AAPM/RSNA Physics Tutorial for Residents: Topics in US


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Techniques of Doppler ultrasonography (US) have been available to clinicians for nearly 40 years. The Doppler effect as developed by sound propagation in human tissues and with the velocities observed for the human vasculature produces shifts in the frequencies of returning echo signals. These signals can be processed in a manner that allows the observer to determine the condition of the blood flow. The instrumentation for Doppler US has evolved to accommodate the expanding clinical use of US. Each development (eg, pulsed-wave Doppler US, color flow imaging) has been motivated by a desire to provide more clinical information about flow in the body. The algorithms used are complex, but increasingly powerful microelectronics have made these methods a reality at a reasonable cost. Users of Doppler US techniques must be aware of the complicated aspects of flow in the body, especially with regard to detection of disease in the human vasculature. The continuing development of US equipment aims to provide a greater understanding of hemodynamics and the relationship between blood flow and various disease processes.

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Introduction
The Doppler effect produced with ultrasonic frequencies has been used in medicine for almost 50 years (1,2). In that time, advances in electronics and processing have made it possible to use the Doppler effect not only to determine whether flow is present but also to visualize flow in two and even three dimensions while providing semiquantitative information. Although the underlying physics of the Doppler effect are fairly simple, the instrumentation and how the information is applied for diagnosis of disease can be complex and involve a number of assumptions.

The purpose of this summary is to present these concepts to an audience of clinical users. Specific topics discussed are the basic physics of Doppler ultrasonography (US), basic Doppler US technology, flow velocities and frequency shifts, types of blood flow and the physics of flow resistance, flow parameters (derivations and meaning), and recent innovations in Doppler US techniques.

Basic Physics of Doppler US

One will detect the Doppler effect by standing next to a street as traffic passes. By noting the pitch of the sound emanating from vehicles (horns, music, engine sounds) and paying attention to the pitch change as the vehicle direction changes from approaching toward the observer to moving away, the Doppler effect is experienced. The degree to which the frequency of sound coming from the vehicle changes pitch is directly proportional to the velocity of the vehicle. For example, the Doppler frequency shifts (up and down) will increase as one moves from a residential street to a highway or a racetrack. Why does this happen?

The shift in frequency is related to the contraction or expansion of wavelengths ahead of or behind the sound-emitting moving object (Fig 1). The wavelength of the sound, \( \lambda \), is the speed of sound propagation, \( c \), divided by the frequency of the sound, \( f \). As sound speed is defined in units of length divided by time and frequency is the number of cycles per unit time, wavelength is expressed in units of length. The velocity of the moving object, \( v \), is also in units of length divided by time. As the sound is emitted from the moving object, the wavelengths observed at points on either side of the object are lengthened (\( \lambda_1 \)) or shortened (\( \lambda_s \)). Because wavelength is inversely proportional to frequency, the observer will detect a frequency different from that emitted by the object when it has zero velocity:

\[
\lambda_1 = \frac{c}{f - \Delta f} = \frac{c}{f} - \frac{c}{\Delta f}
\]

(1)

\[
\lambda_s = \frac{c}{f + \Delta f} = \frac{c}{f} + \frac{c}{\Delta f}
\]

(2)

In Equations (1) and (2), \( f \) is the frequency of the sound emitted by the object and would be detected by the observer if the object were at rest. \( \pm \Delta f \) represents a Doppler effect–induced frequency shift; the sign depends on the direction in which the object is traveling with respect to the observer. Note that these equations apply to the specific condition that the object is traveling either directly toward or directly away from the observer.

In the case of ultrasound scattering back from moving objects (eg, red blood cells) in the body, the derivation of the Doppler equation may follow that of Wells (3). Because ultrasound is used in a transmit-echo approach, there is a Doppler effect with the sound arriving at the scattering object and a Doppler effect as the sound is reflected from that object back toward the ultrasound transducer. In US, the round trip time for sound is related to the depth and the speed of sound in tissue.

As seen in Equations (1) and (2), an inverse relationship exists between frequency and the wavelength of the sound transmitted to the red blood cells. The sound speed is a constant (\( c_{\text{tissue}} = 1,540 \text{ m sec}^{-1} \)) in this relationship. The frequency of the sound incident on the red blood cell is changed because the relative velocity of the
The velocity of red blood cells relative to the position and angle of the transducer depends on the direction of sound propagation and the motion of the particle.

The red blood cell is added to the sound propagation speed. The variables can be rearranged in the form of the transmitted frequency, \( f_0 \), and an object velocity term:

\[
f_0 = \frac{c}{\lambda_0} \rightarrow \lambda_0 = \frac{c}{f_0}.
\] (3)

\[
f_{\text{RBC}} = \frac{c + v_{\text{RBC}}}{\lambda_0} = f_0 + \frac{v_{\text{RBC}}}{\lambda_0}.
\] (4)

In Equations (3) and (4), \( f_0 \) is the frequency of sound produced in a transducer and propagating toward a red blood cell, \( c \) is the speed of sound in tissue (assumed to be 1,540 m sec\(^{-1}\)), and \( \lambda_0 \) is the wavelength of the sound initially sent toward the red blood cells. The frequency of the sound as it is perceived by the red blood cell is defined in Equation (4) as \( f_{\text{RBC}} \), and the velocity of the red blood cell is \( v_{\text{RBC}} \).

The sound is then reflected back toward the transducer in the echo from the red blood cell. The frequency of sound arriving back at the transducer is shifted again in proportion to the red blood cell velocity:

\[
f' = f'_{\text{RBC}} + \frac{v_{\text{RBC}}}{\lambda_{\text{RBC}}}
\] (5)

\[
= f_0 + \frac{v_{\text{RBC}}}{\lambda_0} + \frac{v_{\text{RBC}}}{\lambda_{\text{RBC}}}.f_0.
\]

The equation for the frequency returning back to the transducer, \( f' \), can be rewritten once again with respect to the original frequency, \( f_0 \), and the wavelengths of the incoming and outgoing waves.

These relationships are now substituted into the following equation to put everything in terms of the red blood cell velocity, \( v_{\text{RBC}} \), the original frequency, \( f_0 \), and the speed of sound propagation, \( c \):

\[
f' = f_0 + \frac{v_{\text{RBC}}}{\lambda_0} + \frac{v_{\text{RBC}}}{\lambda_{\text{RBC}}}
\]

\[
= f_0 + \frac{v_{\text{RBC}}f_0}{c} + \frac{v_{\text{RBC}}f_{\text{RBC}}}{c}
\]

\[
= f_0 + \frac{(v_{\text{RBC}})}{c}(f_0 + f_{\text{RBC}})
\]

\[
= f_0 + \frac{(v_{\text{RBC}})}{c}(f_0 + f_0 + \frac{v_{\text{RBC}}}{\lambda_0})
\]

\[
= f_0 + \frac{(v_{\text{RBC}})}{c}(2 \cdot f_0 + \frac{v_{\text{RBC}}f_0}{c})
\]

\[
= f_0 + 2f_0\left(\frac{v_{\text{RBC}}}{c}\right) + f_0\left(\frac{v_{\text{RBC}}}{c}\right)^2.
\] (6)

The Doppler shift frequency, \( f_D \), is the difference between the returned frequency, \( f' \), and the transmitted (or original) frequency, \( f_0 \). Subtracting \( f_0 \) from both sides yields the final result:

\[
f_D = f' - f_0
\]

\[
= f_0 + 2f_0\left(\frac{v_{\text{RBC}}}{c}\right) + f_0\left(\frac{v_{\text{RBC}}}{c}\right)^2 - f_0
\]

\[
= 2f_0\frac{v_{\text{RBC}}}{c} + f_0\left(\frac{v_{\text{RBC}}}{c}\right)^2.
\] (7)

Since \( v_{\text{RBC}} \ll c \),

\[
f_D \approx \frac{2f_0v_{\text{RBC}}}{c}.
\] (8)

The term that is squared at the end of Equation (7) can be approximated as zero if the ratio of red blood cell velocity and sound speed is low. Since \( v_{\text{RBC}} \) is generally less than 0.1% of the sound speed, \( c \), this assumption is reasonable. Whether \( f_0 \) is positive or negative depends on the sign of the velocity term, \( v_{\text{RBC}} \). \( f_0 \) is directly proportional to the velocity (eg, if \( v_{\text{RBC}} = 0 \), then \( f_0 = 0 \)).

Now consider the fact that the relative velocity (to be indicated as \( v_r \)) between the red blood cell and the transducer is dependent on the angle of a straight line (the line down which the sound is traveling) and the red blood cell and the direction of the red blood cell motion (Fig 2). In other words, if the red blood cell were traveling directly toward the transducer (or directly away), the relative velocity would be at a maximum (or the negative maximum). When cosine \( 0^\circ = 1 \), \( v_r = v_{\text{RBC}} \) at \( 0^\circ \); when cosine \( 180^\circ = -1 \), \( v_r = -v_{\text{RBC}} \) at \( 180^\circ \).
Since cosine $90^\circ = 0$, $v_r = 0$ at $90^\circ$. This relationship can also be written as $v_r = v_{RBC} \cdot \cos \theta$, where $\theta$ is the angle indicated in Figure 2.

Plugging this relationship into Equation 8, we obtain the Doppler equation as it is seen in many radiology texts:

$$f_D = \Delta f = \frac{2 \cdot f_0 \cdot v_r}{c} = \frac{2 \cdot f_0 \cdot v_{RBC} \cdot \cos \theta}{c}.$$  \hspace{1cm} (9)

This equation may be rearranged to solve for the velocity of the red blood cells:

$$v_{RBC} = \frac{f_D}{f_0} \cdot \frac{c}{2 \cdot \cos \theta}. \hspace{1cm} (10)$$

where the velocity is estimated by the ratio of the Doppler shift frequency to the original transmit frequency, multiplied by the sound speed and divided by two and the angle correction. It will be seen in the next section that the application of this equation in basic types of US equipment will produce estimates of the blood flow velocity from the frequency shifts. These estimates are obtained with varying algorithms and equipment arrangements. These velocity estimates provide valuable clinical data. However, one should always consider that there are a number of complicating aspects when evaluating these data. Some of these aspects are related to the geometry of the blood vessel and the ultrasound beam; others are related to the varying blood flow velocities across the vessel lumen and the variation of blood flow velocity with the cardiac cycle. In addition, certain compromises are used to achieve an efficient measurement of blood velocity that can be resolved in both time and space. These issues are discussed in the remainder of this article.

**Basic Doppler US Technology**

**Continuous-Wave Doppler US**

Conceptually speaking, the simplest technology for detection of flow with US is continuous-wave Doppler US. The term *continuous wave* means that sound is emitted from a transmitting transducer 100% of the time. Sound that echoes back must then be detected by a second receiving transducer. A continuous-wave instrument has an arrangement whereby the two transducers have some overlap of their beams, resulting in a region...
of interest where the Doppler shifted signals may be detected (Fig 3). The Doppler shifted frequencies are obtained by comparison of the transmitted signal with the received signal.

Figure 3 is a schematic of the continuous-wave Doppler instrument. In this figure, the continuous sound source is transducer #1. The backscattered sound is detected by transducer #2. In the analysis circuit, the outgoing signal frequency, \( f_0 \), is compared with the returning signal to determine the shift in frequency related to moving structures. That comparison is achieved by mixing the transmitted signal with the incoming signal, then filtering out the high-frequency result, leaving the Doppler shifted frequencies to pass through to the remaining instrument processing (e.g., directional detection and a spectrum analyzer). This is referred to as mixing-demodulation (4). Anything that is moving (and causing phase shifts) within the sensitive area of the two transducers will result in some difference signal frequency \( f_D \).

**Quadrature Detection of Flow Direction**

In Equation (9), the difference signal frequency, \( f_D \), may be in either the positive or the negative direction, depending on the direction of flow with respect to the transducers. A quadrature detection scheme is used to produce an output that differentiates between forward and reverse flow (Fig 4). For quadrature detection used in Doppler US, the echo signals are sent through demodulators (5). Similarly to the Doppler shift detection described earlier, the demodulator is a circuit that mixes in a reference signal frequency to the input and filters out the higher-frequency components of the resulting mixed signal. In so doing, this output reflects both the Doppler shifted frequencies and the reference. The quadrature detector uses two of these demodulator circuits, with the difference in the reference signal phase being one quarter of the period (hence the name quadrature detector). The relative phase of the output of the two demodulator circuits depends on whether the input signal, \( f_D \), is greater than or less than the reference frequency, which is sometimes called the pilot frequency. The signal can then simply be segmented into forward or reverse channels. In many instruments, these forward and reverse channels are played on opposing speakers (left and right). The Doppler shifted frequencies (in the audible range) are perceived in stereo.

**Frequency Content and Display of Doppler Information**

The Doppler shifted frequencies returned from a vessel are not of a single frequency. A range of frequencies corresponding to a range of blood flow velocities is present in the backscattered signal. The term spectral broadening is used to refer to the fact that the Doppler shifted frequencies are not confined to a narrow range, but exist over a wide range of frequencies (zero frequency shift to the maximum). This will be discussed in more
detail in the next section. This range of frequencies will also vary over time and the duration of the cardiac cycle. The output of the quadrature detection circuit is typically fed into a spectrum analyzer, which in turn segments the frequency content of the signal according to the relative signal strength at each interval frequency. An instantaneous frequency spectrum is plotted against time (sometimes on a moving display). In Figure 5a, this spectrum is shown in an inverted gray scale, with the Doppler shift frequencies with higher power displayed with darker pixels and those with lower power displayed in lighter shades of gray.

**Spectral Broadening**

The term *spectral broadening* is used to describe an increasing frequency spread under certain conditions. One way to view spectral broadening is to compare the maximum Doppler shift frequency \( f_{\text{max}} \) at a given point in time with the mean Doppler shift frequency \( f_{\text{mean}} \) at the same point in time. A narrower spectral range would result in an \( f_{\text{mean}} \) that is closer in value to \( f_{\text{max}} \). A broader spectral range will result in an \( f_{\text{mean}} \) that is somewhere between \( f_{\text{max}} \) and no Doppler shift frequency (Fig 5b). There are two possible reasons for broadening of the spectrum. One possible explanation is a set of flow velocities existing over a broad range, thereby producing a broad range of Doppler shifted frequencies. Such a condition of flow may have diagnostic implications. The other mechanism that may produce spectral broadening is the geometry of the transducer and the ultrasound beam interrogating the vessel. If the beam is narrow but the transducer is located close to the vessel and has a wide footprint, there are multiple beam angles between the transducer and the vessel. In Equations (9) and (10), it was assumed that there was a single Doppler angle, \( \theta \). However, with multiple angles, the effect that is produced is a range of Doppler shifted frequencies. This effect is sometimes referred to as *intrinsic spectral broadening* and can be mitigated by using a Doppler angle as far away as possible from 90°.

**Pulsed-Wave Doppler US**

Pulsed-wave Doppler US is an extension of the ideas developed for continuous-wave Doppler US, but now pulsed sound is used instead of continuous sound. In B-mode US, the pulse-echo range equation is used to determine the depth of returning echoes for the purpose of mapping these data to a two-dimensional image (10). The distance \( d_{\text{object}} \) between the transducer and the object producing an echo \( d \) at a particular time \( t_{\text{echo}} \) represents the round trip of sound to and from that object:

\[
 t_{\text{echo}} = \frac{2 \cdot d_{\text{object}}}{c} \Rightarrow d_{\text{object}} = \frac{t_{\text{echo}} \cdot c}{2} .
\]  

(11)

The pulse-echo range equation may be used to locate the depth of the Doppler shifted frequencies (6). Although this provides tremendous clinical advantages, because the Doppler shifts are now being sampled over time, the detection of frequency shifts is subject to aliasing due to inadequate sampling. Pulsed-wave Doppler US is also referred to as *duplex Doppler US* (7), which refers to the fact that the instrument shares time and interrogation of acoustic pulses with a B-mode
operation, thereby providing both a B-mode image and pulsed-wave Doppler information.

Figure 6a is a schematic of the pulsed-wave Doppler instrument. A single transducer is used in pulse-echo mode. The other boxes indicate the flow of data through the instrument and the stages of Doppler shift processing for pulsed-wave US. (b) Image display from pulsed-wave Doppler US. Red line indicates the axial line used to interrogate a volume contained within the range gate. The angle correction is also indicated.

![Diagram of pulsed-wave Doppler instrument](image)

**Figure 6.** (a) Schematic of a pulsed-wave Doppler instrument. A single transducer is used in pulse-echo mode. The other boxes indicate the flow of data through the instrument and the stages of Doppler shift processing for pulsed-wave US. (b) Image display from pulsed-wave Doppler US. Red line indicates the axial line used to interrogate a volume contained within the range gate. The angle correction is also indicated.

![Diagram of sample and hold process](image)

**Figure 7.** Sample and hold process of capturing phase shifts in subsequent pulse interrogations within the region of the range gate. Each pulse is launched over the time interval determined by the pulse repetition frequency (PRF). The phase shifts over time are filtered to produce a Doppler shift frequency.

![Diagram of sample gate and Doppler shift](image)

![Diagram of Doppler equation](image)
Aliasing

Aliasing occurs in pulsed-wave Doppler US when the rate at which interrogating pulses are sent to obtain the phase shift information is less than two times \( f_0 \). Aliasing is a phenomenon that occurs in any kind of discrete sampling and is discussed in many texts with regard to digital imaging modalities (8). As can be seen in Figure 7, the data points in the sample and hold circuit are obtained at specific intervals in time (\( \Delta t \)). Because these are discrete samples, Nyquist criteria dictate that there must be at least two samples per period of the maximum frequency. Anything less will result in an artificially lower frequency being reconstructed from the sample and hold data. The sampling rate is dictated by the depth of the range gate in the body. This sampling rate, also known as the pulse repetition frequency (PRF), can reach only a maximum that is allowed by the round trip time between the transducer and the range gate depth (identical to the pulse-echo range equation). This is expressed in the form of an equation:

\[
\text{PRF} = \frac{1}{\text{RTT}} = \frac{c}{2 \cdot \text{range gate depth}}, \tag{12}
\]

where PRF is in samples per second, RTT is the round trip time in seconds, \( c \) is the speed of sound propagation in tissue (1,540 m/sec), and range gate depth is in meters. The maximum Doppler shifted frequency that may be detected without aliasing is \( \frac{1}{2} \) times the PRF:

\[
f_{D_{\text{max}}} = \frac{\text{PRF}}{2} = \frac{c}{4 \cdot \text{range gate depth}}, \tag{13}
\]

where \( f_{D_{\text{max}}} \) is in hertz. Putting this into the context of the Doppler equation (Eq [10]), the maximum velocity that can be detected without aliasing is as follows:

\[
v_{\text{max}} = \frac{c^2}{8 \cdot \text{range gate depth} \cdot f_0 \cdot \cos \theta}. \tag{14}
\]

Equations (12)–(14) may be used to illustrate how the maximum detectable velocity (without aliasing) will change under varying conditions.

Any one of the following changes will allow \( v_{\text{max}} \) to increase (other effects are also indicated):

- Reducing the depth of the range gate will allow an increase in the PRF.
- Reducing the frequency of the transmitted pulse, \( f_0 \), will reduce the Doppler shift frequency.
- Decreasing the angle, \( \theta \), between the beam axis and the vessel axis will reduce the Doppler shift frequency.
- The effect of increasing the PRF is discussed in the next paragraph.

Some instruments may be configured into what is called the high-PRF mode. In this mode, the instrument will increase PRF beyond the limit described by Equation (12). The net effect is that a new transmit pulse will be launched prior to the arrival of the previous echo. The result is that a range gate located deep in the body will have an additional range gate corresponding to the most recently launched pulse at a shallow depth. This additional range gate is indicated on the instrument to allow the operator to avoid shallow vessels and prevent unexpected flow information from being portrayed.

Pulsed-Wave US and Wall Filters

During application, there is often some Doppler signal that originates from regions outside the blood vessel. This signal may be the result of voluntary or involuntary patient motion. These signals are represented by very low Doppler frequency shifts, and wall filters are used to remove these signals. Wall filters are referred to as high-pass filters. This means that frequencies below a threshold frequency are removed from the signal. In some cases, the operator has control over the threshold frequency and might adjust this up or down, depending on the clinical circumstance. In other instruments, the wall filter is automatically set by the anatomic configuration for which the scanner is set. In situations where the flow rates (or velocities) are very low, a wall filter set with too high of a threshold may inadvertently remove these Doppler shift frequencies. Operators must pay attention to ensure that the proper instrument configuration is used.

Color Flow Imaging

Color flow imaging takes the pulsed-wave Doppler concept a step further by using Doppler frequency shift detection over a set of range gates along a number of acoustic lines. The result is a
two-dimensional image depicting flow that is superimposed on the two-dimensional gray-scale image generated from backscattered echoes. Figure 8a shows the basic concept, and Figure 8b is an example of a color flow image. It is desirable to provide real-time color flow imaging, thus approximating the advantages of B-mode imaging. To achieve this, the instrument uses Doppler interrogation pulses that are shorter and an analysis algorithm streamlined to operate in real time. Although the train of pulses used for pulsed-wave Doppler US might be between eight and 20 pulses, for color flow imaging interrogation is performed with only two to four pulses for each line in the color flow image. The result is that color flow imaging is somewhat less sensitive to flow (at low flow rates and small vessels) compared with pulsed-wave Doppler US. Another means to achieve higher frame rates is to reduce the region of interest in color flow imaging to a smaller area than the field of view of the B-mode image.

**Autocorrelation**

Autocorrelation is the method of phase shift accumulation used by color flow imaging instruments (9). Because of the reduced number of pulses, the phase shifts will not represent a complete cycle at the Doppler frequencies. The autocorrelation algorithm compares the phase from two consecutive pulses to compute an average Doppler frequency. This estimate improves with additional successive pulses. The autocorrelation algorithm performs the estimate of the average Doppler shift for a set of pixels along the axis of the beam.

When the estimates are complete and mapped to the image, the next line is interrogated, in the same manner as a B-mode image is acquired (10).

As the region of interest area in color flow imaging is reduced, the frame rate of the color flow imaging display will increase. In addition, most systems use some preset values that alter the number of transmit pulses according to the type of Doppler shift velocities involved. Typically, these changes will not only alter the sensitivity of the color flow imaging mode but also change the frame rates.

**Color Flow Imaging Filtering and Stationary Echo Cancellation**

Some phase shifts detected in the autocorrelation circuits may occur for other reasons than the Doppler shifts produced by red blood cells. Various methods are used to reject phase shifts from sources such as vessel wall motion. Because the autocorrelation methods used for color flow imaging are not amenable to the wall filter techniques used in pulsed-wave Doppler US, different approaches are used. One is to perform a stationary echo cancellation. In this approach, the prior pulse echo data are subtracted from the current pulse echo data. Echoes from structures that are stationary or moving only slightly will be canceled out in this subtraction (Fig 9), while the phase shifted data are allowed to pass through to the autocorrelation circuitry.
Another method used to reduce “flash artifact” from nonblood moving structures is to threshold the echo data (11). Since the echogenicity of blood is approximately 20–40 dB below that of other tissues, the threshold circuit passes through only weaker echo levels to be processed in the autocorrelation circuitry. Another mode of imaging, commonly referred to as Doppler tissue imaging, does the opposite, allowing the stronger echo signals through and portraying tissue motion in color (12).

Power Doppler Imaging

Power (mode) Doppler imaging (also known as energy mode Doppler imaging) is a variation on color flow imaging in which the displayed intensity is not based on the intensity of the Doppler frequency signal (13). This has been likened to summing up all of the Doppler shift frequencies and presenting on the display a pixel intensity based on that summed value, rather than on the mean Doppler shift as in color flow imaging. This is portrayed in Figure 10, where the same flow velocities are shown running in a phantom. The velocities depicted across the diameter of the vessel are based on mean Doppler shifts. In the color flow image, directional information is preserved and the color switches when the flow direction changes with respect to the transducer. However, the power mode display shows only the intensity of the Doppler shift—the velocity information and directional information are not preserved in the Doppler signal. The advantage of power Doppler imaging is that slow flow rates and small vessels are more readily depicted in comparison with color flow imaging. This is due to the fact that all of the phase shifts are lumped together by the summing.

Flow Velocities and Frequency Shifts

Variance in blood flow velocity in both time and space (across vessel diameters and dependent on vessel conditions) can be quite extensive. This results in a range of Doppler frequencies to be displayed to the operator. In pulsed-wave duplex Doppler US, this range of frequencies is typically displayed along the y axis of a moving time strip. The intensity of the gray-scale pixels reflects the intensity of a particular Doppler frequency. In color flow imaging, color mapping is used to de-
Three factors determine the velocity of a fluid moving through a vessel: (a) the viscosity of the fluid, (b) the geometry of the vessel (eg, the cross-sectional area and the shape), and (c) the pressure. A general model shows that the flow rate through a cylindrical vessel (a long tube with a circular cross section) is determined by the pressure difference:

$$P_2 - P_1 = \frac{8l\eta Q}{\pi r^4},$$

where $l$ is the length of the vessel, $\eta$ is the viscosity, $r$ is the radius of the vessel, and $Q$ is the resulting flow rate in volume per unit time. The equation is known as Poiseuille’s law. Figure 11a shows these relationships and how the velocity changes across the cross-sectional diameter of the vessel. The velocity of the fluid at any given point in the vessel is related to the viscous resistance of the surrounding fluid. At the edge of the vessel, this resistance to flow is greater; therefore, the limit of the flow velocity at the edge of the vessel is zero. The flow velocity reaches a maximum at the center of the vessel. Under ideal conditions, the profile of flow is a parabola. These ideal conditions are: (a) a straight vessel with no change in diameter and (b) laminar (or nonturbulent) flow. Laminar flow exists when the product of velocity and mass density divided by the viscosity of the fluid multiplied by the diameter of the vessel remains below a dimensionless value of around 2,000. This number is called the Reynolds number. Beyond this number, the flow becomes turbulent, leading to a blunt flow profile (Fig 11b). Blunt flow is produced by eddies and chaotic motion within the flow. One result is that Poiseuille’s law is no longer valid, that is, resistance to flow increases and therefore a greater pressure difference is needed to produce an increase in flow rate compared with laminar conditions.

Blood is a complex fluid, exhibiting properties that are referred to as non-Newtonian. This term refers to the fact that the viscosity of blood will increase as the flow rate slows down. This increase in blood viscosity occurs because of increased cell-to-cell and cell-to-protein adhesion. Also, the vessel structure itself is very complex. Blood vessels are curved, can branch and change diameter, and are viscoelastic (eg, behave like plastics). In addition, the flow changes over the cardiac cycle (pulsatile flow). The net effect is that laminar flow is only approximated along short lengths in smaller vessels in the abdomen. Furthermore, stenotic changes in the vessel diameter will contribute to disturbance and turbulence in the flow profile, contributing to viscous losses of flow energy and resulting in subsequent loss of flow velocity downstream from the stenosis. Nevertheless, Doppler US measurement of blood flow in the body is capable of demonstrating flow differences between vessels in the body as a result of the differences in pressure and size of the vessel and the distance from the heart. Careful comparisons of different flow conditions and the resultant frequency spectral content will allow the observer to detect the aforementioned turbulence via evidence of spectral broadening. Burns (14) provides a more thorough treatment of these topics.

**Flow Parameters: Derivations and Meaning**

The ability of Doppler US to demonstrate changes in flow has prompted efforts to derive quantitative values that provide diagnostic thresholds for disease conditions in the body (eg, is flow
in the vessel 90% blocked?). Absolute quantification of the flow velocity is difficult because of the angle dependence of the Doppler equation (Eq [10]). In some cases, it is problematic to make an adequate estimate of the Doppler angle.

Calculated flow parameters are useful not because they provide absolute quantification of flow, but because these are derived from the Doppler frequency spectrum. Because the computation of these flow parameters is based on the same Doppler spectrum, errors introduced by the Doppler angle (as well as other possible instrument-dependent effects) are normalized. Two primary parameters are the pulsatility index (PI) and resistivity index (RI) (15). These are defined as follows (Fig 12):

\[
\text{PI} = \frac{v_{\text{max}} - v_{\text{min}}}{v_{\text{mean}}} = \frac{S - D}{\text{mean}}.
\]  

\[
\text{RI} = \frac{v_{\text{max}} - v_{\text{min}}}{v_{\text{max}}} = \frac{S - D}{S}.
\]

The maximum and minimum velocities are defined according to the maximum and minimum Doppler frequency shifts along a cardiac cycle. In some texts, the maximum velocity is referred to as the systolic velocity \((S)\) and the minimum velocity is referred to as the diastolic velocity \((D)\). The mean estimated velocity \((v_{\text{mean}})\) is related through Equation (10) to the mean Doppler shifted frequency shifts \(f_{\text{mean}}\). In this case, however, the mean frequency is determined over the period of time through the cardiac cycle (Fig 12). Some US scanners can be set to automatically calculate \(v_{\text{mean}}\). An algorithm determines the beginning and end of the cardiac cycle and computes \(v_{\text{mean}}\) based on the spectral content of the waveform in that period. The operator will then be left to mark the maximum and minimum using a cursor on the image.

The PI will increase as flow is impeded by a stenosis. Care must be taken when the PI is used. Proximal and distal stenoses as well as natural flow resistance from the vascular bed may affect PI measurement. The RI is easier to calculate and is used to evaluate a number of physiologic conditions (16,17). Both of these indexes are used to assess the resistance to flow in the vascular system. Another index that is a variation of the RI is the systolic-to-diastolic ratio \((S/D)\). However, this index must be used with care when the diastolic velocity approaches zero, causing the \(S/D\) ratio to approach infinity.

**Recent Innovations in Doppler US Techniques**

Doppler US is experiencing technical innovations that also contribute to improved B-mode imaging. One example is extended field of view imaging being adjusted to incorporate color flow imaging (18). This mode of imaging allows the operator to sweep the transducer across the anatomy of interest, creating a panoramic view that extends beyond the width of the field of view in the real-time image. An extended color flow image is superimposed on the same space as the extended B-mode image.

Other examples of recent innovations include

(a) wideband, wide dynamic range systems that serve to improve the sensitivity of Doppler US; and

(b) smaller US instrumentation, achieved primarily via incorporation of custom-designed integrated circuitry. These have made it possible to offer Doppler US processing on even handheld US scanners. Many imaging enhancements described in a companion article (10) contribute to improved Doppler flow detection and color flow imaging.

Another innovation is contrast agents. The original motivation for the development of contrast agents was to enhance the Doppler signals and thereby make flow detection easier. Contrast
agents have become much more than simple “echo enhancers” for Doppler instrumentation. Intravascular contrast agents have been applied in conjunction with Doppler US techniques in an attempt to assess perfusion. Typically, these approaches involve injection of the agent upstream from the region of interest in the body. One such study involved assessment of perfusion in the brain (19). The motivation for development of perfusion assessment methods is the relationship between the vascular bed and many clinical conditions, including cancer, heart disease, and diabetes.

**Conclusions**

Doppler instrumentation has evolved to accommodate the expanding clinical use of US equipment. The basic physics and equations supporting the estimation of flow velocity from Doppler shift frequency are presented. Also presented are various types of instrumentation used to detect and present Doppler shift data to a user. Each development (pulsed-wave Doppler US, color flow imaging) has been motivated by a desire to provide more clinical information about flow in the body. The algorithms used are complex, but increasingly powerful microelectronics have made these methods a reality at a reasonable cost. Users who employ these Doppler-based techniques must be aware of the complicated aspects of flow in the body, especially with regard to detection of disease in the human vasculature. Continued development of US equipment is aimed at a greater understanding of hemodynamics and the relationship between blood flow and various disease processes.

**References**