PRINCIPLES OF DOPPLER ECHOCARDIOGRAPHY AND THE DOPPLER EXAMINATION #1

Joseph A. Kisslo, MD
David B. Adams, RDCS

INTRODUCTION

Doppler echocardiography is a method for detecting the direction and velocity of moving blood within the heart. As will be seen in this program, the technique may be used for detection of cardiac valvular insufficiency and stenosis as well as a large number of other abnormal flows. The current interest in Doppler echocardiography has reached a remarkable level in just the past few years. Doppler methods extend the use of cardiac ultrasound into the evaluation of normal and abnormal flow states and provide quantitative data that are essential in the clinical decision making process concerning patients with heart disease.
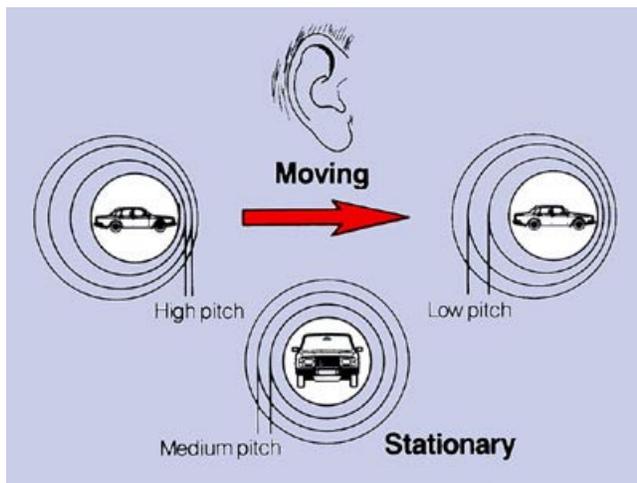

Figure 1.1: An example of the Doppler effect in every day life: the sound emitted from a stationary and moving automobile engine.

Understanding Doppler echocardiography begins with an understanding of the Doppler principle. All readers are familiar with the Doppler effect in every day life. For example, an observer stationed on a highway overpass readily notices that the pitch of the sound made from the engine of a passing automobile changes from high to low as the car approaches and then travels into the distance. The engine is emitting the same sound as it passes beneath, but the observer notices a change in pitch dependent upon the speed of the auto and its direction. **Figure 1.1** demonstrates the changes in the frequency from an approaching and departing sound source (the moving automobile) relative to a stationary sound source.

The first description of the physical principles used in Doppler echocardiography is attributed to Johann Christian Doppler, an Austrian mathematician and scientist who lived in the first half of the 19th century. Doppler's initial descriptions referred to changes in the wavelength of light as applied to astronomical events. In 1842, he presented a paper entitled "On the Coloured Light of Double Stars and Some Other Heavenly Bodies" where he postulated that certain properties of light emitted from stars depend upon the relative motion of the observer and the wave source. He suggested that the colored appearance of some stars was caused by their motion relative to the earth, the blue ones moving toward earth and the red ones moving away.

He drew an analogy of a ship moving to meet, or retreat from, incoming waves. The ship moving out to sea would meet the waves with more frequency than a ship moving toward the shoreline. Interestingly, Doppler never extrapolated his postulations to sound waves.

The Doppler principle is now used in many complex technologies. It is the fundamental principle upon which complex radar weather systems detect the severity of approaching storms and tracks its speed. It is also used by police to determine the speed of fast moving automobiles.

THE DOPPLER PRINCIPLE AND THE STUDY OF CARDIAC FLOWS
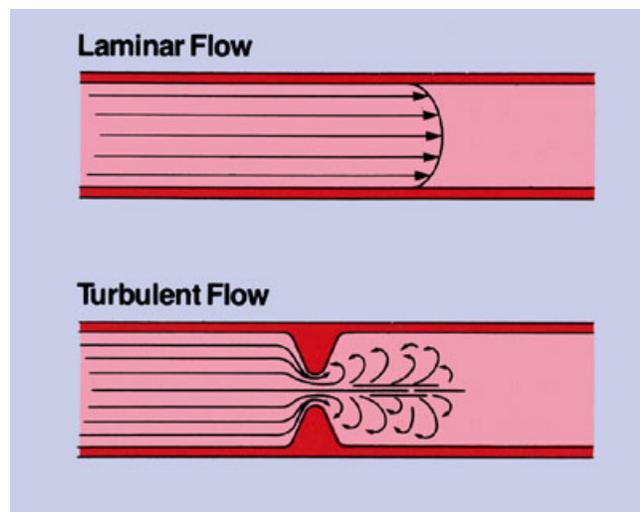
Blood Flow Patterns

Figure 1.2: Diagrammatic representation of normal laminar flow in comparison with turbulent flow that results in whirls and eddies of many different velocities.

Blood flow through the heart and great vessels has certain characteristics that can be measured using Doppler instruments designed for medical use. For the purpose of understanding flow patterns in the heart, it is important to recognize the difference between laminar flow and turbulent (or disturbed) flow. Laminar flow is flow that occurs along smooth parallel lines in a vessel so that all the red cells in an area are moving at approximately the same speed and in the same direction **(Fig. 1.2)**. Due to friction, flow is always slightly slower near the walls of a vessel. With the pulsations of the heart, the red cells generally accelerate and decelerate at approximately the same speed. Flow in most of the cardiovascular system, including the heart and great vessels, is normally laminar and rarely exceeds the maximum velocity of 1.5 m/sec.

In contrast, turbulent or disturbed flow is present when there is some obstruction that results in a disruption of the normal laminar pattern. This causes the orderly movement of red blood cells to become disorganized and produces various whirls and eddies of differing velocities and directions. Obstruction to flow usually also results in some increase in velocity. Thus, turbulent flow is characterized by disordered directions of flow in combination with many different red cell

Figure 1.3: Examples of normal laminar flow through the aortic valve (top) and disturbed or turbulent flow resulting from aortic stenosis (bottom).

velocities. If the obstruction is significant, some of the red blood cells may be moving at higher velocities than normal and may reach speeds of 7 m/sec. Turbulent flow is usually an abnormal finding and is considered indicative of some underlying cardiovascular pathology.
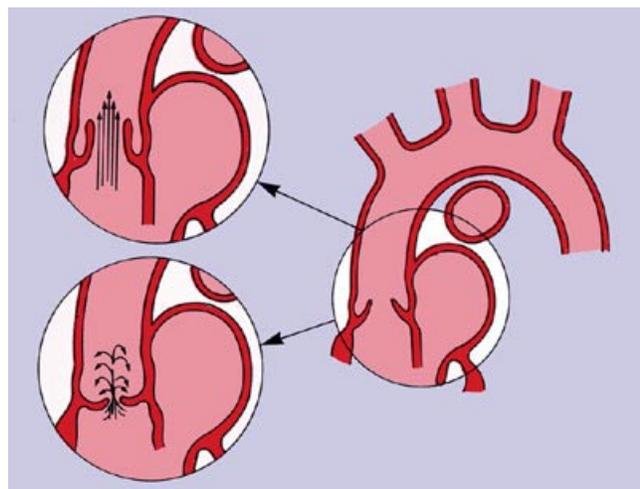
Abnormal flows are therefore generally characterized by turbulence and any increase in velocity. As an example, consider blood flow in the ascending aorta during systole. If the aorta and aortic valve are normal, then this flow is laminar. However, the presence of a valvular stenosis will induce a turbulent flow pattern. **Figure 1.3** shows that a narrowed aortic valve orifice interrupts the parallel lines of normal laminar flow and produces turbulent flow. The resulting jet of blood creates a short segment within the proximal aorta with complex flow and velocity characteristics.

The Frequency of Sound Waves

Conventional two-dimensional echocardiographic systems emit high frequency bursts of sound (ultrasound) into the tissues. In standard echocardiographic imaging a given pulse of ultrasound is transmitted into the body and then reflected back from the various tissues. Since the speed of sound in tissue is known (approximately 1540 m/sec), a standard ultrasound imaging system can wait for a given time for the transmitted pulse to travel to a target (time X) and then back (time 2X) and the given target will be received and recorded. In complex two-dimensional imaging systems this alternating process is repeated in a variety of directions thousands of times each second. The best ultrasound images are made when the target is perpendicular (or specular) to the sound waves.
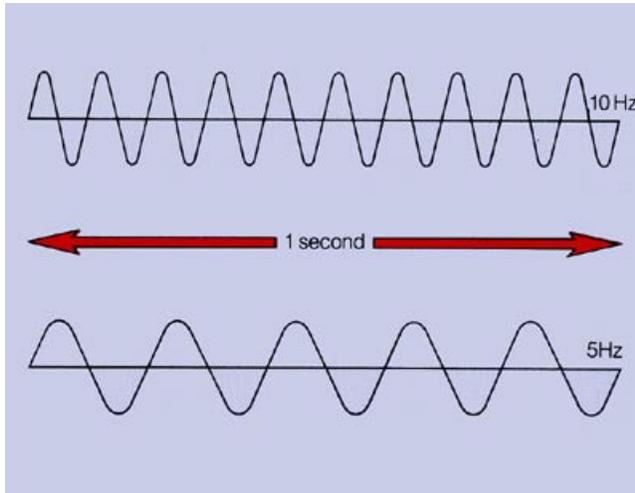

Figure 1.4: Sound is emitted in waves of measurable frequencies.

Frequency is a fundamental characteristic of any wave phenomenon, including sound, and refers to the number of waves that pass a given point in one second **(Fig. 1.4)**. It is usually described in units of cycles per second or Hertz (Hz). Thus, the top of the illustration in **Figure 1.4** shows an example of a waveform of 10 Hz while the one below is 5 Hz. Ultrasound is emitted in waveforms of a known frequency.

Doppler echocardiography, on the other hand, depends entirely on measurement of the *relative change* in the returned ultrasound frequency when compared to the transmitted frequency. Depending on the relative changes of the returning frequencies, Doppler echocardiographic systems measure these characteristics of disturbed flow: direction, velocity and turbulence. This enables examiners to differentiate between normal and abnormal flow patterns and, in some cases, to quantitate those characteristics that are helpful in determining the severity of abnormal flow states.
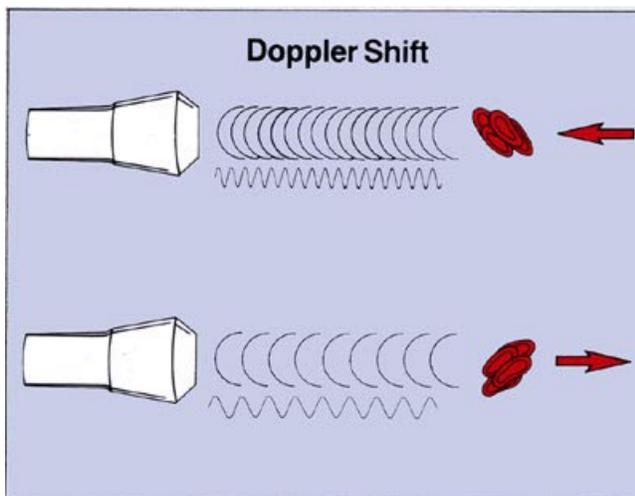

Figure 1.5: For any given transmitted ultrasound frequency, the returned frequency will be higher after encountering red blood cells moving toward the transducer and lower after encountering red cells moving away from the transducer.

Most readers understand frequencies in relationship to the pitch of audible sound. The relationship between pitch and frequency is simple: the pitch of any given sound is proportional to its frequency. As sound wave frequency increases, pitch gets higher; and as frequency decreases, pitch declines.

Doppler systems are totally dependent on the changes in the frequency of the transmitted ultrasound that result from the encounter of the wavefront with moving red blood cells. **Figure 1.5** shows a transducer on the left that is emitting a given frequency of ultrasound toward the right and into the tissues. The transmitted sound waves encounter a group of red cells moving toward the transducer and are reflected back at a frequency higher than that at
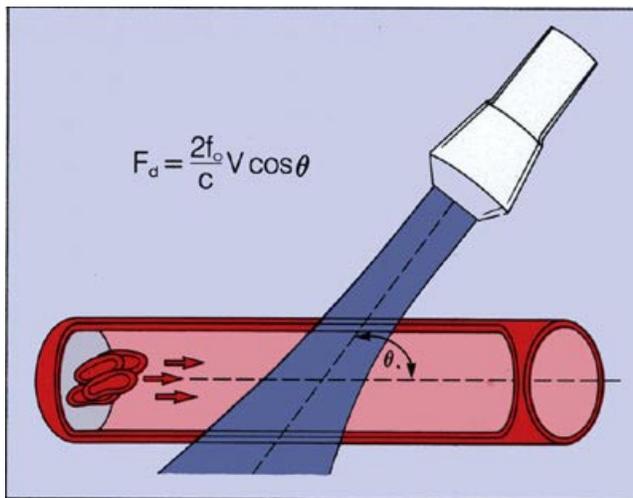
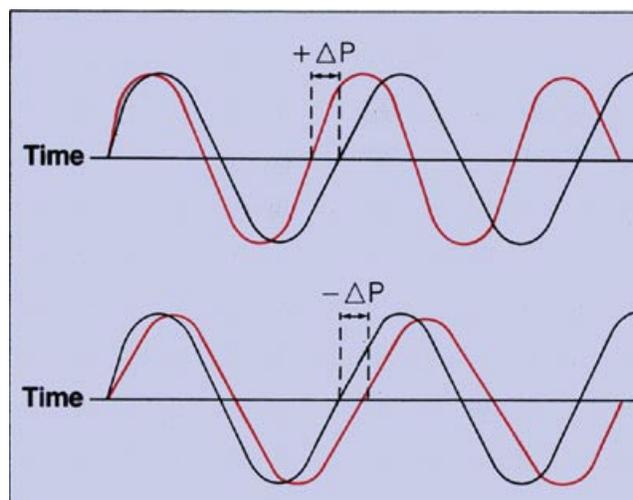*Figure 1.6 : The Doppler equation solved for frequency shift.*

Figure 1:7: The phase shifts of returning frequencies are compared with transmitted frequency within the Doppler system. Positive shifts (top) result from blood moving away from, and negative shifts (bottom) result from blood moving toward, the transducer..
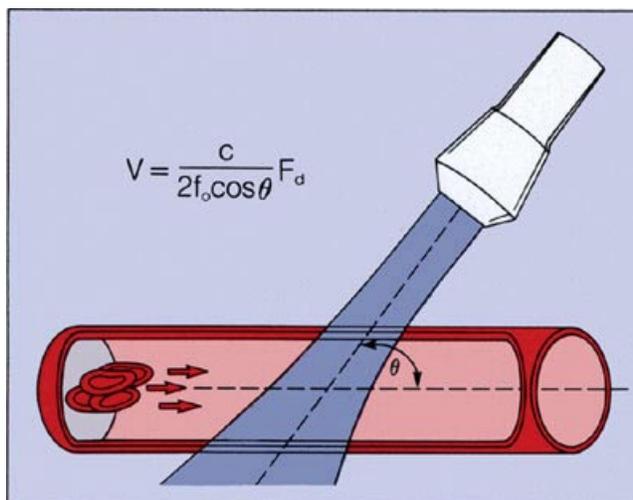

Figure 1.8: The Doppler equation rearranged for determination of velocity. It is velocity that is displayed by the Doppler instrument.

which they were sent producing a positive Doppler shift. The opposite effect occurs when a given frequency sent into the tissues encounters red cells moving away. The result is the return of a frequency lower than that transmitted, and the Doppler shift is negative.

The Doppler Equation

This Doppler effect in tissues maybe expressed as an equation as shown in **Figure 1.6**. Simply stated, the Doppler shift (Fd) of ultrasound will depend on both the transmitted frequency (fo) and the velocity (V) of the moving blood. This returned frequency is also called the "frequency shift" or "Doppler shift" and is highly dependent upon the angle (θ) between the beam of ultrasound transmitted from the transducer and the moving red blood cells. The velocity of sound in blood is constant (c) and is an important part of the Doppler equation.

The Direction and Velocity of Flow

The fact that makes frequency of the Doppler effect more than just an interesting curiosity is that it actually provides a method that is used to measure the direction and speed of moving red blood cells. Clinically we are most interested in measuring velocity since, as mentioned above, it is altered in disease states.

A Doppler system then compares the transmitted waveform with the received waveform for a change in frequency as shown in **Figure 1.7**. These are called "phase shifts" and they are automatically determined within the Doppler instrument. If there is a higher returning frequency (+AP) then the flow is called a "positive Doppler shift" and represented as moving toward the transducer. If there is a lower returning frequency (-AP) then the flow is called a "negative Doppler shift" and represented as moving away from the transducer. All components of the Doppler equation, except velocity, are readily measured by the Doppler instrument. The Doppler equation may be rearranged to solve for velocity of blood movement as shown in **Figure 1.8**. The angle θ (the angle the Doppler beam is incident to flow) may be measured or may be assumed to be parallel

depending upon orientation of the beam by the system operator.

The Doppler device can be regarded as a complex speedometer designed to detect red cell motion (i.e., blood flow) and measure its velocity. What is important to recognize is that:

FREQUENCY SHIFT ➝ DOPPLER EQUATION ➝ VELOCITY DATA

The Doppler Display

All Doppler systems have audio outputs and listening to this is very helpful during a Doppler examination. The changing velocities (frequencies) are converted into audible sounds and, after some processing, are emitted from speakers placed within the machine.

High pitched sounds result from large Doppler shifts and indicate the presence of high velocities, while low pitched sounds result from lesser Doppler shifts. Flow direction information (relative to the transducer) is provided by a stereophonic audio output in which flow toward the transducer comes out of one speaker and flow away from the transducer.

The audio output also allows the operator to easily differentiate laminar from turbulent flow. Laminar flow produces a smooth, pleasant tone because of the uniform velocities. Turbulent flow, because of the presence of many different velocities, results in a commonly high-pitched and whistling or harsh and raspy sound.

The audio output remains an indispensable guide to the machine operator for achieving proper orientation of the ultrasound beam, even when Doppler echocardiography is being used in conjunction with an ultrasound imaging technique. The trained ear can readily appreciate minor changes in spectral composition more readily than the eye, given the same information displayed graphically. The major limitation of audio Doppler outputs is the requirements for subjective interpretation and the lack of a permanent objective record. The audio output from a Doppler machine is not the same as that received by a stethoscope or a phonocardiogram. The sounds detected with a stethoscope are transmitted vibrations or pressure waves from the heart and great vessels that are believed to be the result of rapid accelerations and decelerations of blood. The Doppler audio output, in contrast, is an audible display of the Doppler frequency shift spectrum produced by red cells moving in the path of the ultrasound beam. It is a sound produced by the Doppler machine that does *not* occur in nature and, therefore, it does not originate in the heart. All newer generations of Doppler echocardiography equipment contain sophisticated sound frequency or velocity sp-ectrum analyzers for hard copy recording. Most commercially available Doppler systems display a spectrum of the various velocities present at anytime and are, therefore, called "spectral velocity recordings.

Flow velocity toward the transducer is displayed as a positive, or upward, shift in velocities while flow velocity away from the transducer is displayed as a negative, or downward shift in velocities (**Fig. 1.9**). Time is on the horizontal axis.
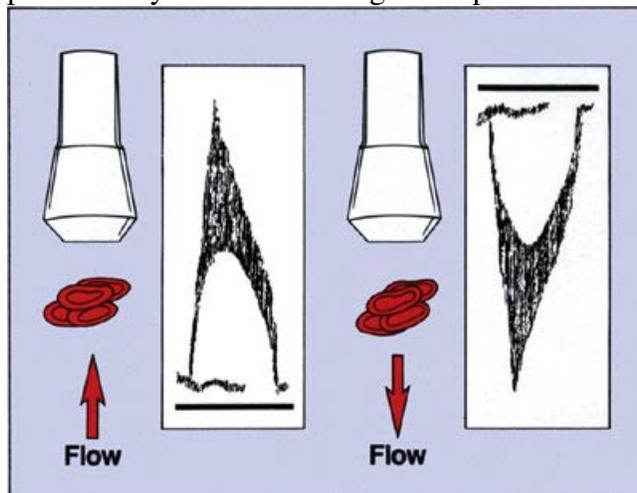

Figure 1.9: Schematic representation of the velocity output of flow. Flow toward the transducer is displayed above the baseline and flow away from the transducer is displayed below the baseline.
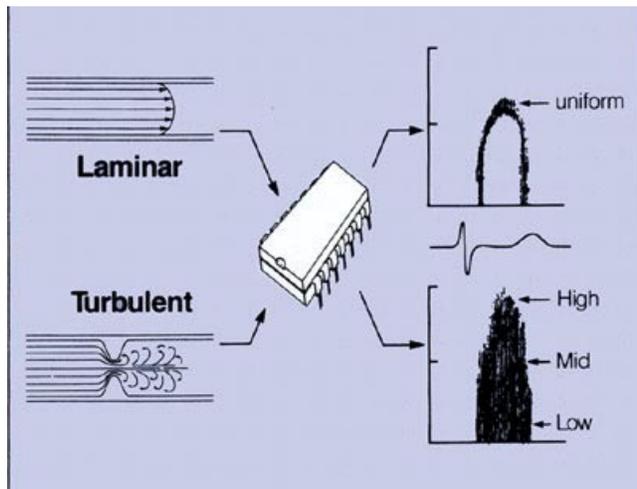
Figure 1.10: The various velocities detected by the Doppler instrument are processed by Fast Fourier Transform (FFT) and the resulting spectrum of velocities present is displayed. Laminar flows are uniform. Turbulent flows show spectral broadening.
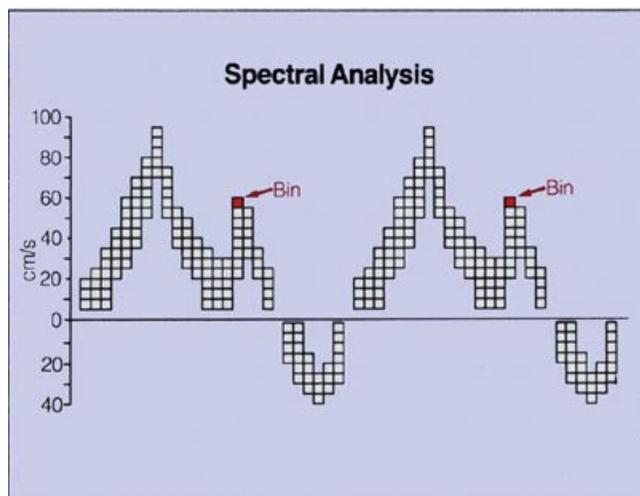

Figure 1.11 : The spectral analysis is created by placing the velocity data into bins that are displayed over time.
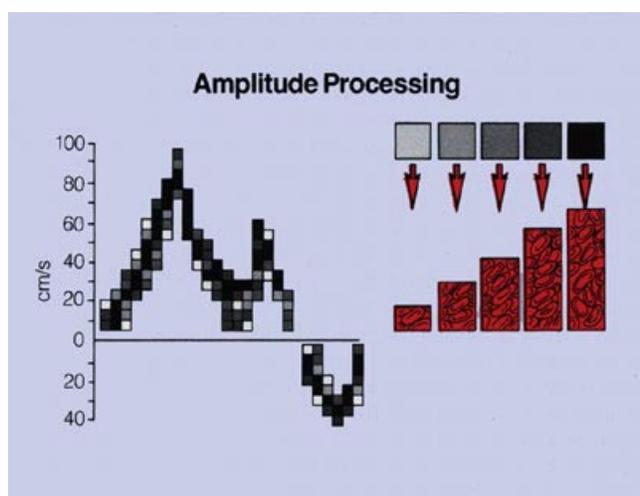

Figure 1.12: The brightness of the signal at any given bin relates to the relative number of red cells detected at that velocity. The term "amplitude" is applied to relative brightness.

The internal working of such systems are complex but the results are rather simple. When flow is laminar and all the red cells are accelerating and decelerating at approximately the same velocities, a neat envelope of these similar velocities is recorded over time **(Fig. 1.10)**. When flow is turbulent, however, there are many different velocities detected at any one time (a wide spectrum of velocities). Such turbulence, produced by an obstruction to flow, results in the spectral broadening (display of velocities that are low, mid and high) and an increase in peak velocity as seen in disease states.

This display of the spectrum of the various velocities encountered by the Doppler beam is accomplished by very sophisticated microcomputers that are able to decode the returning complex Doppler signal and process it into its various velocity components. There are two basic methods for accomplishing this. The most popular is Fast Fourier Transform (FFT) and the other is called Chirp-Z Transform. These are simply ways for deciphering, analyzing and presenting vast amounts of returning data.

A better understanding of the complex creation of a spectral velocity recording helps one in performing and interpreting Doppler studies. The spectral recording **(Fig. 1.11)** is really made up of a series of "bins" (vertical axis) that are recorded over time (horizontal axis). At any given point in time there is a differential speed of movement of red cells with more red blood cells moving at the velocity of the most intense bin than are moving at the other velocities which are represented by the less intense bins **(Fig. 1.12)**. The intensity of any bin refers to the "amplitude" or brightness. Thus, the velocity spectral analysis is really a complex plot of the various velocities over time.

Since this full spectral display is so highly processed there are a variety of other outputs that can be displayed and they are electronically derived from the spectral data **(Fig. 1.13)**. These include mean velocity and maximum velocity. A line drawn as an
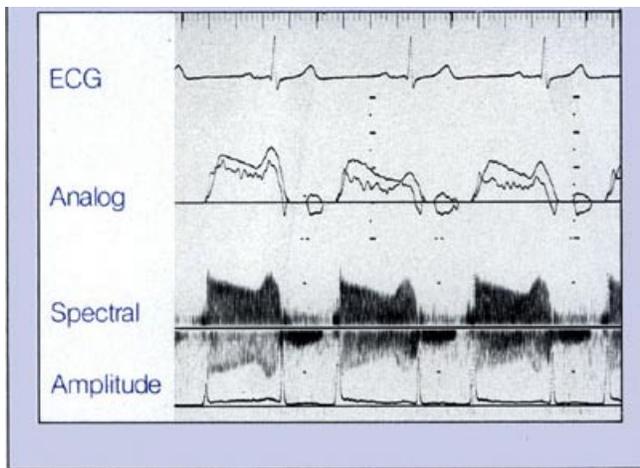
Figure 1.13: An example of the various Doppler displays from a patient with mitral stenosis with the transducer held at the apex. Flow in diastole is toward the transducer. The ECG, analog outputs (maximum and mean velocities), spectral display, and amplitude signals are shown.
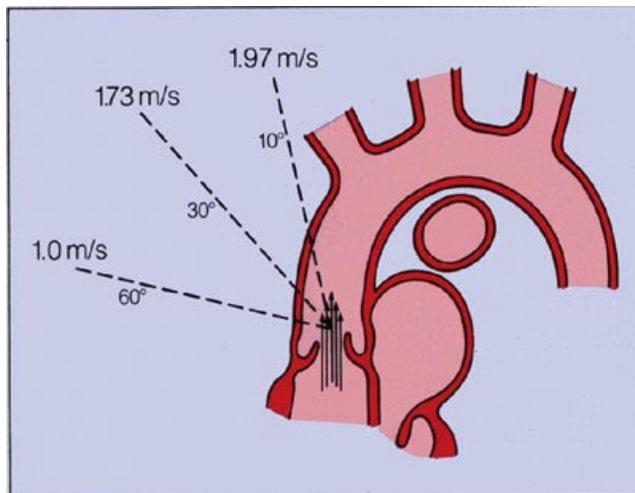

Figure 1.14: Schematic diagram showing the importance of being parallel to flow when detecting flow through the aortic valve. A jet of known velocity (2.0 m/s) emerges from the aortic valve in systole. Moving 60 degrees from parallel only allows a peak velocity of 1.0 m/s to be recorded. The most accurate
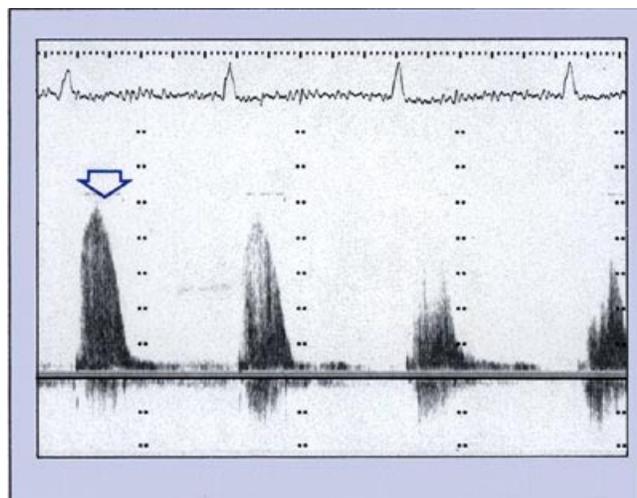

Figure 1.15: The effect of varying angulation in relation to a systolic jet in a patient with aortic stenosis taken from the suprasternal notch with flow toward the transducer. Note that the peak jet is nearly 5 m/s (open arrow). In subsequent beats the flow profile is lost.

envelope around the spectrum at the peak Doppler shift at any point during the cardiac cycle is the peak velocity profile. Mean Doppler shift can be estimated from a line drawn through the darkest part of the spectrum. The brightness, or "amplitude," may also be displayed. For standard clinical purposes the full spectrum is generally used.

The Effect of Angle

The Doppler equation also tells us that the angle the Doppler beam is relative to the lines of flow being evaluated is very important. This angle theta, (written as θ in the Doppler equation), is of crucial importance in the calculation of velocities from Doppler shift data in **Figure 1.14** where the effect of varying angle on the measurement of peak velocity of an aortic stenotic jet is shown.

When the ultrasound beam is directed parallel to blood flow, angle θ (cosine $0^o = 1$) and measured velocity on the recording will be true velocity. In contrast, with the ultrasound beam directed perpendicular to flow, angle $\theta = 90$ degrees (cosine $90^o = 0$) and measured velocity will be zero. Therefore, the smaller the angle, the closer angle cosine θ is to 1.0 and the more reliable is the recorded Doppler velocity. A wider angle will result in a greater reduction in measured velocity compared to true velocity.

Thus, the more parallel to flow the Doppler ultrasound beam is directed the more faithfully the measured velocity will reflect true velocity. For practical purposes, angles of greater than 25° between the ultrasound beam and the blood flow being studied will generally yield clinically unacceptable qualitative estimates of velocity.

A Doppler operator seeking the best quantitative estimates of flow must, therefore, always attempt to orient the beam parallel to flow. This concept is of fundamental importance in the clinical examination. The actual effect of changing angle on a systolic aortic stenotic jet toward a transducer in the suprasternal notch is shown in **Figure 1.15**. The first beat (open arrow) shows the only fully formed profile.
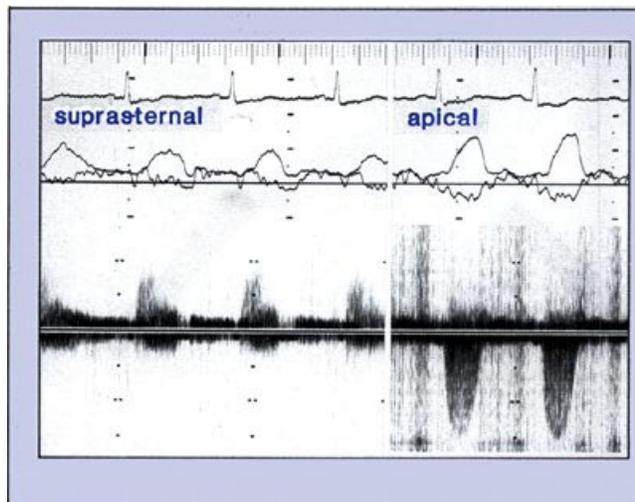
Figure 1.16: Suprasternal (left, with flow toward the transducer) and apical (right, with flow away from the transducer) jet of aortic stenosis. The best profile was taken from the apical position.

Such abnormal jets are often eccentric and their directions cannot always be predicted. A Doppler examiner must therefore interrogate the jet from a variety of angles. Note that the full jet is not seen from the suprasternal area in this patient but is detected from the apical approach. The great importance of this concept in the clinical examination for aortic stenosis is demonstrated in **Figure 1.16**. The need to be parallel to flow leads the Doppler examiner to depend on some windows for examination that may sacrifice the quality of the two-dimensional image. For example, the direction of the ultrasound beam through either the mitral or tricuspid orifices from the apical position offers an excellent Doppler window but one which may allow significant echocardiographic "drop-out" since the imaging beams are parallel to the endocardium.

Simultaneous Imaging and Doppler

Since both two-dimensional and Doppler echocardiography use ultrasound, it is logical to assume that both Doppler and imaging functions can be combined into one ultrasound instrument. It is important to realize that creation of the image takes time as does creation of the Doppler information. The imaging device is already working as quickly as it can to transmit, receive and then display the image data.

Therefore, time is the critical factor for any shared arrangement. These problems are solved by switching off the imaging mode (sometimes with the image held in memory) while the Doppler modes are in operation. This results in either an image or Doppler choice. In some systems a complex time sharing arrangement allows some Doppler to be carried on while the image of the heart is still moving. When this happens, there is always some sacrifice of the quality of the Doppler information and frequently also in the image information. Conventional Doppler and imaging simply do not work to their full capabilities in these interlaced modes.

PULSED AND CONTINUOUS WAVE DOPPLER

There are two main types of Doppler echocardiographic systems in common use today, continuous wave and pulsed wave. They differ in transducer design and operating features, signal processing procedures and in the types of information provided. Each has important advantages and disadvantages and, in our opinion, the current practice of Doppler echocardiography requires some capability for both forms.

Continuous Wave Doppler

Continuous wave (CW) Doppler is the older and electronically more simple of the two kinds. As the name implies, CW Doppler involves continuous generation of ultrasound waves coupled with continuous ultrasound reception. A two crystal transducer accomplishes this dual function with one
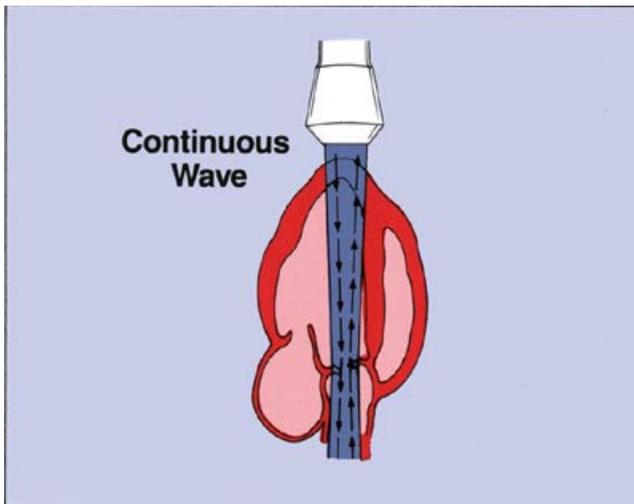
Figure 1.17: In CW Doppler, the transducer is constantly emitting and receiving ultrasound data.

crystal devoted to each function **(Fig. 1.17)**.

The main advantage of CW Doppler is its ability to measure high blood velocities accurately. Indeed, CW Doppler can accurately record the highest velocities in any valvular and congenital heart disease. Since velocities exceeding 1.5 m/sec are frequently seen in such disorders, accurate high velocity measurement is of particular importance for allowing the recognition of the full abnormal flow profile. It is also of considerable importance for the quantitative evaluation of abnormal flows, as will be seen later.

The main disadvantage of CW Doppler is its lack of selectivity or depth discrimination. Since CW Doppler is constantly transmitting and receiving from two different transducer heads (crystals) there is no provision for imaging or range gating to allow selective placing of a given Doppler sample volume in space. As a consequence, the output from a CW examination contains Doppler shift data from *every* red cell reflecting ultrasound back to the transducer along the course of the ultrasound beam.

Thus, true CW Doppler is functionally a stand-alone technique whether or not the capability is housed within a two-dimensional imaging transducer. The absence of anatomic information during CW examination may lead to interpretive difficulties, particularly if more than one heart chamber or blood vessel lies in the path of the ultrasound beam.

It is possible , however, to program a phased array system to perform both two-dimensional and CW Doppler functions almost simultaneously. The quasi-simultaneous CW-imaging uses a time sharing arrangement in which the transducer rapidly switches back and forth from one type of examination to the other. Because this switching is done at very high speeds, the operator gets the impression that both studies are being done continuously and in real-time. During the imaging period, no Doppler data is being collected, so an estimate is generated, usually from the preceding data. During the Doppler collection period, previously stored image data is displayed. This arrangement usually degrades the quality of both the image and Doppler data.

Pulsed Wave Doppler

Pulsed wave (PW) Doppler systems use a transducer that alternates transmission and reception of ultrasound in a way similar to the M-mode transducer **(Fig. 1.18)**. One main advantage of pulsed Doppler is its ability to provide Doppler shift data selectively from a small segment along the ultrasound beam, referred to as the "sample volume". The location of the sample volume is operator controlled. An ultrasound pulse is transmitted into the tissues travels for a given time (time X) until it is reflected back by a moving red
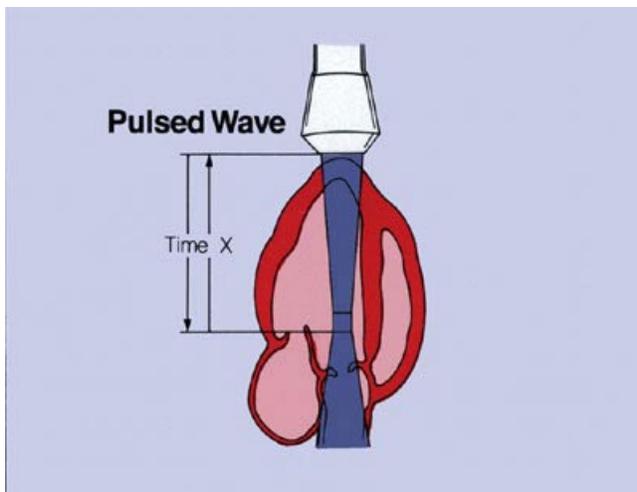

Figure 1.18: In PW Doppler, the transducer alternately transmits and received the ultrasound data to a sample volume. This is also known as range-gated Doppler.

cell. It then returns to the transducer over the same time interval but at a shifted frequency. The total transit time to and from the area is 2X. Since the speed of ultrasound in the tissues is constant, there is a simple relationship between roundtrip travel time and the location of the sample volume relative to the transducer face (i.e., distance to sample volume equals ultrasound speed divided by round trip travel time). This process is alternately repeated through many transmit-receive cycles each second.
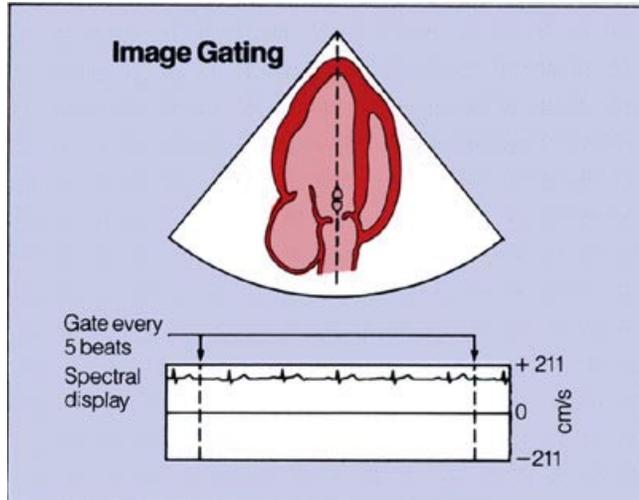

Figure 1.19: When the PW Doppler operates, it causes the two-dimensional image to be held in a frozen frame. The image is periodically updated and will usually appear as a blank on the spectral display (dashed lines).

This range gating is therefore dependent on a timing mechanism that only samples the returning Doppler shift data from a given region. It is calibrated so that as the operator chooses a particular location for the sample volume, the range gate circuit will permit only Doppler shift data from inside that area to be displayed as output. All other returning ultrasound information is essentially "ignored".

Another main advantage of PW Doppler is the fact that some imaging may be carried on alternately with the Doppler and thus the sample volume may be shown on the actual two-dimensional display for guidance. PW Doppler capability is possible in combination with imaging from a mechanical or phased array imaging system. It is also generally steerable through the two-dimensional field of view, although not all systems have this capability.

In reality, since the speed of sound in body tissues is constant, it is not possible to simultaneously carry on both imaging and Doppler functions at full capability in the same ultrasound system. In mechanical systems, the cursor and sample volume are positioned during real-time imaging, and the two-dimensional image is then frozen when the Doppler is activated. With most phased array imaging systems the Doppler is variably programmed to allow periodic update of a single frame two-dimensional image every few beats (**Fig. 1.19**). In other phased arrays, two-dimensional frame rate and line density are significantly decreased to allow enough time for the PW Doppler to sample effectively. This latter arrangement gives the appearance of near simultaneity.
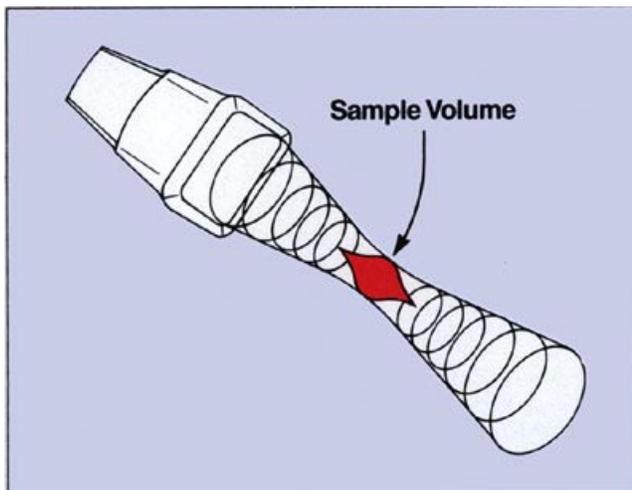

Figure 1.20: The sample volume of PW Doppler is actually a three-dimensional volume that changes in size at its location relative to the transducer is changed. When placed in the far field, it becomes very large.

The sample volume is really a three-dimensional, teardrop shaped portion of the ultrasound beam (**Fig. 1.20**). Its volume varies with different Doppler machines, different size and frequency transducers and different depths into the tissue. Its width is determined by the width of the ultrasound beam at the selected depth. Its length is determined by the length of each transmitted ultrasound pulse.
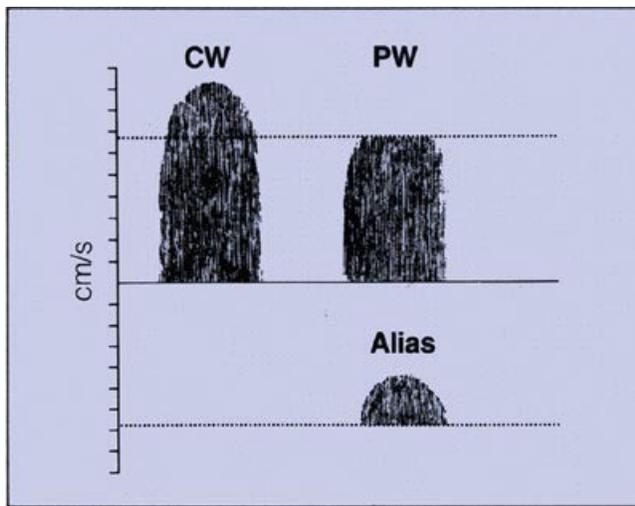
Figure 1.21: Schematic rendering of the full spectral display of a high velocity profile fully recorded by CW Doppler. The PW display is aliased, or cut off, and the top is placed at the bottom.
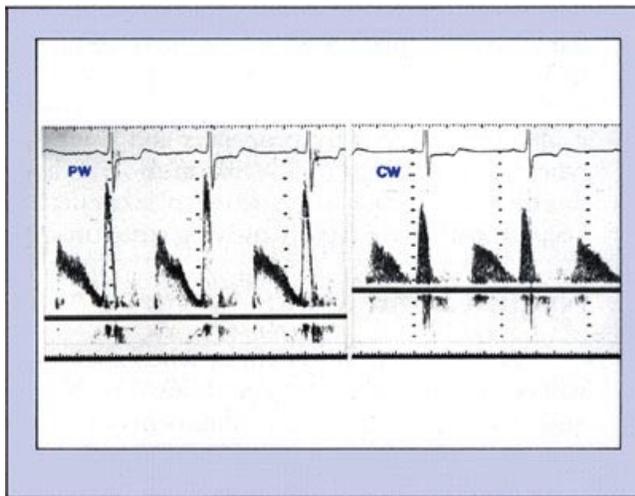

Figure 1. 22: Spectral displays of diastolic flow through the mitral orifice. The transducer is located at the apex and diastolic flow is toward the transducer (positive) Note the laminar appearance of the PW display. The CW does not usually display the same laminar flow pattern as it receives flow information from all portions of the ultrasound beam.

Comparison of Pulsed and Continuous Wave Techniques

	Range Resolution	Limitation on Maximum Velocity
Pulsed Wave	Yes	Yes
Continuous wave	No	No

Figure 1. 23: Table summarizing the advantages and disadvantages of pulsed and continuous wave Doppler echocardiography..

Therefore, the farther into the heart the sample volume is moved, the larger it effectively becomes. This happens because the ultrasound beam diverges as it gets farther away from the transducer.

The main disadvantage of PW Doppler is its inability to accurately measure high blood flow velocities, such as may be encountered in certain types of valvular and congenital heart disease. This limitation is technically known as "aliasing" and results in an inability of pulsed Doppler to faithfully record velocities above 1.5 to 2 m/sec when the sample volume is located at standard ranges in the heart **(Fig. 1.21)**. Aliasing is represented on the spectral trace as a cut-off of a given velocity with placement of the cut section in the opposite channel or reverse flow direction. Because aliasing is so common in disease states, it will be considered in more detail in the next section.

The spectral outputs from PW and CW appear differently **(Fig. 1.22)**. When there is no turbulence, PW will generally show a laminar (narrow band) spectral output. CW, on the other hand, rarely displays such a neat narrow band of flow velocities even with laminar flow because all the various velocities encountered by the ultrasound beams are detected by CW.

It can usually be said that when an operator wants to know where a specific area of abnormal flow is located that pulsed wave Doppler is indicated. When accurate measurement of elevated flow velocity is required, then CW Doppler should be used. The various differences between pulsed and continuous wave Doppler are summarized in **Figure 1.23**.

Aliasing

The aliasing phenomenon occurs when the abnormal velocity exceeds the rate at which the pulsed wave system can record it properly. PW Doppler spectral tracing in **Figure 1.24** from an individual with aortic insufficiency with the transducer positioned at the apex. In this situation, abnormal diastolic flow is detected toward the transducer and recorded in
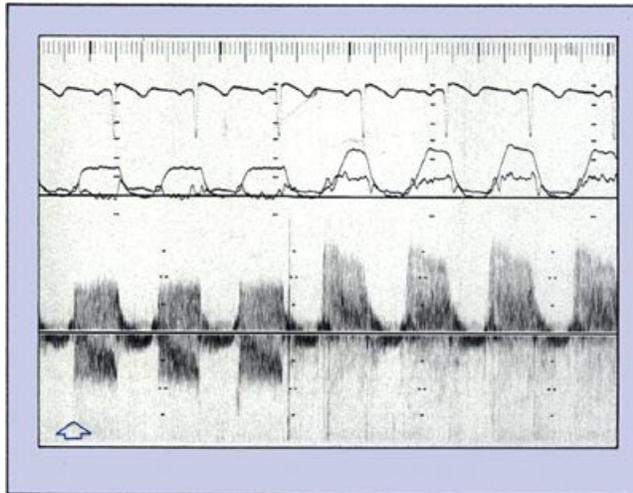
Figure 1. 24: Aliased spectral display of aortic insufficiency (left arrow) in PW mode detected from the ventricular apex. Abnormal flow is toward the transducer. After 3 beats, the system is switched to CW and the full profile is seen.
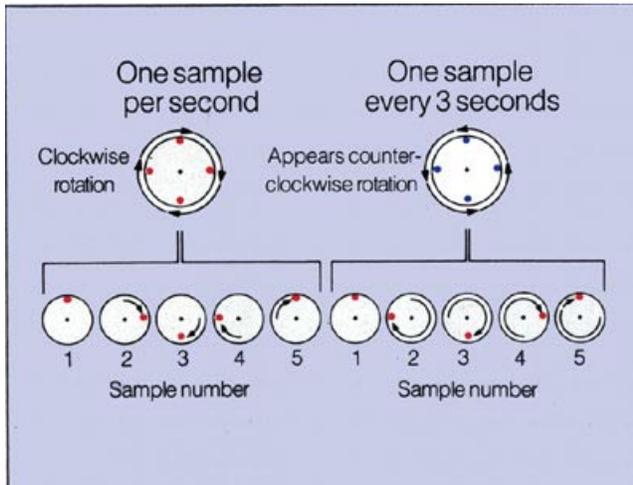

Figure 1. 25: Schematic depiction of the origin of the aliasing phenomenon using a turning wheel. When the Nyquist limit is exceeded, aliasing occurs. For details, see text.

Figure 1. 26: The Nyquist limit is defined by the number of pulses/second divided by two.

a positive, or upwards direction. The system first detects a pulsed (and aliased) spectral profile. After the fourth beat, the system is switched into CW and the full profile is recognized. The aliased portion in the first three beats is cut off the top of the velocity spectrum and replaced at in the reverse channel, or below the baseline (open arrow).

The phenomenon of aliasing is best explained using a simple example (**Fig. 1.25**). A mark is placed on a turning wheel and the wheel rotates in a clockwise fashion at a speed of one turn every four seconds. If the sample rate (or pulse repetition frequency) is one sample per second the mark is recorded at each progressive 90 degree position. The final recording would then show the proper clockwise direction of motion of the wheel (**Fig. 1.25** *left column*).

If the sample rate (or pulse repetition frequency) is slowed to only one sample every three seconds a strange phenomenon occurs (**Fig. 1.25** *right column*). Note that the mark is moving 180 degrees between sampling times and that while actually turning clockwise the recording makes the wheel appear to be moving in the opposite, or counter-clockwise, direction. This is also the reason why propellers and wagon wheels appear to go backwards in movies as the film frame rate is too slow to accurately keep up with these rapidly moving structures.

Nyquist Limit

The Nyquist limit defines when aliasing will occur using PW Doppler (**Fig. 1.26**). The Nyquist limit specifies that measurements of frequency shifts (and, thus, velocity) will be appropriately displayed only if the pulse repetition frequency (PRF) is at least twice the maximum velocity (or Doppler shift frequency) encountered in the sample volume.

It is obviously desirable to use as high a PRF as possible for recording abnormally elevated velocity jets. The problem is that the maximum PRF is limited by the distance the sample volume is placed into the heart.
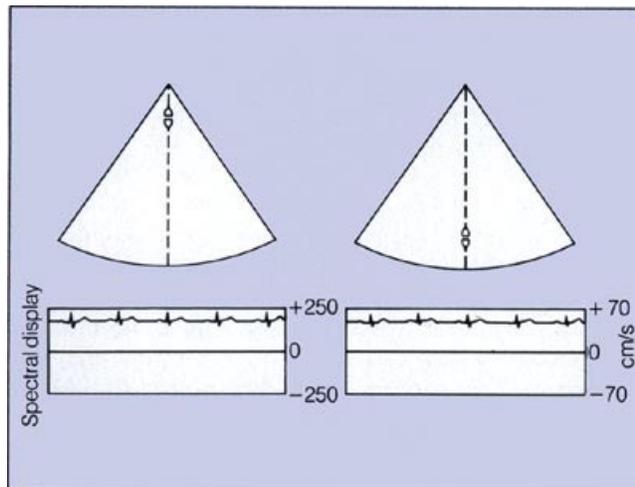
Figure 1. 27: The left panel shows the sample volume placed in the near field. Higher velocities may be recorded with a location near the transducer than when the sample volume is positioned at a farther range (right panel).

The closer the sample volume is located to the transducer the higher the maximum PRF that can be used. Conversely, the farther the sample volume is placed into the heart the lower the maximum PRF becomes. This occurs because the distance (and therefore pulse travel time) to and from the sample volume is much shorter in the near field and, therefore, pulse roundtrip transit time is much less when compared with greater distances. An example of the spectral velocity display with the sample volume located in the near field is shown in **Figure 1.27**. The maximum velocity that can be recorded without aliasing is 2.50 m/sec in either direction **(Fig. 1.27 *left panel arrow*)**. When the sample volume is positioned farther into the field the maximum possible velocity in either direction is reduced to .70m/sec **(Fig. 1.27, *right panel arrow*)**.

Note that with the Doppler system used in the previous example, the scale of the spectral display automatically changes. In some systems, the scale is fixed and the size of the spectral tracing will alter. **Figure 1.28** demonstrates mild aliasing when a high velocity jet is in the near field (A). Progressively more severe aliasing with distortion of the full profile occurs if this same jet is encountered at progressively increasing distances from the transducer face (*B, C and D*). At point D the aliased profile is so distorted as to be unrecognizable.
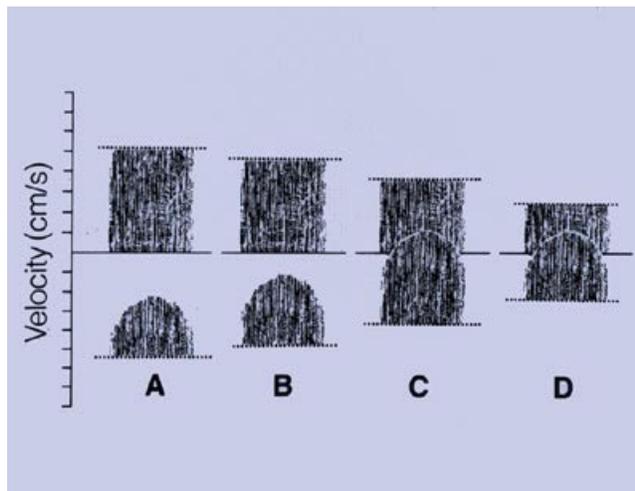

Figure 1. 28: Schematic drawing of the effect of a reduction in the Nyquist limit in PW echo when a high velocity jet is encountered in the near field (A) or successively deeper into the tissues (B, C, and D). Note the progressive and marked aliasing of the spectral signal the farther away the jet is encountered from the transducer.

In a practical sense, the Nyquist limit is a descriptive term which specifies the maximum velocity that can be recorded without aliasing. This limit is controlled by two factors: depth into the tissue and transducer frequency.

When dealing with valvular heart disease, most abnormal jets exceed 1.5 to 2.0 m/sec. Therefore, Doppler beginners should not expect to record easily the full flow profile of these abnormal jets using PW Doppler. In fact, most beginners should simply attempt to recognize the presence of aliasing. With experience, it will become easier to identify an aliased signal. The recognition of aliasing on the audio output is difficult

Control of Aliasing

It is also important to realize that the maximum recordable velocities in any jet relate to the frequency of the transducer used. A lower frequency transducer increases the ability of a PW

system to record high velocities at any given range. Thus, aliasing will be encountered at lower velocities with a 5 MHz transducer when compare to a 2.5 MHz transducer.

The main drawback of lowering the transducer frequency is a reduction in the signal to noise ratio of the resulting Doppler data (and thus the quality of the output). For this reason, most standard PW Doppler systems operate at approximately 2.5 MHz.

The second, and most practical, method to overcome aliasing is to take advantage of the range of velocity available in the opposite channel by moving the baseline (also known as "baseline shift"). In **Figure 1.29** almost the entire profile of a pulmonic insufficiency jet can be seen when the baseline is at the bottom of the display. In the left portion of this velocity spectrum, the top of the pulmonic insufficiency profile is poorly seen. As the baseline is moved downwards to the bottom of the display, the top of the spectral profile becomes obvious. This baseline control may be called "zero shift" or "zero off-set" on some systems. Note that use of the baseline shift control doubles the Nyquist limit at any given depth.
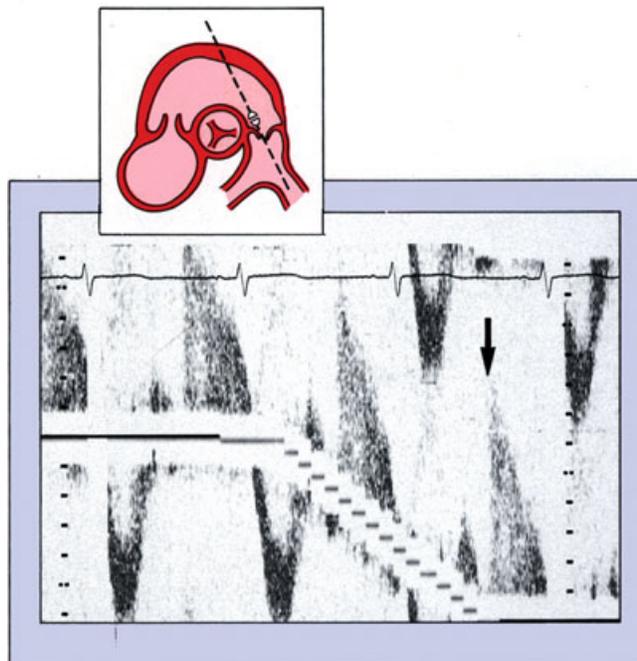

Figure 1. 29: Pulmonic insufficiency is detected by PW Doppler with the sample volume located in the right ventricular outflow tract. The baseline is lowered to reveal the full profile (arrow) of the regurgitant jet toward the transducer.

High PRF Doppler

It seems reasonable to expect that if the problem of aliasing is caused by an insufficiently high PRF, the way to reduce the problem is to find some method of increasing the PRF. Recently, some PW Doppler systems have been introduced which allow the operator to increase PRF above the Nyquist limit and thereby reducing aliasing. High PRF Doppler can use multiples of the PRF corresponding to the Nyquist limit at a given depth.

The basic principle of the so-called "high PRF" Doppler is illustrated in **Figure 1.30**. In this example a given high velocity is located near the mitral valve. If the pulse transit time to the jet and then back to the transducer is one second then the PRF is one pulse per second. Because the velocity being sampled exceeds the Nyquist limit, aliasing is seen in the spectral display. High PRF systems emit multiple pulses without waiting for the original one to be received. In this example multiple pulses are emitted and received each second. This results in an increase in PRF and a spectral display that is not aliased.
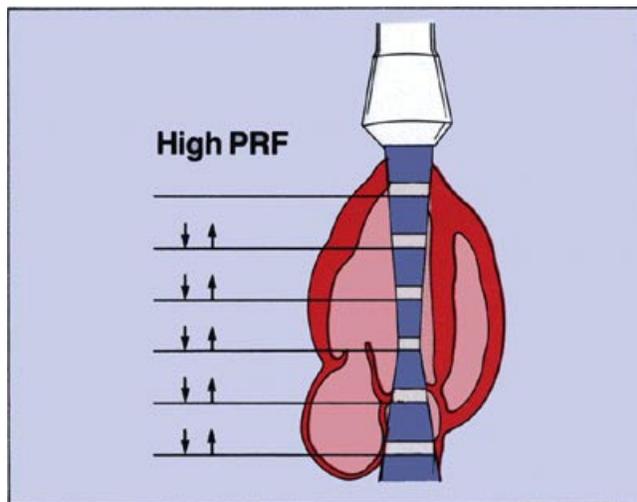

Figure 1. 30: High PRF systems do not wait for a single pulse to return to the transducer before pulsing again. Higher pulse repetition rates are achieved but some range ambiguity exists.

The problem with this approach is that some of the range selectivity used in precisely locating the sample volume is relinquished. As this pulsing sequence is carried on over and over, some data is returned to the transducer the points that are located by the increased PRF in space. If other turbulence were located at any one of these ranges the operator would not be able to tell where the high velocity jet of interest was located, as data from all these volumes are added together. This results in what is called "range ambiguity".

This method of increasing the PRF for the acquisition of high velocity data has been described as programming the machine to "think" that the high velocity jet is much closer to the transducer than it really is. In reality, the machine knows exactly what it is doing and the beginning operator is the one that is fooled.

When using the high PRF mode, it is best first to locate the high velocity jet or turbulence using standard single gate PW Doppler, and to ensure the absence of other areas of turbulence along the path of the ultrasound beam. The high PRF mode can then be used to record the unaliased Doppler signal.
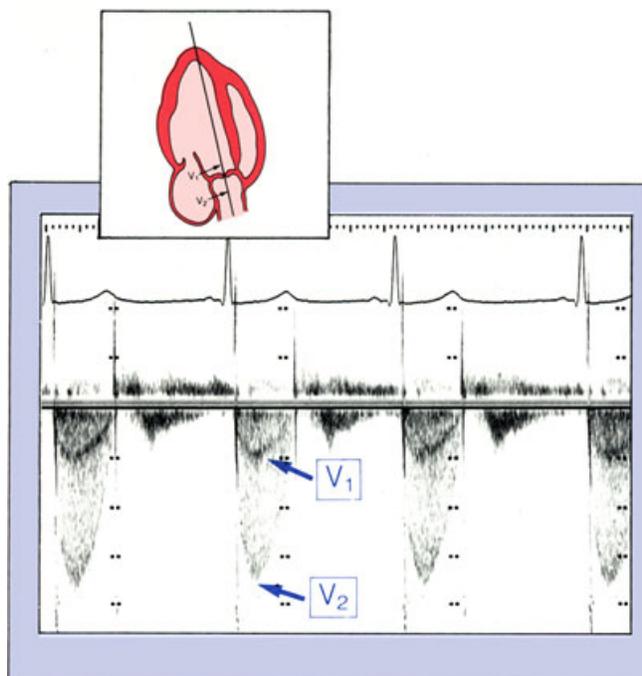
The Bernoulli Equation

There is clear justification for concern over accurate recording of very high velocity jets within the heart. As will be discussed in more detail later, the presence of an obstruction to flow, such as aortic stenosis, will result in a significant increase in velocity across the aortic valve in systole. In practice, these jets attain speeds of up to 7 m/sec.

The Bernoulli equation is a complex formula that relates the pressure drop (or gradient) across an obstruction to many factors, as is seen in **Figure 1.31**. For practical use in Doppler echocardiography this formula has been simplified to: $p_1 - p_2 = 4V^2$

Figure 1. 31: The Bernoulli equation is a complex formula that may be reduced to its simplest expression

As we shall see later, Doppler recordings of velocity may, in certain situations, be used to estimate pressure gradients within the heart. When used for this purpose, it is important to keep in mind that the angle the Doppler beam is incident to any given jet may not be known since these examinations are frequently done blindly by CW. In these cases the operator always tries to orient a beam as parallel to flow as possible so that the full velocity recording is obtained (this assumes cosine $\theta = 1$).

Note that the full Bernoulli equation requires velocity data from below (V_1) and above (V_2) any given obstruction. Since V_1 is normally much smaller than V_2 **(Fig. 1.32)** it can usually be ignored in the calculation of a pressure gradient.

In the example cited, the peak velocity is approximately 3.5 m/sec and this would correspond to an aortic gradient of 48 mmHg by the simplified Bernoulli equation. Obviously, faithful recording of

Bernouilli Equation

$$p_1 - p_1 = 1/2\rho \, (V_2^2 - V_1^2) + \rho_1 \int \frac{d\bar{V}}{dt} \, d\bar{s} + R\,(\bar{V})$$

Convective acceleration — Flow acceleration — Viscous friction

$$p_1 - p_2 \approx 4V^2$$

Figure 1. 32: An example of a CW spectral recording of aortic stenosis. There is a given velocity (V_1) on the ventricular side of the valve that is accelerated (V_2) as blood is ejected through the stenotic orifice. V_1 is usually ignored in the simplified Bernoulli equation.
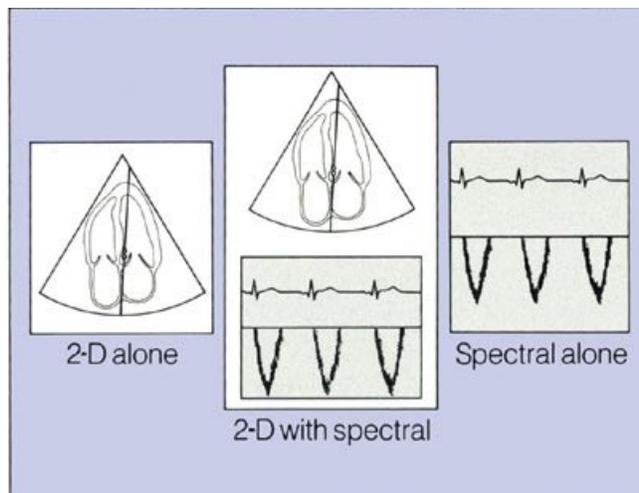
2-D alone

2-D with spectral

Spectral alone

Figure 1. 33: Schematic representation summarizing the various displays available in the combined two-dimensional and Doppler system. Hard-copy spectral recordings are also available in systems with this capability.

Doppler Control Panel

Recording quality — Doppler gain (low/high) — Gray scale (2/16) — Wall filter (less/more)

Pulse sample location — Cursor (L/R) — Sample depth (near/far) — Angle (L/R)

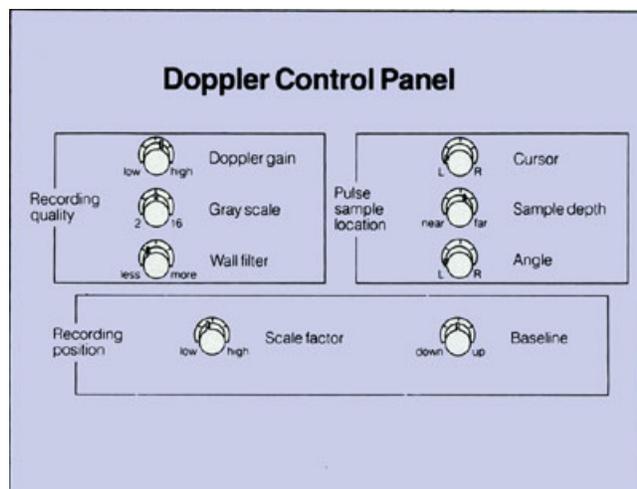Recording position — Scale factor (low/high) — Baseline (down/up)

Figure 1. 34: Mock Doppler control panel showing the most common CW and PW Doppler controls. Almost every Doppler system with pulsed and continuous wave capabilities will contain these controls.

abnormal velocities has great importance, not only for clear identification and recognition of abnormal profiles but also for quantitative purposes.

System Display Capabilities

There are various displays available in Doppler echocardiographic systems **(Fig 1.33)**. The operator sees a normal two-dimensional echocardiographic image when not in the Doppler mode. When the PW Doppler is switched on the cursor and sample volume indicator appear along with a video picture of the spectral display, and the system goes into an automatic gating mode which, in this case, updates the two-dimensional image periodically. Notice the location of the sample volume in the apical four chamber view just below the aortic root. The spectral displays an obviously aliased flow pattern. The scale factors on the spectral display indicate the maximum velocity able to be detected before aliasing occurs.

A hard copy printout of the PW spectral tracing on the screen is also usually available on most machines from a graphic recorder. If a CW examination is then performed, the video screen on the system shows the CW mode spectral display since no imaging is possible in this mode. The full profile of the abnormal jet may then be recorded on the screen or by hard copy strip chart recorder. Of course, higher velocity profiles may be displayed using the CW approach.

THE USE OF THE DOPPLER CONTROLS

Understanding the Doppler controls is very important because improper adjustment of these controls can increase or decrease the quality of the Doppler recordings. A summary of the important controls for Doppler examination of the heart is given in **Figure 1.34**. The schematic drawing of the control
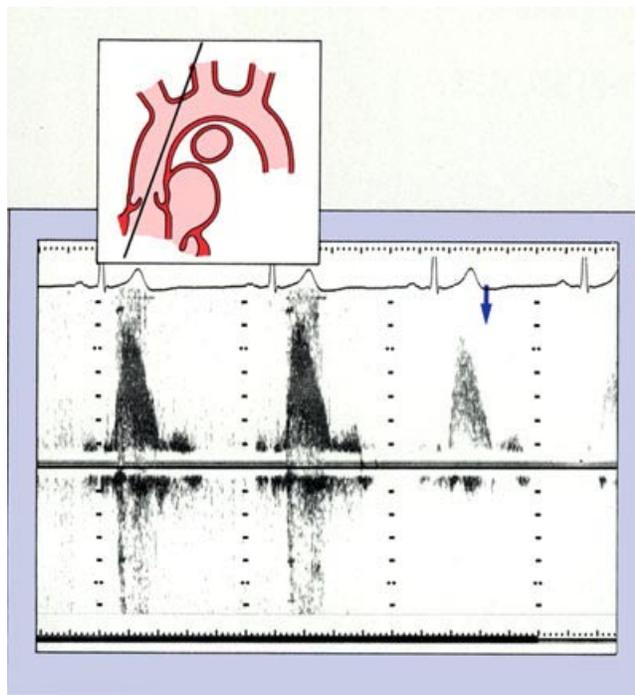
Figure 1. 35: Normal aortic CW spectral trace from the suprasternal window, showing spectral trace with proper gain settings (arrow). Tracings on the left have too much gain while the trace on the right has too little.
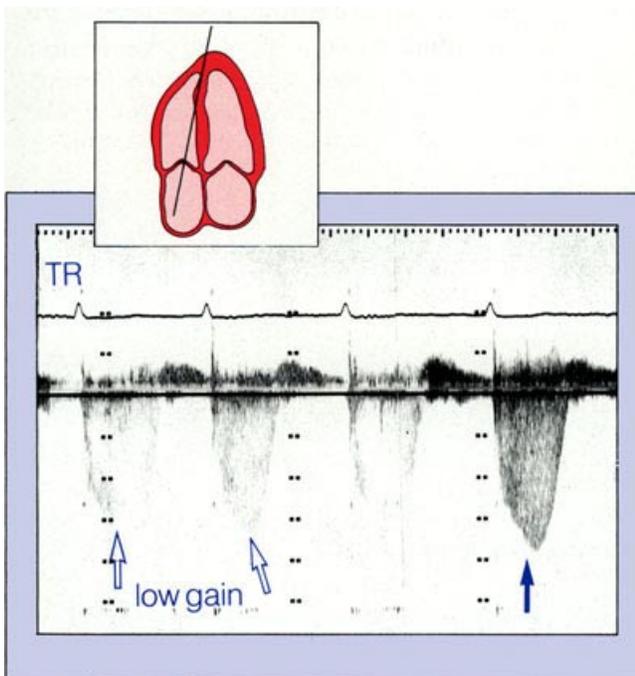

Figure 1. 36: CW spectral velocity trace of tricuspid insufficiency from a transducer located in the apical window. Note the low gain (open arrow) that fails to display the full spectral profile. Proper gain is on the right (closed arrow) and spectral broadening is observed.

panel is generic, and Doppler users should be able to find these controls on their system by the same or a similar name.

In this basic model, the eight controls are divided into three categories. First is the group of controls that influence the quality of the Doppler recording (Doppler gain, gray scale, and wall filter). This group is of importance in both CW and PW examinations. Second are the controls that change the appearance of the graphic display (scale factor and baseline position) and also apply to both CW and PW examinations. The third group are of use *only* for PW Doppler since they relate to the sample volume (cursor, sample depth and angle).

Doppler Gain

The most important Doppler control is the overall gain. Rotating between higher and lower settings will alter the strength of the Doppler signals from the audible output and will be perceived by the operator as a change in the volume of the sound. **Figure 1.35** shows the range in appearance of the velocity spectral recording for excessively high, correct and low gain settings.

As with any ultrasound system, it is prudent to use the lowest gain or power setting that allows the recording of adequate signals. More detailed examination of the recording in **Figure 1.35** shows a normal aortic spectral trace obtained from the suprasternal window. Systolic flow velocity toward the transducer is depicted as an upward profile and is laminar in appearance. The first two profiles show a gain setting that is too high. This results in excessive background noise that makes identification of the clear outline of systolic flow difficult and produces an overflow in the opposite channel represented below the baseline (this is called "mirroring" or "crosstalk"). The third profile **(Figure 1.35, arrow)** is set at an optimal gain setting and displays a clear systolic envelope of flow with minimum background noise, while the fourth profile demonstrates an incomplete spectrum due to an improperly low gain setting.

The practical use of correct gain setting is again shown in **Figure 1.36**. In this CW examination from the ventricular apex, tricuspid insufficiency is encountered as a systolic movement of the

velocity spectrum away from the transducer. In the complexes without adequate gain the full velocity profile is not well seen **(Figure 1.36,** *open arrows***)**. It is not until the gain is increased to an adequate level that true spectral broadening and the true peak velocity are noted. A beginner should first detect some flow signal and then run through all of the possible gain settings to become familiar with the effect of too much or too little gain.

Gray Scale

The gray scale control provides a means of altering the various ranges of gray (from white to black) on the spectral display and has no effect on the audio output of the Doppler system. Different Doppler instruments have from two to more than eight different ranges of gray scale display.
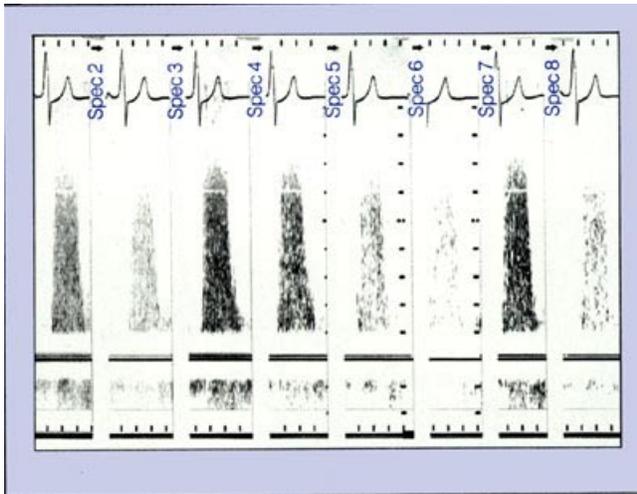

Figure 1. 37: CW spectral velocity recording of eight different gray scale settings. The eight beats are continuous as the gray scale setting changes.
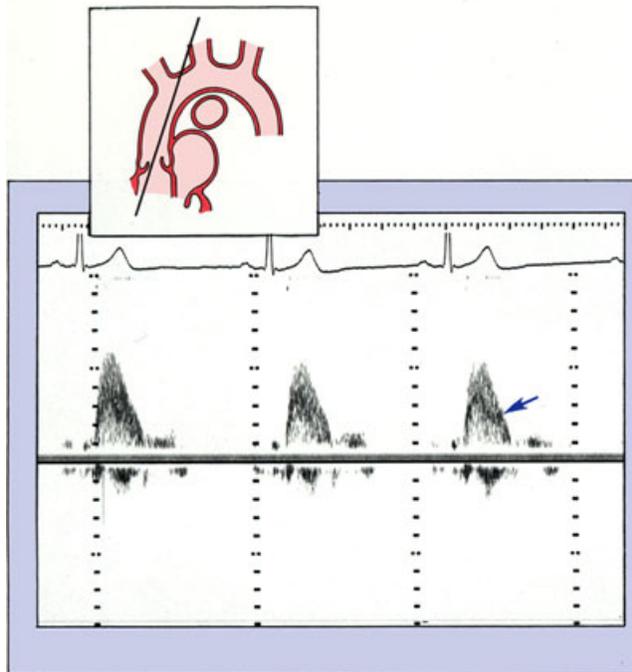
The use of this control at eight different gray scale settings is demonstrated in **Figure 1.37**. This recording was produced from the suprasternal notch and shows eight consecutive beats each with a different gray scale setting. Beginners should practice moving through all available gray scale settings of their Doppler instrument during the learning phase, as the effect of this control is frequently difficult to understand. Ideally, it is desirable to have as many shades of gray as possible in the display. Careful inspection of this tracing in **Figure 1.37** reveals that full range of spectral velocities are not present in the first, second, fifth, sixth or eighth profiles. Adequate gray scale is seen in the third, fourth and seventh profiles.

Since the concentration of velocities within the spectral trace are displayed in varying intensities of gray, full operator understanding of this control is necessary. Lighter shades of gray indicate that there are fewer red cells moving at that velocity in comparison to darker shades of gray or black where many red cells are moving at that velocity.

Figure 1. 38: Proper display of normal aortic flow from the suprasternal notch. Flow is laminar and toward the transducer.

The problem with leaving the gray scale control at the maximum setting is that a light level of gray is assigned to low amplitude background noise in the spectral trace. Thus, there must be a balanced adjustment between the gain control and the gray scale control so that the cleanest spectral trace with the most shades of gray is displayed. A spectral recording of normal aortic flow with properly set gain and gray scale is shown in **Figure 1.38**.

Wall Filter

The low frequency (velocity) of heart motion is easily detected by all Doppler instruments and commonly interferes with clear recording of blood flow profiles. Movement of the heart walls produces easily heard, low pitched "clunking" sounds in the audio output that may obscure the higher frequency flow signals desired.

All Doppler systems have a variable wall filter control that sets the threshold below which low frequency signals are removed from the display. **Figure 1.39** shows improper wall filter settings in a spectral velocity of aortic systolic flow in a patient with aortic stenosis. The arrow points to where there is filtering of the lower frequencies that results in a cleaner baseline, free from low velocity interference. Since we are mainly interested in these high frequency flow signals, it is best to set the wall filter so that most, if not all, the lower frequency signals are attenuated.
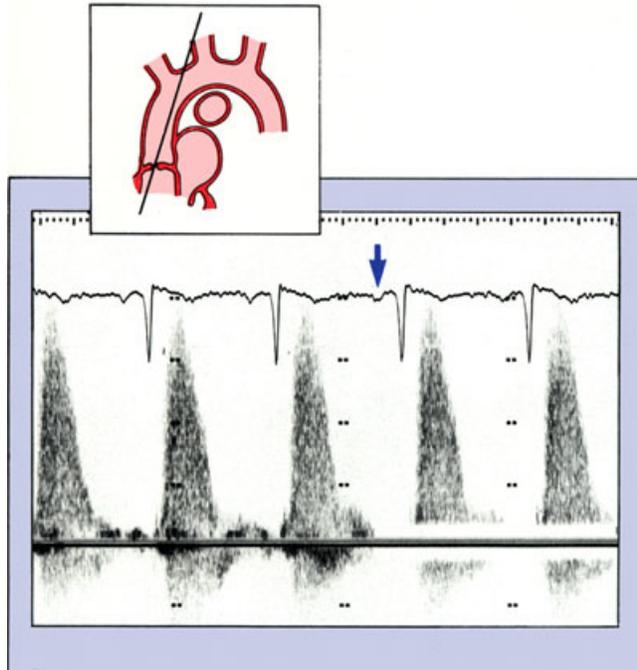
Figure 1. 39: CW spectral velocity recording from the suprasternal notch in a patient with aortic stenosis. Note that the baseline (or wall) filter is turned on (arrow) and the baseline is cleaned of low velocity noise.

Scale Factor

The scale factor control varies the range of velocity that can be displayed on the spectral recording. Altering this control has no effect on the audio output. **Figure 1.40** shows how increasing the scale factor (A vs B) increases the display range of a mitral regurgitation jet using CW Doppler. The velocity of the spectral trace has remained at between 3 and 4 m/sec even though the display size was decreased as well as the velocity calibration markers. The beginner should note that the actual velocity of the spectral trace does not change despite the differing appearances.
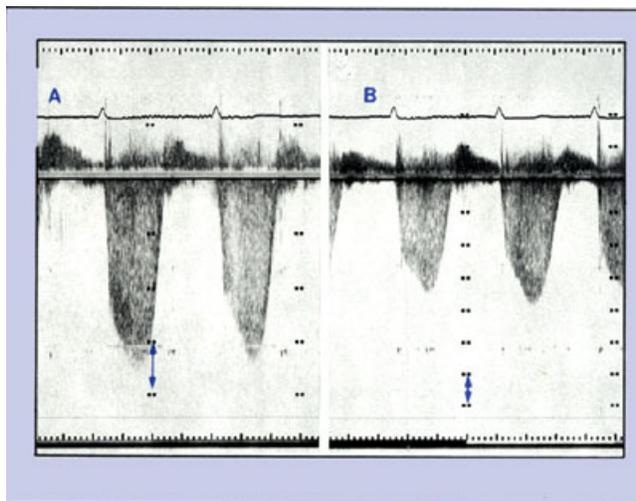
Figure 1. 40: CW spectral velocity recordings of mitral regurgitation from the apical window. The abnormal diastolic jet is away from the transducer. The scale factor is changed between A and B (note arrow).

This control should be set so that the highest velocity spectral trace can be displayed without fear of cutting off a part of the peak velocity. For CW Doppler the operator is usually allowed to move through a full range of scale factors. When operating in pulsed mode, however, many instruments will automatically limit the range of scale factors possible, depending on the depth of the sample volume and the PRF. Scale velocity markers are usually expressed in meters or centimeters per second.

Baseline Control

The baseline control will vary the position of the velocity baseline (zero velocity) within the Doppler display. While altering this control has no effect on the audio output, it is very important for the graphic spectral display. Its use has been previously described in the section discussing aliasing. If an aliased signal is detected it is usually helpful to position the baseline in one of its extreme positions (top or bottom) in order to display as much of the abnormal velocity as possible. This takes advantage of the range of velocity available in the opposite channel.

Pulsed Cursor Position

The Doppler cursor placement is controlled by the cursor switch or paddle (on some machines a joystick or trackball). It is important to remember that the cursor control is operational in both PW and CW mode.
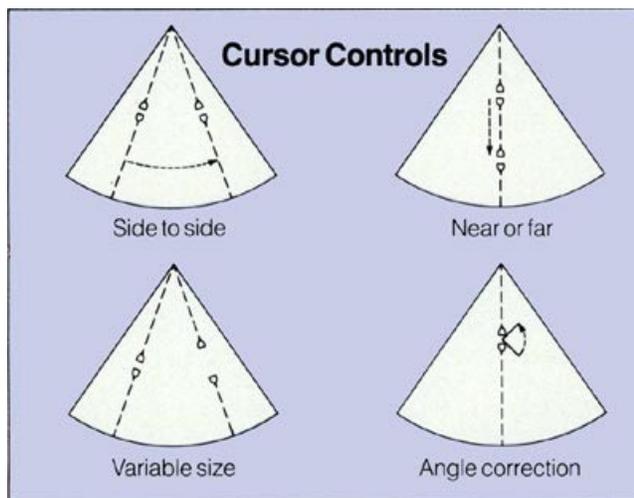

Figure 1. 41: Schematic diagram showing the various PW Doppler sample volume location controls.

This cursor allows movement of the Doppler beam to the left or right within the two-dimensional scan **(Fig. 1.41)**. In common use the movement of this control allows for positioning the pulsed sample volume from one heart chamber to another. Occasionally, this control is used for positioning the Doppler beam as parallel to flow as possible. As will be seen later, the most important use of cursor positioning is in the mapping technique for semi-quantifying regurgitant lesions.

Pulsed Sample Depth

In PW Doppler, the sample depth control selects the depth at which the sample volume gate is placed along the cursor line. As with all cursor and sample volume controls, sample depth has no effect in the CW mode.

Use of this sample depth control is important in properly positioning the sample volume properly for the detection of normal or abnormal flows, particularly in the mapping of various lesions. It should be remembered that deeper positioning of the sample volume into the heart results in an automatic decrease in the Nyquist limit. The farther into the tissues a sample volume is placed the lower will be the velocity at which aliasing will occur.

Other cursor controls include the size of the sample gate (large or small). Most users of pulsed Doppler will leave this gate in a mid-position for routine scanning and decrease or increase the size of the gate depending on the clinical situation.

Pulsed Angle Correction

Most pulsed Doppler systems have some kind of angle correction mechanism. By adjusting according to the direction of assumed flow, it changes the angle calculations in the Doppler equation resulting in different estimates of flow velocity. The use of this control does not actually change the direction of the Doppler beam and its use does not alter the quality of either the audio output or the spectral recording. It also assumes that the operator knows precisely where the jet is

directed. In actual practice it is better to realign the position of the transducer as parallel to perceived flow than to depend upon this angle correction.

There are no standardized methods for carrying out a Doppler examination such as exist for two-dimensional echocardiography. For conventional pulsed and continuous wave Doppler, a partially systematic approach is used by many examiners that begins after the imaging part of the examination is completed. Performing the two-dimensional examination first is very helpful in familiarizing the operator with the spatial positioning of the chambers and valves to be examined by Doppler.

Beam Orientation

It is critical that the operator remembers that the best Doppler information is obtained when the Doppler beam is oriented so that it lines up as parallel as possible to blood flow. This will ensure that the strongest Doppler signals are reflected back to the transducer and that maximum peak velocities are obtained. When using a combined two-dimensional and Doppler machine, the novice must bear in mind the fact that the best two-dimensional pictures usually will not be achieved from exactly the same window that yields the best Doppler tracings. In order to achieve the goal of a small intercept angle to blood flow (as required by the Doppler equation) the operator must try a wide variety of acoustic windows, some of which may not be used for standard M-mode or two-dimensional examinations.
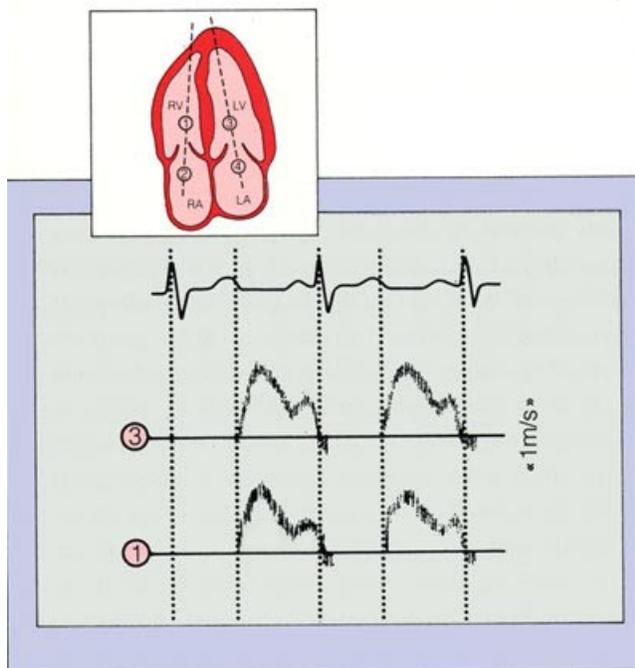
Apical Window

For routine Doppler examination of patients with suspected valvular heart disease, it is usually best to begin by using the apical window. This allows selective orientation of the Doppler beam as parallel as possible to the direction of assumed flow through the mitral and tricuspid valves. It also allows the largest Doppler shift to be recorded and the strongest signals to be reflected back to the Doppler transducer.

The imaging mode of the system may be used to acquire an apical four-chamber view as seen schematically in **Figure 1.42**. The PW Doppler sample volume can then be positioned on the atrial or ventricular sides of the mitral or tricuspid valves. The *right panel* of **Figure 1.42** shows schematic representations of the normal spectral outputs through the mitral (*sample site 3*) and tricuspid valves (*sample site 1*).

Figure 1. 42: Left panel shows a schematic representation of various starting positions for locating the sample volume of a PW Doppler system from the apical four-chamber view. Right panel shows a schematic spectral recording of normal mitral (site 3) and tricuspid (site 1) diastolic flow. Note that peak tricuspid velocity is normally lower than mitral.

In most normal individuals, whether the sample volume is on the atrial or ventricular sides of the mitral and tricuspid valves results in spectral flow outputs that are quite similar. In the presence of valvular disease, however, markedly different flow patterns are encountered depending upon sample volume position.

When in the apical four chamber view, slight superior angulation of the scan plane will allow the operator to encounter the left ventricular outflow tract. An operator can almost always obtain Doppler flow data from the ventricular side of the aortic valve. In some patients data may also be obtained from the aortic root side. A complete PW Doppler evaluation would continue with movement of the sample volume from place to place in the standard, and then intermediate points.

Normal Doppler Velocities

Flow	Children (m/s)	Adults (m/s)
Mitral diastolic	1.00 (0.7-1.4)	0.92 (0.6-1.4)
Tricuspid diastolic	0.62 (0.5-0.9)	0.58 (0.4-0.8)
Pulmonary systolic	0.84 (0.6-1.2)	0.72 (0.5-0.9)
Aortic systolic	1.52 (1.2-1.7)	1.40 (0.9-1.8)

Figure 1. 43: Table of normal peak forward velocities recorded across the various cardiac valves using PW Doppler Echocardiography.

Continued practice repositioning the PW cursor and sample volume in the various portions of the cardiac chambers accessible from the apical four chamber view will eventually provide the novice operator with an appreciation of the spatial locations and directions of normal and abnormal flows. The heart chambers are actually three dimensional structures and an abnormal flow jet may be directed anywhere within this three dimensions. An experienced operator will be able to track an abnormal jet even if it is directed out of a standard two-dimensional plane.

The apical window is also an excellent position for obtaining some initial experience with continuous wave Doppler. Placing the CW transducer directly over the apical impulse (located by palpation) and angling the beam somewhat leftward and posteriorly will usually result in a typical mitral flow profile. It is wise for the beginner to practice locating flow through the mitral valve as the flow profile resembles the
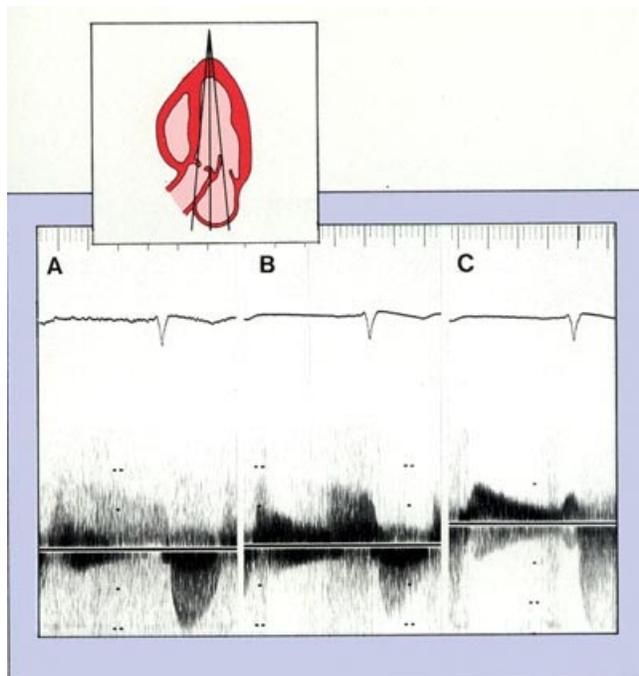

Figure 1. 44: CW velocity spectral recordings from three apical transducer beam angulations. Panel A shows aortic insufficiency and stenosis. In panel B, slight movement of the beam begins to mix the aortic insufficiency with diastolic flow through a mildly stenotic mitral valve. Panel C demonstrates pure mitral diastolic flow and mitral regurgitation.

appearance of the mitral valve on M-mode and is readily recognized. Marked medial redirection of the CW beam from the apex will result in a flow profile through the tricuspid valve. Because pressures are higher on the left side of the heart, velocities are generally higher on the left when compared to the right in normal and diseased states. Exceptions to this rule are encountered in severe pulmonary hypertension or stenosis.

The ranges of normal peak flow velocities are shown in **Figure 1.43**. Notice that the normal flows are slightly higher in children than adults and slightly higher on the left side of the heart in comparison to the right.

After some experience, an operator can become very skillful with minor manipulation of the CW transducer. The difference between the various waveforms obtained from the ventricular apex using a CW transducer is shown in **Figure 1.44**. At first, the beam is directed superiorly to encounter aortic insufficiency and stenosis (**Figure 1.44 *panel A***). The insufficiency is directed toward the

transducer and appears on the spectral display in diastole. The aortic stenosis flow moves away from the transducer in systole. The CW is then angled midway between ventricular outflow and inflow **(Figure 1.44 *panel B*)** and encounters a mixed diastolic profile with mitral inflow superimposed on the aortic insufficiency. Progressive angulation through the mitral valve demonstrates a pure mitral inflow in diastole with mitral insufficiency in systole **(Figure 1.44 *panel C*)**.

The problem of recording flow across mitral and aortic valve simultaneously **(Figure 1.44 *panel B*)** results partly from the fact that the ultrasound beam width is large enough to detect more than one jet. Failure to appreciate this may lead the unwary beginner to diagnose mitral stenosis, for example, when only aortic regurgitation is present.
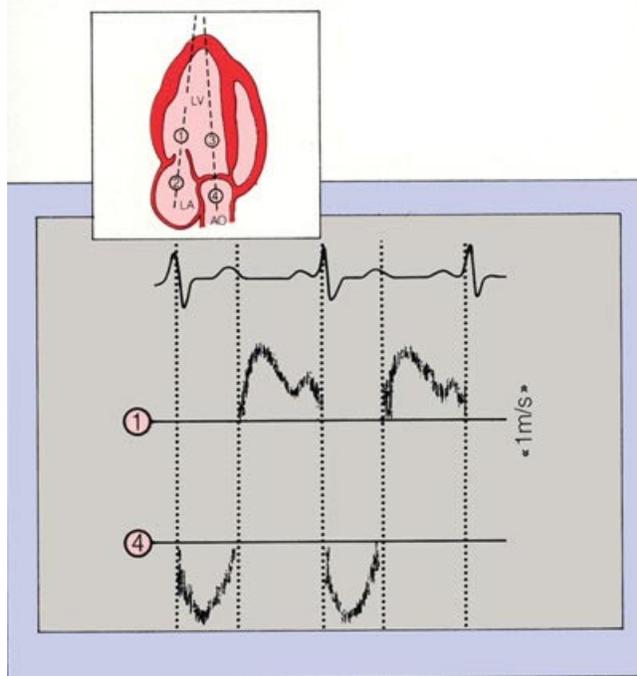

Figure 1. 45: Left panel shows a schematic diagram for location of pulsed Doppler sample volumes from the apical two-chamber view. Right panel shows schematic spectral velocity renditions of flows across the mitral and aortic valves.

The apical window also supplies an opportunity for examination of left sided flow with PW Doppler echocardiography by movement of the imaging plane to the apical two-chamber view **(Fig. 1.45)**. This view is obtained by rotating the two-dimensional transducer counterclockwise and 90 degrees from the apical four chamber view and is particularly suited for examination of both left ventricular inflow **(Fig. 1.45, sample sites 1 and 2)** and outflow **(sample sites 3 and 4)**. Flow velocity profiles through the mitral valve resemble those obtained in the apical four-chamber view and are directed toward the transducer in diastole. It is a little easier to position and hold the pulsed sample volume on the aortic side of the aortic valve using the two-chamber approach rather than the four-chamber with extreme superior angulation.

Suprasternal Window

The suprasternal window provides an opportunity for angulation of the Doppler beam parallel to flow in the ascending or descending aorta. Placement of the PW or CW transducer in the suprasternal notch allows easy access to systolic flow toward the transducer in the ascending aorta and systolic flow away from the transducer in the descending aorta. Positioning a large mechanical or phased array transducer in the suprasternal notch is frequently difficult and we have learned to depend upon CW for this window because the transducer is small and easily maneuvered.

In patients with aortic valve stenosis no such assumption as to direction of systolic aortic flow can be made. In this case, we are most interested in recording the highest peak systolic velocity present. As previously pointed out, the most faithful representation of flow will be obtained when the beam is parallel. The use of multiple positions for the recording of peak systolic aortic velocity is very important in aortic stenosis since this jet may be directed in a wide variety of orientations. When examining for aortic stenosis all available acoustic windows should be used.

Parasternal Windows

The right and left parasternal windows may also be utilized for Doppler echocardiography. Since abnormal jets may be directed anywhere in space, the right parasternal approach should be used in all patients with suspicion of aortic stenosis. This view is best obtained by rotating the patient into a right lateral decubitus position and placing the patient's right hand behind the head to open the intercostal spaces. Practice with the concomitant imaging will help in spatially orienting the beginner to the location of the ascending aorta.
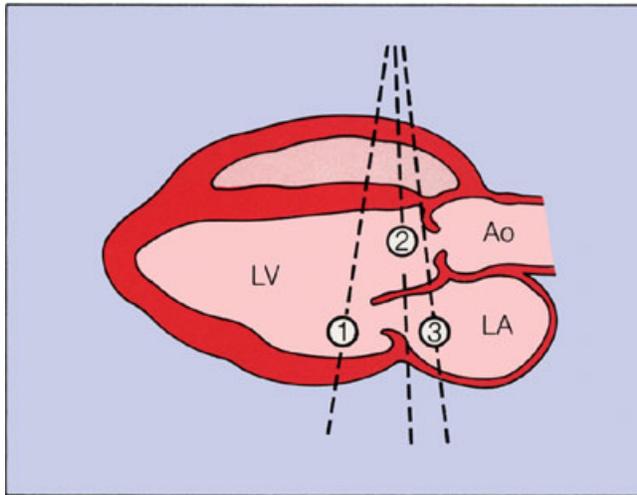

Figure 1. 46: Schematic diagram for sample site locations with PW Doppler from the parasternal long-axis view. Abnormal valvular flows from this view are usually perpendicular to the transducer and are poorly recorded.

The left parasternal window is usually not the best for routine recording of valvular flows through the left heart. As seen in **Figure 1.46**, where the left parasternal long-axis is represented, it is difficult to orient the Doppler beam parallel to flow through the aortic or mitral valves. While some flows may be detected, faithful representation of peak velocities is almost always unrewarding.

When using the left parasternal approach for interrogation of the aortic and mitral valves, it is frequently most profitable to use pulsed wave Doppler in order to provide some range information for interpreting the low amplitude signals. When experience is acquired, one may find this view helpful in specifically localizing turbulence and tracking the spatial orientation of any given jet. Occasionally, the direction of an abnormal jet such as aortic insufficiency will be posterior, making it easiest to record from the left parasternal window.
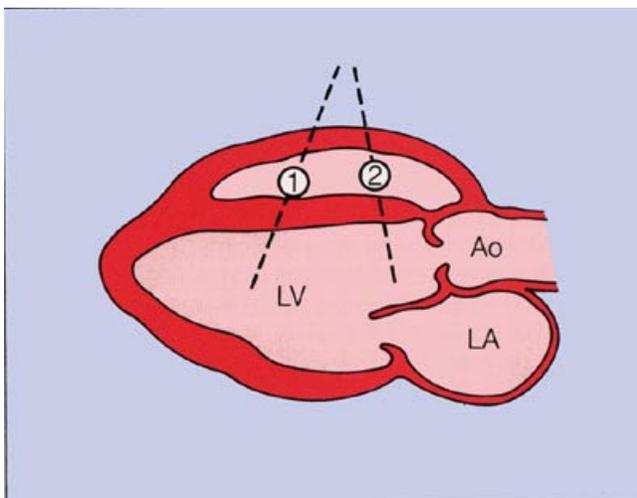

Figure 1. 47: The parasternal long-axis view, is, however good for sampling for the presence of a ventricular septal defect. If present, the abnormal flow is frequently aligned parallel rather than perpendicular to the Doppler beam.

The left parasternal window does have very important use for examining the interventricular septum for ventricular septal defects where flow through a ventricular defect is generally parallel to the interrogating Doppler beam from the left parasternal window **(Fig. 1.47)**. The technique of searching for a ventricular septal defect involves moving the sample cursor along the right ventricular side of the interventricular septum until abnormal flow is detected.

The left parasternal window is also ideal for detecting flow through the pulmonic and tricuspid valves. Using PW Doppler, it is best to start with a short axis view at the level of the aortic root **(Fig. 1.48)**. This shows that Doppler beams directed at the tricuspid or pulmonic valves are parallel to blood flow and will result in a strong Doppler signal. Sample volumes may be positioned on either side of these valves, and the flow profiles resemble the configuration of their left sided counterparts except that the velocities are a little lower.
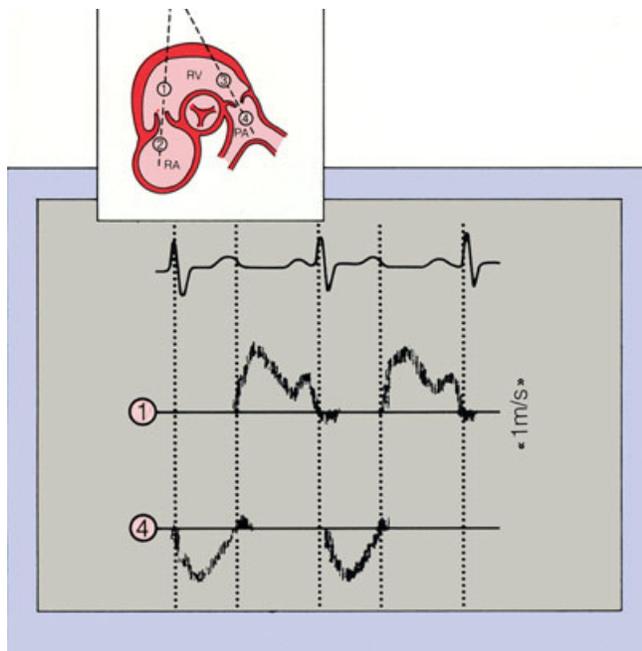
Figure 1. 48: Left panel shows a schematic diagram of possible pulsed Doppler sample sites in the left parasternal short-axis view. Right panel shows schematic renditions of normal tricuspid diastolic (site 1) and pulmonic systolic (site 4) flows.

In some patients the left parasternal view may be the only one where intelligible flow is detected through the pulmonic valve. Beginners will find the use of pulsed Doppler for detection of flow easiest. Note that normal systolic flow in the pulmonary artery is away from the transducer in this position. As the operator gains experience, he or she will also feel comfortable using continuous wave Doppler for recording the full spectral profile of abnormal jets. Skill in examining flow through the pulmonary valve is most useful in the study of congenital heart disease.

Other Windows

It is important note that all other available windows be used for interrogating flow through the heart. These include the lower left parasternal, subcostal and many intermediate windows. We have described a few windows in detail to provide a means for beginners to familiarize themselves with the easiest, and generally most productive, transducer positions and beam directions.

The Doppler Examination

Using this sequential window approach is not the only way to conduct a methodical Doppler examination but it has appeared to us to be very helpful to those with little, or no Doppler experience. When conducting a Doppler examination, the operator should remember that these studies may be very difficult to interpret for an individual not present during the actual examination. A poor quality echocardiographic image of cardiac structures may be identifiable by an experienced observer but a poor quality Doppler tracing may be impossible to identify and interpret.

For this reason we believe that sonographers and physicians interested in acquiring skills in Doppler echocardiography should conduct the examinations jointly during the learning period. We also highly recommend that each laboratory develop an annotation system so that sonographers may convey key orientation information to an interpreting physician not actually present during the examination. This may be accomplished on almost every commercial system by voice labeling (or commenting) on the audio track of the tape recording of the video or audio record. The information should concern the window used, probable beam orientation and suspected lesion encountered. We also suggest written annotation of similar information on the hard-copy paper output of the graphic recorder.

FIGURES:

Figure 1.1: An example of the Doppler effect in every day life: the sound emitted from a stationary and moving automobile engine.

Figure 1.2: Diagrammatic representation of normal laminar flow in comparison with turbulent flow that results in whirls and eddies of many different velocities.

Figure 1.3: Examples of normal laminar flow through the aortic valve (top) and disturbed or turbulent flow resulting from aortic stenosis (bottom).

Figure 1.4: Sound is emitted in waves of measurable frequencies.

Figure 1.5: For any given transmitted ultrasound frequency, the returned frequency will be higher after encountering red blood cells moving toward the transducer and lower after encountering red cells moving away from the transducer.

Figure 1.6: The Doppler equation solved for frequency shift.

Figure 1:7: The phase shifts of returning frequencies are compared with transmitted frequency within the Doppler system. Positive shifts (top) result from blood moving away from, and negative shifts (bottom) result from blood moving toward, the transducer.

Figure 1.8: The Doppler equation rearranged for determination of velocity. It is velocity that is displayed by the Doppler instrument.

Figure 1.9: Schematic representation of the velocity output of flow. Flow toward the transducer is displayed above the baseline and flow away from the transducer is displayed below the baseline.

Figure 1.10: The various velocities detected by the Doppler instrument are processed by Fast Fourier Transform (FFT) and the resulting spectrum of velocities present is displayed. Laminar flows are uniform. Turbulent flows show spectral broadening.

Figure 1.11: The spectral analysis is created by placing the velocity data into bins that are displayed over time.

Figure 1.12: The brightness of the signal at any given bin relates to the relative number of red cells detected at that velocity. The term "amplitude" is applied to relative brightness.

Figure 1.13: An example of the various Doppler displays from a patient with mitral stenosis with the transducer held at the apex. Flow in diastole is toward the transducer. The ECG, analog outputs (maximum and mean velocities), spectral display, and amplitude signals are shown.

Figure 1.14: Schematic diagram showing the importance of being parallel to flow when detecting flow through the aortic valve. A jet of known velocity (2.0 m/s) emerges from the aortic valve in systole. Moving 60 degrees from parallel only allows a peak velocity of 1.0 m/s to be recorded. The most accurate velocities are recorded when the transducer is parallel to flow.

Figure 1.15: The effect of varying angulation in relation to a systolic jet in a patient with aortic stenosis taken from the suprasternal notch with flow toward the transducer. Note that the peak jet is nearly 5 m/s (open arrow). In subsequent beats the flow profile is lost.

Figure 1.16: Suprasternal (left, with flow toward the transducer) and apical (right, with flow away from the transducer) jet of aortic stenosis. The best profile was taken from the apical position.

Figure 1.17: In CW Doppler, the transducer is constantly emitting and receiving ultrasound data.

Figure 1.18: In PW Doppler, the transducer alternately transmits and received the ultrasound data to a sample volume. This is also known as range-gated Doppler.

Figure 1.19: When the PW Doppler operates, it causes the two-dimensional image to be held in a frozen frame. The image is periodically updated and will usually appear as a blank on the spectral display (dashed lines).

Figure 1.20: The sample volume of PW Doppler is actually a three-dimensional volume that changes in size at its location relative to the transducer is changed. When placed in the far field, it becomes very large.

Figure 1.21: Schematic rendering of the full spectral display of a high velocity profile fully recorded by CW Doppler. The PW display is aliased, or cut off, and the top is placed at the bottom.

Figure 1. 22: Spectral displays of diastolic flow through the mitral orifice. The transducer is located at the apex and diastolic flow is toward the transducer (positive) Note the laminar appearance of the PW display. The CW does not usually display the same laminar flow pattern as it receives flow information from all portions of the ultrasound beam.

Figure 1. 23: Table summarizing the advantages and disadvantages of pulsed and continuous wave Doppler echocardiography.

Figure 1. 24: Aliased spectral display of aortic insufficiency (left arrow) in PW mode detected from the ventricular apex. Abnormal flow is toward the transducer. After 3 beats, the system is switched to CW and the full profile is seen.

Figure 1. 25: Schematic depiction of the origin of the aliasing phenomenon using a turning wheel. When the Nyquist limit is exceeded, aliasing occurs. For details, see text.

Figure 1. 26: The Nyquist limit is defined by the number of pulses/second divided by two.

Figure 1. 27: The left panel shows the sample volume placed in the near field. Higher velocities may be recorded with a location near the transducer than when the sample volume is positioned at a farther range (right panel).

Figure 1. 28: Schematic drawing of the effect of a reduction in the Nyquist limit in PW echo when a high velocity jet is encountered in the near field (A) or successively deeper into the tissues (B, C, and D). Note the progressive and marked aliasing of the spectral signal the farther away the jet is encountered from the transducer.

Figure 1. 29: Pulmonic insufficiency is detected by PW Doppler with the sample volume located in the right ventricular outflow tract. The baseline is lowered to reveal the full profile (arrow) of the regurgitant jet toward the transducer.

Figure 1. 30: High PRF systems do not wait for a single pulse to return to the transducer before pulsing again. Higher pulse repetition rates are achieved but some range ambiguity exists.

Figure 1. 31: The Bernoulli equation is a complex formula that may be reduced to its simplest expression.

Figure 1. 32: An example of a CW spectral recording of aortic stenosis. There is a given velocity (V_1) on the ventricular side of the valve that is accelerated (V_2) as blood is ejected through the stenotic orifice. V_1 is usually ignored in the simplified Bernoulli equation.

Figure 1. 33: Schematic representation summarizing the various displays available in the combined two-dimensional and Doppler system. Hard-copy spectral recordings are also available in systems with this capability.

Figure 1. 34: Mock Doppler control panel showing the most common CW and PW Doppler controls. Almost every Doppler system with pulsed and continuous wave capabilities will contain these controls.

Figure 1. 35: Normal aortic CW spectral trace from the suprasternal window, showing spectral trace with proper gain settings (arrow). Tracings on the left have too much gain while the trace on the right has too little.

Figure 1. 36: CW spectral velocity trace of tricuspid insufficiency from a transducer located in the apical window. Note the low gain (open arrow) that fails to display the full spectral profile. Proper gain is on the right (closed arrow) and spectral broadening is observed.

Figure 1. 37: CW spectral velocity recording of eight different gray scale settings. The eight beats are continuous as the gray scale setting changes.

Figure 1. 38: Proper display of normal aortic flow from the suprasternal notch. Flow is laminar and toward the transducer.

Figure 1. 39: CW spectral velocity recording from the suprasternal notch in a patient with aortic stenosis. Note that the baseline (or wall) filter is turned on (arrow) and the baseline is cleaned of low velocity noise.

Figure 1. 40: CW spectral velocity recordings of mitral regurgitation from the apical window. The abnormal diastolic jet is away from the transducer. The scale factor is changed between A and B (note arrow).

Figure 1. 41: Schematic diagram showing the various PW Doppler sample volume location controls.

Figure 1. 42: Left panel shows a schematic representation of various starting positions for locating the sample volume of a PW Doppler system from the apical four-chamber view. Right panel shows a schematic spectral recording of normal mitral (site 3) and tricuspid (site 1) diastolic flow. Note that peak tricuspid velocity is normally lower than mitral.

Figure 1. 43: Table of normal peak forward velocities recorded across the various cardiac valves using PW Doppler Echocardiography.

Figure 1. 44: CW velocity spectral recordings from three apical transducer beam angulations. Panel A shows aortic insufficiency and stenosis. In panel B, slight movement of the beam begins to mix the aortic insufficiency with diastolic flow through a mildly stenotic mitral valve. Panel C demonstrates pure mitral diastolic flow and mitral regurgitation.

Figure 1. 45: Left panel shows a schematic diagram for location of pulsed Doppler sample volumes from the apical two-chamber view. Right panel shows schematic spectral velocity renditions of flows across the mitral and aortic valves.

Figure 1. 46: Schematic diagram for sample site locations with PW Doppler from the parasternal long-axis view. Abnormal valvular flows from this view are usually perpendicular to the transducer and are poorly recorded.

Figure 1. 47: The parasternal long-axis view, is, however good for sampling for the presence of a ventricular septal defect. If present, the abnormal flow is frequently aligned parallel rather than perpendicular to the Doppler beam.

Figure 1. 48: Left panel shows a schematic diagram of possible pulsed Doppler sample sites in the left parasternal short-axis view. Right panel shows schematic renditions of normal tricuspid diastolic (site 1) and pulmonic systolic (site 4) flows.