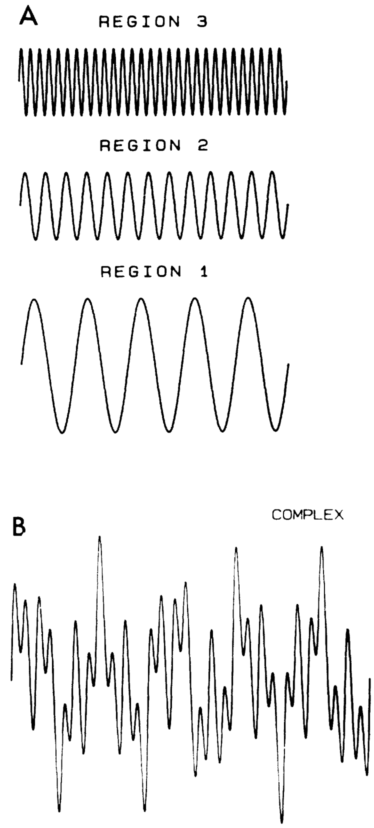

FIG. 15A. Beat frequencies for regions 1, 2 and 3. Region 1 contains twice as many RBC's as the other regions. B. Complex Doppler signal which arises from doubling the number of RBC's in region 1.

power spectrum is now represented by varying the brightness level to indicate the relative importance of each frequency. Consider the example in Figure 19A in which high velocity group of RBC's produces twice the signal as the mid-velocity group which in turn produces twice the signal as the low-velocity group. This information can be converted to points of varying brightness along a frequency axis (Fig. 19B). The distance along the axis represents increas-

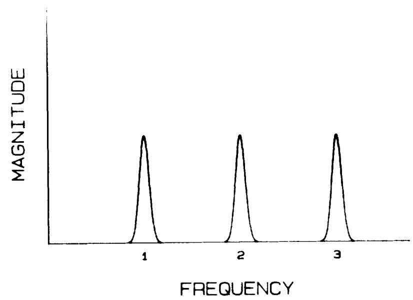

FIG. 14. Power spectrum of the complex Doppler signal in Figure 13. The peak corresponding to each region is indicated.

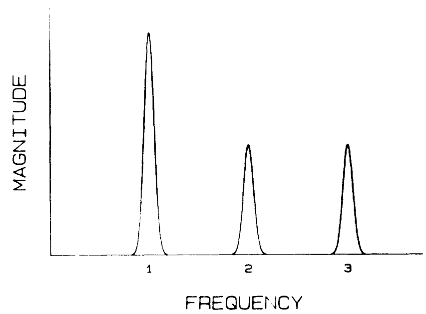

FIG. 16. Power spectrum of the complex Doppler signal in Figure 15.

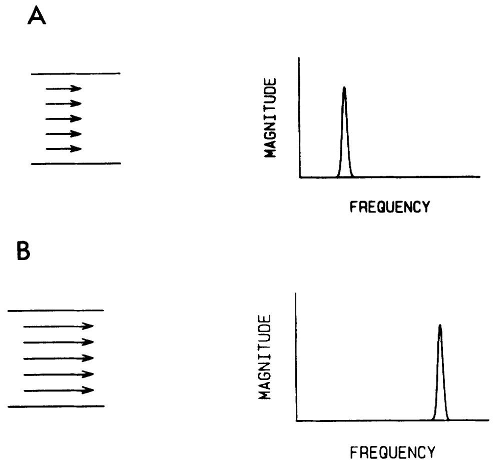

FIG. 17A. Power spectrum of RBC's moving uniformly at slow velocity. B. Power spectrum of RBC's moving uniformly at fast velocity.

ing frequency. Note that in this example the high frequency is brighter than the mid-frequency which, in turn, is brighter than the low frequency.

The dimension of time is obtained by sampling the Doppler signal repeatedly in small time increments on the order of a few milliseconds. The FFT frequency analysis is then applied to each short-time segment of the Doppler signal. The display of these multiple analyses consists of a vertical axis corresponding to frequency, a horizontal axis corresponding to time, and varying brightness levels representing magnitude. Each analysis of a short-term segment of the Doppler signal is presented as a single vertical line. By placing succeeding frequency analyses side by side (the vertical lines are used to build up a pattern), a profile of the time changing velocities within the vessel is obtained (Fig. 20).

ALTERNATIVE METHODS OF SPECTRAL ANALYSIS

With advances in the speed of computer systems, the FFT technique has rapidly become the method

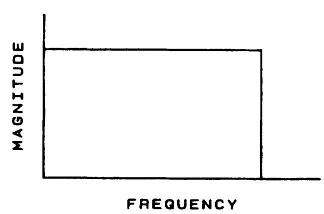

FIG. 18. Power spectrum of RBC's moving with varying velocities.

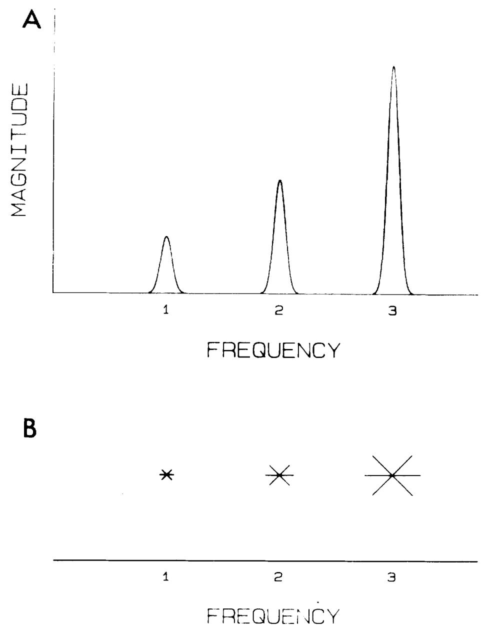

FIG. 19A. Power spectrum of three discrete groups of RBC's in which the signal decreases in importance from high frequency to low frequency. B. Gray level representation of the power spectrum in (A).

of choice for analyzing the Doppler signal. However, the methods of parallel filter bank and time compression have also been employed for frequency analysis. The parallel filter bank analyzer contains a large number of filters in which each filter passes signals within a narrow frequency range. The analysis is performed in real time by simultaneously introducing the complex Doppler signal to each of the filters (Fig. 21). The disadvantage of this system is that the large number of filters required for the simultaneous analysis are very expensive.

In time compression the complex Doppler signal is sampled for a short period of time (8 msec) and then stored in digital memory. Once stored, the Doppler signal can be played back at an accelerated rate (250 times normal speed). The time compressed signal is analyzed during the time that the next Doppler signal is being collected for analysis. A filter with a bandwidth of 150 Hz is swept through a wide frequency range (200 Hz to 15 kHz). That is, the filter is initially set at a baseline of 200 Hz with a window of 150 Hz and the Doppler signal is examined for frequency components in this range. The baseline is then incremented by 150 Hz and the Doppler signal is re-examined for frequency components in this new range. This process is repeated several times until the Doppler signal is analyzed

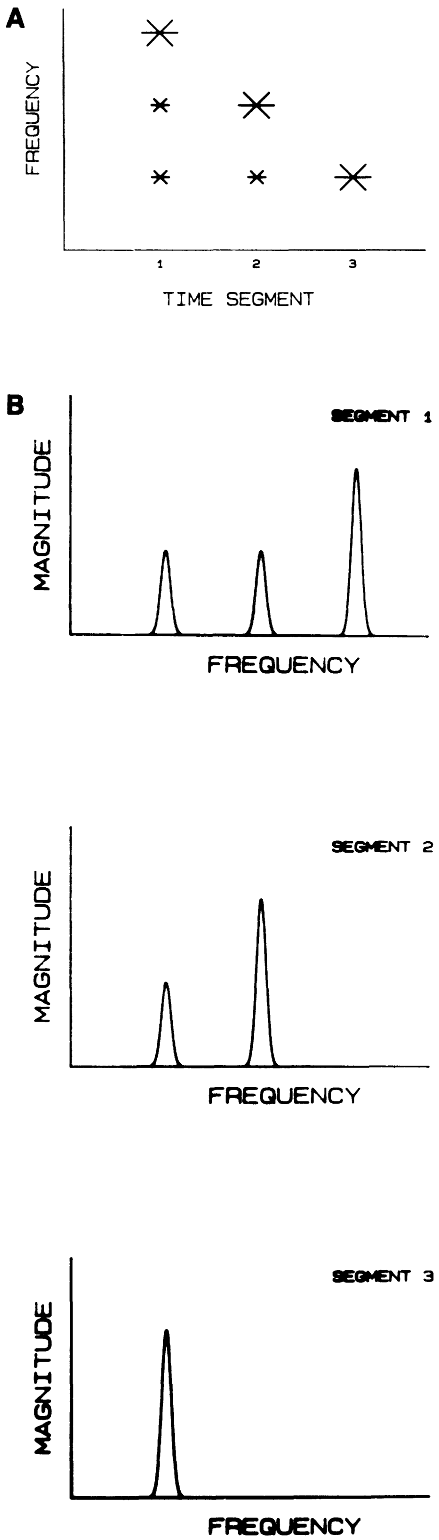

FIG. 20A. Time sequence of gray level representation of varying power spectra which are shown in (B). B. Power spectra corresponding in the three different time segments in (A).

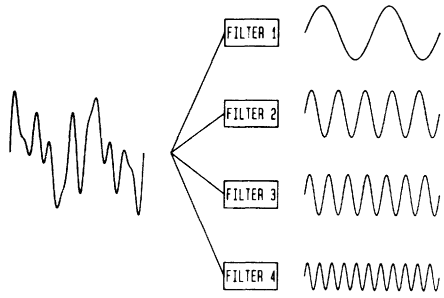

FIG. 21. Parallel filter bank in which four filters analyze the complex Doppler signal with respect to four different frequencies.

throughout the entire frequency range. The accelerated rate of playback is necessary in order to complete the analysis in real time.

ZERO CROSSING DETECTORS

Zero crossing detectors are designed to monitor how rapidly the complex Doppler signal changes back and forth between maximum to minimum values. This provides an indicator of the frequency shifts which make up the Doppler signal.

Each time the Doppler signal crosses zero in one direction (e.g., moving from below zero to above zero), a trigger pulse is generated (Fig. 22). In order to reduce the influence of electronic noise the Doppler signal must exceed a threshold value above zero to produce the trigger pulse. The number of pulses per second gives rise to the zero crossing frequency, which represents an averaging of all the frequencies in the Doppler signal. The zero crossing frequency depends upon flow changes. That is, the

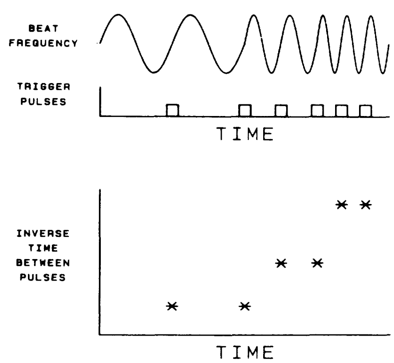

FIG. 22. A trigger pulse illustrated by squares along the time axis is generated whenever the complex Doppler signal crosses zero from below. The time interval histogram corresponding to the Doppler signal is also shown.

zero crossing frequency will increase if the velocity of flow becomes faster.

The output of the zero crossing detector can be modified so that each zero crossing pulse is presented as a dot on a display. The time interval histogram (or TIH) is generated as a function of time by positioning the dot in the vertical direction according to the reciprocal of the elapsed time from the previous pulse (Fig. 22). If two zero crossings occur rapidly one right after the other, then the dot corresponding to the last zero crossing will be located high on the vertical scale. The time dependence of the rate of zero crossings is visualized, which yields a pattern that is similar to that obtained by spectral analysis.

COLOR FLOW IMAGING (REAL-TIME MODE)

Color flow imaging in real time or angiodynography is a relatively new imaging technique that combines two dimensional gray scale images with Doppler flow images in real time (typically 18 frames/sec). Motion designated by the colors red (forward motion) and blue (reverse motion) is depicted throughout the entire field of view. A linear array or phased array transducer is most commonly used for detecting both stationary and moving structures by analyzing the received signal with respect to amplitude (signal strength), phase, and frequency. A change in phase indicates the presence of motion and further, the direction of the motion. The frequency shift gives the velocity of the motion (via the Doppler equation), which is represented by changing the intensity of the color. Stationary structures do not produce a phase shift, and thus are assigned a gray level based on signal strength. Every pixel throughout the image is evaluated in this manner. Dynamic focusing enables pixels of constant size (depending on the transducer frequency) to be depicted throughout the field of view.[4]

The major disadvantage of duplex scanning is that flow is not evaluated simultaneously throughout the field of view, but rather sampled at a particular location as selected by the operator. In order to establish the flow pattern, multiple sites must be sampled, which requires precise positioning of the sample volume. The repetition of the sampling also is a time consuming process.

By displaying the two dimensional spatial distribution of the velocities and the temporal changes in these velocity patterns, real-time color flow imaging enables regions of flow disturbance to be easily visualized. The FFT spectral analysis can then be performed for selected regions of interest. Since areas of flow disturbance are rapidly identified, angiodynography has the potential to greatly reduce patient examination time.

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