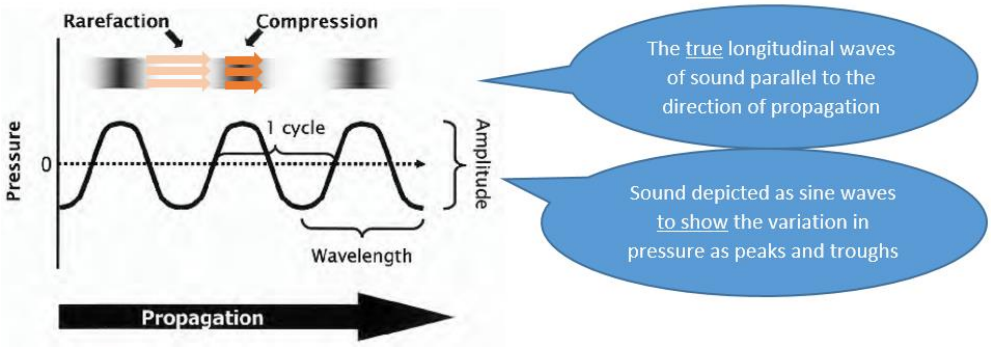
1. ULTRASOUND CHARACTERISTICS

SOUND is a mechanical vibration transmitted through an elastic medium. Unlike electromagnetic waves, such as visible light and radio waves, sound waves need the presence of particles to be transmitted.

ULTRASOUND includes that portion of the sound spectrum having a frequency > 20 KHz. Sound travels as a "longitudinal mechanical wave" that is a cyclic disturbance in which the energy propagates parallel to the direction of particle motion and can be thought of as a series of vibrating particles in a line. Even though, sound can be depicted as a sine wave to show the variation in pressure through the medium as peaks and troughs corresponding to areas of compression and rarefaction, respectively. As sound energy propagates through tissue, the wave has a fixed wavelength that is determined by the frequency and amplitude that is a measure of the magnitude of pressure changes.



WAVELENGTH: the length of a single cycle of the ultrasound wave (e.g. from one pressure peak to the next); a measure of distance, not time. It depends on the frequency and the medium in which the sound wave propagates. Wavelength is usually fixed and can't be changed by the operator

VELOCITY OF THE SOUND WAVE is the product of the wavelength and the frequency. Velocity is directly related to stiffness and inversely related to density. Ultrasound travels faster through a stiff medium, such as bone (*remember*: ultrasound does not penetrate bone or aerated lung well). Within soft tissue, velocity of sound is fairly constant at approximately 1540 m/sec

CALCULATION OF WAVELENGTH AND MAXIMUM DEPTH

Wavelength = Velocity ÷ Frequency
Thus, for a 5 MHz transducer, wavelength = 1540 ÷ 5,000,000 = 0.0,003 m = 0.3 mm

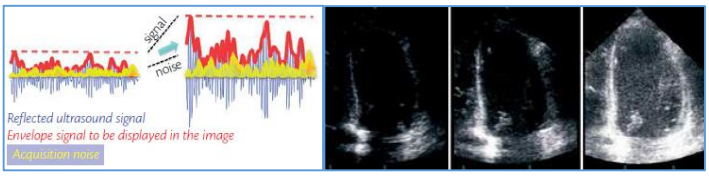
Maximum depth = 200 wavelengths
Thus, for 5 MHz transducers, maximum depth = 200 x 0.3 = 6 cm
N.B. lowering the frequency increases the maximum depth by the same percentage. i.e. lowering the frequency from 5 MHz to 1 MHz will increase the maximum depth by 5 folds (from 6 cm to 30 cm)

AMPLITUDE AND POWER

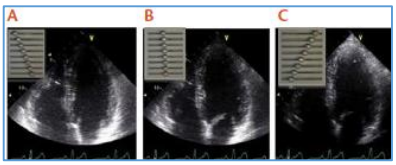
Amplitude (dB) is a measure of the strength of the sound wave (loudness). It is defined as the difference between the peak pressure within the medium and the average value, depicted as the height of the sine wave above and below the baseline. Amplitude is measured in decibels (dB), a logarithmic unit that relates acoustic pressure to some reference value. The primary advantage of using a logarithmic scale to display amplitude is that a very wide range of values can be accommodated and weak signals can be displayed along side much stronger signals. Of practical use, an increase of 6 dB is equal to a doubling of signal amplitude, and 60 dB represents a 1,000-fold change in amplitude. Amplitude delivered to the patient can be adjusted by the sonographer by changing the echo machine's transmit power to achieve better signal to noise ratio (SNR), but on account of adverse mechanical & thermal effects. Acoustic energy output is usually expressed in dB relative to the maximal energy output available on the system (100% output = 0dB; 50% reduction = −6dB).

Power (watts) is a parameter closely related to amplitude, which is defined as the rate of energy transfer to the medium, measured in watts. For clinical purposes, power is usually represented over a given area (often the beam area). Intensity (watts/cm^2) is analogous to loudness. It represents the concentration of power within an area. Intensity diminishes rapidly with propagation distance and has important implications with respect to the biologic effects of ultrasound**.** More energy → better signal → better image quality (i.e. better signal-to-noise ratio: SNR). Too much acoustic energy can result in tissue damage due to: heating (monitored by thermal index, TI, which should be <2) and cavitation (i.e. formation of small gas bubbles with subsequent bubble collapse associated with high pressures/temperatures locally, monitored by mechanical index, MI, which should be < 1.9).

GAIN AND DEPTH (TIME) GAIN COMPENSATION



Gain: amplitude of the returning signal to the machine is altered through gain. Increased gain amplifies the received signal to increase the brightness of the displayed spectral trace, but equally amplifies the noise → SNR remains identical. Neither the resolution nor the contrast change by altering the gain settings.



Depth gain compensation, also called time gain compensation (TGC) is depth-specific amplification of the echo signals to compensate for attenuation. It is either:

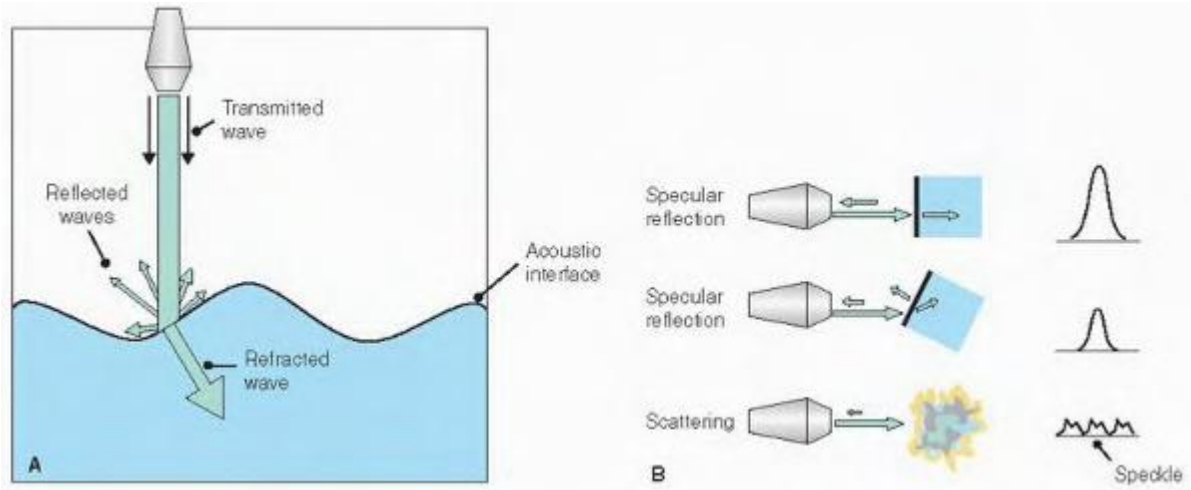
- **Automatic**: amplifies signals from deeper structures
- **Manual**: allows correction of the automatic compensation. It is recommended to start each examination with the sliders in their neutral (i.e. centre) position.

2. INTERACTION BETWEEN ULTRASOUND AND TISSUE

ACOUSTIC IMPEDANCE- the product of the density of the medium and the velocity of sound. Acoustic impedance is independent of frequency and cannot be changed by the operator as it is a property of the tissue itself.

ABSORPTION- the transfer of ultrasound energy to the tissue during propagation

REFLECTION AND REFRACTION- when the beam crosses a boundary between two tissues, a portion of the energy is reflected, a portion is refracted, and a portion continues in a relatively straight line.



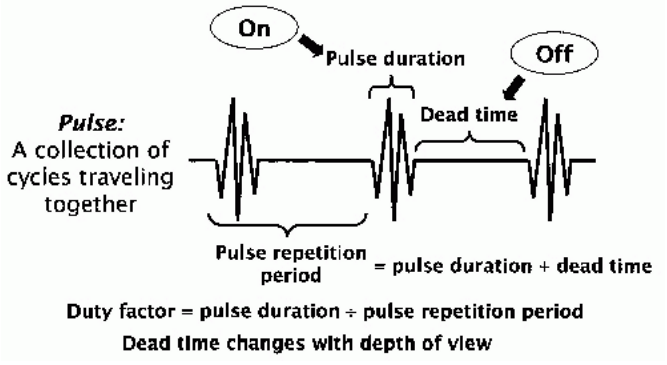
A transmitted wave interacts with an acoustic interface in a predictable way. Some of the ultrasound energy is reflected at the interface and some is transmitted through the interface. The transmitted portion of the energy is refracted, or bent, depending on the angle of incidence and differences in impedance between the tissues. **B:** The interaction between an ultrasound wave and its target depends on several factors. A **specular reflection** occurs when ultrasound encounters a target that is large relative to the transmitted wavelength. The amount of ultrasound energy that is reflected to the transducer by a specular target depends on the angle and the impedance of the tissue. Targets that are small relative to the transmitted wavelength produce a **scattering** of ultrasound energy, resulting in a small portion of energy being returned to the transducer. This type of interaction results in "speckle" that produces the texture within tissues.

As the red blood cells are much smaller than the wavelength of the ultrasound, red blood cells are effective scatterers and this form the principle of Doppler flow studies. This scatter is referred to as Rayleigh scattering, in which the ultrasound is uniformly scattered in all directions.

ATTENUATION- As ultrasound transverses tissue, it will gradually lose energy, a process known as attenuation. Attenuation always increases with depth and the higher the frequency, the more rapidly it will attenuate. Thus, at higher frequency the penetration depth decreases due to increased attenuation. Attenuation is greater with increasing mismatch in acoustic impedance, since a large proportion of the energy will be reflected rather than transmitted.

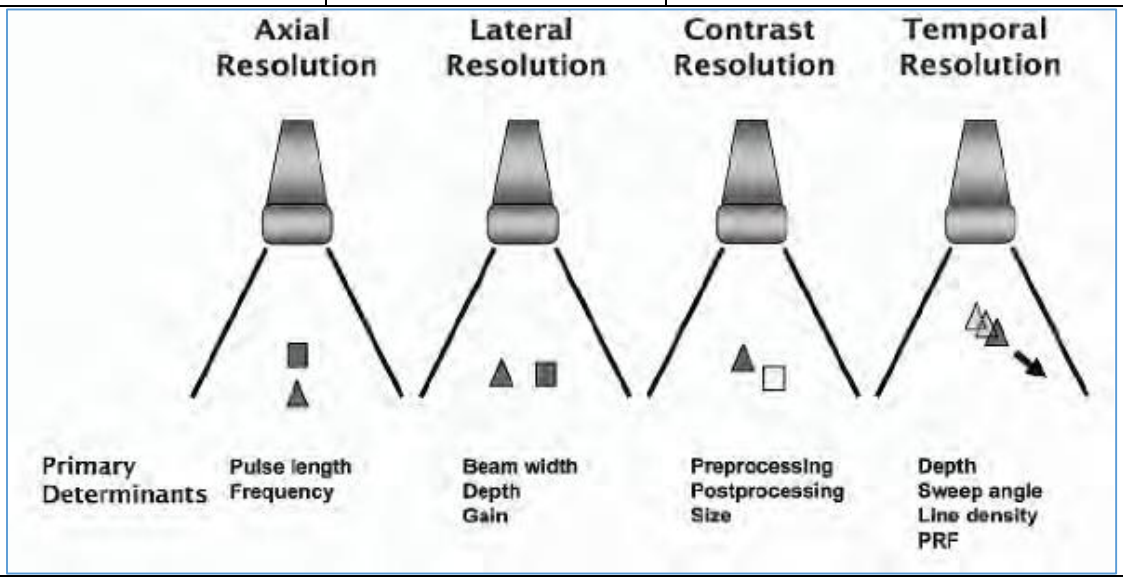
PULSE REPETITION

Ultrasound energy is usually emitted from the transducer in a series of pulses, each one representing a collection of cycles. Each pulse has a duration, or length and is separated from the next pulse by the dead time. The diagram is not drawn to scale. In reality, dead time is much greater than pulse duration.

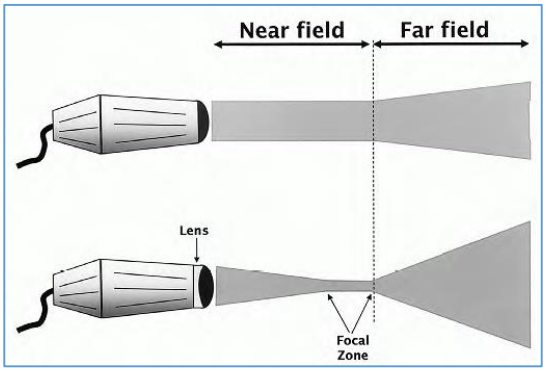


RESOLUTION- is the ability to distinguish between two objects in close proximity. Resolution has at least two components:

Types of resolution	Components	The primary determinants (how to improve it)
Spatial resolution (scan density)- is the ability to differentiate two structures lying close to each other. It is expressed as the smallest distance between 2 structures that allow the system to distinguish between them *(decreased resolution distance implies improved resolution)*. It has 2 components: Axial resolution and lateral resolution.	**Axial resolution:** the ability to differentiate 2 structures lying along the axis of the ultrasound beam (i.e., one behind the other)	*Reduced length of the wave &/or the pulse of waves* 1.↑**Frequency**: higher frequency is associated with shorter wavelength, and the size of the wave relative to the size of the object determines resolution. N.B: lowering frequency will activate harmonic imaging → reduced axial resolution, but also reduced noise (better S/N ratio) 2.↓**Pulse length/duration**: the shorter the train of cycles, the greater the likelihood that two closely positioned targets can be resolved.
	Lateral resolution: the ability to distinguish 2 structures lying side by side relative to the beam	1.↓**Beam width and depth**: lateral resolution diminishes as beam width and depth increase 2.↓**Gain**: when gain is low, weaker echoes from the edge of the beam may not be recorded and the beam appears relatively narrow. However, ↑ gain improves the signal brightness, but also increases the noise (no change in S/N ratio). Also ↓ frame rate improves the lateral resolution.
Temporal resolution (frame rate)- is the ability to distinguish events that occur close in time (i.e. the ability to track moving targets over time. It depends on the number of image frames that can be generated each second. N.B: higher the frame rate, the higher the temporal resolution, but the lower the lateral resolution		1.↓**Beam width and depth:** increased beam width and depth increases the time taken to collect all the data required to create one image and, hence, decreases the frame rate. 2.↑**PFR:** with sampling rates (frame rates) of 1,000 to 2,000 images per second, temporal resolution of M-mode is much higher than that of two dimensional imaging. 3.**Appropriate Line density** 4.**Appropriate Sweep angle**



ULTRASOUND FOCUSING- The shape and size of the ultrasound beam can vary, depending on several factors, including the design of the transducer and the inherent characters of ultrasound waves. Initially, the beam is columnar, but it gradually becomes divergent (less focused), and this divergence results in deterioration of the image quality. The shape of the beam can be altered to have a focal depth (narrowest point) by changing the surface of the piezoelectric crystal to be concave or by adding an acoustic lens. Focus can be fine-tuned with phased-array transducers and should be adjusted for each view so that the beam is focused on the region of interest. Around this region spatial (lateral) resolution is optimal. Generally, the focus should be placed at the level of the deepest structure of interest. Modern echo machines has the option of dual focus, which allows optimization of focus at 2 levels but at the expense of lowering the frame rate.

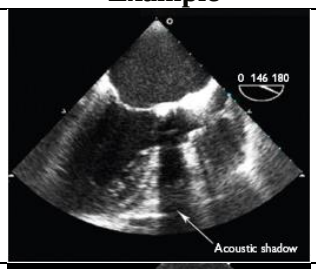
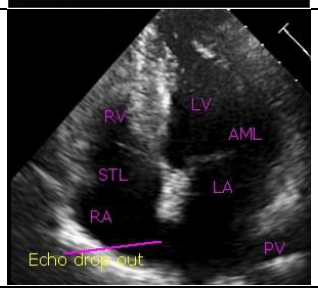
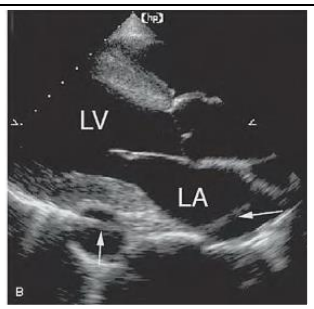
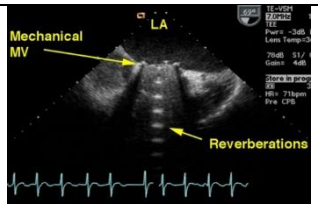


FRAME RATE (fps) is the number of frames created per second. Higher frame rate (better temporal resolution) will result in less lines in the image and thus worse spatial (lateral) resolution. The less the sector size width and the lower the depth, the higher the frame rate. It is recommended to Keep frame rate at its default value unless modifications are required for specific processing methodologies (e.g. speckle tracking analysis)

TRANSMIT FREQUENCY can be reduced (for the same transducer) to achieve better penetration, but at the expense of worse spatial (axial) resolution. Lowering the transmit frequency will activate harmonic imaging.

HARMONIC IMAGING is a technique that employs the resonance characteristics of tissue. It was discovered by accident when scientists working on the development of microbubbles for contrast found that a tissue image was created when the receiver was tuned to receive at twice the transmitted frequency, without injecting the contrast agent. This meant harmonics frequencies arises from soft tissue as well as from the contrast agent. When an ultrasound wave passes through the body, the tissue resonates (in a way similar to the contrast agent) and generates "harmonic ultrasound waves". The resonance frequency is typically a multiple of the transmitted (fundamental) frequency. The harmonic frequencies are reflected back to the transducer, where they are used to create the image. Specifically, the second harmonic frequency is used to create the image. All other harmonic frequencies and the fundamental component are filtered out. The received frequency of modern transducers is two-fold higher than the sending frequency. Tissue harmonic imaging allows emitting lower frequency (fundamental frequency), thus achieving better penetration, while processing the returning higher frequency (second harmonic frequency), achieving better image quality (better border definition) and fewer artefacts (better SNR), but at the expense of worse spatial (axial) resolution (valves appear thicker); therefore, the fundamental imaging is preferred for very fine structures such as valve leaflets. Another disadvantages of second harmonic imaging is that it requires a higher power output. Doppler recordings are not affected when using harmonic 2D imaging. It is generally recommended to (1) use harmonic mode as default setting and (2) keep the transmit frequency equal to the centre frequency of the probe unless (1) penetration is insufficient and no other probe is available or (2) switching between fundamental and harmonic imaging is required.

ARTIFACTS

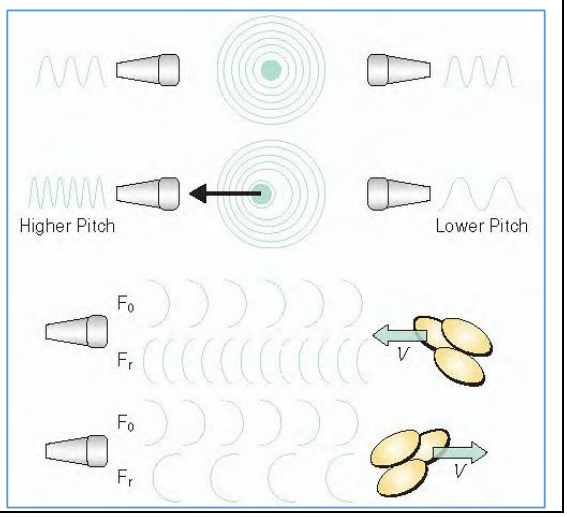
	Mechanism	Outcome	Example
1.Acoustic Shadowing *(You don't see behind)*	Strong reflector (prosthetic valve or calcium accumulation) → blocks ultrasound from penetrating any further.	echo drop-out in the far field.	
2.Parallel alignment *(You don't see behind)*	Parallel alignment → very little reflection back to the transducer (typically seen of the atrial septum in apical 4-chamber view)	echo drop-out	
3.Beam width artefact *(You see smear)*	Strong reflectors at the edge of the beam → displayed as though they arise from the centre of the beam. This can be reduced by focusing the ultrasound beam to minimize its diameter.	'smearing' the displayed echo	
4.Side lobes artefact *(You see ghost lateral)*	Portion of the ultrasound beam propagates radially to the side of the central beam → echo recorded via the off-axis side lobes will be interpreted by the machine as having arisen from the central beam and, therefore, will be displayed lateral to the true location of their source. A prerequisite is that the structure reflecting the side lobes must be a fairly strong reflector (the atrioventricular groove, the fibrous skeleton of the heart). The mechanism is similar to beam width artefact, but due to unavoidable side lobes. In contrast to grating lobes, side lobes are common to all transducers.	edge effect	 Bright echoes within the pericardium produce a linear artifact that appears within the descending aorta and the left atrium (*arrows*).
5.Grating lobes *(You see ghost very lateral)*	Signals returning from the grating lobes are displayed at large angles from the true location of their source (common to phased-array transducers only)		
6.Reverberations *(You see ghosts behind)*	Two strong reflectors (prosthetic valves, aortic plaques) → Ultrasound rebounds several times between before returning to the transducer. The time spent 'rebounding' delays the return of the signal to the transducer, and so the processing software misinterprets the returning signal as having originated further away that it really has.	'Ghost' images occurring in the far field (comet-tail of diminishing reverberations below the reflectors) that can be recognized because they move in tandem with the structure that caused the reverberation	

PRINCIPLES OF DOPPLER ULTRASOUND

THE BASIC PRINCIPLES OF THE DOPPLER PHENOMENON

	Two-dimensional Echocardiography	Doppler
Ultrasound target	Tissue	Blood
Goal of diagnosis	Anatomy	Physiology
Type of information	Structural	Functional
Optimal alignment between beam and target	Perpendicular	Parallel
Preferred transducer frequency	High	Low

The basic principles of the Doppler phenomenon are illustrated. **(Top):** Stationary source of sound produces a given pitch or frequency. If the sound is moving toward a recorder, the pitch appears increased and if the sound is moving away from a recorder, the pitch appears decreased. **(Bottom):** This same concept is applied to blood flow. If the red blood cells are moving toward the transducer at a given velocity (*v*), the reflected frequency (*F*r) will be higher than the emitted frequency (*F*0). If the red blood cells are moving away from the transducer, the opposite will occur.



CALCULATION OF DOPPLER FREQUENCY SHIFT (ΔF)

$$\text{Doppler frequency}(f_d) = \frac{2 \cdot f_t \cdot V \cdot \cos\theta}{c}$$

f_d = doppler shift
f_t = transmitted beam
c = speed of sound in tissue
V = velocity of blood flow
θ = angle of incidence between the ultrasound beam and the direction of flow.

CALCULATION OF BLOOD VELOCITY (V) using the Doppler frequency shift equation:

$$V = \frac{\Delta F \times 1540}{F_t \times 2 \cos\theta}$$

Simplified formula**:**
Velocity = (Frequency shift x 1500) / (Frequencytransducer x 2) … for angle 0º
Velocity = (Frequency shift x 1500) / (Frequencytransducer x 1) … for angel 60º
Thus, for a 5 MHz transducer at an angle of 60° to blood flow, the Doppler frequency shift is 10 kHz:
V = (10,000 x 1540) ÷ (5,000,000 x 2 x ½) ≈ 3 m/sec.
Remember:
cos0 = 1 → optimal velocity
cos90 = 0 → 0 velocity
cos60 = ½

DOPPLER MODALITIES

The main clinically relevant types of Doppler techniques for cardiac applications are CW Doppler, PW Doppler, colour Doppler, and tissue Doppler.

CW DOPPLER AND PW DOPPLER display graphic information on direction and velocity of flow against time. Velocity is displayed on the vertical axis with flow toward the transducer above the baseline and flow away from the transducer below the baseline. Under physiologic conditions, blood flow in the cardiovascular system is essentially laminar (blood cells are traveling at approximately the same speed, although velocity tends to be higher in the centre of a vessel and lower near the vessel wall). In pathologic situations, flow tends to become turbulent, often with abnormally high velocity. Purely laminar flow has a narrow envelope of velocities, indicating that most of the blood cells travel over a narrow range of velocity. With increasing turbulence, both the direction and the range of velocities increase, and this leads to a widening of the spectral pattern.

CW Doppler uses two crystals, one continuously transmit an ultrasound signal and the other continuously receive the returning signal. Since the sound is sent continuously there is no way to determine the time between emitted and reflected sound, therefore it can not separate velocities out (spatial localization of the abnormal velocities is lacking) and is not useful for assessing mid-cavity gradients, but this modality identifies highest (peak) velocity throughout the ultrasound without aliasing or Nyquist limit.

PW Doppler uses piezo-electric crystals to transmit intermittent bursts of ultrasound in a fashion similar to that of 2-D and 3-D imaging. The pulsed Doppler instrument "listens" only to returning signals from one specific distance from the transducer, a process called range resolution and, therefore, PW Doppler is used to identify velocity at varying depth, which the sonographer chooses by placing a sample volume (indicated by two parallel lines perpendicular to the main cursor line) at the point of interest. The optimal size of the sample volume is often in the range of 5-7 mm. Wider sample volume will resolve more velocities, but at lower spatial resolution and, hence, taking measurements might be difficult; whereas narrower volume will give good spatial resolution (crisp signal), but at lower velocity resolution and, hence, may not resolve all the velocities that go through that sample volume. Sample volume should be positioned at the tips of the (open) valve leaflets (for MV inflow).

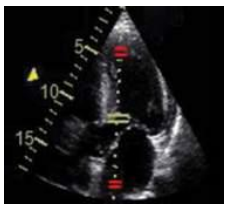
Optimal Doppler recordings require:
1. Proper alignment of the Doppler line with the flow direction (< 20° off-axis). It is recommended to reposition and angulate the probe under colour Doppler guidance to obtain optimal alignment.
2. Proper velocity scale (also referred to as Nyquist velocity or PRF). Too low scale results in aliasing, and too high scale results in sub-optimal velocity resolution (i.e. increase the smallest measurable difference between two different velocities). Velocity scale should be as low as possible without aliasing.
3. Average over five cardiac cycles in atrial fibrillation excluding extreme cycle length.

PRF is the number of pulses transmitted from a Doppler transducer each second. PRF is independent of ultrasound frequency and is determined by the source. PRF can be increased by reducing the depth of imaging (PRF = 77,000 ÷ depth**)**. The higher the pulse repetition frequency, the higher the frame rate and the better the temporal resolution.

Aliasing- is the inability of a PW Doppler to properly depict high velocities/frequencies exceeding the Nyquist limit. This occurs when the velocity/frequency exceeds the Doppler sampling speed. An analogy may be made with viewing a rotating fan. As long as the fan is rotating at a speed below the eye's natural sampling speed, it appears to be rotating clockwise. However, when the fan accelerates so that is rotating faster than the eye's natural sampling speed, it appears to be rotating counterclockwise.

Nyquist limit- the upper limit of frequency that can be detected by a given pulsed Doppler system, which is defined as one half the PRF, as according to Nyquist's theorem, in order to unambiguously measure wavelength, a waveform must be sampled at least twice in each cycle).

Measures to reduce aliasing:
- Shifting the zero velocity baseline (as much as possible)
- Increasing the Doppler velocity scale (the range of velocities displayed in the graph) as far as possible. This will reduce the size of the velocity curve and thus reduced aliasing, but should not be increase to the extent that does not allow accurate measurement of the velocity.
- Sampling at the lowest possible distance from the transducer
- Decreasing the transmitted frequency
- Increasing the angle of incidence.
- Ultimately it may prove necessary to switch to CW Doppler instead, where possible. One further alternative is to use high-PRF PW Doppler.



High PRF Doppler imaging- is a technique that combines features of both PW and CW Doppler imaging. With repeated manual increase of PW Doppler scale, the machine will at some point automatically shift to high PRF mode in order to achieve the desired range (On some machines, High PRF has to be enabled in the setup menus). High PRF Doppler allows PW to sample more than one site along the ultrasound beam, and velocity information from these different depths can all be analysed during a single listening event . The use of the shallower sample volume depth is associated with a higher PRF, which allows higher velocities to be analysed without aliasing. High PRF Doppler introduces some range ambiguity as to the exact location of the blood flow being sampled and so, the sonographer needs to try to eliminate ambiguity by avoiding the placement of the cursor in line with areas where higher velocities are likely to be present.

Wall filter: is a way of filtering out low or high frequency Doppler signals. It is usually used to filter out very low frequencies that may add noise to a spectral Doppler waveform. A typical use is removing the low frequency reverberation of an arterial wall. A potential pitfall is that the wall filter may filter out genuine low-frequency slow flow, such as venous blood flow. If the goal is to evaluate very weak signals such as venous blood flow, make sure the wall filter is set as low as possible. Generally, Wall filter should be as low as possible while avoiding pollution by myocardial velocities.

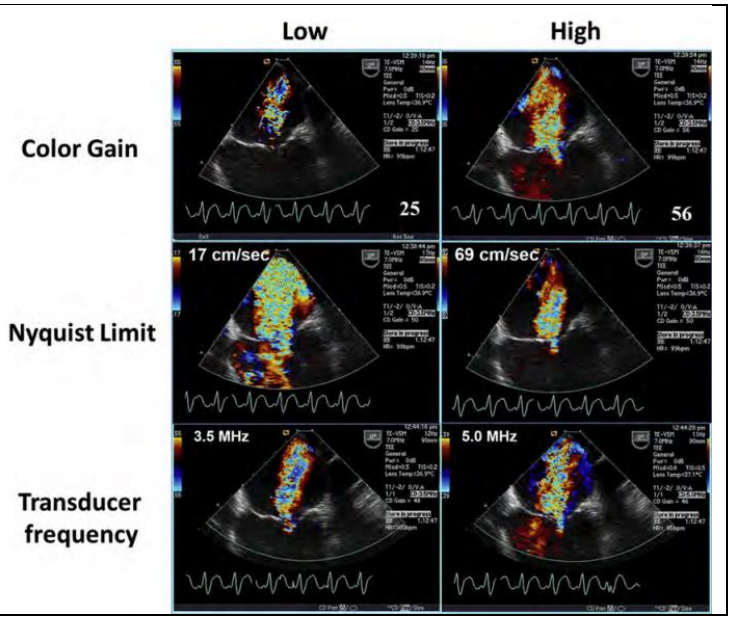
Sweep speed is the refresh rate of the velocity spectrum (i.e. speed of display of the image), similar to the speed of the ECG tracing (25 mm/sec. vs. 50 mm/sec.); the faster one (e.g. 100 mm/sec) spreads out the tracing and make it easier to mark the relevant timepoints and take measurements, while the slower one (e.g. 25 mm/sec) compresses the tracing and is useful for evaluation of respiratory variation in suspected tamponade or constrictive pericarditis. As a general rule, always use a sweep speed of 100 mm/s unless looking for inter-beat variations.

COLOUR DOPPLER - is based on the principles of pulsed Doppler echocardiography. However, rather than evaluating one sample volume depth along a single beam, it evaluates multiple sample volumes along multiple sampling lines. Moreover, instead of grayscale spectral analysis, the received backscatters are displayed on a 2-D image in a color-coded pattern, with flow toward the transducer shown in red and flow away from the transducer shown in blue. Because Doppler colour flow imaging is a pulsed Doppler modality, aliasing occurs, and velocity above the Nyquist limit would show as bright yellow/white with colour reversal. Velocity scale will control the range of velocities displayed in the colour box and, similar to PW Doppler, increasing the colour Doppler scale (also called aliasing velocity or Nyquist limit) may overcome aliasing but the jet will appear smaller; whereas lowering the colour scale will emphasize lower velocities and, thus, the jet will appear larger. Velocity scale should be as low as possible without aliasing.

The colour jet appears larger with (1) higher gain, (2) higher frequency and (3) lower colour velocity scale (aliasing velocity/Nyquist limit)

Colour gain controls amplification of the colour Doppler signals and should be as high as possible, without noise appearance. Set the colour gain by turning it up until noise is encountered and then backing off until the noise just clears from the image. (remember: wall filter should be kept as low as possible without pollution, and scale should be kept as low as possible without aliasing)

Scale (also called aliasing velocity or Nyquist limit)
For jet area and VC use aliasing velocity 40-70 cm/s
For PISA radius use aliasing velocity 20-40 cm/s



COLOUR DOPPLER M-MODE- uses the same principles as colour Doppler, but instead of overlaying the colour data on a 2-D display it instead overlays it on an M-mode display. It can be useful for precisely timing the occurrence of colour jets, and is commonly used for measuring the width of a jet of AR in relation to the diameter of the LVOT.