

VETERINARY ECHOCARDIOGRAPHY

Second Edition

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THE PHYSICS OF ULTRASOUND

Cardiac ultrasound, echocardiography, permits noninvasive and nonionizing visualization of the inside of the heart including the aorta, the ventricles and atria, the auricular appendages, and all of the cardiac valves. Dynamic images of the contracting heart are created with two-dimensional and motion mode (M-mode) images while blood flow through the heart can be seen and measured with Doppler ultrasound. Tissue Doppler imaging allows analysis of myocardial motion. Defects including valvular lesions, cardiac shunts, cardiac and thoracic masses, pleural and pericardial effusions, myocardial diseases, and stenotic lesions can be seen. More importantly it allows assessment of cardiac chamber sizes, cardiac function, blood flow, and myocardial motion, which provides information on hemodynamic status and extent of the disease process.

Uses of Echocardiography

- See internal cardiac structures.
- Evaluate function.
- Evaluate size.
- See defects.
 - Valvular lesions
 - Shunts
 - Myocardial abnormalities
 - Masses
 - Effusions
 - Stenotic lesions
- Evaluate blood flow.
- Assess myocardial motion and function.

All of this is possible because of sound. Sound is sent into the body and reflected from soft tissue structures. The reflected sound waves are analyzed, and an image is generated on a monitor. Sending out many sound waves side by side will produce an image with depth

and width. The result is a two-dimensional image (Figure 1.1). When the sound waves are continuously and rapidly sent out in sequence, many two-dimensional images can be generated per minute, and a moving image of the heart is made called real-time or B-mode ultrasound. By sending out only one sound beam instead of many, only the structures associated with that one beam are seen, producing an M-mode image (Figure 1.2). The structures associated with that one line through the heart keep scrolling on the screen as the heart continues to contract and relax. The M-mode image displays depth on the vertical axis and time along the horizontal axis.

Doppler is used in diagnostic ultrasound to provide information on blood flow (spectral and color-flow Doppler) or myocardial motion (tissue Doppler imaging [TDI]) of the heart and its vessels. Specific locations within the heart can be selected and a spectral display of blood flow or muscle motion is created. As in M-mode the horizontal axis represents time while the vertical axis represents velocity (Figure 1.3).

This chapter deals with the physical principles of sound waves that allow ultrasound to be used as a diagnostic tool. The physics of ultrasound involves an understanding of the basic properties of sound waves and how these properties affect transducer selection, image quality, and diagnostic interpretation. Only the principles needed to make knowledgeable technical decisions and diagnostic interpretations are presented in this chapter. More detailed information can be found in books dedicated to the physics of diagnostic ultrasound. Selected references are listed at the end of the chapter.

BASIC PHYSICS

Cycles and Wavelengths

Sound waves travel in longitudinal lines within a medium. The molecules along that longitudinal course of movement are alternately compressed (molecules

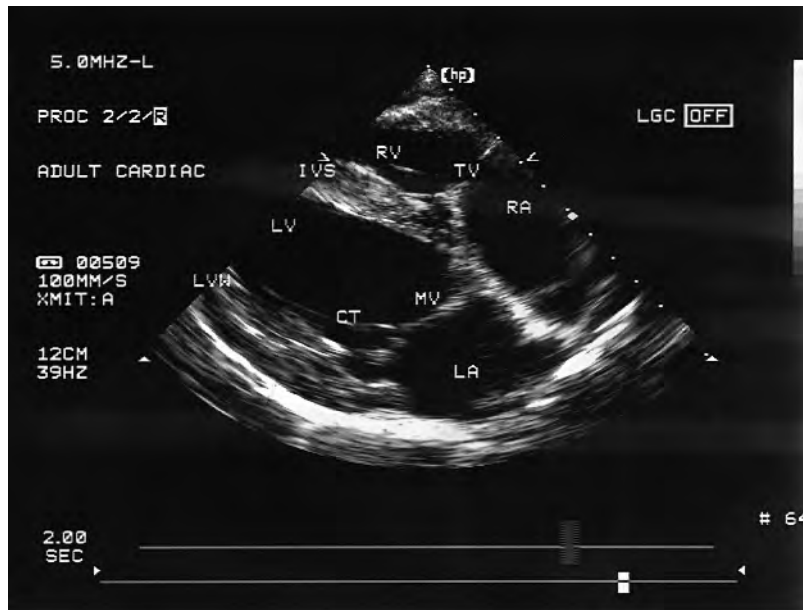


Figure 1.1 Many sound waves sent into the body side by side will create an image with depth and width. The result is a two-dimensional echocardiographic image. This is a right parasternal long-axis four-chamber view of the heart. RV = right ventricle, TV = tricuspid valve, RA = right atrium, IVS = interventricular septum, LV = left ventricle, LVW = left ventricular wall, LA = left atrium, MV = mitral valve, CT = chordae tendineae.

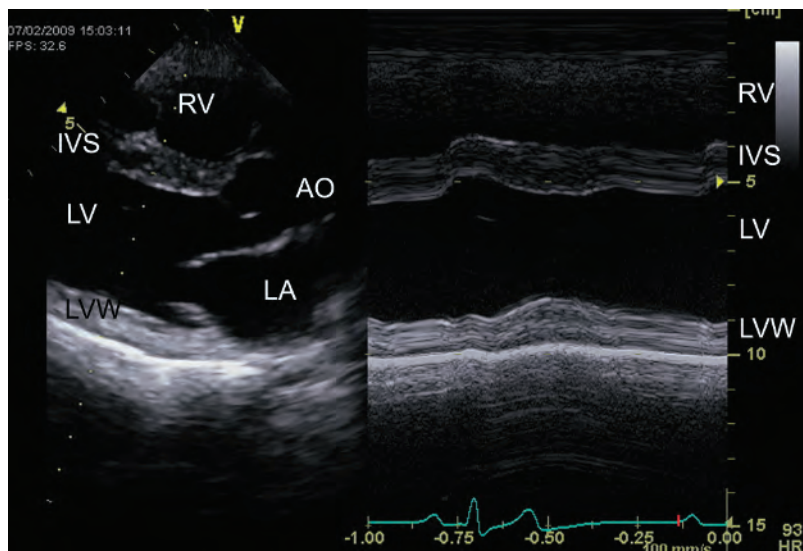


Figure 1.2 When one of the sound beams used to create the two-dimensional image is selected to generate an M-mode image, only the structures associated with that one beam are seen. The two-dimensional image at the left side of this figure shows a cursor representing the one beam. A one-dimensional M-mode image is created from the structures the cursor crosses. The vertical axis represents depth while the horizontal axis represents time. Here the structures of the left ventricle are seen on the M-mode to the right as they change throughout a cardiac cycle. RV = right ventricle, IVS = interventricular septum, LV = left ventricle, LVW = left ventricular wall, AO = aorta, LA = left atrium.

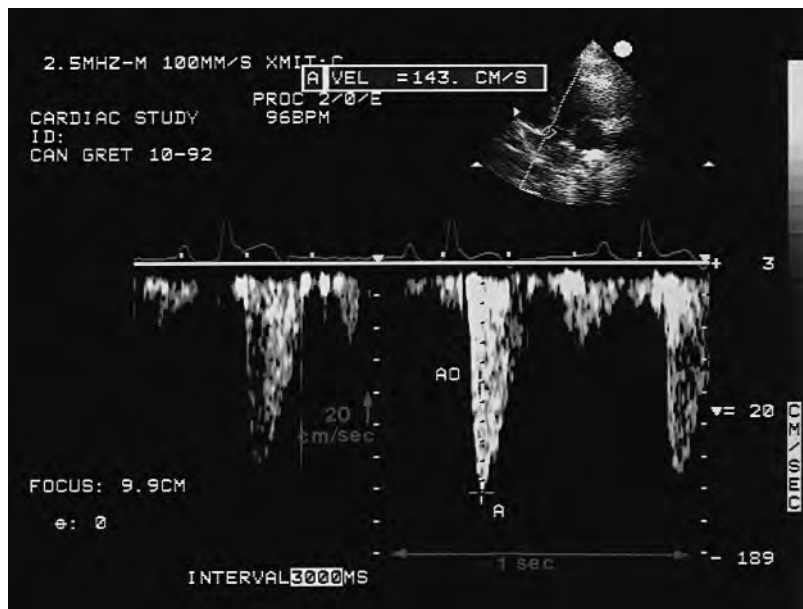


Figure 1.3 Doppler images display flow velocities on the vertical axis and time on the horizontal axis. Blood flow for specified areas in the heart is seen as it accelerates, reaches a maximum velocity, and then decelerates throughout the cardiac cycle. This CW Doppler tracing of aortic flow (AO) in a dog has a velocity of 143 cm/sec (A).

move closer together) and rarefacted (molecules are spread apart). The time required for one complete compression and rarefaction to occur is one cycle (Figure 1.4). The distance in millimeters that the sound wave travels during one cycle is its wavelength.

Sound Waves

- Alternately compress and spread apart the molecules in their pathway.
- 1 cycle = one complete compression and expansion
- Wavelength = the distance traveled during 1 cycle

The source of the sound wave determines the length of a cycle. Transducers generate the sound in diagnostic ultrasound. They will be discussed in detail later, but for any given transducer the wavelength is constant.

Frequency

The number of cycles per second is the frequency of the sound wave (Figure 1.5). Frequency is measured in Hertz (Hz), where 1 Hz equals one cycle per second. Ultrasound has a frequency greater than 20,000 cycles per second, and is beyond the range of human hearing.

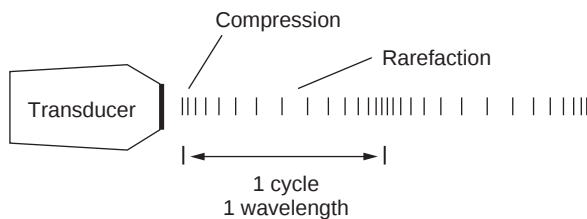


Figure 1.4 Sound waves cause compression and rarefaction of the molecules along their path. The time for one complete compression and rarefaction to occur is called a cycle. The distance sound travels during one cycle is measured in millimeters and is its wavelength.

Since frequency is the number of complete cycles per second, the higher the frequency of the sound wave the shorter the wavelength must be.

Frequency

- The number of cycles per second = frequency
- High frequency = shorter wavelengths
- Low frequency = longer wavelengths

A 5.0-megahertz (MHz) transducer transmits 5 million cycles per second at 0.31 millimeters (mm) per cycle, while a 2.0-MHz transducer transmits only 2 million cycles per second at 0.77 mm per cycle. Table 1.1 lists wavelengths for sound generated at various frequencies.

Speed of Sound

The speed of sound (V) depends upon the density and stiffness of the medium through which it is traveling. Increased density allows sound to travel faster. The velocity of sound does not change within a homogeneous substance and is independent of frequency (Figure 1.6). Table 1.2 lists the speed of sound in various tissues. The speed of sound through air is very slow because of its low density, while bone allows sound to travel at relatively high speeds.

The average velocity of a sound wave in soft tissue is 1,540 meters per second regardless of transducer frequency (Figure 1.7). Velocity is calibrated into the ultrasound machine, which then calculates the distance (D) to cardiac structures based upon how long it takes to receive reflected echoes (T):

$$D = V \times T/2 \quad \text{Equation 1.1}$$

Transducer frequency does not affect the speed of sound in tissues.

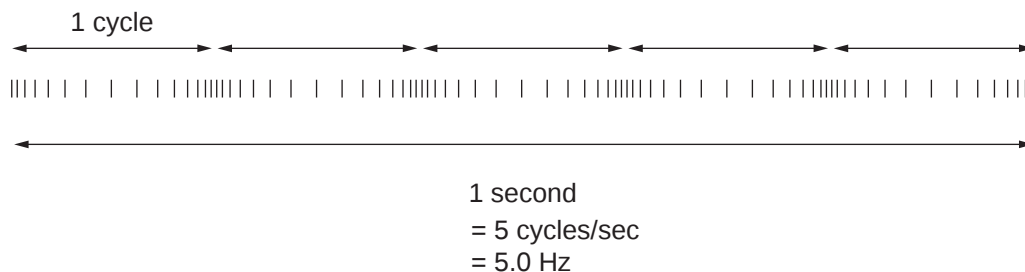


Figure 1.5 The number of cycles per second is the frequency of the sound wave. Frequency is measured in Hertz (Hz). One Hz equals one cycle per second.

Table 1.1

Wavelength of Sound at Commonly Used Frequencies

Frequency (MHz)	Wavelength (mm)
2.0	.77
3.5	.44
5.0	.31
7.5	.21

Table 1.2

The Speed of Sound in Soft Tissues

Tissue	Speed(m/sec)
air	330
fat	1,440
brain	1,510
liver	1,560
kidney	1,560
muscle	1,570
blood	1,570
bone	4,080

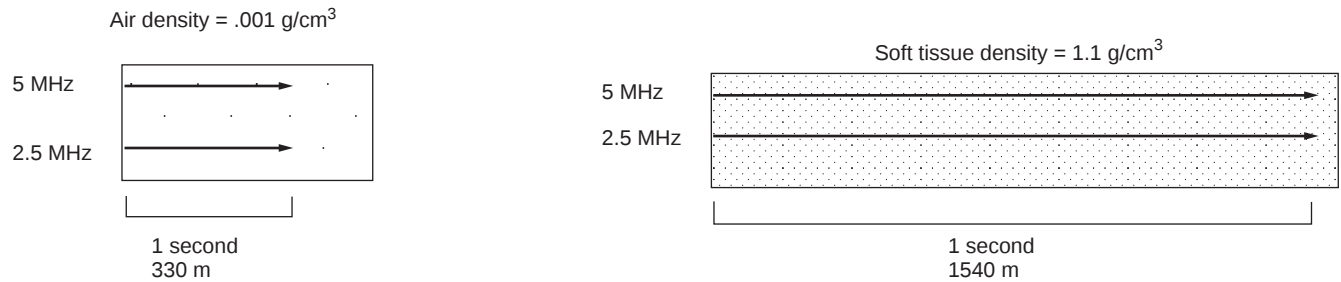


Figure 1.6 Increased tissue density allows sound to travel faster. Sound generated by a 2.5-MHz transducer and a 5.0-MHz transducer will have the same velocity within the same tissues since the speed of sound is not affected by frequency.

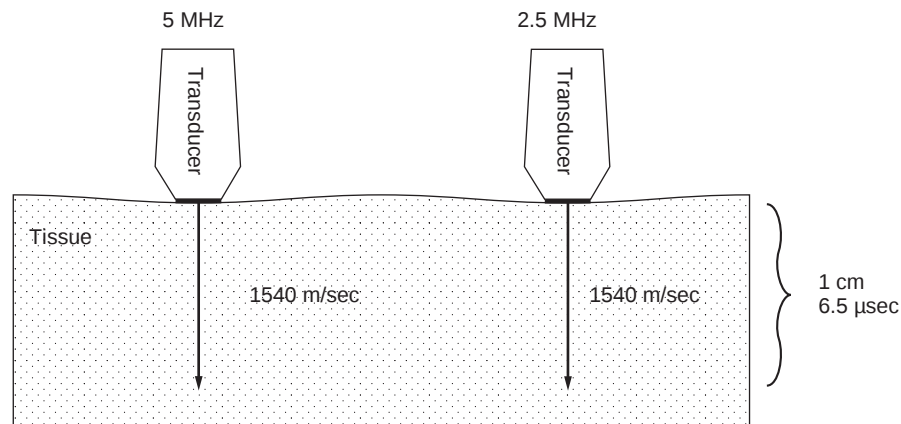


Figure 1.7 Sound travels through soft tissues at an average velocity of 1,540 m/sec regardless of transducer frequency. The time required to travel 1 cm at 1,540 m/sec is 6.5 microseconds (μsec) one way and 13 μsec round trip.

The time (T) required to travel 1 cm is 6.5 microseconds or 13 microseconds round trip. Even though sound must travel through various tissues with slightly different velocities during an echocardiographic exam, the equipment is calibrated for the average speed of sound in soft tissues (1,540 meters per second). Structures are displayed on a monitor at the calculated depth, and an image of the heart is created. This always creates some degree of error in calculating true structure depth, but the error is generally negligible.

Acoustic Impedance

Acoustic impedance is the opposition or resistance to the flow of sound through a medium. Impedance depends upon the density and stiffness of the medium and is independent of frequency. Very stiff or hard materials are hard to compress and rarefact. Therefore, although increased density increases the speed of sound, if the ability to compress and rarefact a sound wave is limited, the impedance or resistance to sound transmission is high.

Table 1.3
Acoustical Impedance of Various Tissues

Tissue	Impedance (g/cm ² sec)
bone	7.80×10^5
muscle	1.70×10^5
kidney	1.62×10^5
blood	1.61×10^5
brain	1.58×10^5
fat	1.38×10^5
air	$.0004 \times 10^5$

acoustic impedance = density $\times$ speed **Equation 1.2**

Contradictory as it sounds, the higher the density and the greater the velocity of sound through a medium, the greater the resistance is to sound transmission. Table 1.3 lists the acoustical impedance of sound in various tissues. Because of its stiffness and inability to compress and rarefact molecules easily, bone has high impedance, while air, because its molecules are easily compressed and rarefacted, has low impedance. High acoustical impedance is what produces a high degree of sound reflection at bony or air interfaces, creating a shadow on the ultrasound image beyond the bone or air due to lack of further sound transmission.

Reflection of Sound

- Depends upon acoustical mismatch
 - The greater the difference in acoustical properties the greater the degree of reflection.
- Depends upon angle of incidence
 - Sound striking an organ perpendicularly will have a large amount of sound reflected straight back to the transducer.
- Depends upon reflecting structure's size
 - Must be at least 1/4 size of the wavelength
 - Higher frequency transducers can reflect sound from smaller structures.

Reflection, Refraction, and Scattering

Reflection is sound that is turned back at a boundary within a medium. These reflected echoes are called specular echoes. When an interface between two tissues with different acoustical impedances is reached, a portion of the sound is reflected back to the transducer. The rest continues on through the tissues. The greater the difference in acoustical impedance the

greater the degree of reflection. For the same reason, if two boundaries have little or no acoustical mismatch they will not be identified as two different tissues. Therefore interfaces between muscle and fluid, as in the heart, will reflect sound at different intensities while the cells within the homogeneous muscle itself will reflect sound with similar strengths.

All interfaces between muscle and blood-filled chambers in the heart have slightly brighter boundaries on the ultrasound image because of this increased reflection. The interface between tissue and air has an even greater difference in acoustical impedance, and therefore the pericardial sac around the heart is always one of the brightest structures on the ultrasound image. The gel placed between the transducer and skin surface is used to prevent the large degree of reflection ordinarily seen between a tissue and air interface.

The angle at which sound strikes the reflective surface (the angle of incidence) determines the angle of reflection. The angle of reflection is equal to the angle of incidence (Figure 1.8). When sound is directed perpendicular to a structure the angle of incidence is zero and the sound is reflected straight back to the transducer. If the angle of incidence is 50°, then the angle of reflection will also be 50°. When the angle of incidence is 90° or parallel to the interface, no sound will be reflected back to the source. This principle tells us that the best two-dimensional and M-mode cardiac images are obtained when sound is directed perpendicular to the tissues.

Not all sound is reflected however, and some continues on through the tissues. These sound waves are refracted if the two tissues are different (Figure 1.8). Refraction is the change in direction of sound as it travels from one medium to another. This is similar to what happens when light waves in water create a distorted image. The greater the mismatch in acoustical impedance between the two tissues the greater the degree of refraction. As the refracted sound beam travels in a new direction, the angle of reflection with respect to the original source is different, and positional errors can result since the transducer thinks the received sound is coming from the same direction as the sound waves it generated earlier.

The errors produced by refraction during an examination create few problems unless the refracted beam has to travel a great distance. An angle of 1 or 2 degrees at the top of the refracting tissue can result in a several millimeter error in position by the time it reaches the far side of a deep structure. When the two mediums differ enough to create a refractive angle of greater than 90° (as with soft tissue and bone) then an image is not generated beyond the second structure.

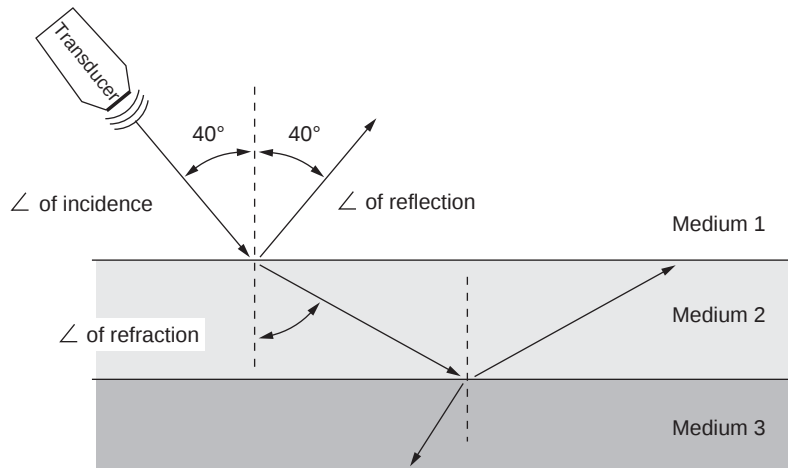


Figure 1.8 The angle of reflection is equal to the angle at which sound strikes the tissue. Sound that is directed perpendicular to the tissue is reflected straight back to the transducer producing the best images. Sound is refracted when it crosses a boundary between two different tissues. The greater the difference in acoustical properties between the two tissues, the greater the degree of refraction.



Figure 1.9 Structures that are small and irregular with respect to the wavelength cause sound to be scattered in all directions. Some of this scattered sound will be directed back to the transducer for image generation. Scattered sound is important in tissue characterization.

Reflection of sound is not only dependent upon the acoustical mismatch of two tissues but also upon the structure's size. The structure must be at least one-quarter the size of the wavelength for reflection to occur. The short 0.21-mm wavelengths of a 7.5-MHz transducer are reflected from structures that are as small as 0.05 mm in thickness, while structures must be at least 0.19-mm thick for the 0.77-mm wavelengths of a 2.0-MHz transducer to be reflected. High frequency transducers then, provide higher resolution images since smaller structures reflect their sound waves.

↑ Frequency = ↓ Wavelength = ↑ Resolution
↓ Frequency = ↑ Wavelength = ↓ Resolution

Structures that are small and irregular with respect to the sound wave do not reflect sound but rather scatter it in all directions without regard for the angle of incidence (Figure 1.9). Some of this scattered sound is directed back to the sound source and is what allows ultrasound to give us information about tissue character. Scattered sound is important for the generation of images from objects with large angles of incidence to the sound beam or small structure-like cells.

Attenuation

Sound traveling through a medium is weakened by reflection, refraction, scattering, and absorption of heat by the tissues. This loss of energy is called attenuation. High frequency sound attenuates to a greater degree than lower frequency sound because its wavelength allows it to interact with more structures. This is the reason the deep bass sounds of an orchestra carry farther than the high-pitched sounds. The large degree of attenuation with high frequency sound leaves less energy available for continued transmission through the medium.

↑ Frequency = ↓ Depth
↓ Frequency = ↑ Depth

The half-power distance of a tissue is the distance sound will travel through it before half of the available sound energy has been attenuated. Table 1.4 lists the half-power distances of various tissues at two different frequencies. The data in this table clearly show that low frequency sound waves are able to penetrate tissues deeper than higher frequency sound waves.

Air attenuates half of the sound energy within 0.05 centimeter (cm) when a 2.0-MHz transducer is used.

Table 1.4
Half-Power Distances of Various Tissues

Tissue	Distance (cm)	
	2.0 MHz	5.0 MHz
blood	15	3
soft tissue	1.5	.5
muscle	.75	.3
bone	.1	.04
air	.05	.01

Therefore, although the density of air creates less impedance for sound, little sound energy is left for image generation from soft tissues after 0.05 cm. Gel is used to eliminate the air between the transducer and skin, which would otherwise attenuate sound dramatically.

Tissue Harmonic Imaging

When ultrasound is transmitted at one frequency and returned at twice or more the transmitted frequency, it is called tissue harmonic imaging. Sound waves change from their sinusoidal shape as they travel through tissues to nonsinusoidal waves. This is caused by changes in pressure, with the higher pressure portion of the sound waves traveling faster than the slower portions. These nonsinusoidal waves contain additional frequencies in multiples of the fundamental or originating frequency. These even and odd multiples of the fundamental frequency are called harmonic frequencies. When using harmonic imaging the fundamental frequency is filtered out and second harmonic waves are used to generate the ultrasound image. This imaging mode is used to enhance the definition of endocardial borders and reduces the generation of artifacts, especially in patients with poor acoustic windows. Poor image quality is usually the result of factors like fat, muscle, and fibrosis that are present before the sound beams have even entered the tissue of interest. These factors create variations in the speed of sound and create distortion of the sound beam and the resulting ultrasound image. The harmonic frequencies are created in the chest from the reflected sound and not at the chest wall where most artifacts originate; this results in the alleviation of imaging artifacts especially side lobe artifact. It also enhances contrast resolution of the ultrasound image. The result is improved image quality with reduced artifact generation, enhanced endocardial details, improved contrast, and decreased noise. There are some patients in which harmonic imaging does not

improve image quality because of frequency dependent attenuation of sound.

TRANSDUCERS AND RESOLUTION

Transducers are the source of sound in diagnostic ultrasound. Transducers contain piezoelectric crystals that are deformed by electrical voltage and generate sound. These crystals, often called elements, are also able to receive sound and convert it back into electrical energy. The thickness of the crystal dictates the basic operating frequency of the transducer. Wavelength will be one-half of the element thickness so decreased crystal thickness produces shorter wavelengths and higher frequencies.

Pulse Repetition Frequency

Transducers used in pulsed-echo applications do not transmit sound continuously. They send sound waves out in short bursts and receive sound the remainder of the time. This is called pulsed ultrasound. The number of pulses per second is referred to as the pulse repetition frequency (PRF). PRF is measured in Hz. The PRF for example would be 10 Hz if there are 10 pulses per second (Figure 1.10). Each pulse may have any number of cycles, but in diagnostic ultrasound, there are generally two or three cycles per pulse. The number of cycles per pulse is controlled by damping materials within the transducer.

The duration of a pulse, measured in microseconds, and pulse length, measured in mm, decreases if the frequency of the sound wave increases since the wavelengths are shorter (Figure 1.10). By the same token, lower frequency sound waves have longer wavelengths so pulse duration and length are increased. Accurate ultrasound images can only be generated if all reflected and scattered echoes are received at the transducer before the next pulse is generated. The transducer assumes that the echoes it receives are products of its last burst. If an echo has not been received before the next burst and it arrives at the transducer shortly after the second burst, then the instrument “thinks” very little time has elapsed since it was transmitted and received. Since time is used along with the speed of sound in tissues to determine structure depth, a structure that is actually deeper will be displayed closer to the body surface (Figure 1.11). Pulse repetition frequency must decrease as deeper structures are imaged for accurate depth assessment.

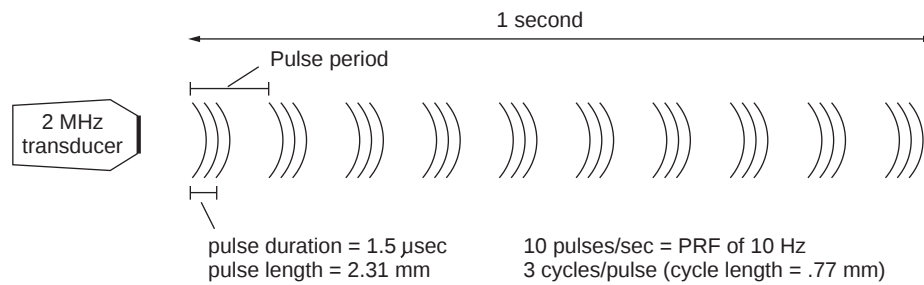
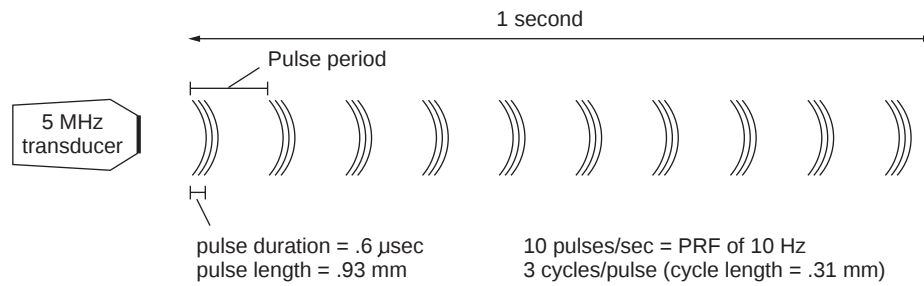
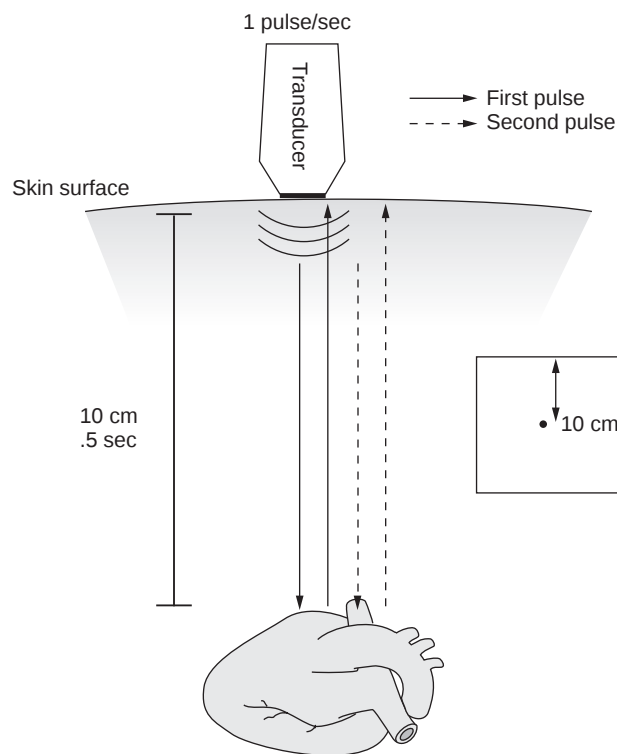
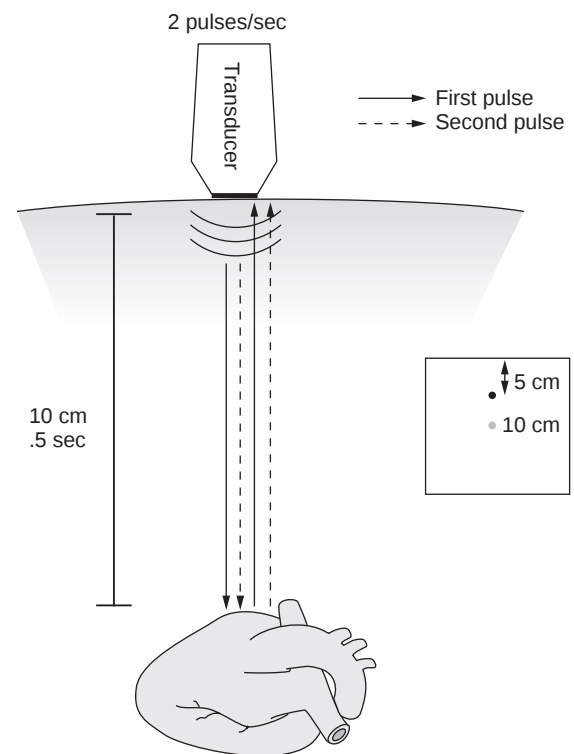


Figure 1.10 Transducers send out sound waves in short bursts called pulses. The number of pulses per second is called the pulse repetition frequency (PRF), measured in Hz. Each pulse has a duration based on the wavelength and number of cycles.



Sound from first pulse is received before next pulse is sent out. Structure is accurately plotted at 10 cm.



First pulse reflected at same time second pulse is sent out; received 0.5 seconds after second pulse. Structure incorrectly plotted at 5 cm.

Figure 1.11 A sound wave must be transmitted, reflected, and received by the transducer before the next pulse is generated. The number of pulses per second is the pulse repetition frequency. Pulse repetition frequency must decrease for accurate structure localization when interrogating deeper structures.

Sound Beams

Sound beams generated by transducers are three-dimensional. They not only have pulse length and duration but they also have beam widths and thicknesses. Beam diameter determines the width within the scan plane and the thickness perpendicular to the scan plane.

Sound beams do not remain the same width as they travel through a medium. In an unfocused transducer the sound beam starts out with a width equal to the transducer diameter and, as it travels through the tissues, it diverges (Figure 1.12). The distance from the transducer element to where it diverges is the beam's near field. The area beyond the near field is the far field. Near field length is directly proportional to the beam diameter and inversely proportional to wavelength (Figure 1.12). For two transducers of the same frequency, the near field will be longer for the transducer with the larger diameter. For two transducers with the same diameter, the near field will be longer for the higher frequency transducer.

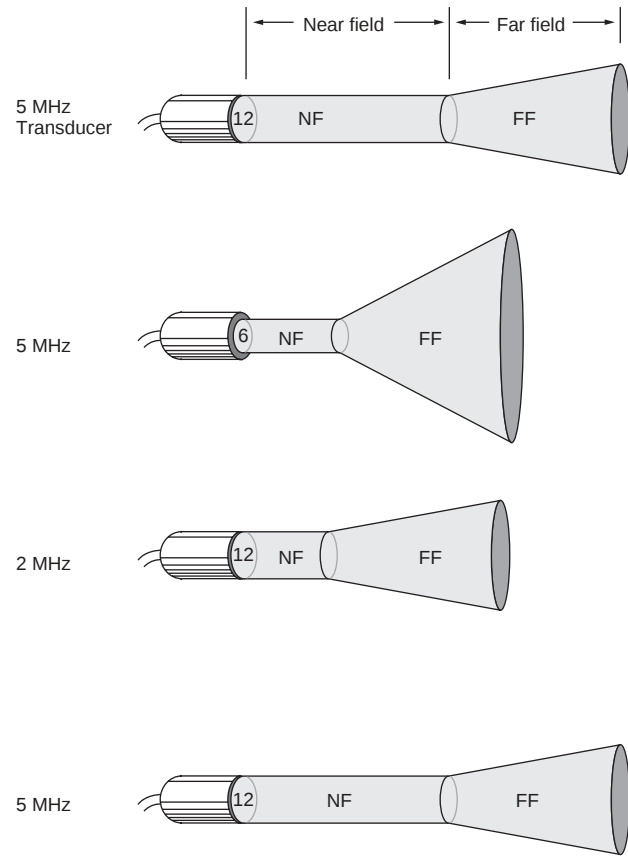
- Near field = $\text{radius}^2 / \text{wavelength}$
- Larger beam width = longer near field
- Shorter wavelength = longer near field

Far field divergence is also dependent upon transducer size. Larger diameter transducers produce less divergence in the far field. High frequency transducers with large diameters therefore produce the longest near field and the narrowest far field (Figure 1.12).

When a curved element or lens is used, the beam can be focused and beam width will decrease throughout the entire near field and create a focal zone, but beam width will diverge rapidly beyond this focal point (Figure 1.13). Many transducers today have variable focal zones that the examiner can set.

If multiple pulses are generated and each pulse is set to a different focal zone, then an elongated focal zone can be created. The transducer simply ignores echoes returning from depths other than the focal depth for any given pulse.

Up to this point only single sound beams have been considered. A single sound beam is used to generate an M-mode image of the heart. This beam travels through the cardiac structures and a one-dimensional image is generated. B-mode or two-dimensional imaging uses an array (group) of crystals that are electronically triggered to generate sound waves. It is important to recognize that each sound beam gener-



$$\text{Near Field} = r^2 / \text{wavelength}$$

∴ ↑ beam diameter = longer near field for transducers of the same frequency

∴ ↑ frequency = longer near field for transducers of the same frequency

∴ ↑ diameter = less far field divergence for same frequency

Figure 1.12 Sound beams have a diameter equal to transducer diameter and diverge as they travel out through a tissue. The distance from a transducer element to where the beam diverges is referred to as the near field. The area beyond that is the far field.

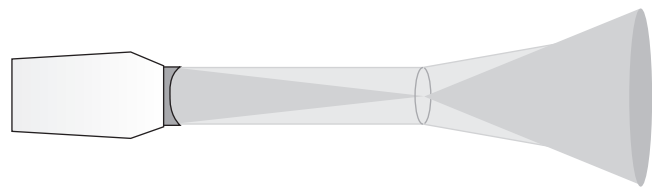


Figure 1.13 A sound beam can be focused by using a curved element or lens. This decreases beam width within the near field.

ated by a transducer is affected by pulse length, beam width, focal length, and PRF.

Linear array transducers have multiple elements arranged in a row. Sequences of elements are electronically stimulated at one time (i.e., elements one through four, then elements two through five, etc.) with each

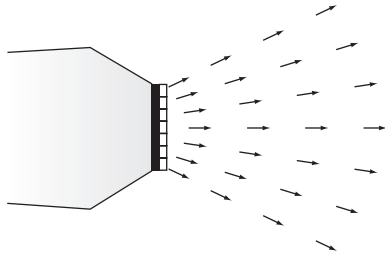


Figure 1.14 Phased array transducers have elements that are stimulated in sequence creating a slightly different angle of transmission through the tissues. This produces a rapidly moving two-dimensional sector image.

group producing one scan line. This produces a high quality image with increased line density within the generated image. Linear array transducers can be modified into curvilinear formats. Phased array transducers stimulate each crystal with a small time interval (less than a microsecond) between them and they are directed through the tissues at slightly different angles (phased) (Figure 1.14). This produces a sector image and these transducers are often called electronic sector transducers. Rapidly stimulating these elements over and over again in sequence produces the moving cardiac images we call real-time ultrasound.

Axial Resolution

Resolution is the ability to identify two objects as different. Pulse length, beam width, beam diameter, focal length, and PRF are important physical aspects of transducers that affect the axial, lateral, and temporal resolution of ultrasound images.

Resolution

- Axial
 - Ability to differentiate between two structures along the length of the sound beam
- Lateral
 - Ability to resolve two structures in the plane perpendicular to the sound beam
- Temporal
 - Ability to resolve structures with respect to time, keeping up with the actual events

Axial resolution is the ability to differentiate between two structures along the length of the sound beam. Axial resolution is also called depth or longitudinal resolution. The smaller the axial resolution is, the better the detail of the image.

Transducer frequency plays an important role in axial resolution. Axial resolution is equal to half the pulse length, that is, two structures cannot be closer than half the pulse length to each other in order to be distinguished as two separate things. Remember that pulse length depends upon the wavelength of the sound and upon the number of cycles per pulse. When one or both of these is reduced axial resolution improves (Figure 1.15). Wavelength decreases as frequency of sound increases, so axial resolution is better with 7.5-MHz frequency sound than with 3.5-MHz frequency sound. Pulse length and duration are shortened by adding damping materials within the transducer or electrical damping within the equipment.

Pulse Length

- A pulse may have any number of cycles (generally two to three in echocardiography).
- Pulse length decreases with higher frequency sound because of shorter wavelengths and increases with lower frequency sound.

Axial Resolution

- Better axial resolution = better image detail
- Equal to half the pulse length
- Higher frequency transducers have better axial resolution.

↑ Frequency = ↑ Resolution
↓ Frequency = ↓ Resolution

Lateral Resolution

Lateral resolution is the ability to resolve two structures as distinct and different in a plane perpendicular to the sound wave. Lateral resolution is equal to beam width and improves with smaller beam widths. Beam width is affected by 1) focusing the sound waves generated by a transducer, 2) transducer diameter, and 3) transducer frequency.

The narrower the beam width the better the ability to differentiate between two structures in a plane perpendicular to the sound beam (Figure 1.16). Beam width varies along the length of the sound wave but is at its narrowest at the focal zone in focused transducers. Lateral resolution is best (smallest) at the focal zone. Two structures that are side by side within the

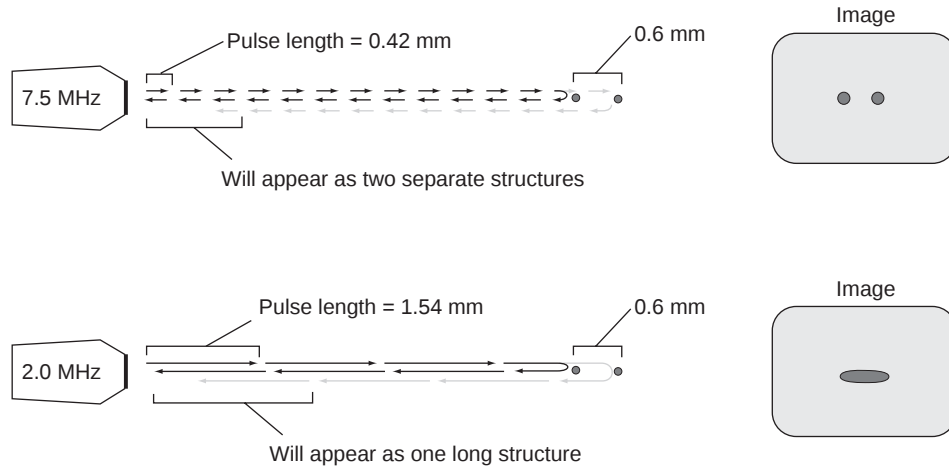


Figure 1.15 Axial resolution improves with increased frequency and decreased pulse length. Two things must be farther apart than one-half the pulse length to be identified as two different structures.

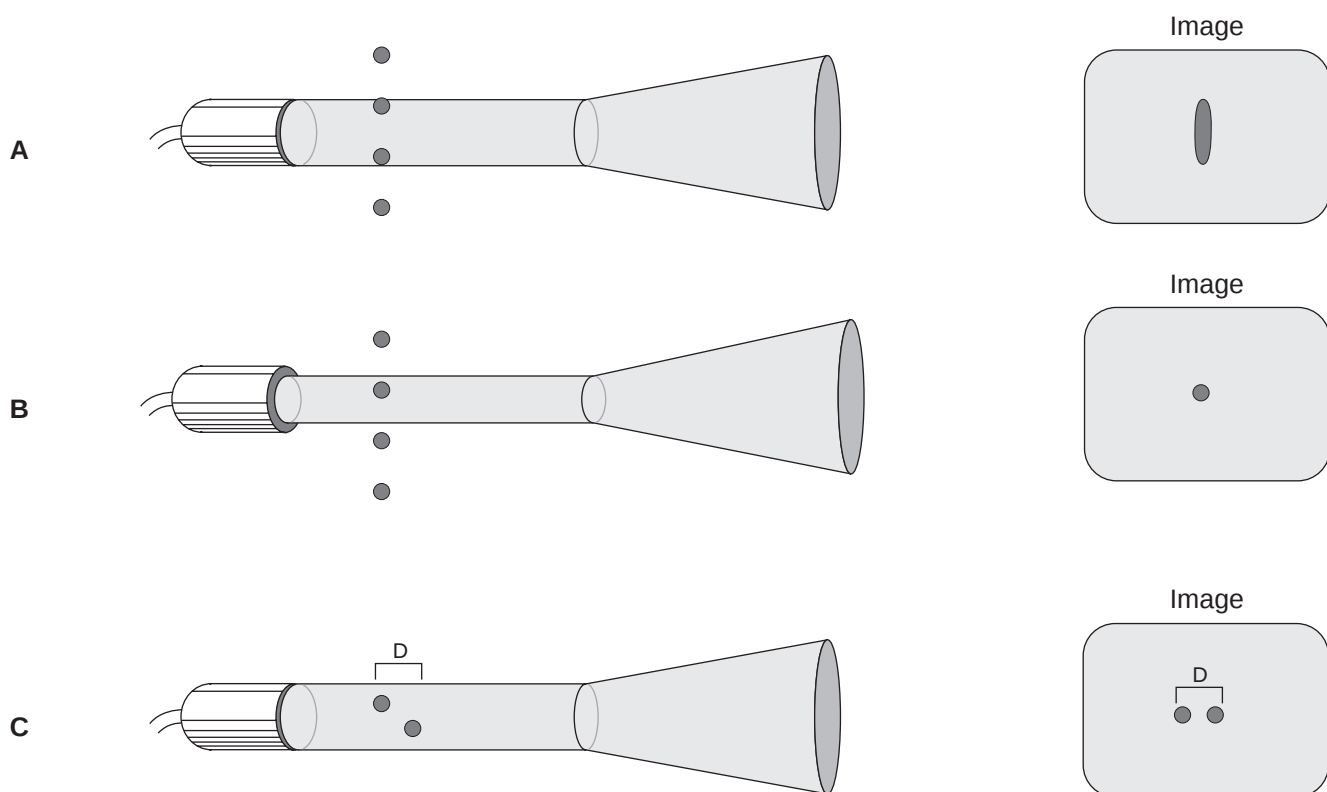


Figure 1.16 The ability to resolve two structures as different in a plane perpendicular (lateral resolution) to the sound beam depends upon beam width. (A) Two structures that fall within beam width will not be differentiated (B) while two structures that are farther apart than the beam width will be identified as separate. (C) The axial resolving powers of the system may differentiate two structures that fall within beam width when they are offset in depth (D).

boundaries of the beam width will not be resolved as two different structures (Figure 1.16). If they are offset a little in depth however, they may be resolved as two different structures based upon axial resolving powers of the transducer (Figure 1.16).

Lateral Resolution

- Improves with narrower beam widths
- Usually narrowest at focal zones of focused transducers
- Best within near field where beam width is narrowest

Lateral resolution of an image is also best within the near field where beam width is narrowest. A high frequency transducer will have better lateral resolution than a lower frequency transducer of the same size because of its longer near field. Long narrow near fields allow more specific areas of the heart to be imaged, creating less ambiguity about the source of returning echoes (lateral position errors).

Longer near field length, focused transducer beams, and less far field divergence also improve image quality by increasing beam strength. Stronger beams increase the degree of reflection and can travel farther before all the sound is attenuated.

Temporal Resolution

The number of real-time images produced per minute is referred to as the frame rate and is dependent upon the PRF. The faster the frame rate, the faster the pulse repetition frequency (PRF). Faster frame rates produce better temporal resolution (resolution with respect to time). Logically, rapidly moving structures require fast frame rates in order to prevent slow motion or freeze frame images of cardiac motion.

Temporal Resolution

- Dependent upon frame rate
 - Reduce sector width in order to improve the frame rate
 - Reduce image depth

Sector transducers that emit multiple pulses with varying focal zones per scan line must wait until all sound has returned before generating the next set of pulses otherwise range or depth ambiguity

results. In doing so the frame rate and temporal resolution of the generated two-dimensional image is reduced. Interrogation of deep structures also requires a slower frame rate and less temporal resolution is possible. It is possible to increase PRF in both of these settings by reducing sector width and/or image depth since less time is required before the next frame can be produced.

DOPPLER PHYSICS

Doppler has dramatically increased the diagnostic capabilities of cardiac ultrasound. This modality allows detection and analysis of moving blood cells or myocardium. It tells us about the direction, velocity, character, and timing of blood flow or muscle motion. The hemodynamic information provided by Doppler echocardiography allows definitive diagnosis in most cardiac examinations.

Doppler Ultrasound

Allows detection and analysis of moving blood cells or myocardium and provides hemodynamic information about:

- Direction
- Velocity
- Character
- Timing

Four types of Doppler used during an echocardiographic exam will be discussed in this text: pulsed-wave (PW) Doppler, continuous-wave (CW) Doppler, color-flow (CF) Doppler, and tissue Doppler imaging (TDI). Pulsed-wave Doppler is site specific. In other words it can be directed and set to sample flow at very specific places within the heart. It is, however, limited in its capacity to detect higher frequency (velocity) shifts. Continuous-wave Doppler has the ability to detect high frequency shifts and therefore can record high-flow velocities with virtually no limits. As you will see, since sound is continuously transmitted and received in CW Doppler, it is not possible to select and interrogate at specific depths within the heart. Although this may sound like a disadvantage, the information provided by CW Doppler is very valuable. Color-flow Doppler is a form of pulsed-wave Doppler. Frequency shifts are encoded with varying hues and intensities of color. Flow information is very vivid, and detection of abnormal flow is easier with color-flow Doppler although quantitative information is limited. Tissue

Doppler imaging uses pulsed-wave Doppler to interrogate myocardial motion and velocities. It is used to assess both systolic and diastolic myocardial function. These various forms of Doppler ultrasound and the factors that influence them are explained and discussed in the following sections.

Pulsed-Wave Doppler

- Allows flow to be examined at very specific sites.
- It is limited in the maximum velocity that it can accurately record.

Continuous-Wave Doppler

- There is no limit to the maximum velocity it can record.
- It is not site specific; blood cells are examined all along the sound beam.

Color-Flow Doppler

- This is a form of pulsed-wave Doppler.
- It color codes the various velocities and directions of flow.

Tissue-Doppler Imaging

- This is a form of pulsed-wave Doppler.
- It records myocardial velocity.
- It is used to assess systolic and diastolic function and synchronicity.

The Doppler Shift

Christian Johann Doppler (1803–1853), an Austrian physicist and mathematician, was the first to describe the Doppler effect. He found that all types of waves (light, sound, etc.) change in wavelength when there is a change in position between the source of the wave and the receiver of the wave. Using sound, if you were moving toward a sound source, the pitch or frequency of that sound would increase, and if you were moving away from that sound source, the frequency would decrease. The change in frequency between sound that is transmitted and sound that is received is the Doppler shift.

The Doppler Shift

There is a change in wavelength (pitch and frequency) when there is a change in position between the sound source and the reflecting structure (blood cells in this case).

When the source and the reflecting surface are both stationary, the transmitted (incident) and reflected wavelengths are equal (Figure 1.17). When the reflecting structure is moving toward the source, sound waves are encountered more often, resulting in an increased number of waves ($\uparrow$ frequency) being reflected back toward the source. When the reflecting structure is moving away from the source, they travel ahead of the transmitted wave front and sound waves are encountered less frequently resulting a decreased number of sound waves ($\downarrow$ frequency) reflected back to the source.

Cells moving toward the transducer reflect an increased number of sound waves, and so the received frequency is greater than the transmitted frequency. This is a positive frequency shift.

Cells moving away from the transducer reflect fewer sound waves, and the received frequency is less than the transmitted frequency. This is a negative frequency shift.

Everyday examples of Doppler shifts include any loud sound moving toward or away from you such as sirens, trains, marching bands, etc. The sound of a siren as it approaches you will increase in pitch (frequency increases) and then as it passes you, the pitch will decrease (frequency decreases). Doppler radar uses this principle when policemen determine the speed of your car, since, as you will see, the frequency shift is used to determine velocity. Doppler radar is also used in forecasting weather. The Doppler shift as we utilize it in diagnostic ultrasound is the difference in frequency transmitted by the transducer and received frequency reflected from blood cells.

The Doppler Tracing

The Doppler-derived frequency shift (f_d) is equal to reflected frequency minus transmitted frequency, therefore, objects moving toward the source result in positive frequency shifts while objects moving away

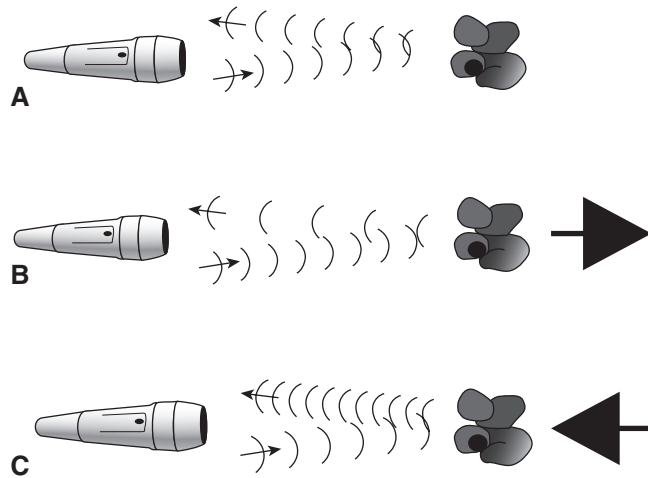


Figure 1.17 The change in frequency between sound that is sent out and sound that is reflected is called the Doppler shift. (A) Sound reflected from stationary blood cells will have the same frequency as the transmitted sound. (B) Reflected sound encounters the transmitted wave front less often and a decrease in frequency is perceived when blood cells move away from the transducer. (C) Sound reflected from blood cells moving toward the transducer will have a higher frequency than what was sent out because the reflected waves encounter the incident waves more often.

from the source result in negative frequency shifts. The site (gate) for Doppler flow interrogation is selected by the examiner and is represented on the Doppler display as a line (baseline). Positive frequency shifts (flow moving toward the transducer) produce waveforms up from the baseline while negative frequency shifts (flow moving away from the transducer) produce downward deflections on the Doppler tracing (Figure 1.18). These images are called spectral tracings. Velocity scale is displayed along the side of the spectral image. The velocity range is split between the positive and the negative directions of flow. When the baseline is located in the middle of the spectral display, the total velocity range is displayed equally above and below the baseline (Figure 1.19). When the baseline is moved all the way to the top of the image, the entire velocity range is allocated to downward flow. When it is moved to the bottom of the image, the entire velocity range is allocated to upward flow.

Pulsed-Wave Doppler

Pulsing the sound waves allows a transducer to act as a receiver for the signal only during the time interval

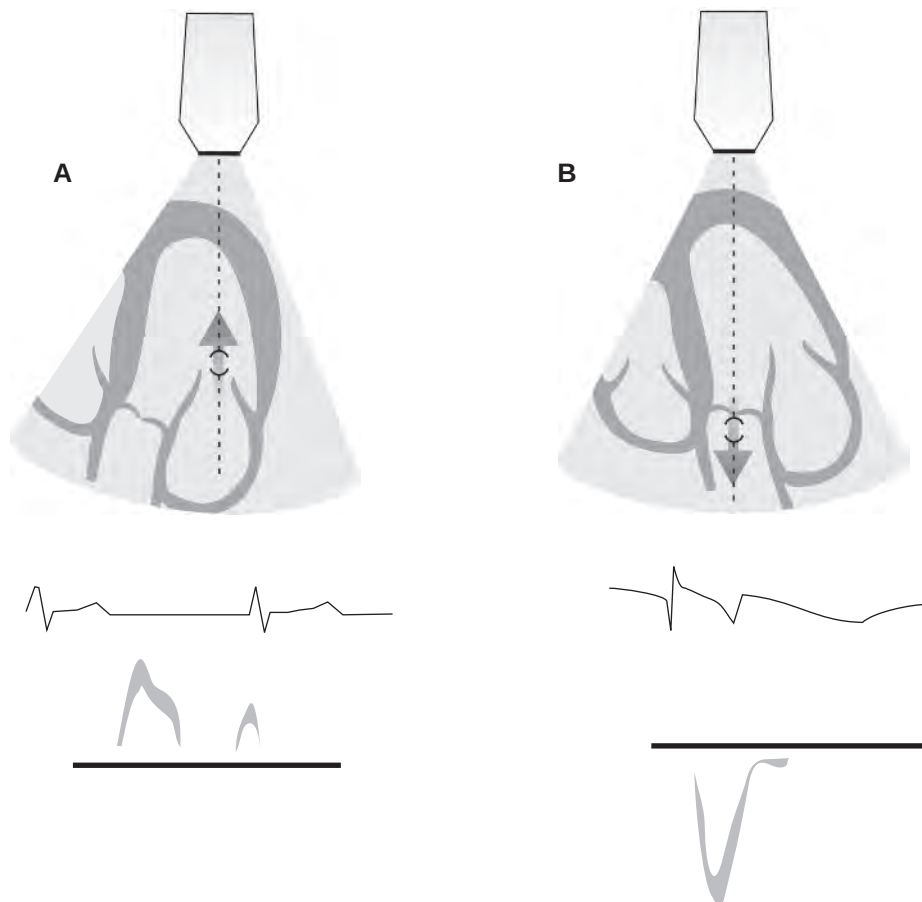


Figure 1.18 The baseline in pulsed-wave Doppler represents the sampling gate. Flow moving toward the transducer creates a positive frequency shift, and velocity will be plotted above the baseline. (A) Mitral valve flow in this apical five-chamber view is toward the transducer and its flow profile is seen above the baseline. (B) Aortic flow, moving away from the transducer in this apical five-chamber view, creates a negative frequency shift, and its flow profile is shown below the baseline.

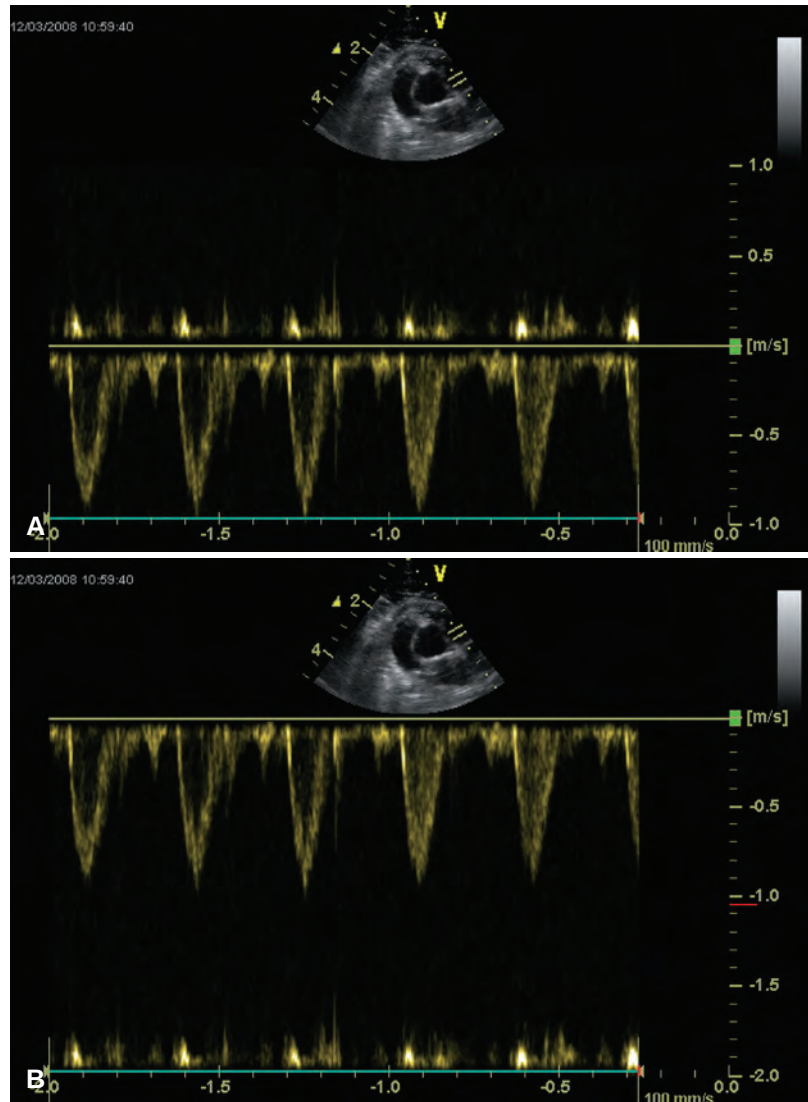


Figure 1.19 (A) When the baseline is positioned in the middle of the spectral display, the velocity range is displayed equally above (1.0 m/sec) and below the baseline (1.0 m/sec). (B) Moving the baseline to the top of the spectral display allocates all of the velocity range (2.0 m/sec) below the baseline.

specified by a sample depth. With pulsed-wave Doppler the transducer will record frequency shifts only during the time interval dictated by the depth of the sample site ignoring all other returning echoes (Figure 1.20). New sound waves will not be transmitted until the transducer has received the echoes from the previous burst. The ability to measure velocity within a small cell at a specified depth along the ultrasound beam is referred to as range resolution, and the site at which sampling is set to occur is referred to as the gate. The gate is manually set by the examiner while watching a two-dimensional image.

Continuous-Wave Doppler

Continuous-wave (CW) Doppler, as the name implies, continuously sends out sound and continuously receives sound. It is not possible to range gate CW

Doppler because the transducer has no way of detecting the depth of the reflected signal. CW Doppler detects frequency shifts all along the ultrasound beam with no range resolution. CW Doppler is steered in one of two ways. Imaging CW systems use a cursor representing the Doppler sound beam. The cursor is placed over the two-dimensional image and frequency shifts are calculated all along the beam. Non-imaging CW systems use a dedicated CW probe without the luxury of a two-dimensional image. These systems require recognition of characteristic flow profiles.

Velocities along the beam vary, and a full spectrum of frequency shifts is detected with CW Doppler. When CW Doppler is used properly, the highest velocities along the line of interrogation are recorded (Figure 1.21). The highest flow velocities are generally what is of interest and diagnostically important. Lower velocity flows found along the Doppler line of interrogation

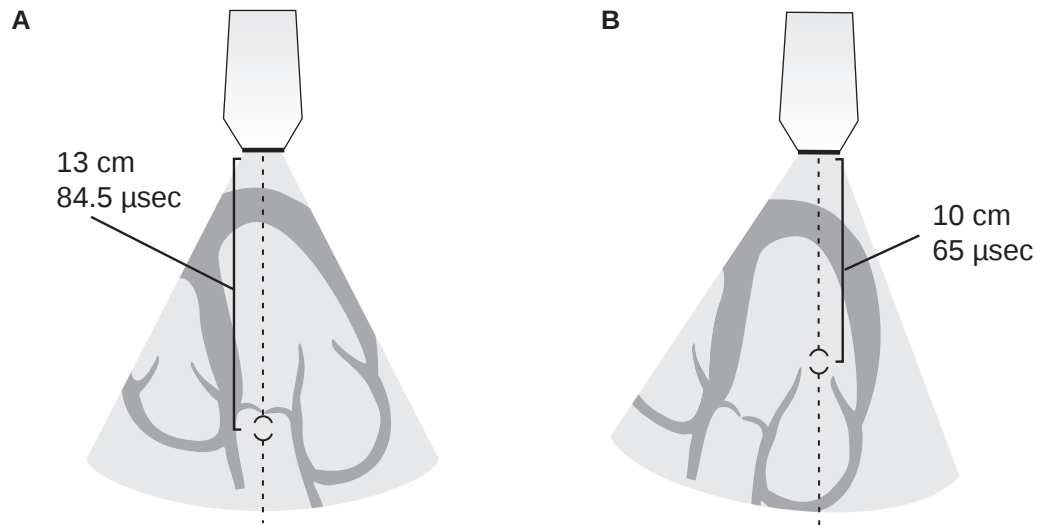


Figure 1.20 Frequency shifts are recorded only during the time interval indicated by the depth of the sample gate. Deeper gates require more time. (A) A gate depth of 13 cm requires 169 μ/sec while a gate of 10 cm (B) requires only 130 μ/sec for sound to return to the transducer. Lower pulse repetition frequency is required for deeper structures.

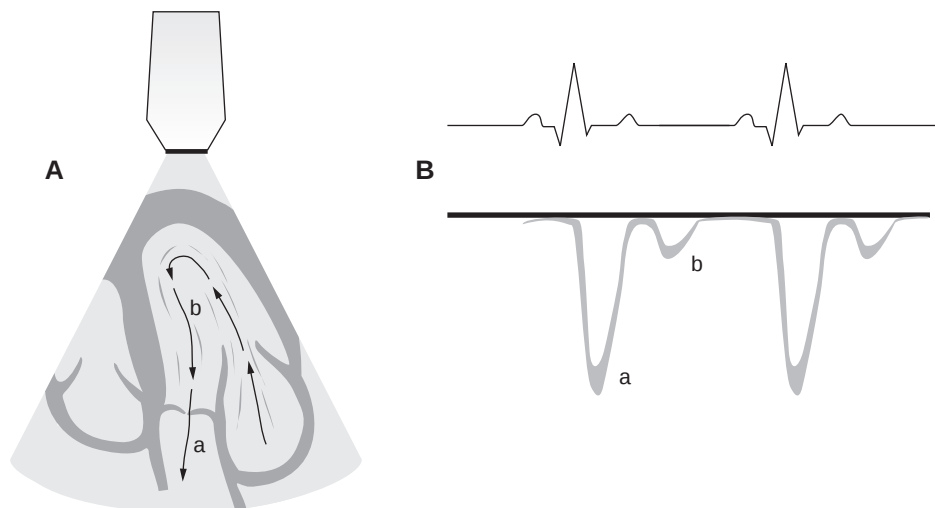


Figure 1.21 Continuous-wave Doppler detects frequency shifts all along the Doppler sound beam. All velocities are recorded. (A) The highest velocity during systole is flow out the aorta; lesser negative velocities during this time period are recorded but hidden within the aortic flow profile. (B) The CW sound beam also records flow during diastole near the apex of the heart.

are hidden within the higher flow profiles. Flow patterns for the various valves and vessels in the heart are very characteristic and usually are easily identified with both PW and CW Doppler.

The Doppler Equation

Doppler ultrasound can determine blood cell velocity within the heart or in peripheral vessels based upon the Doppler shift. Blood cell velocity (V) is determined using the following formula:

$$V = \frac{C \times f_d}{2(f_o) \times \cos \theta} \quad \text{Equation 1.3}$$

Equation 1.3 shows that V is equal to the speed of sound in tissues (C) times the frequency shift (f_d) in kHz, divided by the transmitting frequency of the transducer, f_o (2.5, 3.5, 5.0, etc.), times the cosine of θ , where θ is equal to the intercept angle of the ultrasound beam with respect to the blood flow.

The speed of sound in tissues is a constant (1,540 m/sec), leaving the interrogation angle, θ , and transducer

frequency as variables that can be controlled. Let's consider these two variables and how they affect the way a Doppler exam should be conducted and interpreted.

Velocity Measurement

Accurate measurements are affected by:

- Transducer frequency
- Intercept angle

Angle of Interrogation

An important part of the Doppler equation is the cosine of the intercept angle. The closer to parallel the transmitted wave is with the direction of blood flow being interrogated, the more accurate the velocity measurement (Figure 1.22).

When the Doppler equation is changed to calculate for the frequency shift, you can see that the cosine of the intercept angle directly affects the frequency shift (f_d).

$$C = \frac{V \times 2(f_o) \times \cos \theta}{f_d} \quad \text{Equation 1.4}$$

Since the speed of sound in tissues (C) and the transmitting frequency (f_o) are known, the calculated frequency shift and therefore the calculated flow velocity is directly dependent upon the cosine of the intercept angle. The cosine of 0° is one. The value

of cosine decreases as the angle of interrogation increases, and by the time an angle of 90° is reached, the cosine is zero. Table 1.5 lists the cosines for several angles. Larger intercept angles and cosines of less than one falsely decrease the recorded frequency shift of blood flow. Generally, interrogation angles greater than 15° – 20° are considered unacceptable. The graph in Figure 1.23 shows the relationship between the cosine of the angle of incidence with respect to blood flow and the calculated velocity for a 5.0MHz transducer.

Intercept Angle and Velocity Measurements

Velocity cannot be overestimated, just underestimated, when interrogation angles with respect to flow become larger than zero.

Table 1.5
Cosines of Selected Angles

Angle (θ)	Cosine θ
0	1.000
5	.996
10	.985
15	.966
20	.940
25	.906
30	.866
50	.643
75	.259
90	.000

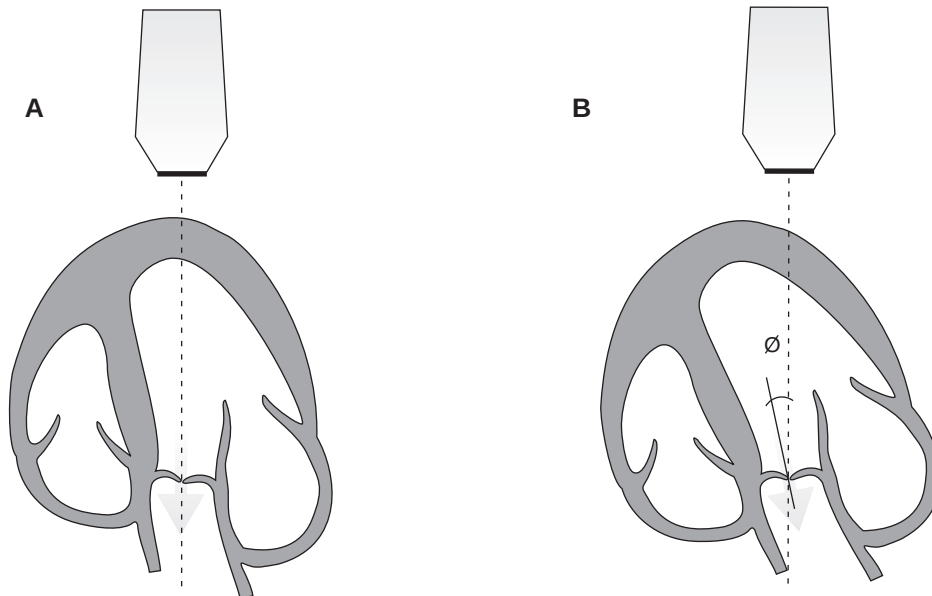


Figure 1.22 Frequency shifts are directly dependent upon the cosine of the interrogation angle (θ). (A) The closer to parallel the transmitted Doppler signal is to blood flow the more accurate the velocity will be since the cosine of 0° is one. (B) As the intercept angle, θ , deviates from zero, velocity will be underestimated.

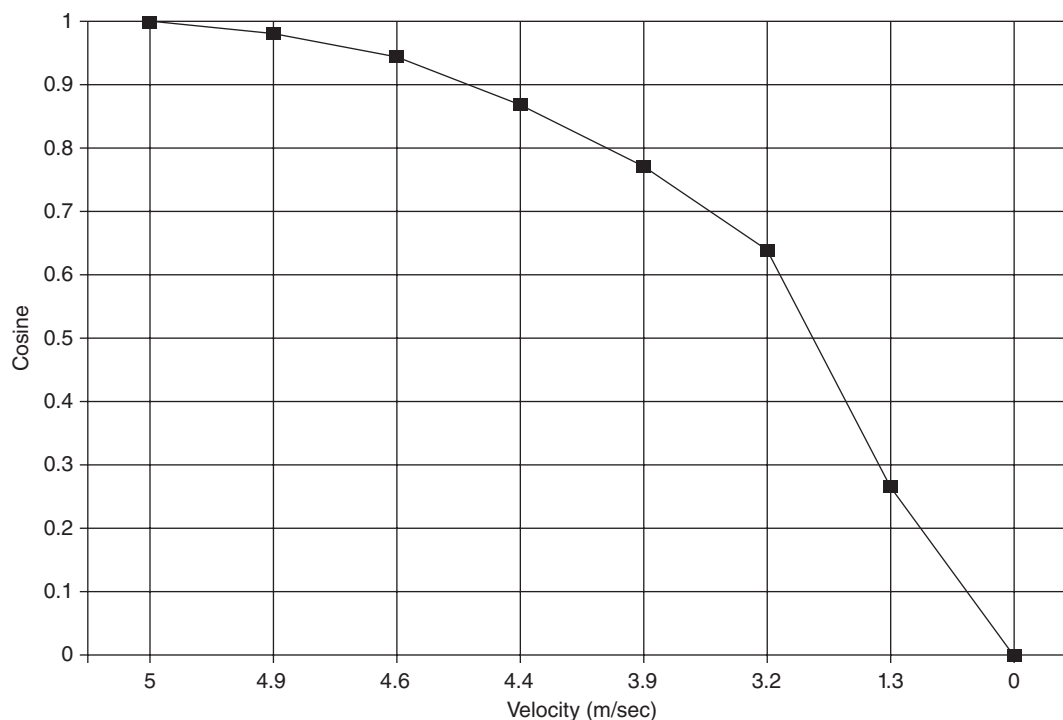


Figure 1.23 The effect of interrogation angle on maximum velocity is displayed on this graph. Cosine becomes smaller than one as intercept angles increase and falsely decrease recorded blood velocities. Angles greater than 15°–20° are considered unacceptable because they greatly underestimate the true velocity.

Effect of Transducer Frequency

Pulsed-wave (PW) Doppler measures the frequency shift at very specific locations within the heart. Just like two-dimensional and M-mode imaging, the reflected signal must be received before the next pulse is transmitted or there will be ambiguity in the recorded signals. The time interval between pulses must be two times the sample depth and is also referred to as the pulse repetition frequency (PRF). The time between pulses must increase as sample depth increases resulting in decreased PRF. Decreased PRF decreases the Doppler frequency shift that can be accurately measured. Figure 1.24 shows how sampling frequency affects your perception of events. As the sampling frequency decreases, information is lost. Time on the clock in Figure 1.24 is perceived correctly until the sampling frequency decreases to two times per minute. At that rate it is not possible to determine whether the hand on the clock is moving clockwise or counterclockwise. At an even lower PRF of three times every 2 minutes, the hand seems to be moving counterclockwise. This is similar to what happens in movies when wheels on vehicles appear to rotate backwards. The sampling rate (PRF) must be at least two times the frequency shift for unambiguous flow information to

be received by the transducer. Equation 1.5 states the maximum Doppler shift that can be recorded accurately is equal to one-half the PRF.

$$\text{Doppler shift} = 1/2 \text{ PRF} \quad \text{Equation 1.5}$$

Transducer Frequency and Velocity Measurement

The best Doppler recordings at any given depth are obtained with lower frequency transducers.

One-half of the PRF is referred to as the Nyquist limit. When the Nyquist limit is exceeded signal ambiguity results. This ambiguity is called aliasing. Figure 1.25 shows an aliased Doppler display. When the Nyquist limit is barely exceeded, the flow profile merely wraps around the image. This can be corrected by moving the baseline up or down on the monitor, allowing the entire profile to be recorded accurately. When the Nyquist limit is exceeded by larger degrees, the aliased signal no longer displays the characteristic flow profile and direction can no longer be determined (Figure 1.26). Switching to CW Doppler allows flows velocities that exceed the Nyquist limit to be recorded

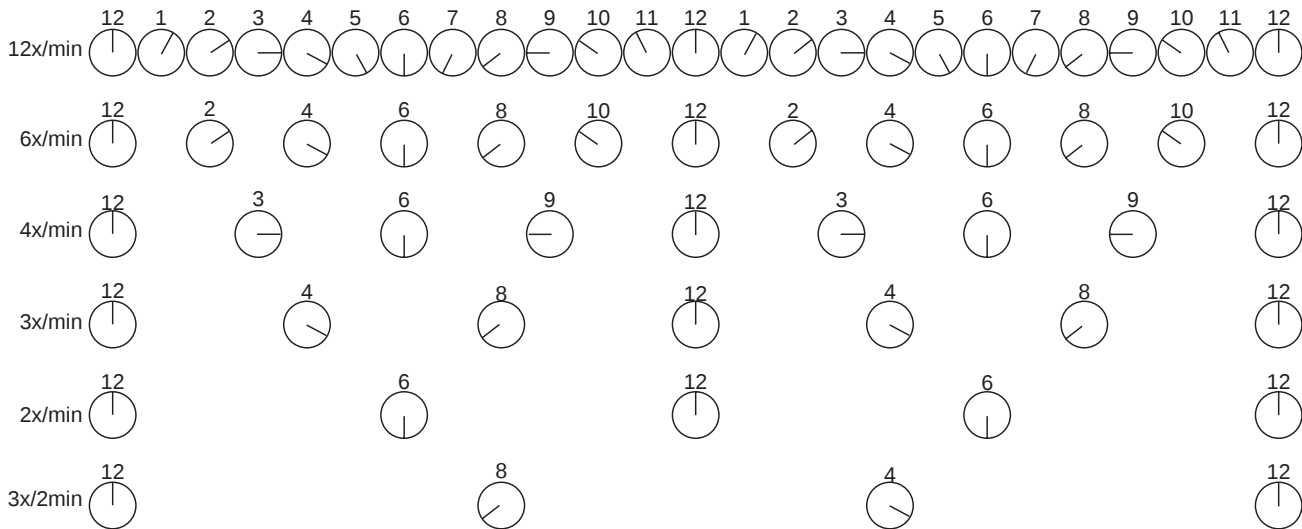


Figure 1.24 When sampling frequency (PRF) is not enough, the information obtained may not accurately represent what is occurring. In this diagram the top line has a sampling frequency of 12 times per minute, and the time is accurately displayed. As sampling frequency decreases to 2 times per minute, it is not possible to decide if time is moving clockwise or counterclockwise. With an even lower sampling frequency of 3 times every 2 minutes, the events are erroneously perceived as moving counterclockwise. This is similar to what happens in PW Doppler when sampling is not rapid enough creating an aliased signal. This creates ambiguity in the perceived direction of flow.

Table 1.6

Maximum Velocities Detectable at Specific Gate Depths for Several Transducer Frequencies

Freq	V at 7 cm	V at 12 cm	V at 20 cm
7.5	.56	.33	.20
5.0	.85	.49	.30
3.5	1.21	.71	.42
2.5	1.70	.99	.59

V = velocity in meters/sec

accurately. Equation 1.6 is used to determine the maximum velocity a pulsed-wave system can record accurately without aliasing for a given transducer frequency (f_0) and sampling depth (D).

$$V_{\max} = \frac{(1,540 \text{ m/sec})^2}{8(f_0) \times D} \quad \text{Equation 1.6}$$

The equation shows that the maximum velocity that can be recorded at any given depth with no ambiguity is inversely proportional to transducer frequency.

The best recordings of higher velocity jets at any given depth are obtained from a lower frequency transducer. This is opposite of what produces the best M-mode and two-dimensional exams where higher frequency transducers produce the best images. Table 1.6 lists the maximum velocities that can be accurately

Gate Depth and Velocity Measurement

For any given transducer frequency, the less the gate depth the higher the velocity that can be measured.

recorded at a variety of depths and transducer frequencies.

Effect of Sampling (Gate) Depth

Equation 1.6 also shows that the maximum velocity that can be recorded without aliasing is inversely proportional to depth for any given transducer frequency. The Nyquist limit is exceeded far sooner at deeper gates for a given interrogation frequency. Table 1.6 lists the maximum velocity that can be accurately recorded for a given transducer frequency at varying depths.

What To Do About Aliasing

- Move the baseline up or down.
- Find an imaging plane where less depth is necessary.
- Use a lower transducer frequency.
- Switch to CW Doppler.

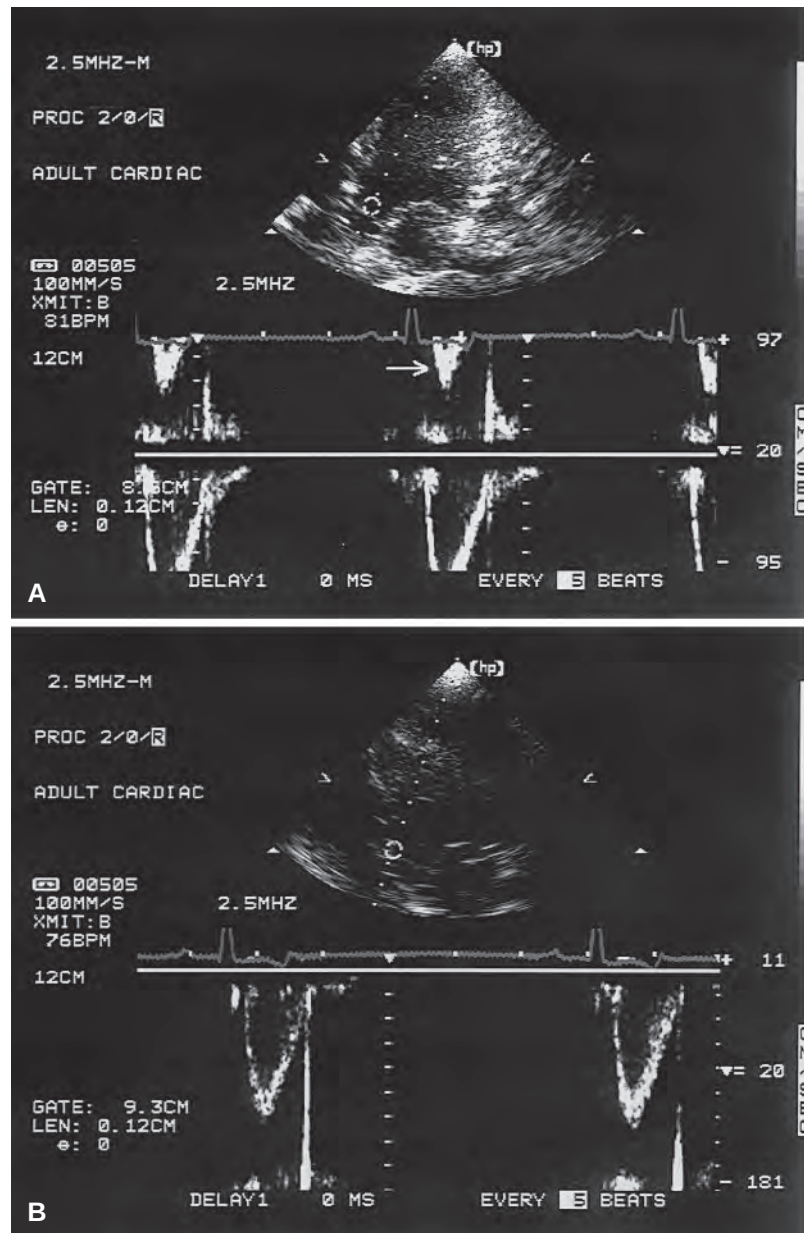


Figure 1.25 When the Nyquist limit is exceeded, aliasing occurs. (A) When the Nyquist limit is not exceeded by a great degree, we can still see the typical flow profile; however, it wraps around the image. (B) This type of aliasing can be eliminated by moving the baseline up or down in order to see the entire flow profile.

Blood Flow

Normal blood flow is typically laminar. All blood cells within a vessel, outflow tract, or chamber are moving in the same direction with very similar flow velocities. Vessel and chamber walls do create friction for the blood cells moving adjacent to their surface and velocities are generally somewhat slower along the periphery of the flow stream than in the center of the flow stream. Nevertheless velocities are similar

enough that a velocity profile is produced that has little variance.

Pulsed-wave Doppler always appears hollow with little spectral broadening when the following occurs: flow is laminar, intercept angles are close to zero, and the Nyquist limit is not exceeded (Figure 1.27). Spectral broadening is the filling in of the typically hollow waveform (Figure 1.28). Spectral broadening in a pulsed-wave signal may be due to improper gain settings, a large intercept angle, or non-laminar (turbulent) flow.

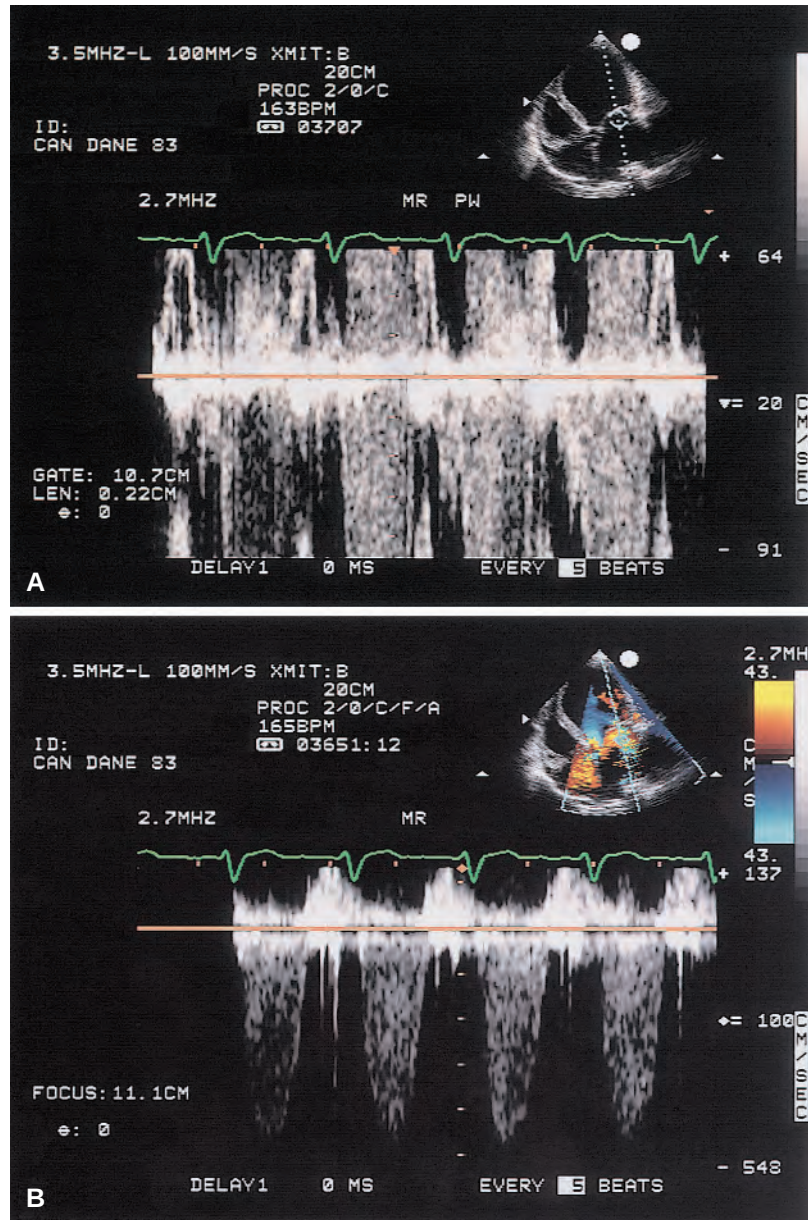


Figure 1.26 (A) When velocities dramatically exceed the Nyquist limit, normal flow profiles are lost and it is impossible to determine flow direction or velocity. (B) Switching to CW Doppler allows the high velocity flow of mitral regurgitation to be recorded accurately. PW = pulsed-wave Doppler, MR = mitral regurgitation.

Spectral Broadening in a PW Signal

- Due to improper gain settings
- Due to large intercept angle
- Due to nonlaminar (turbulent) flow

When flow becomes abnormal it is generally turbulent. Turbulent flow has blood cells moving in many directions and at variable velocities. This kind of flow is seen with stenotic lesions, shunts, and valvular regurgitation. Doppler signals produced from turbulent flow have a lot of spectral broadening because of

the many velocities and flow directions present in the jet. Continuous-wave Doppler always shows spectral broadening even when flow is laminar because flow velocities detected all along the transmitted sound beam vary tremendously (Figure 1.28).

Color-Flow Doppler

Color-flow Doppler is a form of pulsed-wave Doppler. Real-time images and color-flow mapping are done at the same time with alternating scan lines dedicated toward real-time image generation and Doppler signals.

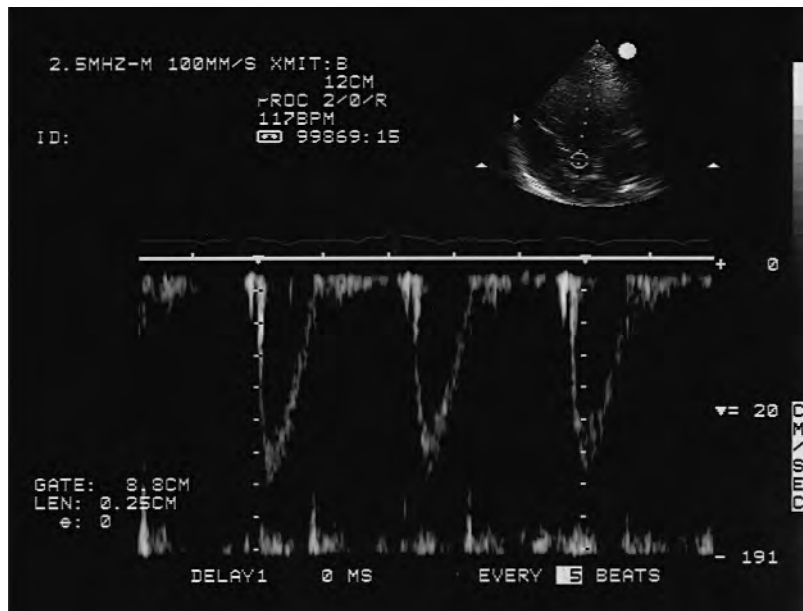


Figure 1.27 Pulsed-wave Doppler signals are hollow when flow is laminar because there is little variance in velocity.



Figure 1.28 Spectral broadening is the filling in of a Doppler flow profile. Continuous-wave (CW) Doppler displays always show spectral broadening because of the many velocities detected along the CW sound beam. Pulsed-wave Doppler may show spectral broadening when gain is too high, when intercept angles are large, or when flow becomes turbulent.

Remember that pulsed-wave Doppler is range gated in that a specific sampling site is chosen and the ultrasound machine ignores signals that come back from any other point along the line of interrogation. This can be done by knowing the speed of sound in tissues and the depth of the gate. Color-flow mapping involves the analysis of information all along hundreds of interrogation lines, each with hundreds of gates, until a wedge is filled with color. Each gate sends frequency shift information back to the transducer (Figure 1.29). This frequency shift information is sent to a processor,

which calculates the mean velocity, direction, and location of blood cells at each gate. Information from each gate is assigned a color and position on the image.

Blood flow in color mapping is perceived by the machine as either moving toward the transducer or away from it via a negative or positive frequency shift. By convention, flow moving toward the sound source is plotted in hues of red, and flow moving away from the transducer is mapped in shades of blue although this can be changed by the operator (Figure 1.30). No flow generates no frequency shift, and no color is

assigned. Enhanced color maps, available in most equipment, display flow velocity information as well as direction. Colors range from deep red for slow flow to bright yellow for rapid blood flow toward the transducer. Slow blood flow away from the transducer is mapped in deep blue colors while more rapid flow

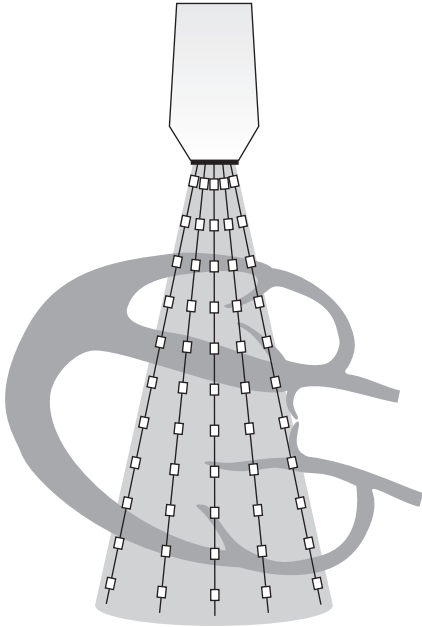


Figure 1.29 Color-flow Doppler involves a sector filled with many lines of interrogation. Each line of sound contains a multitude of gates, each of which send frequency information back to the transducer. Color is then assigned to each gate based on direction and velocity of flow.

away from the transducer is mapped in shades of light blue and white.

Color-flow Doppler quality is dependent upon two important factors: pulse repetition frequency and frequency of the transducer. As with spectral Doppler the frequency of the sound source dictates the maximal velocity, which can be accurately mapped at any given depth before aliasing occurs. Aliasing in color-flow Doppler involves a reversal of color and the result is a mosaic or mixing of the blue and red hues (Figure 1.31). Aliasing can occur while using high frequency transducers when in actuality there is normal flow and the aliasing is only a function of transducer frequency. The aliasing would be eliminated if a lower frequency transducer were used.

Color-Flow Doppler

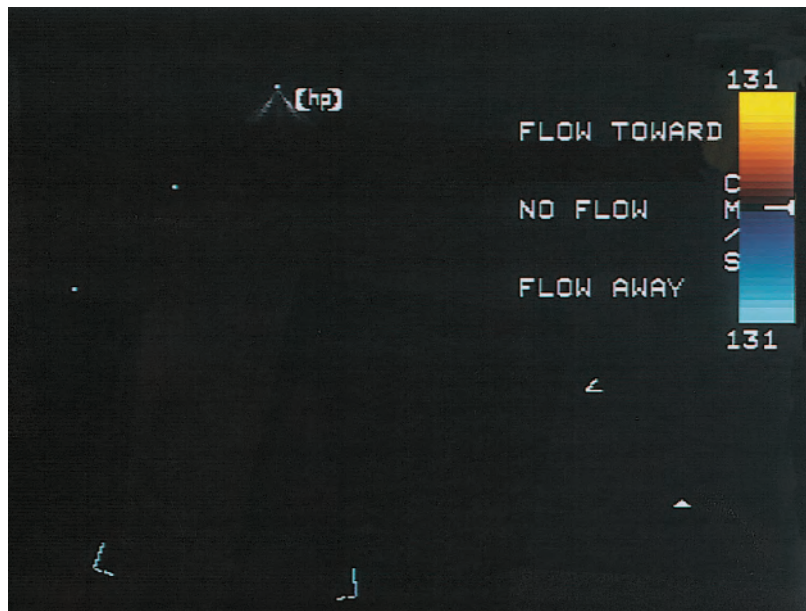
Aliasing occurs at lower velocities due to sampling time requirements.

- Therefore aliasing may be seen even when flow is normal.

Variance maps are found in many ultrasound machines (Figure 1.32). These machines map turbulent flow in hues other than blue or red, typically green. All color-flow images reproduced in this book use enhanced and variance color maps.

Frame rate refers to the number of times a B-mode or color-flow image is generated per minute. A frame rate of at least 15 times per minute is required for

Figure 1.30 The conventional color-flow map is blue away red toward (BART map). In this enhanced map, blood flow moving toward the transducer will be mapped in hues of red and yellow, in which deep red represents slower flow and bright yellow represents faster flow. Blood moving away from the transducer is mapped in hues of blue to white, in which deep blue represents slower flow and bright white represents faster flow. No flow or flow that is perpendicular to the interrogation line has no color assigned and will appear black.



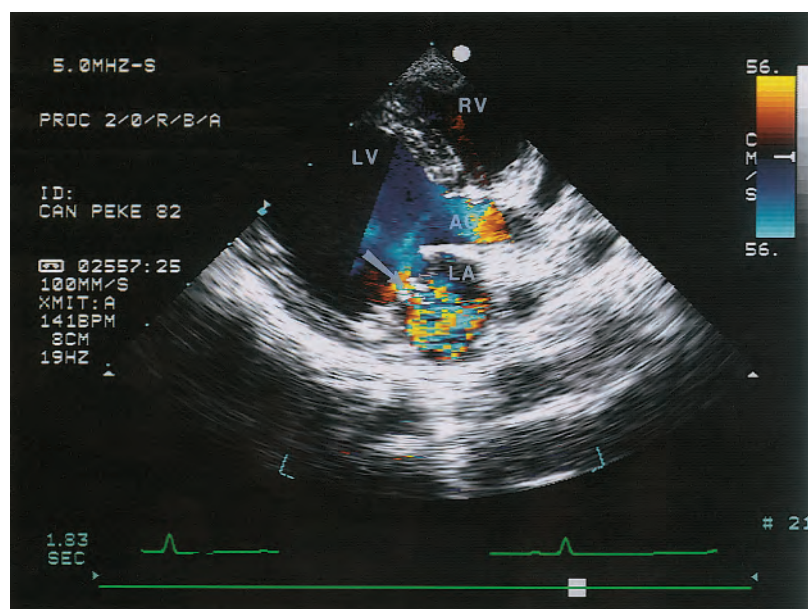


Figure 1.31 Aliasing secondary to high velocities or turbulent flow is displayed as a mosaic of color in color-flow Doppler. The turbulent flow of mitral regurgitation is shown as a multiple colored jet (*arrow*) within the left atrium (LA) on this parasternal long-axis left ventricular outflow view through the heart. RV = right ventricle, LV = left ventricle, AO = aorta.

smooth transitions and the appearance of a continuously moving image. Color-flow information is superimposed upon a two-dimensional image as a sector. Frame rate in color-flow Doppler is equal to PRF divided by scan lines per color sector. The width of this sector can be altered by the operator. Decreasing the color wedge decreases the amount of time necessary for sampling and increases the frame rate (Figure 1.33). The operator can also eliminate the real time image, which extends beyond the width of the color sector. This also decreases the time necessary for image generation and enhances color-flow mapping.

Many machines allow the operator to decrease the depth of the color wedge. This typically has no effect on frame rate on most ultrasound machines since total image depth is unchanged. It merely decreases the information the mind has to process by eliminating processed information from the display.

The number of times a line of sound is sampled is referred to as its packet size (Figure 1.34). Increasing packet size improves image quality and fills in the color display, but this is at the expense of frame rate. Packet sizes can be selected by the operator on some equipment. Decreasing packet size will increase your frame rate but decrease sampling time. Information may be lost with very short sampling times. This may be necessary however with rapid heart rates. Increasing packet size will increase the time required for sampling and decrease the frame rate, but it will be able to map velocities and color with greater color filling.

Optimize CF Imaging

- Decrease transducer frequency.
- Decrease color sector width.
- Eliminate real-time image.
- Increase packet size.
 - Decreases frame rate however.
- Decrease packet size.
 - Decreases sampling time and good for high heart rates but may lose information.

Tissue Doppler Imaging

Tissue Doppler imaging (TDI) or tissue Doppler echocardiography (TDE) involves acquiring myocardial velocities. While blood cells reflect low amplitude signals at high velocity, myocardial motion has high-amplitude signals but low velocity. Standard Doppler interrogation of blood flow filters out low velocity signals. TDI however bypasses the low velocity filter. TDI can employ pulsed-wave signals only or can be used in conjunction with color-flow Doppler. Color TDI uses a narrow sector of color (to keep frame rates high) placed over a section of myocardium. A pulsed-wave gate can be placed anywhere over the color sector after the fact from stored video loops (Figure 1.35). When using pulsed-wave Doppler TDI, the spectral gate is placed over a color Doppler sector in the area of interest on the myocardium and a spectral trace of myocardial motion is displayed in real

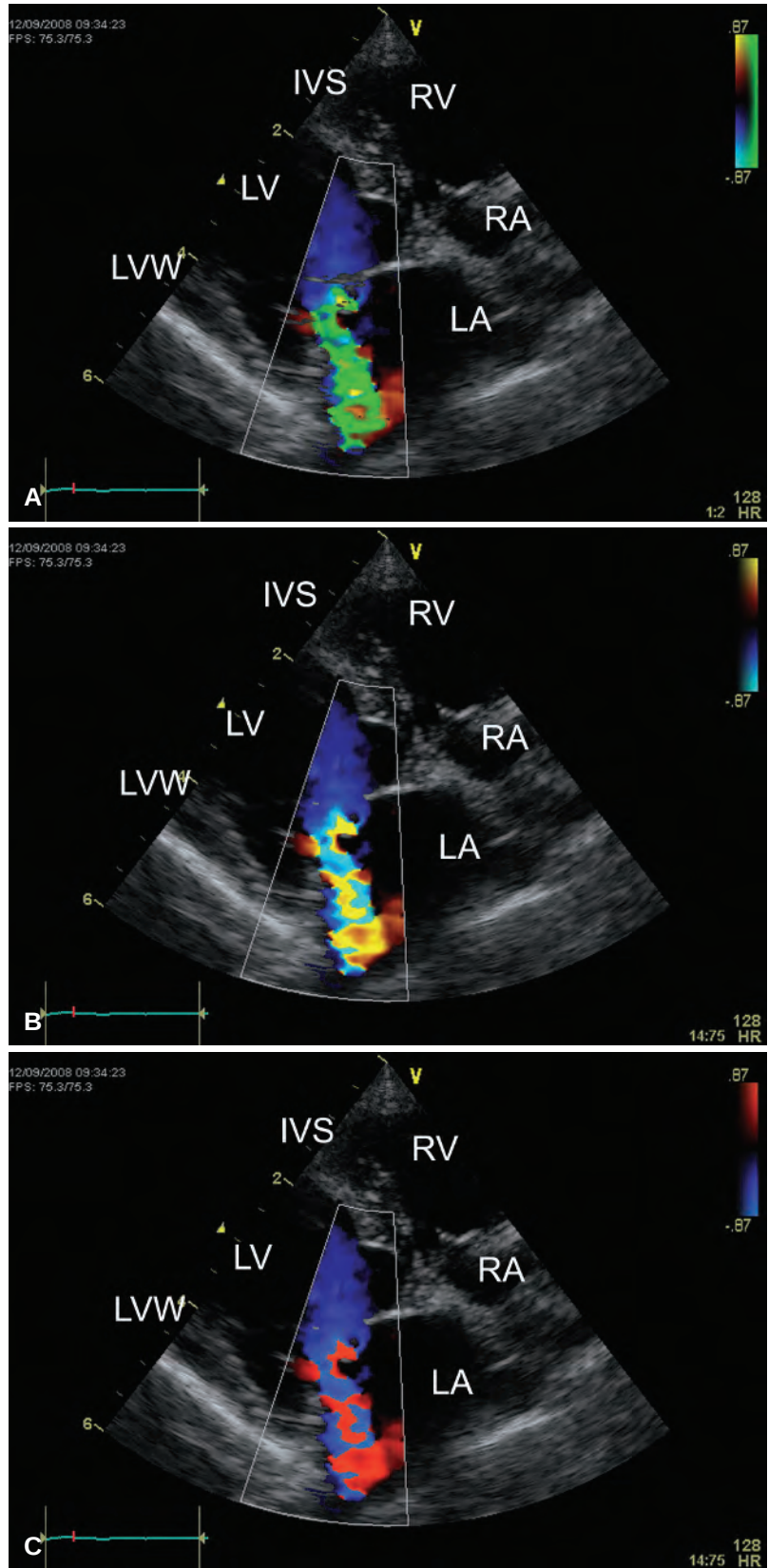
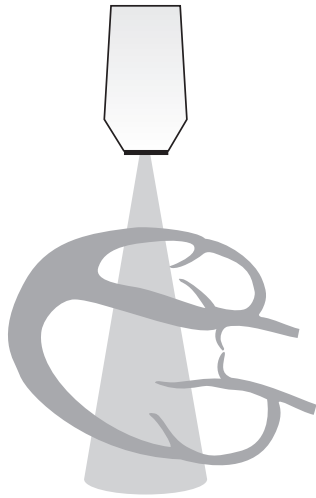
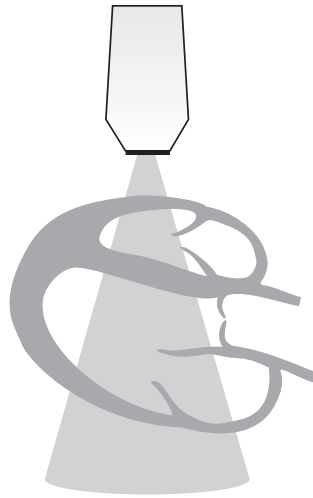


Figure 1.32 This mitral regurgitant jet is displayed with different color flow maps. (A) Variance maps display turbulent (aliased) flow by adding green to the mapped flow. (B) Enhanced maps mix and reverse color and all the hues when flow becomes turbulent. (C) This map shows only pure deep blue and red and mixes these colors when turbulent flow is displayed. RV = right ventricle, IVS = interventricular septum, LV = left ventricle, LVW = left ventricular wall, LA = left atrium, RA = right atrium.



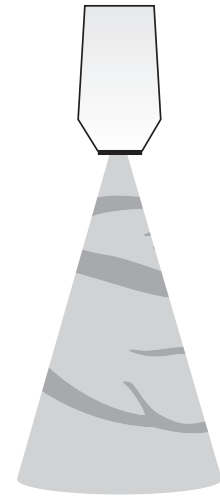
Small color sector = less time required

= ↑ frame rate



Large color sector = more time required

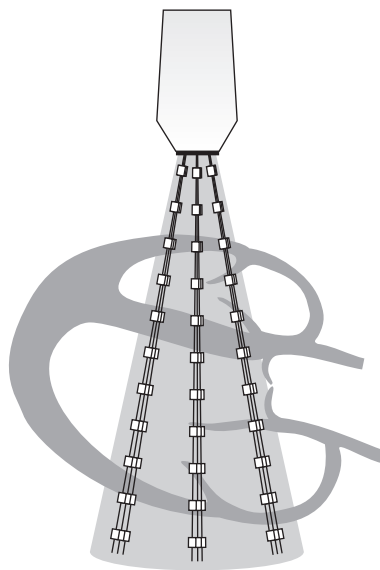
= ↓ frame rate



Eliminate image
outside color sector = less time required

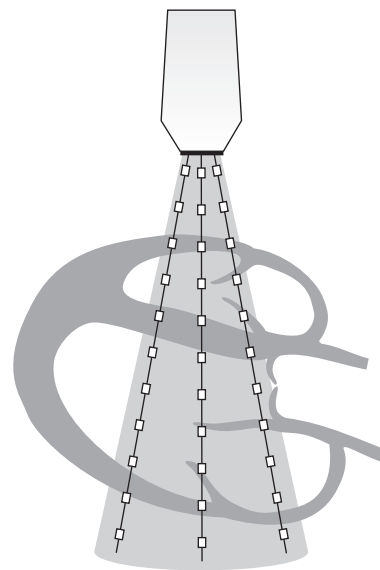
= ↑ frame rate

Figure 1.33 Decreasing the size of the color sector reduces the time necessary for color sampling and increases the frame rate. Better temporal resolution is then possible. Eliminating the two-dimensional image outside the color sector also decreases the amount of time necessary to generate an image and improves color flow mapping.



Large packet size = more time required

= ↓ frame rate



Small packet size = less time required

= ↑ frame rate

Figure 1.34 Packet size is the number of times each line within a color sector is sampled. Large packet sizes produce better color images since more samples can be taken. This is at the expense of frame rate, however, since more time is necessary. Smaller packet sizes decrease the number of times each scan line is sampled so color information is not as complete but frame rate is higher.

Figure 1.35 Color-tissue Doppler uses a narrow sector of color placed over myocardium. Here color is placed over the middle of a transverse left ventricular image. Off line analysis allows gates to be placed over the myocardium anywhere within the color sector. Myocardial velocity corresponding to the selected gate or gates is displayed.

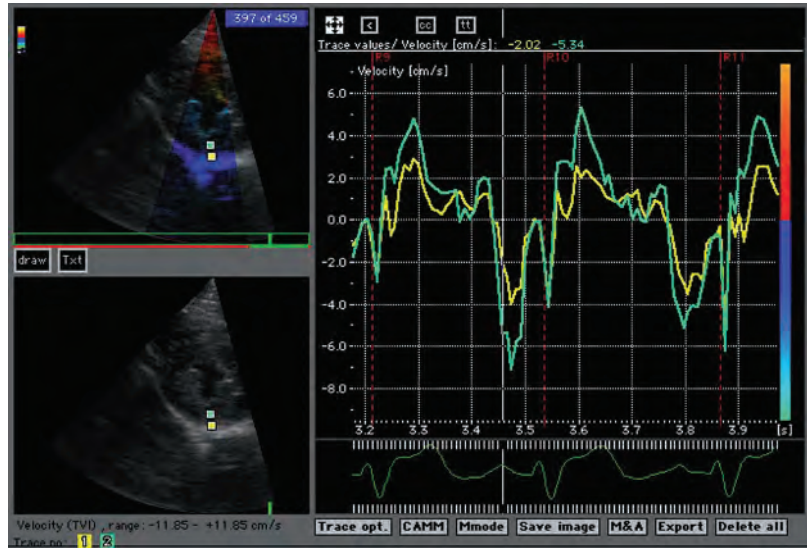
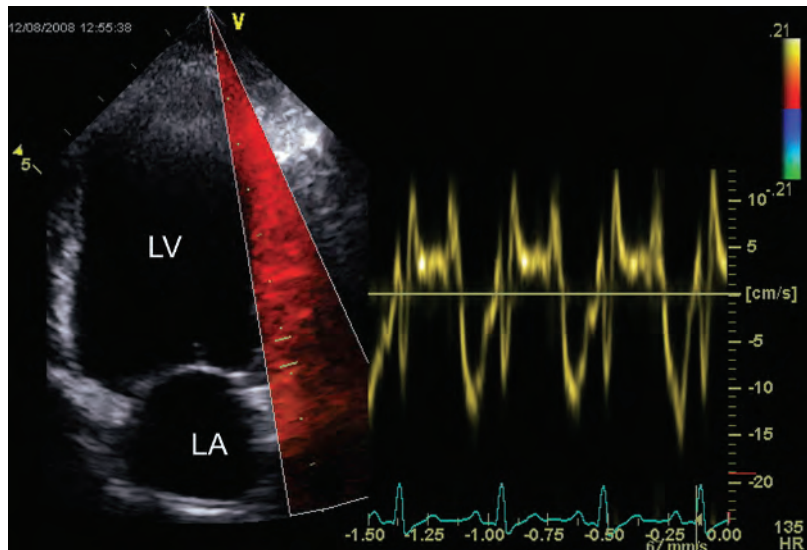


Figure 1.36 Pulsed-wave (PW) tissue Doppler also uses a narrow sector of color placed over myocardium. Here color is placed over the lateral wall of an apical four-chamber image. A PW gate is placed over the myocardium anywhere within the color sector. Instantaneously myocardial velocity corresponding to the selected gate is displayed. Myocardial velocity from only one gate at a time can be displayed.



time (Figure 1.36). The advantage to color TDI is that myocardium anywhere under the color sector can be interrogated after the fact all at the same time and compared. Pulsed-wave TDI is limited to recording myocardial velocities taken in real time under the gate. Pulsed-wave TDI however provides the highest temporal and velocity range resolution.

Tissue Doppler imaging uses both apical, parasternal, and transverse images of the heart. A typical TDI display shows systolic myocardial motion directed centrally into the ventricular chamber and diastolic motion directed away from the center of the chamber on transverse or long-axis parasternal images. On an apical four-chamber view of the heart, systolic motion is directed upward to the transducer, while during

diastole, myocardial motion is directed away from the transducer.

ARTIFACTS

An artifact is anything seen on the echocardiographic image that is in the wrong place, the wrong size, the wrong shape, or is missing. Several artifacts have already been discussed including the inability to resolve two structures as separate entities both in the lateral and longitudinal planes, timing artifacts related to temporal resolution, and aliasing. Selecting the most appropriate transducer for the application will minimize these artifacts. Using harmonic imaging during the echocardiographic examination helps eliminate

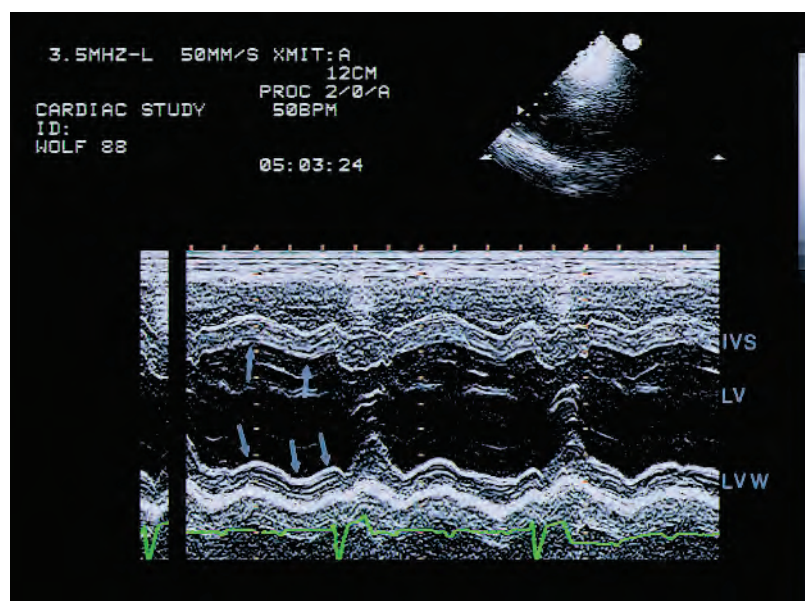


Figure 1.37 Transducer movement secondary to breathing causes structures on this left ventricular M-mode to show excessive motion (*arrows*). Movement of the transducer closer to and farther away from the heart is interpreted by the equipment as cardiac motion. IVS = interventricular septum, LV = left ventricle, LVW = left ventricular wall.

many artifacts. Other artifacts are created as a result of patient movement, and respiration, improper gain settings, peripheral sound beams, or strong reflectors. These are discussed here.

Patient Movement and Breathing Artifacts

Patient movement is a common problem when imaging animals since cooperation is sometimes difficult to obtain. Experience has shown that spending a few minutes calming the small animal while in position prior to the exam is often all that is necessary to quiet the patient. Even puppies and kittens often fall asleep during the exam. A quiet dark room with minimal distractions also helps create a calming atmosphere.

The luxury of telling a patient to hold their breath or exhale deeply and hold it for a few seconds is nonexistent in veterinary medicine, so respiratory artifact is a common problem. Hold a panting animal's mouth shut or blow in his face for a moment or two while obtaining an echocardiographic image in order to obtain good information. Allow the animal to pant between image acquisitions for continued cooperation. An extremely dyspneic animal often needs to be examined while in sternal position or standing.

Respiratory motion moves the transducer farther and closer to the cardiac structures. The resulting M-mode image will have excessive cardiac motion simply due to this perceived cardiac motion by the equipment when in actuality it is only movement of the transducer with respect to the heart (Figure 1.37).

Temporal artifact is the inaccurate timing of color-flow Doppler information. Temporal inaccuracy is often created secondary to respiratory movement. Color is encoded for a chamber or vessel at a location on the screen that has moved before the information is processed (Figure 1.38). This usually appears as a sheet of color. Frame rates that are too slow for sampling in hearts with rapid heart rates will result in color being placed over the heart during the wrong phase of the cardiac cycle.

Side Lobe Artifact

All transducers produce a central beam, which has been the basis of our discussion. They also all produce peripheral beams. These peripheral beams are directed laterally with respect to the central beam. When they intercept structures and are reflected back to the transducer, the equipment does not have the ability to recognize that the information presented did not come from the central beam. Lateral structures are then superimposed upon centrally located structures and are called side lobe artifacts (Figure 1.39). Side lobes are much weaker than central beams and thus returning echoes are also weaker. Often the potential artifact is not even seen since main beam structures have stronger reflections. When other structures are not there to overshadow them, however, they are easily visible. The most common place to see side lobe artifact is within a dilated left atrium or ventricle. This empty space allows weaker side lobe echoes to be displayed (Figure 1.40). When side lobe reflections are

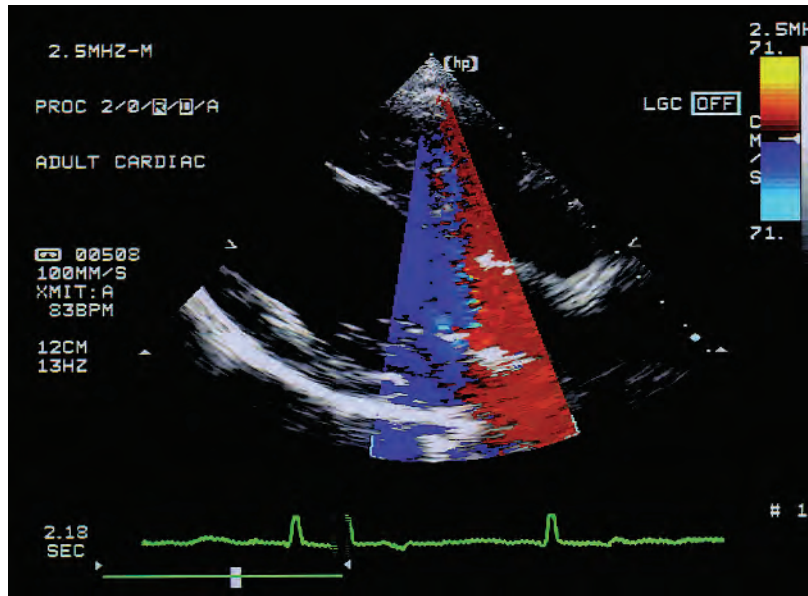


Figure 1.38 Color-flow Doppler will be ambiguous when there is respiratory movement. An ambiguous sheeting effect is created as structures move before flow information is processed accurately.

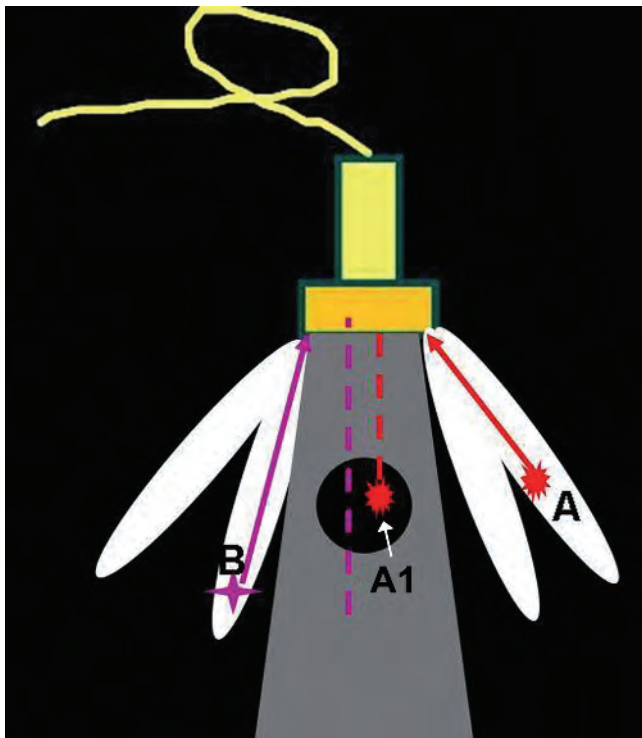


Figure 1.39 All transducers generate sound beams lateral to the central beam. Reflections from structures along these side lobes reach the transducer but are erroneously thought to have been generated by the central beam. Weak reflector (B) in the periphery does not interfere with the true image because it is hidden by structures that truly reflected in the central beam. Reflector (A) that is stronger than what is in the central beam or that would be placed within an echo-free space (like a chamber) will create an artifactual structure on the real-time image, A1.

seen, they typically appear as a curved line extending from the side of the sector into a fluid-filled space or an irregular density within a chamber. Make sure you see a structure in several imaging planes to eliminate possible misinterpretation of these reflections.

Reverberation Artifact/Mirror Image Artifact

Reverberation artifact occurs when strong reflectors are encountered within the thorax. These structures send such strong echoes back to the transducer that the sound is both received by the transducer and reflected from it. The same sound beams travel through the heart again, and when they are sent back to the transducer for a second time they are perceived as having taken twice as long. This produces a mirror image below the first one (Figure 1.41). This occurs often in cardiac imaging when strong reflectors like the pericardium and lung interface cause echoes to bounce back and forth. Reverberation artifact can be minimized by making sure depth settings are adequate for the heart size and not so deep as to see a double image.

Reverberation artifact can also be produced between two strong reflectors within the thorax or heart. The sound may bounce back and forth between these two highly dense structures one to several times before traveling all the way back to the transducer. Multiple images of the same structures are created, each equally spaced and deeper in the image.

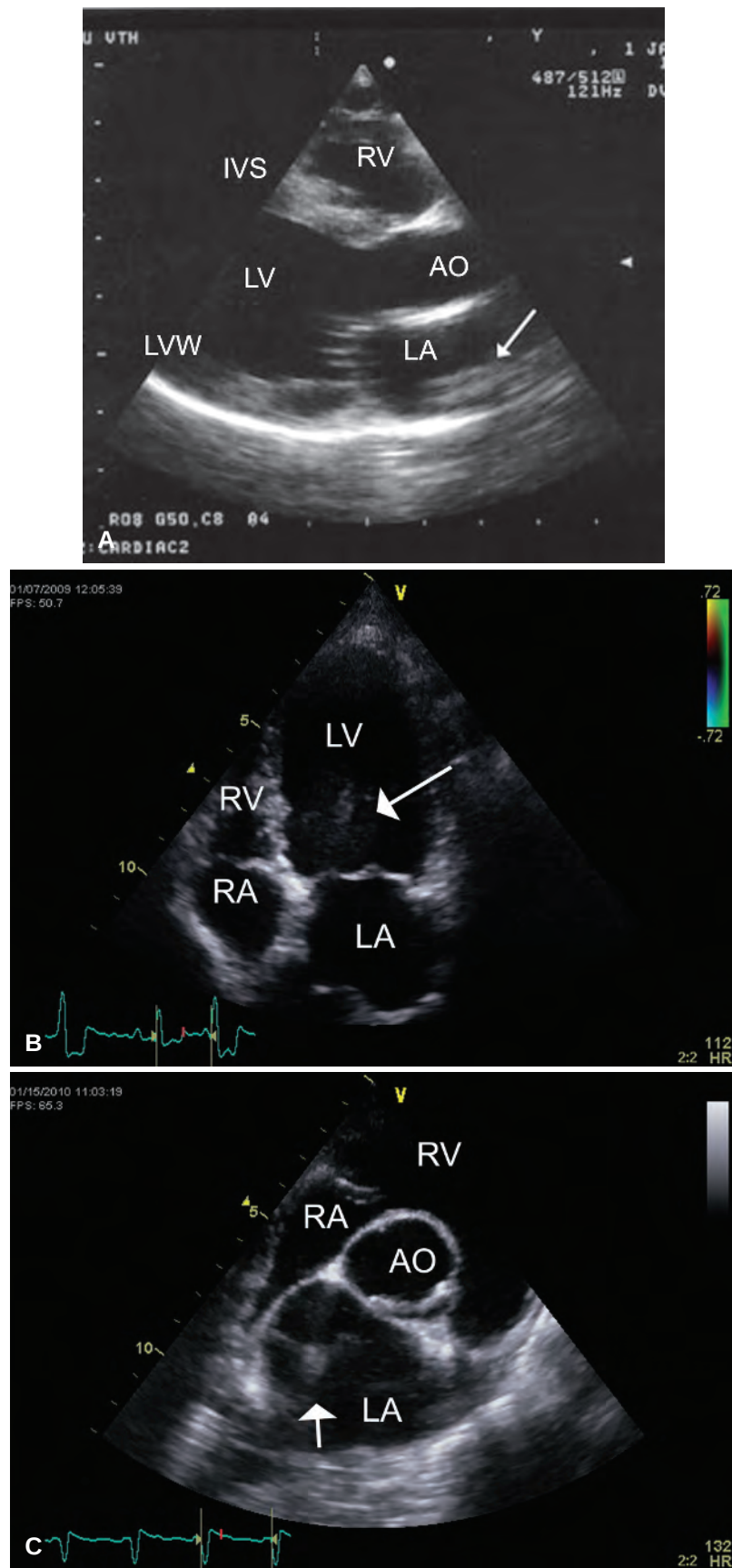


Figure 1.40 Side lobe artifacts appear to be real structures within the fluid-filled chambers of the heart. (A) A side lobe artifact (*arrow*) is seen within the left atrium of this right parasternal long-axis inflow outflow view of the heart. (B) A side lobe artifact (*arrow*) is seen in the left ventricular chamber on this apical four-chamber view. (C) Side lobe artifact (*arrow*) is located within the left atrium on this transverse view of the heart base. (D) This side lobe artifact (*arrow*) seems to fill almost half of the left atrial chamber in this long-axis four-chamber view. RV = right ventricle, IVS = interventricular septum, RA = right atrium, LA = left atrium, LV = left ventricle, LVW = left ventricular wall, AO = aorta.

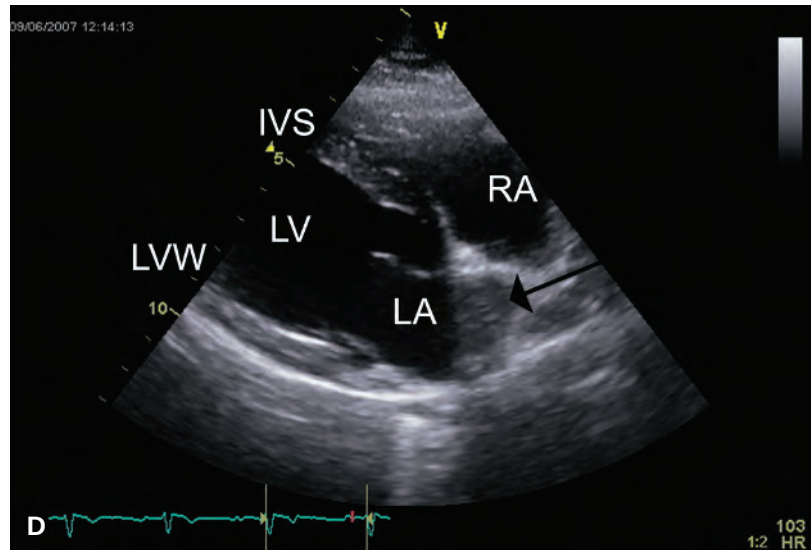


Figure 1.40 *Continued*

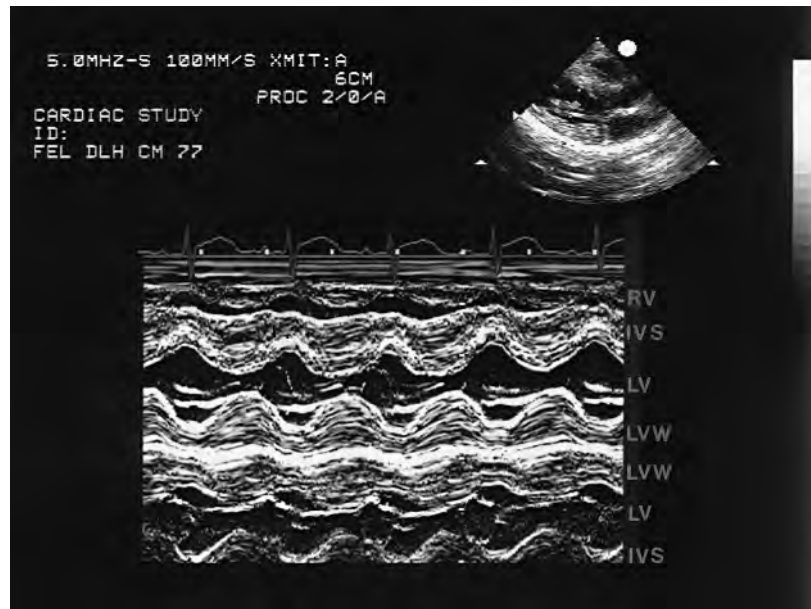


Figure 1.41 Strong reflections may bounce off the transducer and travel through the heart again. A duplicate image of the heart is created below the first one for the sound waves that return from the second reflections since twice as much time has elapsed since the sound beam was originally generated. RV = right ventricle, IVS = interventricular septum, LV = left ventricle, LVW = left ventricular wall.

When spectral Doppler flow shows up on both side of the baseline, it is referred to as mirror image artifact (Figure 1.42). Depending upon where blood flow is being sampled, it is still possible to decide which flow direction is the correct one. This artifact is created by high gain settings creating a situation similar to reverberation artifact. It may also be produced as a result of large angles of incidence with respect to blood flow.

SUMMARY

Transducers send out sound waves that travel in cycles. The number of cycles per second determines

transducer frequency. High frequency transducers generate more cycles per second and so have shorter wavelengths.

The average speed of sound in soft tissue is 1,540 m/second. This speed is independent of transducer frequency. Acoustic impedance increases with increased tissue density and stiffness. This is also independent of transducer frequency. An acoustical difference between two tissues causes sound to be reflected back to the transducer. Bone, being very stiff and dense, impedes the flow of sound tremendously and reflects almost all sound. Scattered sound generates the images from within homogeneous tissues where acoustical properties are similar and structures are small and irregular with respect to wavelength.

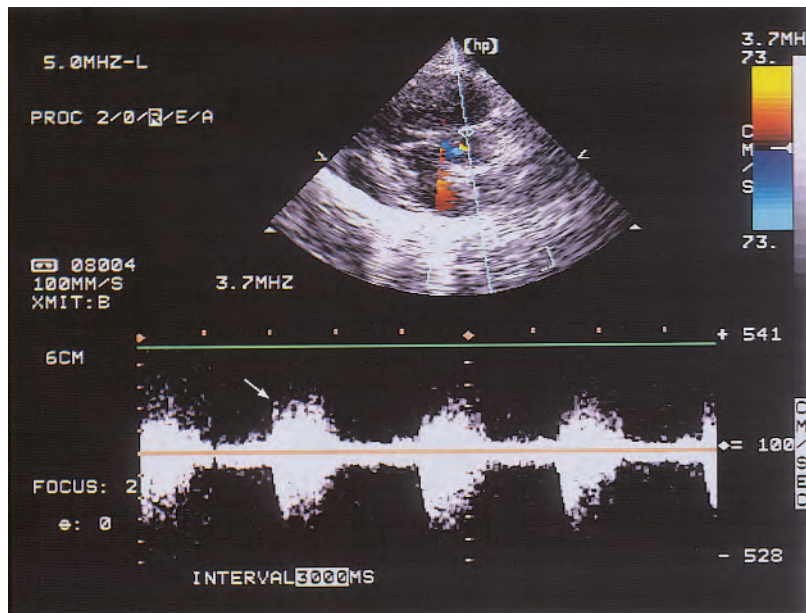


Figure 1.42 Mirror image artifacts (*arrow*) are common in spectral Doppler tracings when gain settings are too high or intercept angles are large.

A structure must be at least one-quarter the size of the wavelength to be reflected. High frequency sound with shorter wavelengths can reflect sound from smaller structures and produce better images. Sound attenuates rapidly with these short wavelengths, however, since they interact with more structures. High frequency transducers therefore create high-resolution images but lose strength rapidly and cannot penetrate as far into tissues as sound from low frequency transducers.

Transducers contain piezoelectric crystals, which function to generate sound and receive sound. The sound is sent out in pulses and the number of pulses per second is the pulse repetition frequency of the transducer. One of the factors affecting axial resolution is the pulse length of a transducer. Higher frequency transducers will have better axial resolution since their short wave and pulse lengths can differentiate smaller structures. Lateral resolution is dependent upon several factors one of which is beam width. The wider the beam width the poorer the lateral resolving power since structures must be farther apart than the beam width to be differentiated. Therefore, lateral resolution is best with smaller diameter transducers. High frequency transducers have longer near fields before the beam diverges, and this enhances the lateral resolution of deeper structures.

Current technology allows multiple focal points along each sound beam in a phased array or annular array transducer. Correct timing of cardiac motion and flow is dependent upon temporal resolution. As multiple focal zones are used or as deeper structures are

imaged, the pulse repetition frequency is decreased. This may not allow fast enough processing of information, and temporal resolution will suffer. Decreasing the width of the real-time sector will increase the PRF as will decreasing the depth of interrogation and using no more than one focal zone.

The change in frequency between sound transmitted by the transducer and sound received by the transducer is the Doppler shift. Blood cells moving toward the transducer will create a positive frequency shift and be displayed above the baseline, while blood moving away from the transducer produces a negative frequency shift and flow profiles below the baseline.

Pulsed-wave Doppler is range gated in that it samples blood at indicated sites within the heart. It is however limited in the highest velocity it is capable of measuring accurately. Continuous-wave Doppler samples blood flow all along the sound beam and while not site specific, it has no Nyquist limit.

The Nyquist limit is dependent upon PRF and thus transducer frequency. The Nyquist limit for any transducer frequency decreases as blood flow is sampled at increasing depths. Lower frequency transducers are capable of accurately recording higher flow velocities at any given depth before aliasing occurs. Accurate velocity measurements are also dependent upon having an angle of interrogation that is parallel with flow. Deviations of the Doppler beam away from parallel result in underestimation of flow velocities.

Laminar flow creates a Doppler signal with little variance in velocity and little spectral broadening.

Spectral broadening is seen when flow becomes turbulent or when CW Doppler is used since many frequency shifts are received by the transducer.

Color-flow Doppler is a form of PW Doppler. Frequency shift information is encoded with color. The most commonly used color map uses a blue away and red toward (BART) format. Aliasing typically occurs at lower velocities in CF Doppler because of the increased time necessary for flow analysis at multiple gates. Aliasing in CF Doppler results in a mosaic of color. As in spectral Doppler, lower frequency transducers will increase the Nyquist limit.

Artifacts are common in diagnostic ultrasound and may be created by the physical properties of transducers, patient-related factors, tissue characteristics, or operator-related errors. Selecting the most appropriate transducer for the exam, calming the patient, and realizing the limitations of diagnostic ultrasound will help eliminate many of these problems or allow intelligent decision making.

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