Clinical Echocardiography

Physics of ultrasound

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Basic sound and ultrasound physics

Unlike light waves, which can propagate through vacuum, sound waves can only propagate through a physical medium. This medium may consist of any matter, e.g air, water, metal, or tissue and fluids in the human body. **Sound waves** arise when a **sound source** generates mechanical vibrations in the particles of the medium. These vibrations continue to propagate through the medium at the speed of sound, thus forming a sound wave.

A familiar example is human speech. Humans speak by setting their vocal cords in motion. When vocal cords vibrate, they generate vibrations in the

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surrounding air and these vibrations propagate in the form of a sound wave. If the sound waves encounter a new medium, some sound waves will be reflected while others will transfer the mechanical (pressure) energy to the new medium, which may also start to vibrate (Figure 1).

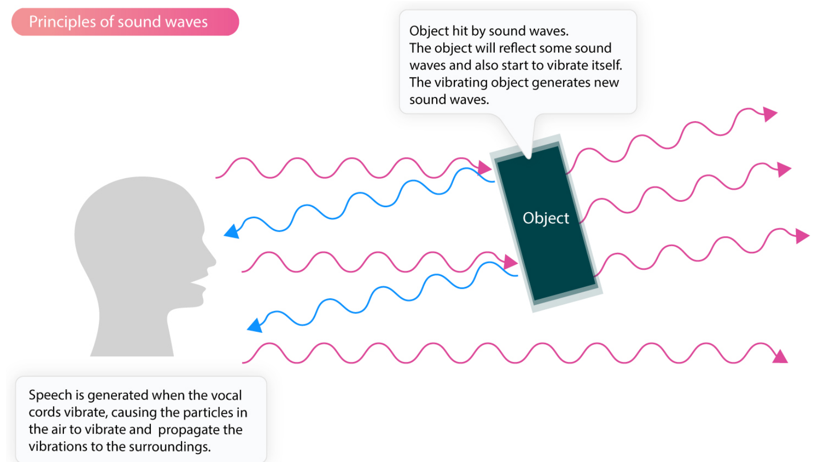

Figure 1. The principle of how sound waves are **generated**, **propagated** and **reflected**. Red waves represent sound waves that are generated when the vocal cords vibrate. Blue waves represent sound waves reflected by the object.

Although sound waves travel through time and space, the particles of the medium do not travel with the sound wave. The particles merely vibrate and transmit the vibrations to neighboring particles in the medium.

Mathematically, sound waves can be described by a **sine curve**. This curve is characterized by the following variables: *wavelength*, *amplitude*, *frequency*, *speed*, and *direction*. The underlying mathematical principles are simple and important to understand. Figure 2 illustrates the wavelength and amplitude of sine curves.

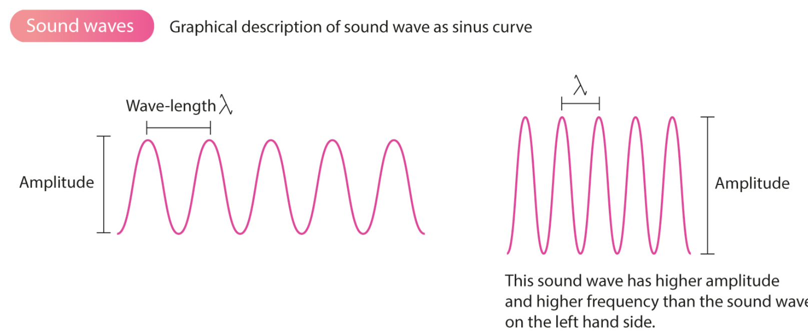

Figure 2. Sound waves can be described mathematically as sine curves.

The peaks and lows of the sine curve correspond to the maximum and minimum pressure, respectively, in the medium. This is illustrated in Figure 3.

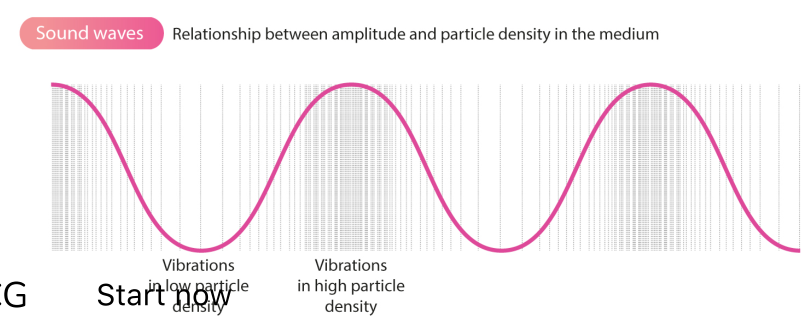

Figure 3.

Wavelength

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Wavelength is defined as the distance between two points (along the sound wave) with equal amplitude (i.e pressure). It is easy to measure the distance between two peaks (maximum) or two lows (minimum). However, the distance between any two points can be measured, provided there is no pressure difference between them. In Figure 2, the wavelength is measured as the distance between two peaks.

The wavelength of the sound waves of human speech is between 17 millimeters (mm) and 17 meters (m). The wavelength is indicated in the unit **m (meters)** and is denoted by the letter **λ (lambda)**.

Note that The International System of Units is used throughout this book. This includes the base units metre (length), kilogram (mass), second (time), ampere (electric current) and Kelvin (temperature). This system is recommended globally.

Amplitude

Amplitude describes the strength of the sound waves, which corresponds to the height of the sine curve (Figure 2). High amplitude equals loud sound and *vice versa*. In Figure 2, two sound waves of different amplitudes are demonstrated. Note that amplitude actually describes the pressure difference between the highest and lowest particle density along the sound wave (Figure 3). Loud sound is characterized by large pressure differences along the sound wave, while low sound has small pressure differences along the sound wave. Amplitude is denoted in the unit **decibels (dB)**.

Frequency

Frequency is the number of wave cycles per second. The unit of frequency, which is denoted by the letter **f**, is **Hertz (Hz)**. In Figure 2, the two sound waves have different amplitudes and different frequencies. If the right sound wave in Figure 2 was recorded during a second, then the frequency would be 5 Hz (since 5 wave cycles are seen in 1 second). If a sound wave has 1000 Hz, then 1000 wave cycles pass every second.

Audible sound and ultrasound

The human ear can perceive sound waves with frequencies between 20 Hz and 20,000 Hz (20,000 Hz can also be written as 20 kHz). **Sound waves with frequency above 20,000 Hz (20 kHz) can not be perceived by the human ear and these sound waves are called ultrasound**. Hence, ultrasound is inaudible to the human ear.

It should be noted that there is a large individual variation in the range of audible sound. The vast majority of humans can not hear sound with frequency above 15 kHz. Younger individuals, however, can hear very high frequencies (sometimes >20 kHz), especially if the amplitude is high.

Ultrasound used for clinical diagnostics, e.g echocardiography, has a frequency of between 2 and 10 million Hz (2-10 MHz), which is far beyond audible sound for humans.

The speed of sound

The speed describes how fast sound waves propagate through the medium. This speed depends on the density of the medium. Sound waves propagate faster in high-density media. The higher the density, the higher the speed. The speed of sound is approximately 300 m/s in air, and 1540 m/s in the human body (which consists mostly of water). The speed is denoted by the letter **c** and is indicated by the unit **m/s**.

Direction of sound waves

Direction simply describes the direction of sound waves in the medium.

Mathematical equations

There is a simple mathematical relationship between speed (c), wavelength (λ) and frequency (f):

$$c = f \bullet \lambda$$

According to the formula, the speed of the sound wave is the product of the frequency and wavelength. Using this formula, we can calculate the wavelength (λ) for ultrasound with frequency 3 million Hz (3 MHz), which is used in ultrasound diagnostics:

$$\lambda = 1540 / 3000000 = 0{,}000513 \text{ meter}$$

0.000513 meters is 0.513 mm (millimeter). Thus, the wavelength of the ultrasound is very short, which is desirable in echocardiography, and ultrasonography in general, because it allows detailed visualization of small structures (i.e the resolution becomes high).

The ultrasound transducer

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The ultrasound transducer & piezoelectric crystals

The ultrasound transducer generates ultrasound (ultrasonic) waves. The transducer is held with one hand and its position and angle are adjusted to send ultrasound waves through structures to be visualized.

Ultrasound waves are emitted rapidly from the transducer. These sound waves travel through tissues and fluids. Some of the sound waves are reflected back to the transducer. By analyzing the reflected sound waves, the ultrasound machine creates an image of the tissues. Thus, the principle of ultrasound imaging is simple: sound waves are sent into the tissue and the reflected waves are used to create an image of the tissue (**Figure 1**).

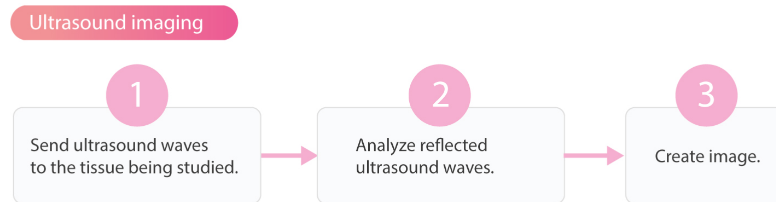

Figure 1. The principle of ultrasound imaging and echocardiography.

Piezoelectric crystals

The ultrasound waves are generated by ceramic crystals exhibiting **piezoelectric properties** (i.e **piezoelectric crystals**). Thousands of piezoelectric crystals are attached to the front of the transducer (Figure 2). The crystals are connected to the ultrasound machine via electrodes.

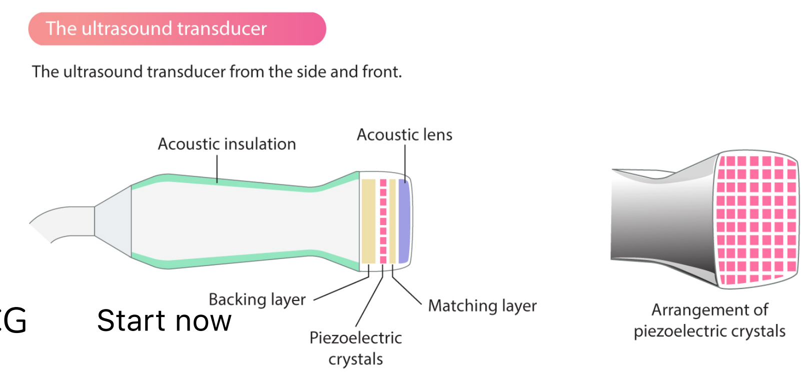

Figure 2. The ultrasound transducer and the piezoelectric crystals that generate and receive ultrasound waves.

Piezoelectric crystals have unique electromechanical properties. When an electric current is applied to a piezoelectric crystal, it starts to vibrate and these vibrations generate sound waves with frequencies between 1.5 and 8 MHz (i.e ultrasound). Thus, piezoelectric crystals can convert electric currents into ultrasound waves. The crystals can also do the opposite; when the crystals are hit by reflected ultrasound waves, they begin to vibrate and these mechanical vibrations are converted into electric current that is sent back to the ultrasound machine, where the electrical signal is interpreted and translated into an image (Figure 3).

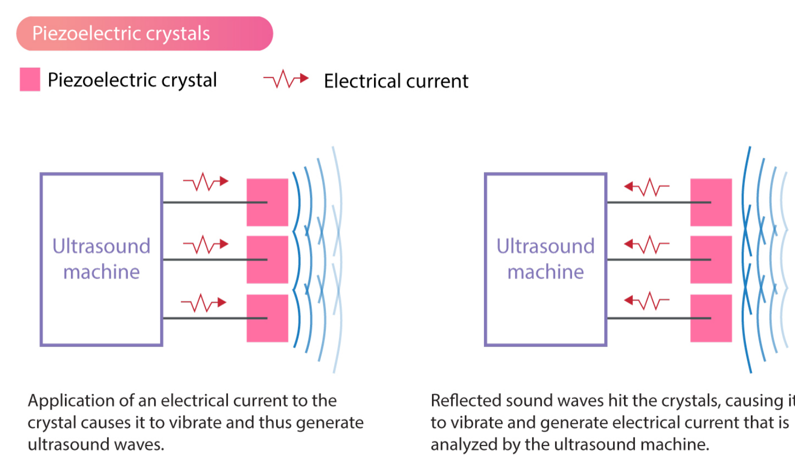

Figure 3. Piezoelectric crystals.

As can be seen from Figure 2, the ultrasound transducer contains several components. The transducer contains *acoustic insulation* that ensures that no other sound waves affect the transducer. The crystals are supported by a *backing layer* that suppresses the vibrations of the crystals, allowing sound waves to be sent out in shorter pulses and this improves resolution (discussed below). In front of the crystals are materials (*matching layer*) that reduce the difference in impedance between the crystals and the tissue to be studied. Without this layer, the impedance difference becomes large, which causes too much of the sound waves to be reflected (leaving fewer sound waves to penetrate the tissues). At the front of the transducer is an *acoustic lens*. This is the hard rubber that focuses the ultrasound waves, which results in less scatter of the waves and thus increase the resolution of the image.

From the transducer, ultrasound waves are sent out in pulses. Each pulse consists of a few sound waves emitted in 1 to 2 milliseconds. These sound waves travel through the skin, chest, pericardium, myocardium, etc. In the transition between each medium (tissue, blood, etc.), a significant portion of all sound waves will be reflected back to the transducer. When the reflected sound hits the piezoelectric crystals, they begin to vibrate and generate electric currents, which are transmitted to the ultrasound machine for analysis.

The reflected sound waves will have the same speed as the emitted sound waves, but the amplitude, frequency and angle of incidence may differ from the emitted sound waves. The ultrasound machine utilizes variations in the amplitude, frequency and timing of the reflected sound waves to create an image of the medium (tissue).

Technical aspects of the ultrasound image

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Generating the ultrasound image

The ultrasound transducer generates short bursts (pulses) of ultrasound waves. Reflected ultrasound waves are analyzed by the machine during the brief pauses between the pulses. Thus, the machine analyzes (“listens to”) reflected sound waves immediately after it emits sound waves (Figure 1).

Ultrasound pulses

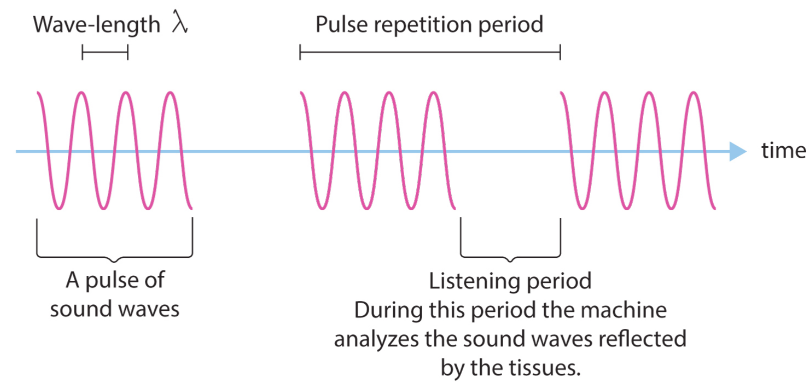

Figure 1. the ultrasound machine sends pulses of ultrasound, and listens to reflected ultrasound between the pulses.

To create a reliable real-time image of the tissue, the ultrasound machine must overcome the following technical obstacles.

1. **The ultrasound machine must know which sound waves are reflected and from where they are reflected**. Since the sound waves are sent out in pulses and the velocity in the tissue is constant (1540 m/s), the machine can calculate where the sound waves were reflected (i.e the machine can calculate the *reflection point*). This is done by analyzing the time it takes for the sound to return to the transducer and thus calculating the distance to the structure that reflected the wave. Structures located near the transducer will reflect

the sound waves early and thus the time interval will be short. Structures located far from the transducer will reflect the sound waves later and it will take longer to reach the transducer.

2. **Ultrasound waves reflected from the same structure can reach the different crystals at different time points**. To solve this, there is a built-in function, called **dynamic focusing**, which calculates which ultrasound waves that originate from the same reflection point.
3. **Reflected ultrasound waves have altered properties (e.g altered amplitude)**. This is exploited to give the reflected sound waves, based on their amplitude, different nuances on the ultrasound image. The tissues in the ultrasound image are drawn with varying shades of one color (usually gray). This is possible because the vibrations in the piezoelectric crystals, and thus the electric current they send back to the machine, vary with the amplitude of the reflected sound. The stronger the reflections, the higher the amplitude and the whiter the color of the tissue on the ultrasound image.
4. **Moving structures (myocardium, blood flow) will alter the characteristics of ultrasound waves (e.g. frequency)**. This is actually exploited to calculate the direction and speed of tissue and fluid movements.

All structures in a medium can reflect ultrasound waves. However, the greatest reflections occur in the interfaces between two media. Hence, in the transition from blood to myocardium, many sound waves will be reflected and this will result in a clearly depicted border zone between blood and myocardium on the echocardiogram. Ultrasound waves will also be reflected as the waves travel through the myocardium, but to a lesser extent, and therefore the myocardium does not shine as clearly on the ultrasound image (Figure 2).

Ultrasound waves are primarily reflected in the interface between media (tissues, fluids, etc) of different density. The greater the difference in density, the more ultrasound waves are reflected. This explains why tissue borders appear as brighter structures on the ultrasound image.

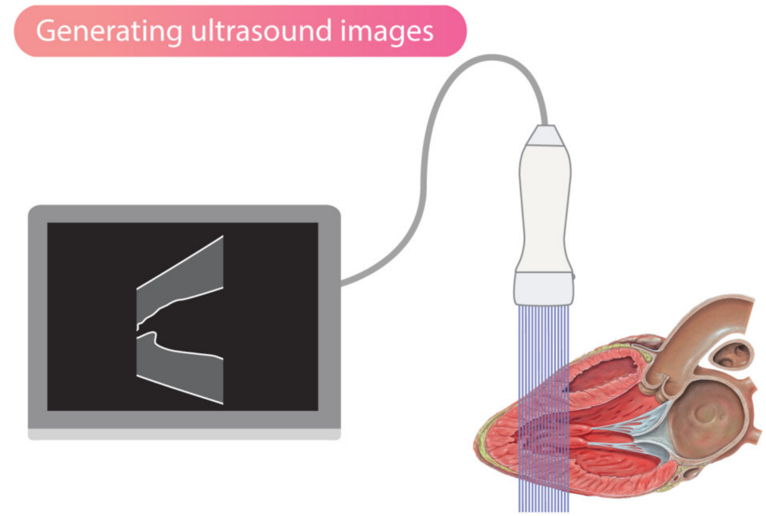

Figure 2. Visualization of reflected ultrasound waves. It is in the transition from one tissue to another that most sound waves are reflected and this gives tissue borders brighter color on the ultrasound image. This schematic image shows brighter colors when going from pericardium to epicardium and when going from endocardium to the ventricular cavity.

Directing and focusing ultrasound waves

The direction and focus of ultrasound waves can be adjusted by varying the sequence of activation of the piezoelectric crystals (Figure 3). By activating all crystals simultaneously, the resulting sound wave travels in a straight direction (Figure 3A). If activation starts on one side, for example from right to left, then the wavefront will be directed to the left (Figure 3B). If the activation starts at the ends and proceeds towards the center, then the ultrasound beam will be focused as illustrated in Figure 3C.

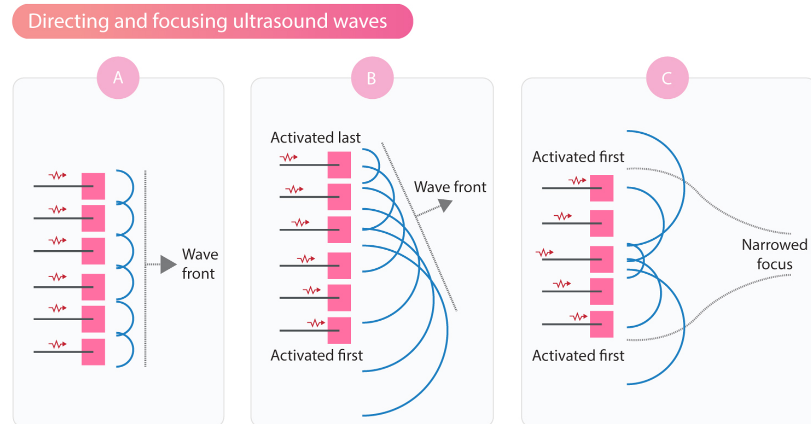

Figure 3A, 3B, 3C. The ultrasound machine can vary the sequence of activation of the piezoelectric crystals, which adjusts the direction of the wavefront and the focus of the ultrasound beam.

Modern ultrasound machines include advanced software that handles the activation of thousands of piezoelectric crystals. Using sophisticated software and hardware, it is possible to obtain high-resolution two-dimensional (2D) and three-dimensional (3D) echocardiograms.

Reflection of ultrasound waves

As mentioned previously, ultrasound waves are primarily reflected in the interface between media (tissues, fluids, etc) of different density. The greater the difference in density, the more ultrasound waves are reflected. For example, the difference in density between skin and bone is very large, which explains why most of the ultrasound waves are reflected when they hit bone. Structures located behind bone can therefore not be visualized using ultrasound (since very few sound waves go through the bone). Similarly, the difference in density between the air-filled lungs and the pericardium, explains why much of the ultrasound is reflected on the pericardial surface (which therefore shine brightly on the echocardiogram).

> The larger the proportion of sound waves reflected, the fewer sound waves remain to study the rest of the tissue (deeper structures). Air-filled spaces (*i.e* lung) and hard surfaces (*i.e* bone) pose a special challenge. It is therefore important to place the transducer and direct the sound waves such that collision with bone and passage through lung tissue is minimized.

When ultrasound waves travel through soft tissues and fluid-filled spaces (e.g ventricular cavity, atria, larger vessels), a relatively small proportion of sound waves is reflected. This is due to the small difference in density within the tissue or fluid.

For an ultrasound wave to be reflected at an unchanged angle (compared to the angle of incidence), the object reflecting the ultrasound wave (*i.e* the reflector) must have a smooth surface, perpendicular to the direction of the sound waves. Human tissues consist of more or less irregular structures, which results in sound waves always being reflected at a slightly altered angle. However, the change in the angle is generally small and most of the reflected sound waves hit the transducer. This type of reflection is called **mirror reflection**. The ultrasound waves that are not reflected at the interface between two media will continue through the second medium with slightly altered angle, a phenomenon called **refraction**.

Although most of the reflected sound waves are reflected in the border zone between two media (tissues/fluids), some waves are also reflected during passage through homogeneous tissue, such as the myocardium. Otherwise, the myocardium would not have been visible on the echocardiogram. However, reflections within the tissues are more scattered. The more irregular the structure of the tissue, the more scattered the reflections.

Erythrocytes are particularly good at spreading the ultrasound waves; they spread the waves in all directions. Thus, only a minority of reflections return to the ultrasound transmitter.

Ultrasound waves are attenuated (weakened) as they travel through the body. The attenuation is due to the reflection of sound waves and the transformation of mechanical energy into heat (which is absorbed by the tissues).

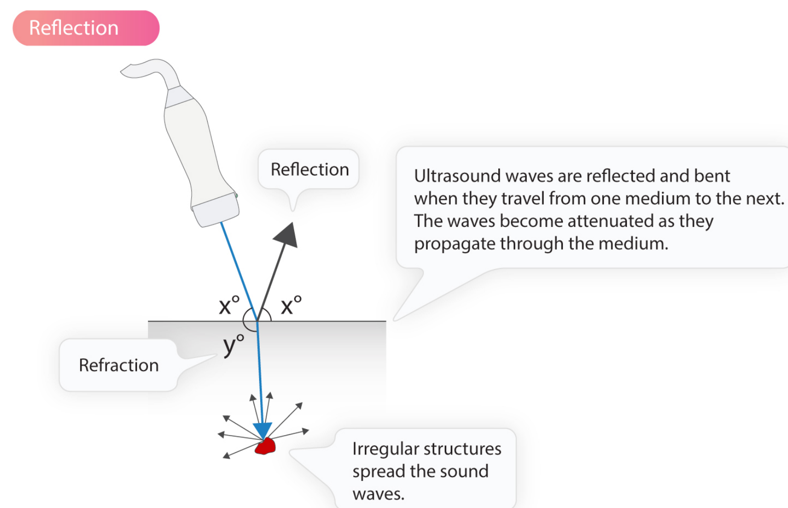

Figure 4. Reflection and refraction of ultrasound waves.

Resolution and penetration of ultrasound waves

Obtaining high-resolution ultrasound images is essential for diagnostic accuracy. The image resolution can be defined as the possibility to distinguish two adjacent objects. Studying small structures, particularly moving structures, requires high-resolution images. The lower the image resolution, the more difficult to distinguish smaller and neighboring objects.

The image resolution depends mainly on the wavelength of the ultrasound waves. As discussed previously (Physics of ultrasound) the wavelength is inversely proportional to the wave frequency according to the following formula:

$$\lambda = c / f$$

This implies that high-frequency waves have short wavelengths and vice versa. The shorter the wavelength, the smaller structures will be able to reflect the sound wave and thus become visible on the ultrasound image. Thus, the higher the frequency, the higher the resolution. It may, therefore, seem reasonable to increase the frequency to the limit of the ultrasound machine. However, the penetration of ultrasound waves diminishes with increasing frequency, meaning that high-frequency waves have poorer penetration. Visualization of deeper objects, therefore, requires waves with lower frequency.

Ultrasound waves with low frequency have a long wavelength, which provides lower resolution but better (*i.e* deeper) penetration. Image quality for distally located objects can thus be improved by using lower frequency; the increase in penetration typically outweighs the loss in resolution.

The maximum resolution is about half the wavelength; *e.g* a frequency of 2.5 MHz yields a resolution of 0.3 mm. Objects smaller than 0.3 mm are not distinguishable at a frequency of 2.5 MHz.

Axial and lateral resolution

The **axial resolution** is the ability to distinguish two *objects located parallel to the ultrasound wave*. This resolution is constant along the ultrasound wave. The axial resolution is fundamentally dependent on the frequency of the sound waves. The higher the frequency the greater the axial resolution.

The **lateral resolution** describes the ability to distinguish two *objects that are perpendicular to the ultrasound waves*. This resolution decreases with the distance from the transducer because the ultrasound waves diverge as the distance increases.

Axial and lateral resolution

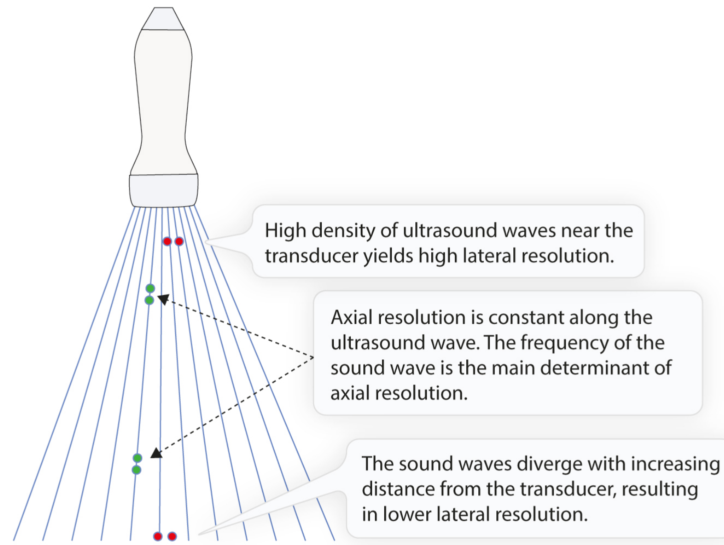

Figure 5. Axial and lateral resolution of the ultrasound image.

Temporal resolution

Temporal resolution (also see **Frame rate** below) is the ability to describe the movement of objects over time. Ultrasound imaging in general, and echocardiography in particular, requires continuous analysis of reflected ultrasound waves to create a 2D or 3D film. The film is created based on individual ultrasound images that appear one after another. In order to generate a film with high temporal resolution, it is critical to produce individual images rapidly. The time it takes to create one image determines the temporal resolution. The more images that can be produced and presented per unit of time, the greater the temporal resolution.

Fundamental and harmonic imaging

The ultrasound transducer generates sound waves with a specific frequency. This frequency is called the *base note*. When the sound waves pass through the tissues, the sound waves are deformed which creates harmonics.

Sound waves are deformed as they pass through tissues. When the high-pressure portion of the sound wave (the highest point of the sine curve, see Physics of Ultrasound) encounters tissue with higher density, the tissue will be compressed and the speed of the sound wave is increased. When the low-pressure part of the sound wave (the lowest point of the sine curve) passes through the tissue, the opposite will occur: the tissue expands, tissue density decreases and the speed of the sound wave also decreases.

Thus, the sound wave is distorted as it passes through tissues. This distortion results in the occurrence of sound waves whose frequency is multiples of the base note. These sound waves are called harmonics. Thus, the ultrasound transducer emits waves at frequency of 3 MHz and then sound waves with frequency 6 MHz (second harmonic), 9 MHz (third harmonic), 12 MHz (fourth harmonic), *etc*, arise. These harmonics are also reflected back to the transmitter. In fact, it is possible to create an ultrasound image using only reflected harmonics. This results in images with improved resolution. Modern ultrasound machines are therefore programmed to primarily analyze reflected harmonics (mostly the first harmonics).

> The ultrasound image is created by listening to one harmonic and filtering out all other frequencies (both the base note and all other harmonics). This imaging method is called **harmonic imaging**.

Harmonic imaging is standard in ultrasound diagnostics and echocardiography. The method makes it possible to send out sound waves of low frequency (allowing deeper penetration of the tissues), but listen to sound waves of high frequency (yielding higher resolution). Harmonic imaging also reduces artifacts in the ultrasound image. The disadvantage of harmonic imaging is that some texture is lost. This is not a significant issue but it may result in heart valves appearing thicker than they really are.

The opposite of harmonic imaging is **fundamental imaging**, which implies that the machine listens for sound waves at the same frequency as it itself generated. For example, if the transducer emits sound waves at a frequency of 3 MHz, it only listens for reflected sound waves that have 3 MHz. This gives poorer resolution and lower penetration. There are, however, situations in which fundamental imaging is useful.

Two-dimensional (2D) echocardiography

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Echocardiography in 2D

Two-dimensional (2D) ultrasound is the most commonly used modality in echocardiography. The two dimensions presented are width (x axis) and depth (y axis). The standard ultrasound transducer for 2D echocardiography is the *phased array transducer*, which creates a sector shaped ultrasound field (Figur 1).

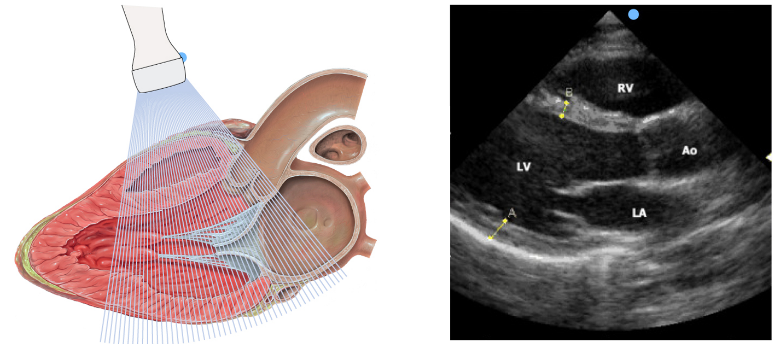


Figure 1. Two-dimensional echocardiogram. This view is called parasternal long axis view (PLAX). Structures that are closest to the transducer are placed at the top of the image. RV = right ventricle. LV = left ventricle. LA = left atrium. Ao = aorta. The thickness of the interventricular septum (B) and the inferolateral wall (A) have also been measured.

The image sector is created using sequential activation of the piezoelectric crystals. The crystals are activated from one side to the other, as illustrated in Figure 2. The sequence of activation goes from right to left and then from left to right, and this is repeated rapidly. To create an image sector of 90° width and 15 cm depth, approximately 200 ultrasound lines are required and this takes about 40 milliseconds (ms) to generate. As illustrated previously (Axial and lateral resolution of the ultrasound image) line density diminishes with increasing distance from the transducer.

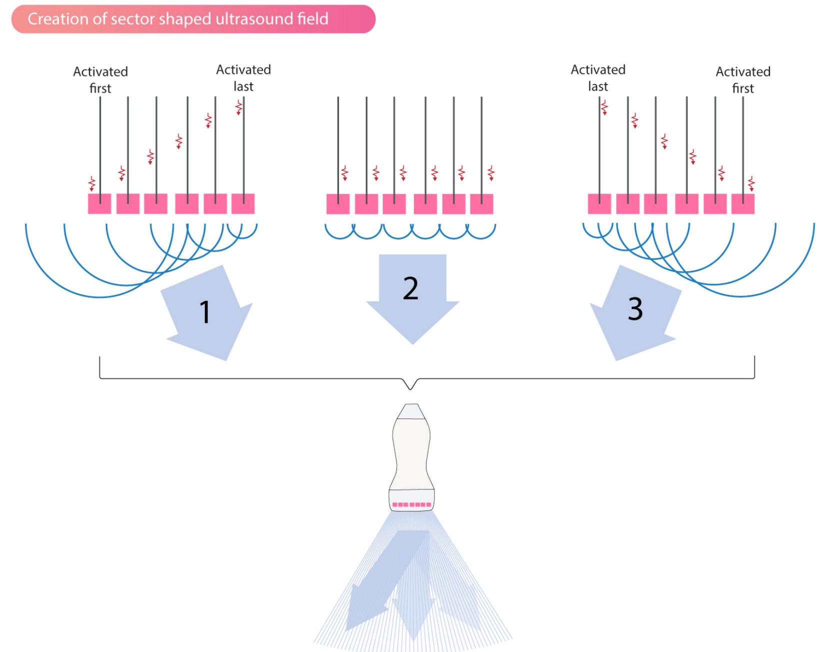


Figure 2. The phased array transducer creates a sector shaped ultrasound field.

The phased array transducer and its sector shaped ultrasound field are suitable for echocardiography since the ultrasound field can pass the ribs and then spread over a larger area. The focus can be adjusted by varying the sequence of activation of the piezoelectric crystals. The density of the ultrasound lines decreases with increasing distance from the transducer; this affects the lateral resolution as explained in Figure 3.

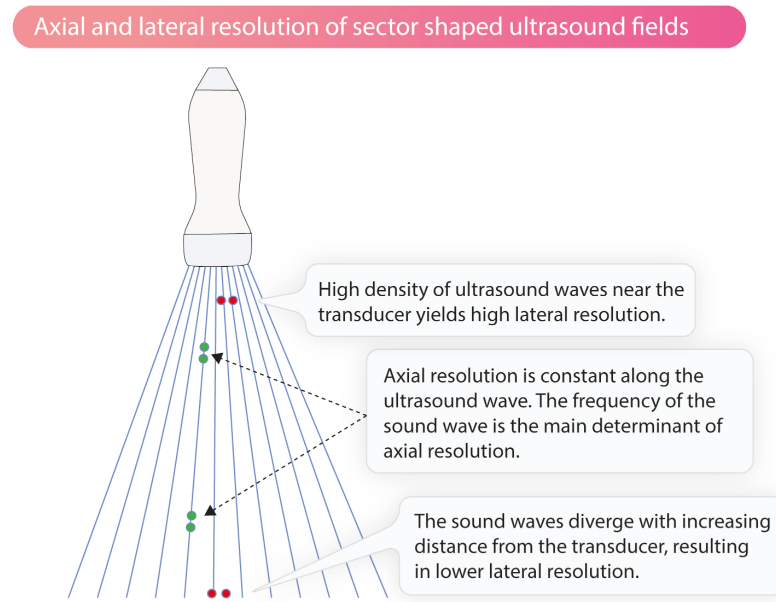


Figure 3. Axial and lateral resolution of the ultrasound image. Axial resolution is constant along the ultrasound lines. Lateral resolution depends on line density, which decreases with increasing distance from the transducer.

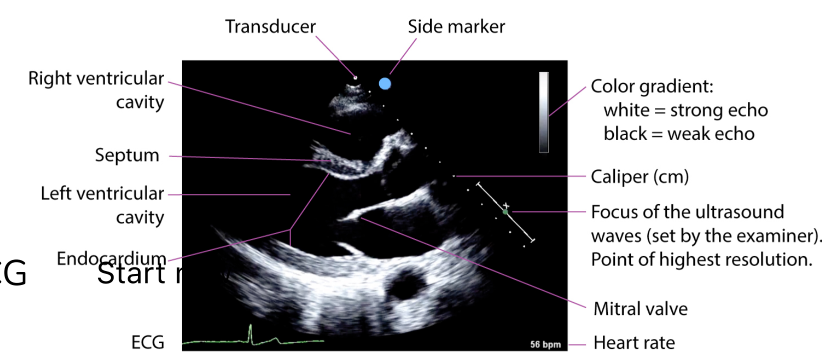
Frame rate

The two-dimensional image must be updated rapidly and continuously in order to obtain a movie. The speed at which the images are updated is crucial for producing a high-resolution movie. The rate of update is described by the technical term **frame rate**, which is the number of images (*frames*) displayed per second. High frame rate (i.e many frames per second) is desirable because it provides better temporal resolution.

Frame rate depends on several factors. The elapsed time for all ultrasound waves to be emitted, reflected and processed in the machine determines the frame rate. As mentioned above, using 200 ultrasound lines to create a 90° wide and 15 cm deep image requires approximately 40 milliseconds (ms). Increasing the number of ultrasound lines or image depth will reduce the frame rate since more time is required to complete each frame. Thus, the temporal resolution is reduced by enlarging the image sector. The opposite is also true; frame rate, and thus temporal resolution, can be increased by reducing image depth or reducing the width of the sector. To achieve the highest resolution possible, the depth and width of the image should be kept as small as possible. The ultrasonic machine has controls to adjust width and depth. It is also possible to adjust the frame rate to a certain extent.

The ultrasound image

The ultrasound image



Start learning ECG

Figure 4. Example of ultrasound image. This view is called parasternal long axis view.

The ultrasound image contains several important parameters (Figure 4). The ECG signal is presented at the bottom and it is used to identify diastole and systole, which is necessary to perform various measurements. The transducer itself is not seen in the image but the contour of the lens is seen at the very top of the sector field (the dark area). Figure 4 shows a blue (the color may vary depending on manufacturer and user settings) circle next to the transducer; this is the side indicator that assists the examiner to identify left and right in the image. This indicator corresponds to the indicator on the transducer.

Focus of the ultrasound beams

The highest resolution on the ultrasound image is located where the width of the ultrasound beam is narrowest and this point is called **focus**. It is possible to adjust the position of focus without moving the transducer; focus is shfited by modifying the sequence of activation of the piezoelectric crystals.

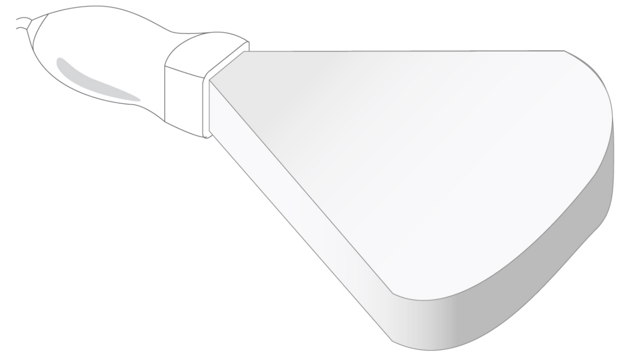


Figure 5. The thickness of the ultrasound field is illustrated here.

Although two-dimensional ultrasound images (e.g Figure 4) suggest that the ultrasound beam is flat, in reality, the ultrasound beam is 2 to 10 mm thick (Figure 5). The ultrasound image presented is a flattened version of the original three-dimensional ultrasound beam. Hence, structures that are not actually located next to each other can be placed next to each other on the two-dimensional image.

Optimization of the ultrasound image

Chapter contents [Show]

Principles of image optimization in echocardiography

In order to obtain optimal ultrasound images, it is necessary to adjust several parameters continuously during the examination. Typically, examinations in each echocardiographic view (also called *window*) are initialized by identifying an overview image. Starting in the overview image the depth is reduced as much as possible. Reducing the depth results in increased frame rate and thus better image resolution. If possible, the width of the image is also reduced, which likewise results in increased image-resolution.

It is also possible to zoom in on regions of interest; e.g the aortic valve can be zoomed in to study its anatomy and function. Zooming in improves the resolution in a particular area. Alternatively, it is possible to place the focus at the level of the region of interest. The difference between zooming in and shifting focus is that the zoom encloses a specific region of the image, whereas shifting focus simply adjusts the location (along the ultrasound beam) with the best resolution. If the ultrasound image is too dark, it is possible to increase the gain. This amplifies the incoming (reflected) ultrasound waves such that each object appears whiter on the image. Increasing gain excessively results in lower resolution and difficulties discerning tissue borders. These are the main adjustments made to improve image quality.

Adjusting image depth and zoom

Examinations in each echocardiographic view are initialized by identifying an overview image. Starting in the overview image the depth is reduced as much as possible, without excluding regions of interest. Reducing the depth results in an increased frame rate and thus enhanced image resolution. If a particular region is of interest, that region can be zoomed in. Note that the image becomes more grainy as the zoom is increased.

Gain: signal amplification

The ultrasound machine amplifies all incoming (reflected) ultrasound waves. However, the examiner can further increase the amount of gain applied to incoming sound waves. This is done using **gain control** or **time-gain compensation (TGC)**.

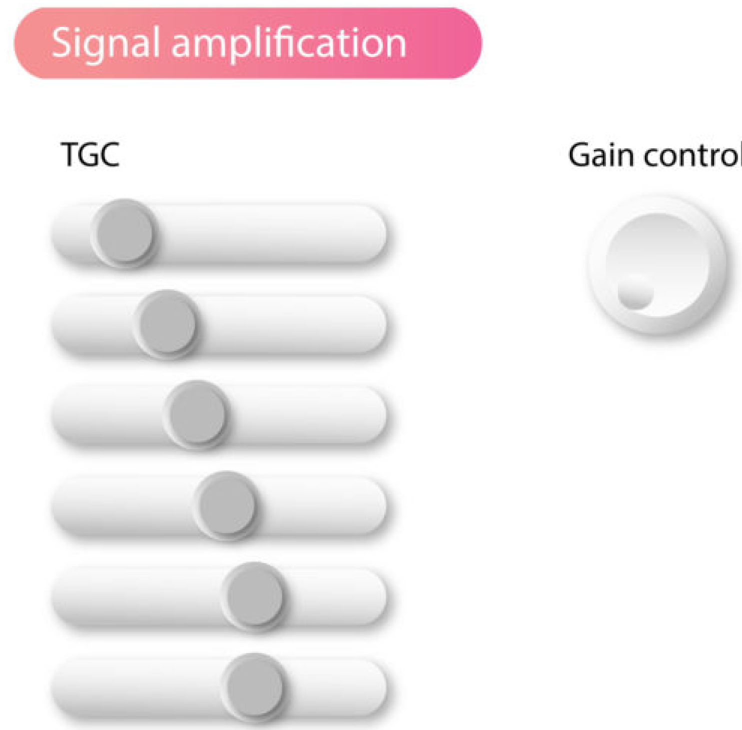


Figure 1. TGC (time gain compensation) and gain control on the ultrasound machine.

Gain control: overall gain

Gain control regulates the *global (overall) gain*. Increasing the overall gain will increase the gain for all reflected sound waves, making all objects in the image whiter. This may clarify some tissue borders but excessive use of gain results in deterioration of image quality.

Time gain compensation / control (TGC)

Time-gain control / compensation (TGC) adjusts the gain at specific levels along the ultrasound field. The purpose of TGC is to gradually increase the amount of gain as the depth increases; this compensates for the attenuation that occurs with increasing depth. TGC is adjusted using multiple controls that each represent a specific depth in the image (Figure 1). The bottom control adjusts gain at the bottom of the image *etc*. TGC is generally increased at the bottom of the image since the ultrasound lines have the lowest density there (and thus the lowest image resolution). TGC at the top of the image is usually kept at low levels.

Frequency of ultrasound waves

Low ultrasound wave frequency provides high tissue penetration and low image resolution. High-frequency waves provide good image resolution but worse penetration. Visualizing objects located in close proximity to the transducer, therefore, is done using high-frequency waves. The frequency of the ultrasound wave must generally be reduced in order to visualize objects located far away from the transducer. Hence, using low-frequency waves to visualize distant objects is motivated by the advantages of greater tissue penetration of such waves.

Recommended: Physics of Ultrasound

Image focus

The focus is placed at the level where the one examines is located. One can choose to place one or more focus (multiple focus lowers the image update rate).

The direction and focus of ultrasound waves can be adjusted by varying the sequence of activation of the piezoelectric crystals (Figure 2). If the activation starts at the lateral crystals and proceeds towards the center, then the ultrasound beam will be focused (Figure 2C). The focus can be placed anywhere along the ultrasound field.

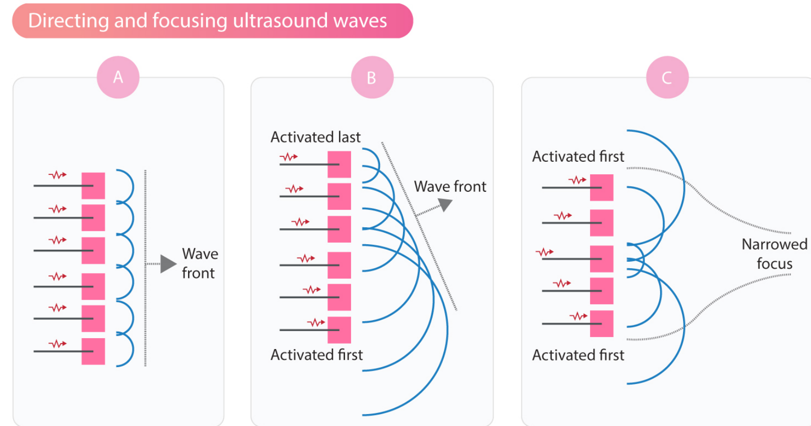


Figure 2. Directing and focusing the ultrasound waves. (2A) and (2B) illustrates how sound waves can be directed by varying the activation sequence. (2C) illustrates how the focus of the sound waves is adjusted.

Frame rate

Temporal image resolution is the ability to describe the movement of objects over time. Echocardiography requires high temporal resolution to study the detailed movements of relatively small objects. In order to produce recordings with high temporal resolution, it is critical to produce images rapidly. The more images that can be produced and presented per unit of time (i.e **frame rate**), the greater the temporal resolution.

It is possible to manually increase the frame rate (of any obtained image) to a certain extent. Frame rate is always increased when reducing the width and depth of the ultrasound image.

M-mode (motion mode) echocardiography

Chapter contents [Show]

M-mode echocardiography

M-mode was previously the dominating modality in echocardiography. Although it has now largely been replaced by 2D echocardiography, it is still used in clinical practice. M-mode provides a one-dimensional view of all reflectors (i.e structures reflecting ultrasound waves) along one ultrasound line. Hence, the M-mode image displays all structures along one line (Figure 1).

M-mode (Motion mode)

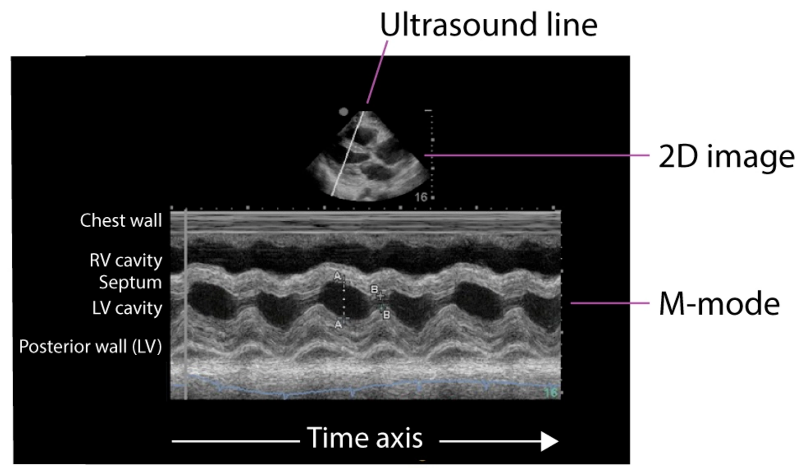


Figure 1. M-mode.

M-mode images are acquired by manually placing an ultrasound line in the 2D image (Figure 1). The line is placed along the structures to be studied. The image will display all structures along that line over time (the x-axis displays time). Since M-mode only analyzes a single ultrasound line, its temporal and axial resolution is very high, as compared with 2D echocardiography. M-mode is useful for quantifying the mobility of structures and measuring dimensions. In order to obtain representative measurements, it is pivotal to align the M-mode line such that it does not overestimate distances; e.g measuring the thickness of the left ventricular walls requires the line to be placed perpendicular to the long axis of the left ventricle, as illustrated in Figure 1.

One obvious disadvantage of M-mode is that it only displays a single ultrasound line. Moreover, the ultrasound line is fixed to the tip of the transducer, frequently making it difficult to obtain representative sections of structures of interest.

M-mode can be combined with Doppler techniques (Color Doppler, Tissue Doppler).

Usage of M-mode

M-mode is commonly used to complement the examination of the following structures:

- Left ventricular (LV) dimension and function.
- Right ventricular (RV) dimension and function.
- RV function can be assessed using TAPSE (Tricuspid Annular Plane Systolic Excursion).
- Study the movement and opening of the aortic valve.
- Study the movement and opening of the mitral valve.
- Left atrial dimension.

TAPSE (Tricuspid Annular Plane Systolic Excursion)

TAPSE is traditionally measured by placing the M-mode cursor at the lateral tricuspid annulus from the apical four-chamber view. TAPSE provides a rough estimate of RV function by measuring the longitudinal shortening of the right ventricle. TAPSE does not, however, account for radial shortening of the right ventricle.

Doppler effect and Doppler echocardiography

Chapter contents [Show]

The Doppler effect

When sound waves hit objects some of the sound waves are reflected back to the sound source. If the reflector (i.e the object reflecting the sound waves) is stationary, then the reflected sound waves will have the same frequency as the sound waves emitted by the sound source. If the reflector is in motion, however, then the frequency of the reflected sound waves will differ from the emitted sound waves. The change in frequency is referred to as the **Doppler effect.**

The Doppler effect was first described in 1843 by the Austrian astronomer Christian Doppler. It can be illustrated by studying how the frequency of reflected sound waves are modified by the direction of movement of the sound source. Figure 1 presents three trumpets; one placed on a table, and two are mounted on ambulances driving towards and away from the observer. When the sound source moves towards the observer, the sound waves are compressed, which leads to a shortening of the wavelength and thus increased frequency. When the sound source moves away from the observer, the sound waves are stretched out, which results in increased wavelength and decreased frequency.

The Doppler principle is primarily used to study blood flow and myocardial motion.

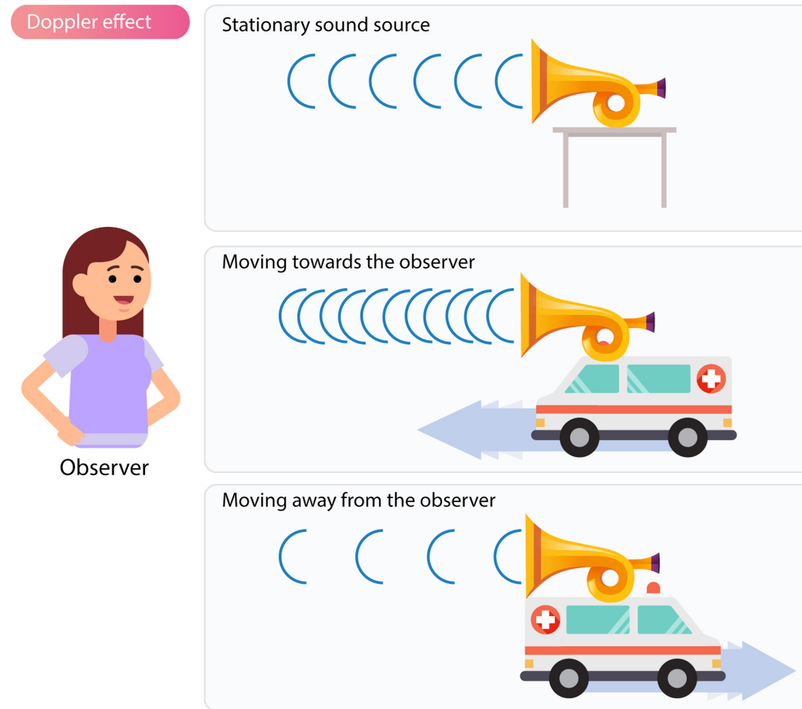


Figure 1. The Doppler effect. When the sound source moves towards the observer, the sound waves are compressed, which leads to a shortening of the wavelength and thus increased frequency. When the sound source moves away from the observer, the sound waves are stretched out, which results in increased wavelength and decreased frequency. The same principles can be applied to blood flow and tissue motions.

The sound source in echocardiography (i.e the transducer) is stationary. The moving objects are instead the blood cells (primarily erythrocytes) and tissues (primarily myocardium). The Doppler principle, however, remains unchanged: when the sound source and reflectors move towards each other, sound waves are compressed and *vice versa*.

Erythrocytes reflect ultrasound waves. Because erythrocytes are small, round and have an irregular surface, the reflected sound waves are scattered in all directions (Figure 2). Although only a fraction of the sound waves are reflected back to the transducer, the billions of erythrocytes in the blood will collectively generate enough reflections to be detected and analyzed by the ultrasound machine.

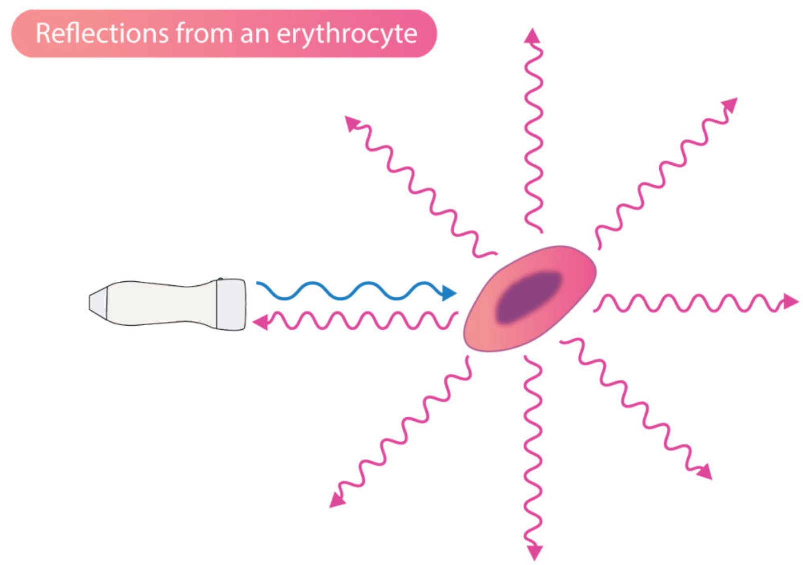


Figure 2. Reflections from an erythrocyte.

Flowing erythrocytes will alter the frequency of reflected sound waves. Erythrocytes flowing towards the transducer will reflect the sound waves with higher frequency. Erythrocytes flowing away from the transducer will reflect sound waves with reduced frequency (Figure 3).

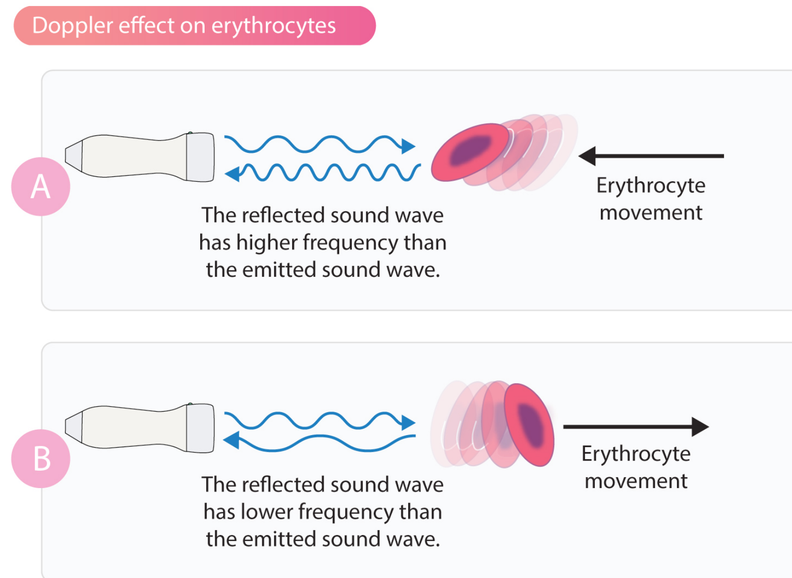


Figure 3.

Doppler effect occurs when reflectors (structures reflecting sound waves) move towards or away from the transducer. Objects moving towards the transducer will compress the sound waves and reflect them at a higher frequency. Objects moving away from the transducer will generate reflections with lower frequency.

Doppler shift

The Doppler effect is utilized to calculate velocity and direction of moving objects. To calculate the velocity of blood flow, the frequency difference between emitted and reflected ultrasound waves is analyzed. This difference is called **Doppler shift**. The Doppler shift depends on the velocity of blood flow (v), the frequency of the emitted ultrasound (f_u), the frequency of the reflected ultrasound (f_r), the ultrasound velocity in the tissue (c) and the cosine of the angle between the direction of blood flow and the reflected ultrasound wave (cos θ). The Doppler equation follows:

$$v = [c \cdot (f_r - f_u)] / [2 \cdot f_u \cdot \cos \Theta]$$

Significance of the angle of insonation

Doppler calculations are highly dependent on the angle of insonation. It is crucial that the ultrasound waves are directed parallel to the direction of blood flow or tissue motion. Ideally, there should be no angle (0°) between the ultrasound beam and the direction of blood flow or tissue motion.

When the ultrasound waves and the direction of movement are parallel, the angle is 0° and cosine 0° is equal to 1. If the angle increases, then the cosine of the angle will be less than 1, which will lead to an underestimation of the velocity. Thus, all angle errors lead underestimation of velocities (Figure 4).

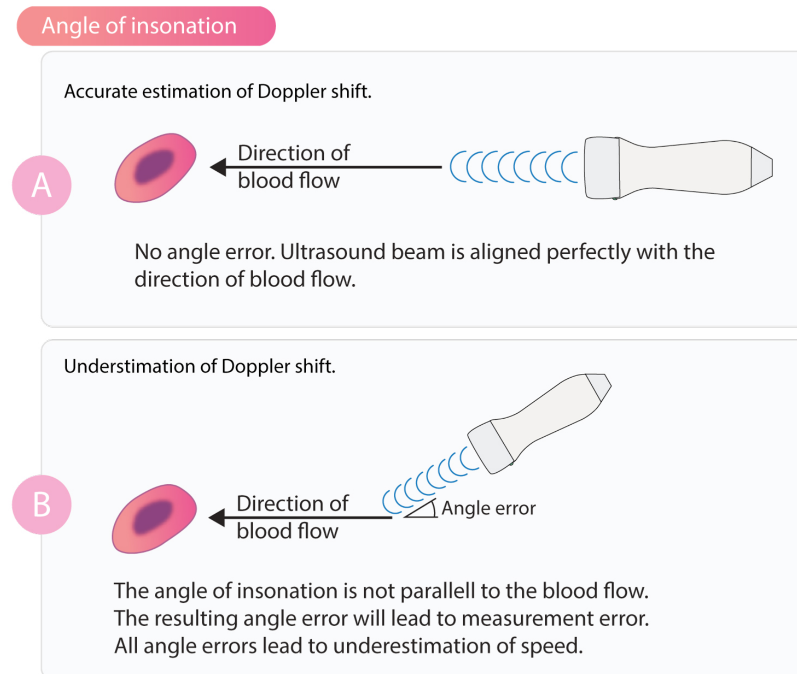


Figure 4.

In clinical practice, it is frequently difficult to obtain an ideal angle. However, small angle errors are without significance. For example, cosine 10° is equal to 0.98, and cosine 20° is 0.94. This implies that small angle errors have a negligible impact on the calculations.

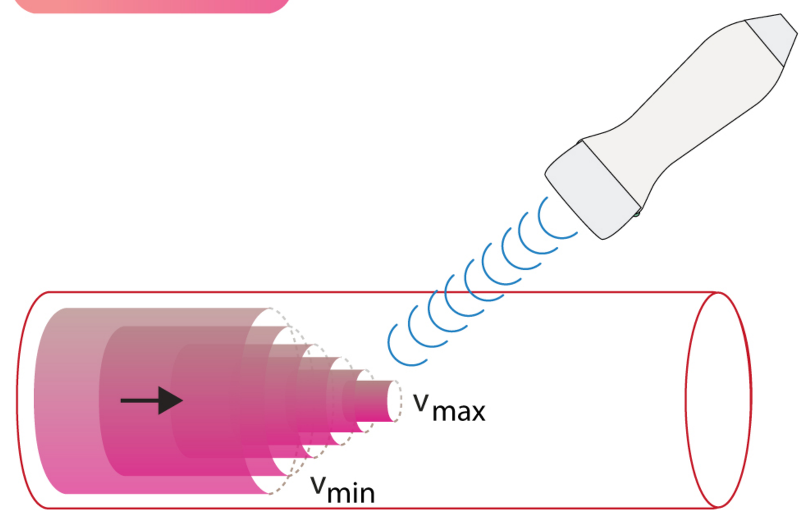
The 2D image is used to correctly align the ultrasound beam along the direction of movement. This is not, however, always straight forward. There may be a discrepancy between the 2D image and the optimal

Doppler signal; the best 2D image may offer a poor angle of insonation for Doppler measurements and vice versa. In such situations, one should prioritize the quality of the Doppler signal (i.e the amplitude of the signal and the angle of insonation).

Spectral Doppler analysis

Laminar blood flow

Blood flow is laminar throughout the circulatory system. This implies that blood flows in concentric layers with varying velocities. The highest velocity (v_{max}) is found in the center of the vessel. The lowest velocity (v_{min}) is found along the vessel wall. This yields a parabolic flow profile, as illustrated in Figure 6. Laminar flow is most pronounced in long, straight blood vessels, under steady flow conditions.



The advantage of laminar flow is its preservation of kinetic energy. The concentric layers and the parabolic flow profile reduces the energy losses by minimizing viscous interactions between the adjacent layers and the wall of the vessel. Disruption of laminar flow leads to turbulence and increased energy losses.

Doppler spectrum

Due to the laminar flow, erythrocytes passing any section of a vessel have different velocities. Moreover, blood flow is pulsatile, peaking during systole and reaching a minimum during diastole. Laminar flow and pulsatility result in reflected waves displaying large variations in Doppler shifts. This variation is called the **Doppler spectrum**.

On the echocardiogram, the Doppler signal is presented with a colored band or area (Figure 7). The colored area contains all the velocities recorded in a selected area during a specific phase of the cardiac cycle. The stronger the Doppler signal, the denser the spectral curve on the echocardiogram.

Figure 7. (A) Doppler recording in the left ventricular outflow tract (LVOT). Doppler signals are recorded in one point and (B) displays the resulting spectral Doppler, which shows all velocities recorded in the measuring point.

Presentation of the spectral curve

Figure 7 shows the presentation of Doppler signals on the ultrasound image. The type of Doppler shown in Figure 7 is called pulsed wave Doppler (discussed later). It is conventional that velocities (*i.e* blood flow or myocardial movements) in the direction towards the transducer yields a signal above the baseline and velocities away from the transducer are depicted with signals below the baseline. The x-axis displays time, and the y-axis displays velocity (m/s). As also shown in Figure 7, it is necessary to manually direct the Doppler line. This is done using the 2D image to align the Doppler cursor.

Figure 7. Presentation and interpretation of Doppler signals.

The Doppler shift is audible

Although ultrasound is not within the audible range for humans, it is possible to hear the Doppler shift. This is due to the fact that the Doppler shift, i.e the difference between the emitted and reflected sound waves, falls within the frequency range that humans can hear. The Doppler shift is the swishing sound from the speakers of the ultrasound machine.

The next chapter discusses different types of Doppler studies.

Pulsed Wave Doppler

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Pulsed Wave Doppler

The pulsed wave Doppler (PW Doppler) sends short pulses of ultrasound and analyzes reflected sound waves between the pulses. This is accomplished by using the same piezoelectric crystals to send and analyze sound waves. The crystals alternate rapidly between sending and analyzing ultrasound. Therefore, emitted sound waves can be associated with reflected sound waves, making it possible to determine the distance of the reflector (*i.e* the structure reflecting the sound wave).

The pulsed wave Doppler can analyze sound waves reflected from a specific location. This is the main advantage of pulsed wave Doppler, namely its ability to determine the location of the measured velocities. However, the pulsed wave Doppler requires time to analyze reflected sound waves. This is due to the fact that the same piezoelectric elements are used to send and analyze sound waves. This reduces the maximum velocity that can be measured using pulsed wave Doppler. Generally, velocities above 1.5 m/s to 1.7 m/s cannot be measured correctly.

Pulsed wave Doppler vs. continuous wave Doppler

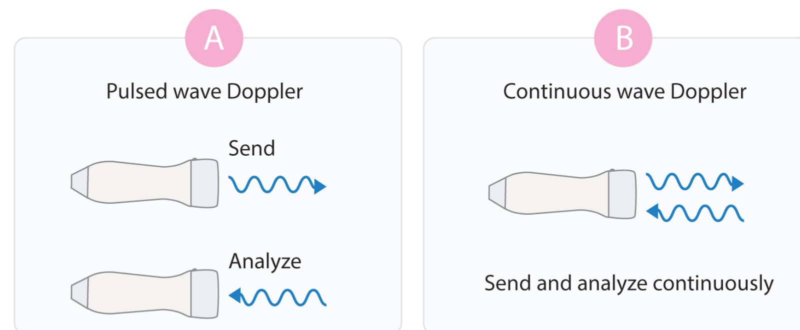


Figure 1. The difference between pulsed wave (PW) and continuous wave (CW) Doppler. PW Doppler sends short pulses of ultrasound and analyzes reflected sound waves between the pulses. CW Doppler sends and analyses ultrasound continuously.

Sample volume (SV)

The major advantage of pulsed wave Doppler is the ability to specify where (along the Doppler line) to measure velocities. This is possible because the pulsed wave Doppler sends and analyses sound waves sequentially. The ultrasound machine is programmed to ignore all signals, except those reflected from a certain depth. The depth can be determined since the speed of ultrasound is constant in the body. The investigator specifies where the measurement should be performed by moving the **sample volume (SV)** along the Doppler line. The sample volume is depicted with two lines perpendicular to the Doppler line (Figure 2).

Pulsed wave Doppler

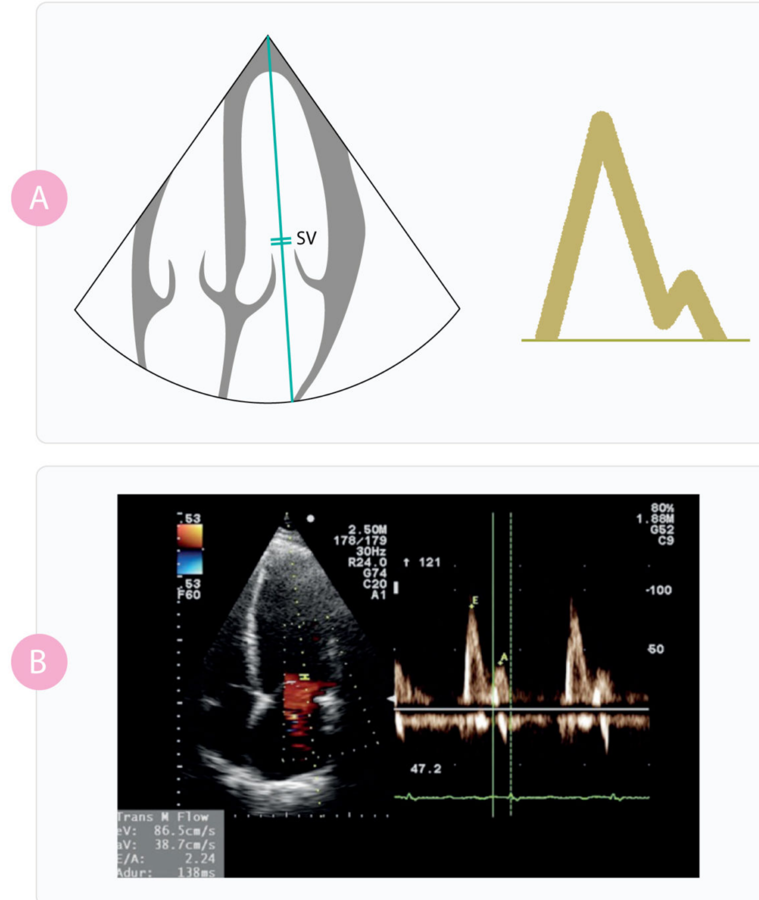


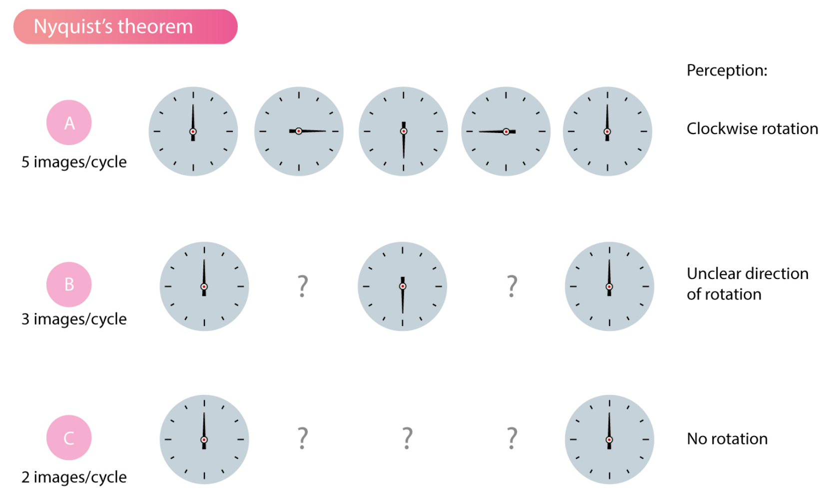
Figure 2. Location of sample volume (SV) and the resulting spectral curve (pulsed wave Doppler).

Pulse repetition frequency (PRF)

The number of ultrasound pulses sent per second is called **pulse repetition frequency (PRF)**. PRF is determined by the speed of sound and the distance it must travel. Since the speed of sound in the human body is constant (1540 m/s), the PRF only depends on the distance the sound waves must travel. The longer the distance, the more time required for sound waves to travel back and forth, which results in lower pulse repetition frequency (fewer ultrasound pulses can be sent per second).

> Pulse repetition frequency is inversely related to the distance the sound waves must travel. Visualizing distant structures result in lower pulse repetition frequency and consequently lower resolution. Visualizing proximally located structures enables the use of greater pulse repetition frequency, which results in greater resolution.

PRF must be high in order to assess the velocity and direction of blood flow, otherwise, the calculations will be uncertain. This is explained by the fact that each ultrasound pulse generates a snapshot of blood flow. The greater the number of snapshots per unit of time, the more accurate the description of blood flow. This is illustrated in Figure 3, which depicts a clock observed 5, 3 and 2 times during one cycle. As shown in Figure 3A, it is possible to determine with certainty the direction of the rotation with 5 observations per cycle. Using 3 observations per cycle, it is not possible to determine the direction of the rotation. With 2 observations per cycle, it appears that there is no rotation. This example illustrates the significance of high pulse repetition frequency in order to obtain accurate assessments of blood flow and myocardial movement.



Figur 3A-3C.

Nyquist's theorem and Nyquist limit

The importance of high PRF is explained mathematically by **Nyquist's theorem** (Harry Nyquist), which demonstrates that a wave must be sampled (*i.e* recorded) at least twice per cycle in order to be reliably measured. For pulsed wave Doppler, this implies that PRF must be at least twice the Doppler shift. Recall that the Doppler shift is directly related to the velocity of blood flow; the greater the velocity, the greater the Doppler shift. Thus, the maximum velocity that can be determined is half the PRF and this limit is called the **Nyquist limit**.

> The maximum velocity that can be determined has a Doppler shift that is half the PRF. Hence, PRF must be at least twice the Doppler shift.

Aliasing phenomenon

Aliasing occurs if the velocity of blood flow exceeds the Nyquist limit. This implies that the ultrasound machine cannot determine the velocity and direction of the flow. On the ultrasound image, the velocities exceeding the Nyquist limit will be presented on the opposite side of the baseline. Positive velocities (i.e velocities normally shown above the baseline) exceeding the Nyquist limit will be shown as negative velocities and *vice versa* (Figure 4 and Figure 5).

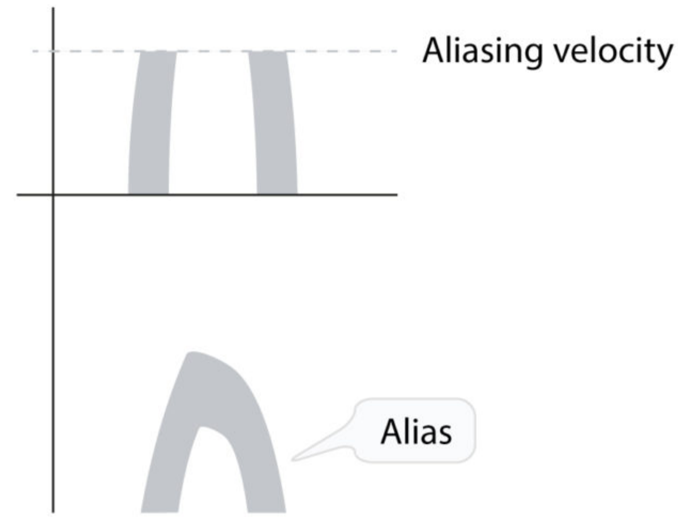
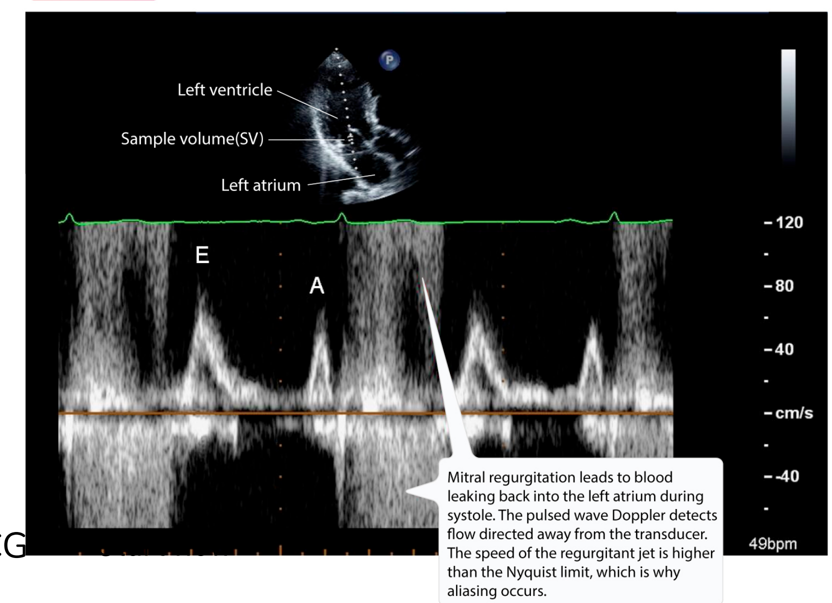


Figure 4.

Aliasing



Start learning ECG

Figure 5.

As mentioned above, the PRF depends on the depth being investigated. The depth is set by moving the sample volume along the Doppler line. The deeper the structures studied, the lower the PRF and, thus, the lower the maximum velocities that can be measured correctly, and *vice versa*.

Aliasing speed

It is straightforward to calculate the maximum velocity that can be measured using pulsed wave Doppler. Aliasing occurs when the velocity exceeds this maximum velocity (which is therefore referred to as the *aliasing speed* or *aliasing velocity*).

For example, at 15 cm depth, using ultrasound waves with a frequency of 3 MHz, the following equation calculates the time elapsed for sound waves to travel back and forth:

$$(0.15+0.15)/1540 = 0.0001948 \text{ seconds}$$

Where 0.15 is the one-way distance in m; 1540 is the speed of sound (m/s) in the human body

The pulse repetition frequency (PRF) is calculated as the number of sound waves that can be transmitted and reflected per second:

$$PRF = 1/0.0001948 = 5133 \text{ soundwaves per second} = 5133 \text{ Hz}$$

The Nyquist limit (the maximum Doppler shift that can be detected) is half the PRF:

$$5133/2=2566 \text{ Hz}$$

To calculate what flow velocity this corresponds, we use the Doppler equation:

$$v = (c \cdot (f_r - f_e)) / (2 \cdot f_e \cdot \cos \Theta)$$

We assume that the measurement is performed without any angle error, such that **cos θ** can be ignored. **f_e** is the frequency of the emitted sound waves and **f_r** is the frequency of the reflected sound waves. **f_r-f_e** equals the Doppler shift. **c** is the speed of sound (m/s) in the human body. The calculation follows:

$$v = (1540 \cdot 2566) / (2 \cdot 3000000) = 0.66 \text{ m/s}$$

The maximum speed that can be measured is 0.66 m/s. If we increase the frequency of the emitted sound waves to 5 MHz, the maximum velocity that can be measured becomes:

$$v = (1540 \cdot 2566) / (2 \cdot 5000000) = 0.40 \text{ m/s}$$

It follows that we can reduce the frequency of the emitted sound waves to increase the aliasing speed; then aliasing occurs at higher velocities. It is also possible to adjust (by lowering or elevating) the baseline of the ultrasound image to reduce aliasing; doing this will adjust the PRF.

> Aliasing can be remedied by reducing the frequency of the ultrasound or increasing the PRF.

Extended range doppler (High PRF Doppler)

Pulsed wave Doppler analyzes reflections from a specific location (*i.e* the sample volume) along the Doppler line. The maximum velocity that can be calculated is determined by the pulse repetition frequency (PRF), which is determined by the distance between the sample volume and the transducer. By using multiple sample volumes, the pulse repetition frequency is increased (the pulses from different sample volumes are added) and thus the aliasing speed is increased. This is referred to as **high PRF Doppler** or **extended range Doppler**. The advantage of high PRF Doppler is that greater velocities can be measured. Unfortunately, using high PRF Doppler makes it difficult to determine the location of the velocities recorded. To alleviate this problem, sample volumes are usually placed in areas known to have low flow velocities, making it possible to determine the location of higher velocities. Use of several sample volumes are illustrated in Figure 6.

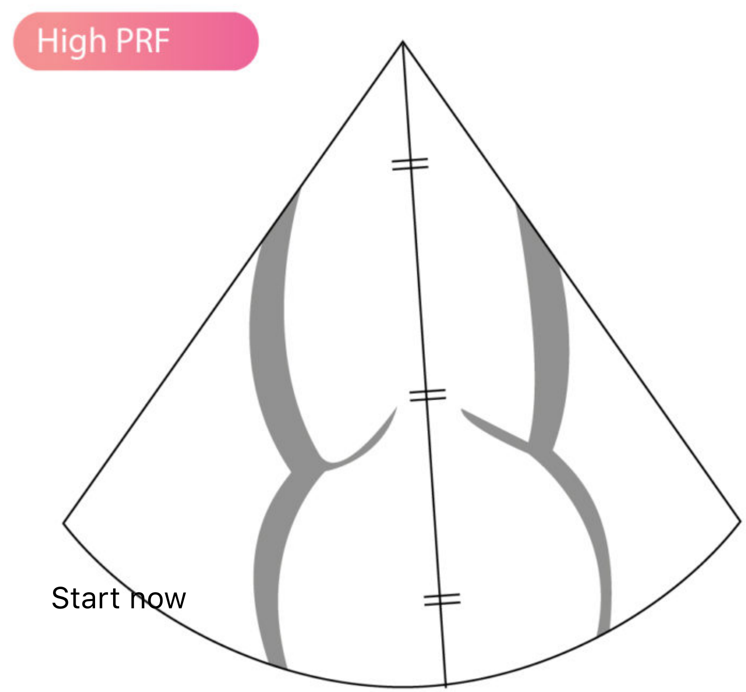


Figure 6. Extended range PRF (High PRF) with 3 sample volumes, two of which are located in areas with low flow rates.

Continuous Wave Doppler (CW Doppler)

Continuous Wave Doppler

In continuous wave Doppler (CW Doppler), ultrasound waves are continuously emitted from the transducer and the reflections of these waves are analyzed continuously (Figure 1). This is possible by using two different sets of piezoelectric crystals; one set for sending ultrasound and the other for analyzing reflected sound waves. The cardinal difference between continuous wave Doppler and pulsed wave Doppler is that ultrasound is emitted and analyzed continuously in the former. This allows much higher velocitites to be measured. Continuous wave Doppler is therefore not limited by pulse repetition frequency (PRF).

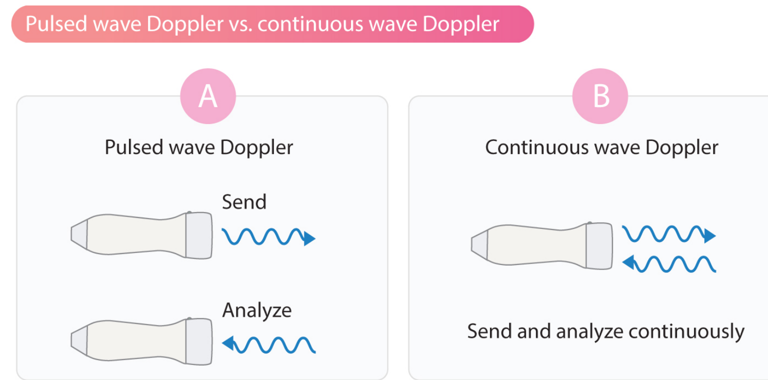


Figure 1. The principles of continuous wave Doppler and pulsed wave Doppler.

The disadvantage of continuous wave Doppler is that it is not possible to determine where, along the Doppler line, the velocities are recorded. The continuous wave Doppler yields a filled spectral curve (Figure 2), which is explained by the fact that all velocities (from zero to maximum) are recorded along the Doppler line.

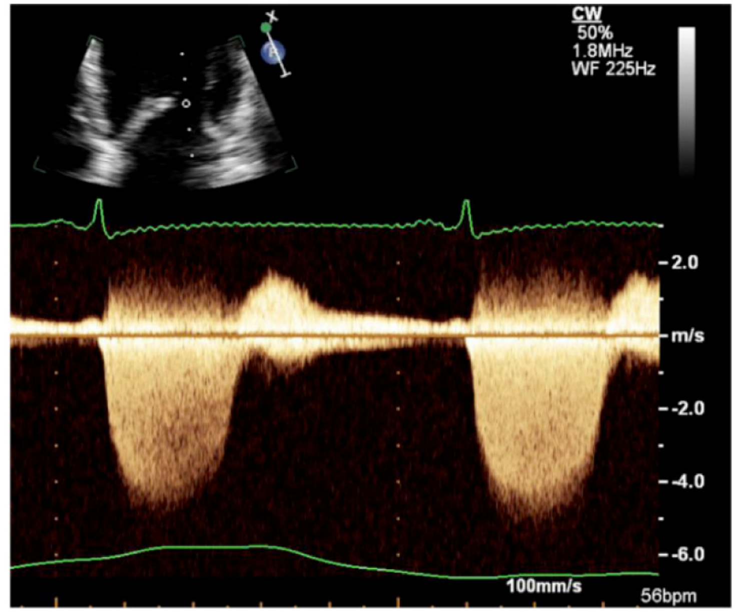


Figure 2. Continuous wave doppler along the left ventricle, the opening of the mitral valve and the left atrium. Recall that velocities (flows) directed away from the transducer are displayed below the baseline and velocities towards the transducer are displayed above the baseline. This image shows a fast and distinct flow (almost 5 m/s) directed away from the transducer during systole (systole starts at the peak of the R-wave on the electrocardiogram). This Doppler signal represents a pronounced mitral regurgitation.

In summary, the continuous wave Doppler cannot determine the location of the maximum velocity recorded, but it allows for the recording of very high velocities along the Doppler line.

Color Doppler

Color Doppler

Velocities recorded in a sample volume of the pulsed wave Doppler can be presented with a color. A color scale from blue to red is conventionally used. Blue color implies velocities (movement) away from the transducer and red color implies velocities (movement) towards the transducer. If many sample volumes are placed along several Doppler lines, then all velocities in the area can be presented with colors. The brighter the color, the higher the velocity. As shown in Figure 1, the Doppler sector is superimposed on the 2D image to facilitate the interpretation of the Doppler signals.

Color Doppler

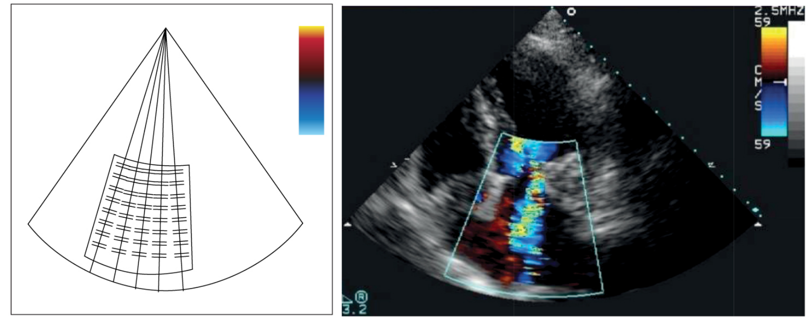


Figure 1. The image on the left shows many sample volumes within a Doppler sector. The picture on the right shows color Doppler located across the mitral valve and the left atrium.

The main advantage of color Doppler is that it allows for rapid visualization of flows, velocitites and volumes. This is useful for detecting valvular regurgitation and defects in the atria or ventricles (Figure 2). In addition, color Doppler can be used to align the continuous Doppler.

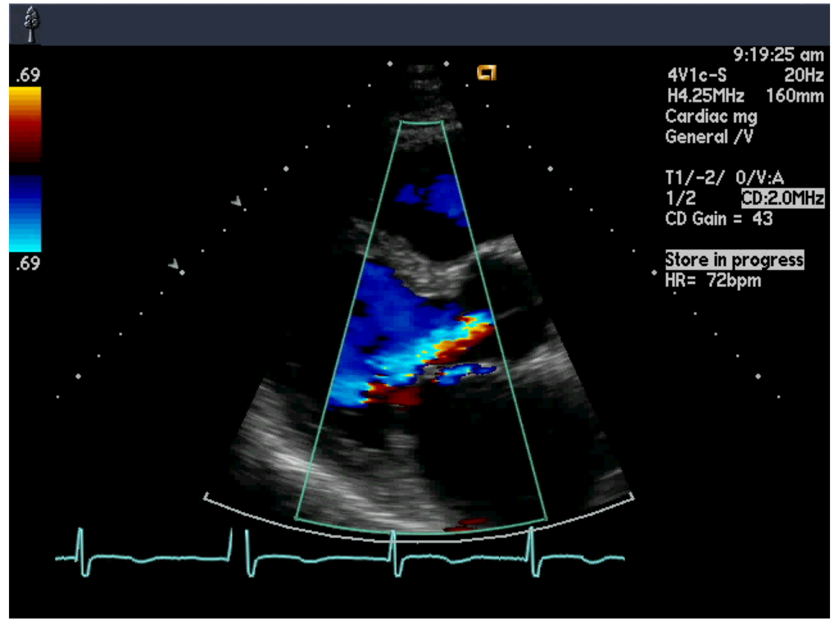


Figure 2. Color Doppler reveals the direction and extent of aortic valve regurgitation.

Because color Doppler is a type of pulsed wave Doppler, it is limited by the Nyquist limit. In fact, color Doppler is limited more by the Nyquist limit (as compared with standard pulsed wave Doppler), which is explained by the fact that the pulse repetition frequency (PRF) is reduced when obtaining both a 2D image and Doppler signals simultaneously. If the blood flow velocity exceeds the Nyquist limit, aliasing occurs and the signal changes color (blue turns red, and red turns blue). Aliasing usually occurs at speeds above 0.5 m/s. Aliasing can be reduced by minimizing the distance between the transducer and the color sector (where sample volumes are recorded) and by using the smallest possible sector.

Note that color Doppler presents the average velocity in each sample volume (*i.e* not the maximum velocity). Large variations in velocities recorded within a single sample volume indicate turbulent flow; the ultrasound machine is programmed to depict such flows with green color to indicate that the flow is turbulent. Figure 1 displays green areas within the Doppler sector.

Tissue Doppler (Tissue Velocity Imaging)

Chapter contents [Show]

Tissue Doppler (Tissue Velocity Imaging)

Previous chapters on Doppler imaging have all focused on measurements of blood flow. However, the Doppler effect can also be used to study myocardial motion. Myocardial motion during systole and diastole alter the frequency of ultrasound waves that are reflected back to the transducer. There are two fundamental differences between blood and myocardium in terms of the reflected ultrasound waves. (1) The velocity of myocardial motion is significantly lower than the velocity of blood flow.

Therefore, ultrasound waves reflected from myocardium will have a lower Doppler shift, as compared with waves reflected from erythrocytes. (2) Whereas reflections from erythrocytes have low amplitude, sound waves reflected from myocardium have high amplitude. This is due to the high density of myocardium. It follows that myocardium generates reflections with high amplitude and low Doppler shift. To analyze reflections from the myocardium, the ultrasound machine filters out all other reflected sound waves, such that only those representing myocardium are recorded.

Pulsed tissue Doppler

Pulsed tissue Doppler uses a sample volume where the velocity is recorded (Figure 1).

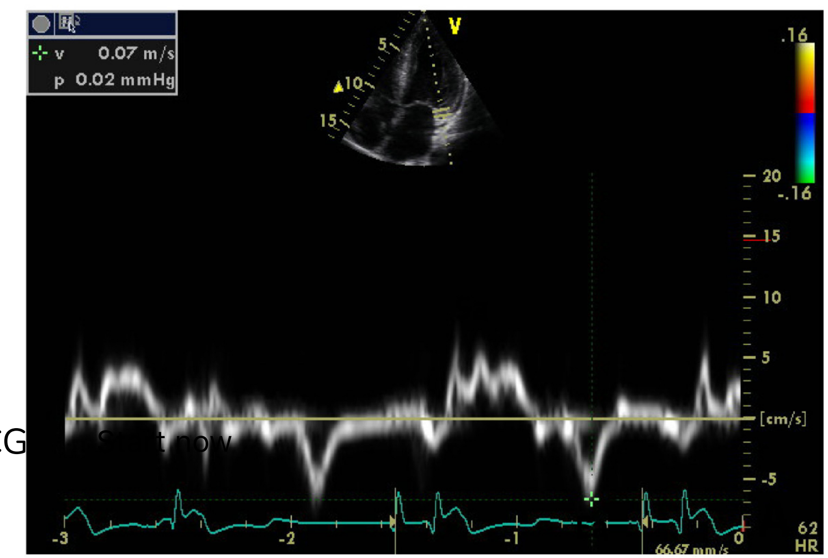


Figure 1. Pulsed tissue doppler with sample volume in the mitral valve plane.

Color Tissue Doppler

Color tissue Doppler analyzes myocardial velocities within a color sector. Myocardium moving towards the transducer is colored red, and myocardium moving away from the transducer is colored blue. The advantage of color tissue Doppler is that all myocardium is analyzed simultaneously, which makes it possible to compare myocardial regions (Figure 2).

It is important to note that the apex of the heart is fixed to the diaphragm via pericardium and connective tissue. Therefore the apex does not move much during the cardiac cycle, despite the fact that cells in the apex contract as much as the cells in the basal parts. Since the apex is fixed and there are longitudinal muscle fibers stretching from the apex to the basal parts, it appears as if basal regions are pulled down towards the apex. The longitudinal muscle fibers generate the *longitudinal contraction* (or *longitudinal shortening*) during systole.

It is also important to note that the velocity recorded in one region does not depend on the function in the region, but rather the function (contractility) of all myocardium located apically to the point of measurement.

Figure 2. Tissue Doppler where velocity is measured in four points.

Tissue Tracking

Tissue tracking is used to calculate the distance that the myocardium moves during the cardiac cycle. The predominant technique for tissue tracking is speckle tracking, which is discussed later.

Artifacts in ultrasound imaging

Chapter contents [Show]

Ultrasound artifacts

The following artifacts are common in ultrasound imaging:

- The ultrasound image displays non-existing structures.
- The ultrasound image does not display existing structures.
- The ultrasound image misrepresents the echogenicity of structures.

Echogenicity is defined as the intensity of reflected sound waves. Structures with high echogenicity will reflect more ultrasound and appear brighter on the image. Structures with low echogenicity reflect less ultrasound and become darker in the image.

Acoustic shadowing

Some structures have very high echogenicity (e.g skeleton, calcifications, mechanical heart valves) and reflect virtually all sound waves, leaving too few waves to explore the area behind the reflector. This results in dark areas, which are referred to as *acoustic shadows*. Figure 1 shows acoustic shadowing bellow a gallstone.

Acoustic shadowing

Figure 1. Gallstone causing acoustic shadowing.

Reverberations artifacts

Ultrasound waves can be reflected multiple times between dense structures (*i.e* structures with high echogenicity). For each reflection, a portion of the sound waves return to the transducer and produce a copy of the reflector on the image. Hence, the ultrasound image may display multiple copies of a dense structure. Such artifacts are referred to as *reverberations*.

Reverberations can also occur within a structure that has boundaries with high echogenicity. Then the sound waves can be reflected multiple times between the boundary layers, as illustrated in Figure 2.

Reverberation artifact

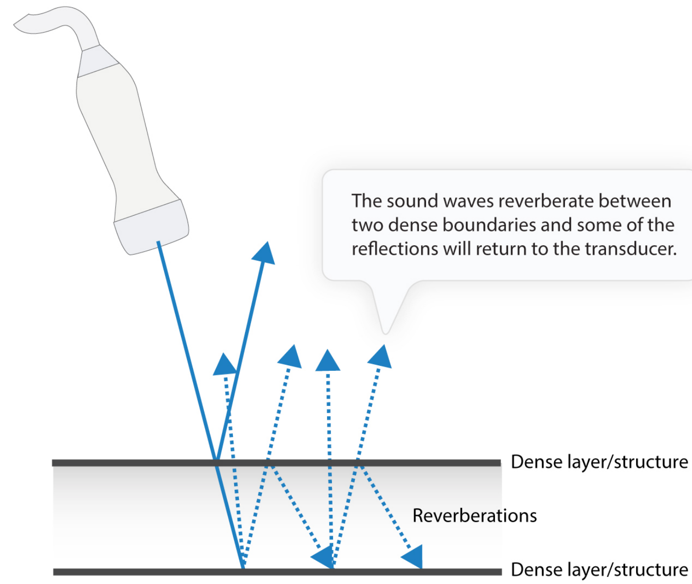


Figure 2. Reverberation artifact.

Reverberations can also occur if sound waves returning to the transducer are reflected back to the tissue.

Reverberations are common when examining lung tissue; the double-layered pleura produce reverberations, which are referred to as **A-lines**.

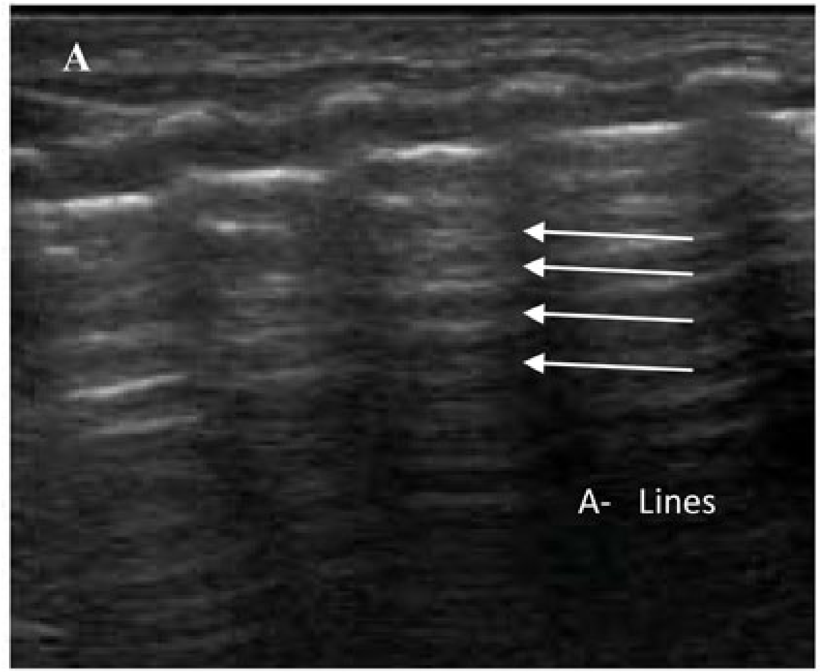


Figure 3. A-lines.

Mirror image artifact

Mirror image artifacts occur under a strong reflector that acts as a mirror. Behind the mirror, a copy of a structure appearing in front of the mirror is shown. The mechanism behind mirror image artifacts is similar to that of reverberations.

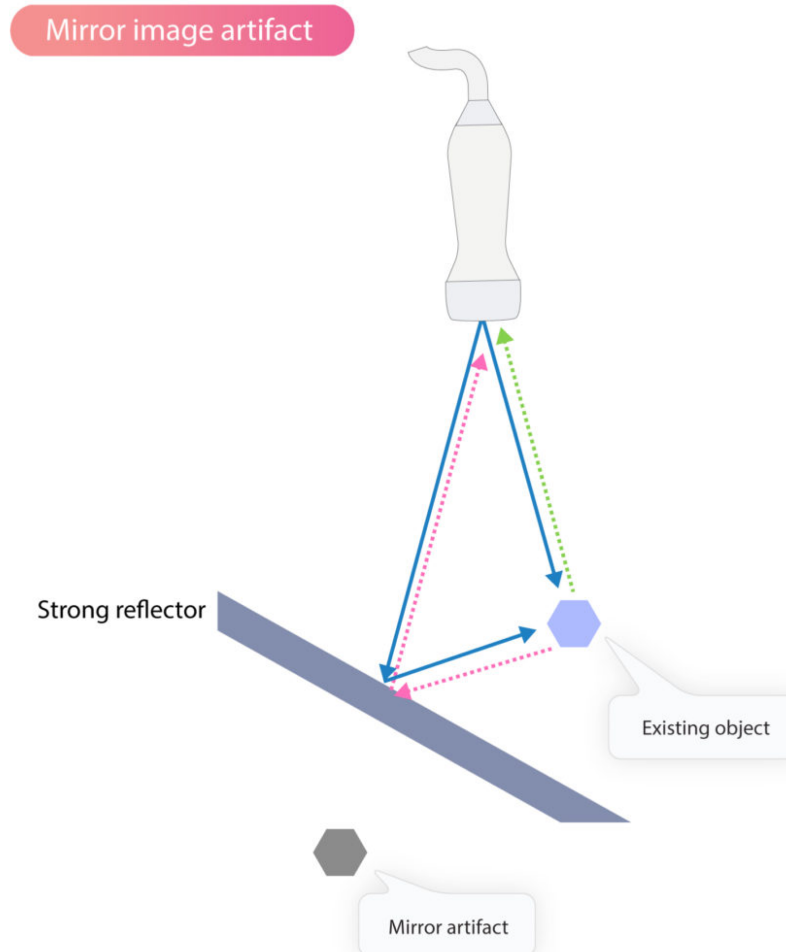


Figure 4. Mirror image artifact.

Side lobe artifact

A 2D image is formed by allowing the ultrasound beam to sweep back and forth within a defined sector. The transducer registers reflections originating from the central ultrasound beam (main beam). However, some ultrasound waves may travel off-axis in so-called *side lobes* (Figures 5A and 5B). Ultrasound energy in side lobes is mostly dissipated in the tissue without generating significant reflections. However, when side lobes encounter strong reflectors (calcifications, pericardium, mechanical heart valves, wires, *etc*), they may generate significant reflections which are detected by the transducer. These reflections are interpreted as originating from the main beam. As the ultrasound beam sweeps back and forth, multiple side lobe artifacts can be generated on both sides of the true reflector. If many side lobe artifacts are generated, they may appear as a continuous structure, as illustrated in Figure 5C.

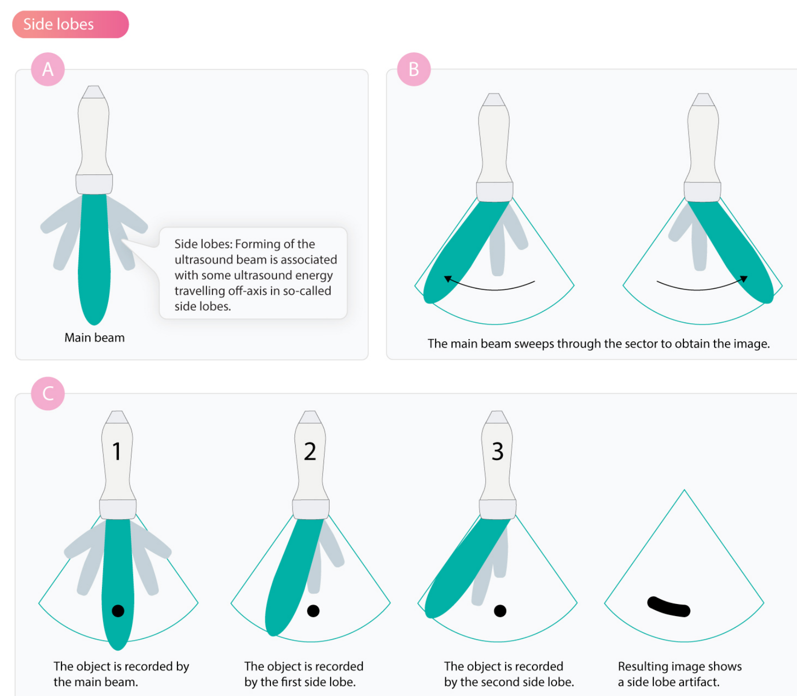


Figure 5A – 5C. Side lobes and side lobe artifacts.

Refraction artifact

Reflection and refraction occur when ultrasound passes tissue boundaries. Tissue boundaries represent acoustic reflectors at which some of the ultrasound energy is *reflected* and the remainder continues through the tissues. Depending on the difference in acoustic impedance between the tissues, the angle of the ultrasound wave may be altered. This is referred to as *refraction*. The larger the difference in acoustic impedance the greater the refraction.

Refraction artifacts occur when ultrasound travels through tissue that behaves as a lens which causes significant refraction, directing the ultrasound to an area interrogated simultaneously by other sound waves (Figure 6). The refracted ultrasound is then reflected back to the lens, from where it is re-refracted to the transducer, resulting in a duplicate of the reflector. The duplicate will be depicted along the original path of the sound wave. Structures behind the lens may be invisible in the image; this is due to the fact that sound waves never reach them and instead they are overwritten by the duplicate.

Refraction artifacts are generally easy to recognize because they create implausible image findings, such as duplication of the ventricles or atria. Fat, pleura and pericardium are among tissues that can behave as lenses that cause refraction. Switching the image window or adjusting the angle of the transducer may remedy refraction artifacts.

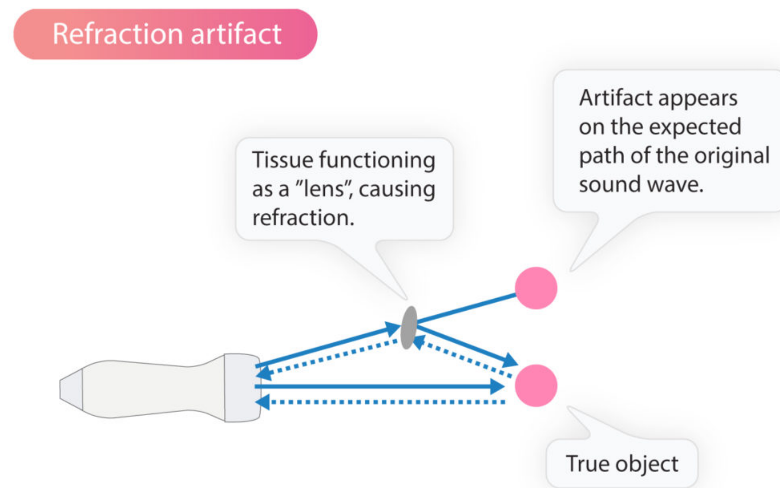


Figure 6. Refraction artifact.