Ultrasound physics

What is Ultrasound?

High frequency sound of frequencies 2-50 MHz is used in medical ultrasound systems. This is well above the human hearing range of 20-20 kHz. Sound is propagated through a medium as the molecules are vibrated by pressure waves with areas of compression and rarefaction. The waves have the characteristic features of wavelength, frequency and amplitude.

Wavelength (λ) is the distance between two areas of maximal compression (or rarefaction). The importance of wavelength is that the penetration of the ultrasound wave is proportional to wavelength and image resolution is no more than 1-2 wavelengths.

Frequency (f) is the number of wavelengths that pass per unit time. It is measured as cycles (or wavelengths) per second and the unit is hertz (Hz). It is a specific feature of the crystal used in the ultrasound transducer. It can be varied by the operator within set limits – the higher the frequency, the better the resolution but the lower the penetration.

The magnitude of pressure change is given by the **amplitude**. It is expressed in decibels on a logarithmic scale.

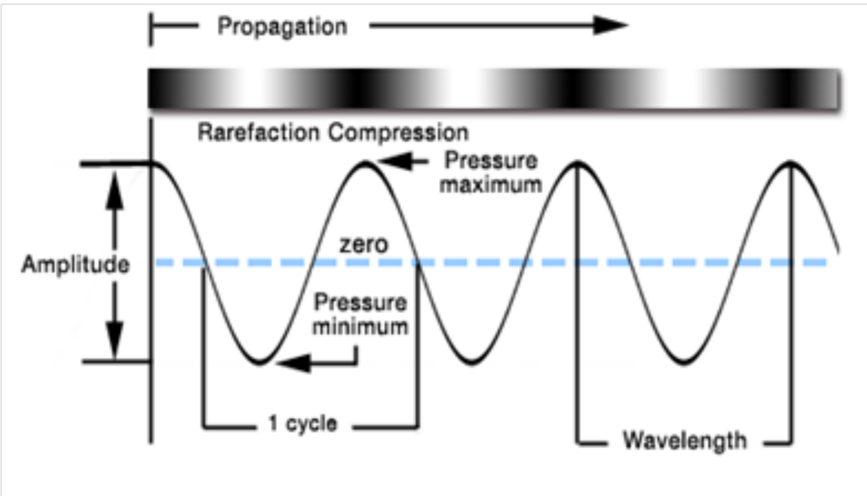

Figure 1. Sound wave characteristics

The sound waves travel at a **Propagation Velocity (v)** dependent on the density and compressibility of the medium.

The relationship between these variables is expressed by the Wave Equation -

$$v = \lambda f$$

In soft tissue propagation velocity is relatively constant at 1540 m/sec and this is the value assumed by ultrasound machines for all human tissue. Hence wavelength is inversely proportional to frequency.

Acoustic Power is the amount of acoustic energy generated per unit time. Energy is measured in joules (J) with joules being the amount of heat generated by the energy in question. The unit is the Watt (W) with 1W = 1J/sec. The biological effects of ultrasound in terms of power are in the milliwatt range.

Intensity is the power density or concentration of power within an area expressed as Watts/m2 or mW/cm2. Intensity varies spacially within the beam and is greatest in the centre. In a pulsed beam it varies temporally as well as spatially

How is Ultrasound produced?

Pierre Curie discovered that when an electrical current is applied to a quartz crystal its shape changes with polarity. This causes expansion and contraction that in turn leads to the production of compression and rarefaction sound waves. This was termed the piezo-electric effect. The reverse is also true, an electrical current being generated on exposure to the pressure of returning echoes. Ultrasound transducers contain and array of piezoelectric crystals functioning as transmitter and responder. The frequency of the generated wave is a function of the crystal used.

Modern transducers use multiple small elements to generate the ultrasound wave. If a single small element transducer is used the waves radiate from it in a circular fashion as do ripples in a pool. If multiple small elements fire simultaneously however the individual curved wave fronts combine to form a linear wave front moving perpendicularly away from the transducer face. This system, that is multiple small elements fired individually, is termed **phased array**. The transducer contains a backing block and acoustic insulator to prevent reverberation within the probe. Quarter wave matching and electronic focusing allows the sound profile produced to be controlled. Some machines allow the focus point to be used defined whilst in others it is set automatically according to scan depth

Near field and Focusing

In a standard disc shaped transducer the beam shape is cylindrical. Initially the beam is of comparable diameter to the transducer as the series of ultrasound waves that make up the beam travel parallel to each other. This is the **near-field** or **Fresnel zone**. At some point distal to the transducer however the beam begins to diverge which will reduce the ability to distinguish two objects close together (resolution). Beyond the point at which the beam begins to diverge is the **far-field** or **Fraunhofer zone**. It is possible to focus the ultrasound beam to cause convergence and narrowing of the beam thus improving (lateral) resolution. Focusing can be achieved by either mechanical or, in a phased array element, by electronic means. If the transducer face is concave or a concave acoustic lens is placed on the surface of the transducer, the beam can be narrowed at a predetermined distance from the transducer. The point at which the beam is at its narrowest is the focal point or **focal zone** and is the point of greatest intensity and best lateral resolution.

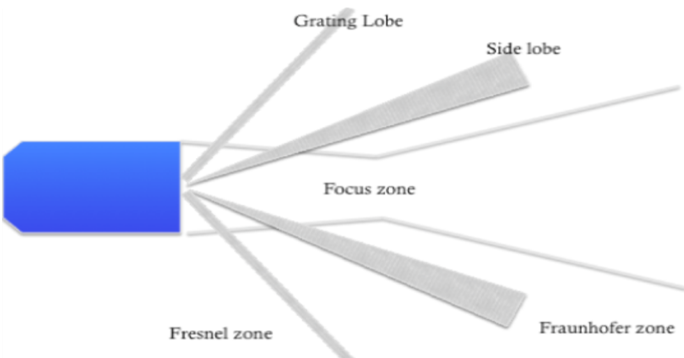

Figure 2. Schematic of beam characteristics. Multiple side lobes may be present.

Resolution

The ability of ultrasound to distinguish between two objects is the resolution. The resolution may be spatial, discrimating distance between objects or temporal, discrimation movement in time. Spatial resolution can be further divided into axial resolution and lateral resolution.
Axial Resolution is the ability to recognise two different objects at slightly different depths from the transducer along the axis of the ultrasound beam.

$$\text{Axial resolution} = \frac{\text{spatial pulse length (SPL)}}{2}$$

$$\text{where SPL} = \lambda * \text{no. of cycles}$$

It is improved by higher frequency (shorter wavelength) transducers but at the expense of penetration. Higher frequencies therefore are used to image structures close to the transducer.
Lateral Resolution is the ability to distinguish objects that are side by side. It is generally poorer than axial resolution and is dependent on beam width. As the ultrasound machine assumes that any echo originates from the centre of the beam then two objects side by side cannot be distinguished if they are separated by less than the beam width.

Beam width is determined by:-

- Transducer frequency (beam width increases with lower frequency transducers)
- Focusing of the beam
- Gain (increased gain will increase the beam width and reduce resolution)

To optimise lateral resolution therefore:

- Use the highest frequency (but note that this occurs at the expense of penetration).
- Optimise the focal zone.
- Use the minimum necessary gain (see the knobology section).

Temporal Resolution is dependent on frame rate. It is improved by:

- Minimising Depth – Ultrasound is sent in a pulse of 2-3 cycles in length, the **spatial pulse length** and then the transducer assumes a receive mode listening for echoes. The depth of scan alters the time required for sound wave to travel to and from the transducer. Thus a shallow scan reduces the distance and time taken and allows more pulses per second. This is known as the pulse repetition frequency (PRF).
- Narrowing the Sector to the area of interest - narrowing the sector angle.
- Minimise Line density (but at the expense of lateral resolution).

Image production

For 2D or B mode ultrasound, the US probe sends pulses of sound waves from piezoelectric crystal arrays. Returning echoes are received by the crystals, each echo having an energy level which is displayed as brightness, and a depth under the probe calculated by the time delay between send and receive. The depth calculation assumes a velocity in tissues of 1540 metres/sec.

The likelihood of reflection and echo creation is dependent on **Acoustic impedance**, which is defined as the opposition of the tissue to the passage of sound waves, and defined by the product of tissue density (ρ) and acoustic velocity (ν).

$$\text{Acoustic impedance} \qquad Z = r\,V \;\; (kg^{-2}s^{-1})$$

Material	Density ($kg.m^3$)	Velocity ($m.sec^{-1}$)	Acoustic impedance ($kg^{-2}s^{-1}$)
Air	1.2	343	429
Water	1.1	1525	1.43
Fat	0.92	1450	1.33
Muscle	1.04	1580	1.64
Cortical Bone	1.9	4040	7.68

Table 1. Acoustic Characteristics of materials

As sound waves strike a surface between two tissues of differing acoustic impedance some of the sound is reflected. The strength of the returning echo is dependent on the difference in acoustic impedance between two tissues,. Two principal types of reflector are described, a specular reflector where the sound beam is reflected in a single direction and a scattering reflector where the energy is reflected in multiple directions. The former usually has a surface width of more than 1 wavelength while the latter has a surface width less than one wavelength. The image created in this way is dependent on the direction from which the sound encounters a barrier, and is directionally dependent; a phenomenon known as Anisotropy.

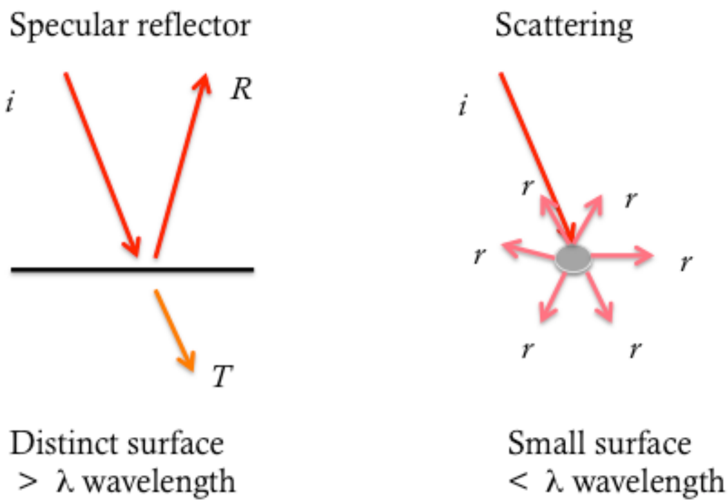

Figure 3 Specular and scattering reflectors. Incidence sound wave (i) with single reflection (R) from a specular reflector, and multiple reflections (r) from a scattering reflector. The acoustic impedance difference between tissues determines the relative amount of reflected to transmitted sound (T)

Attenuation

As sound passes through tissues the energy losses or attenuation occurs due to absorption and reflection. Attenuation is defined as the decrease in amplitude or intensity of a sound wave as it passes through a medium. The magnitude of this loss varies between tissues, and is defined by the attenuation coefficient (a) according to the formula:-

Attenuation (dB) = a (dB) x 2 x depth of reflector(cm) x Frequency (MHz)

It therefore increases with depth and higher frequencies.

Material	Attenuation coefficient (dB/cm)
Water	0.002
Soft tissue	0.3 – 0.8
Fat	0.5 – 1.8
Bone	13 - 26
Air	12

Table 2. Attenuation coefficients for tissues

IMAGING MODES

A Mode   This mode represents the time taken for a single echo to return to the transducer –represented by a by a blip on the screen. The distance the sound travels is twice the depth of the echo structure so that the depth can be calculated from the propagation velocity. This is the principle first used in SONAR.

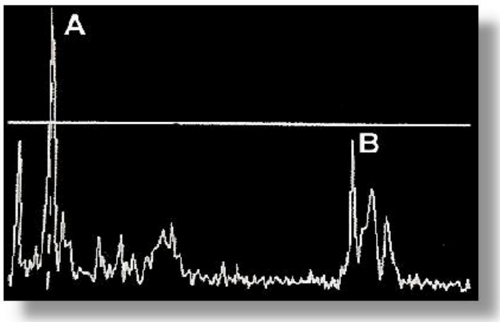

Figure 4. A mode scan

B Mode or 2D Echo: Multiple transducers send and receive echoes in one plane to build a 2D image based on time and echo strength. The strength of the returning echo is represented by brightness (B) of the displayed image and depth of reflection by the time delay.

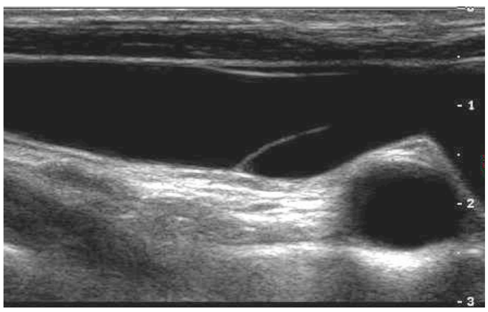

Figure 5. B mode (2d) image of jugular valve

M Mode
Multiple A mode images are created and represented over time creating a map demonstrating movement (M) of reflectors with time. The upper part of the screen below shows the structure being imaged in 2D. A cursor is placed along one part of this image and the relative movement of the component structures is shown against time in the lower part of the screen. In the image below the Left ventricular walls are shown allowing systolic and diastolic dimensions to be measured.

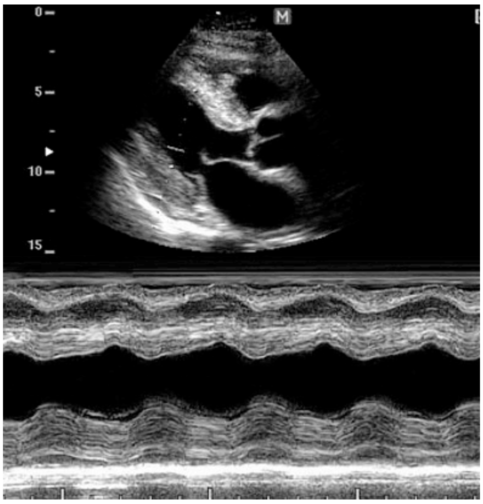

Figure 6. Parasternal long axis view (2D and M mode)

The Doppler effect is a frequency shift caused by relative movement between a sound source and sound receiver. If the two move together the frequency increases and if they move apart the frequency decreases.

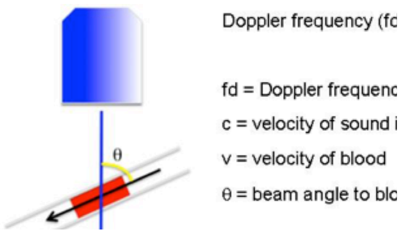

Figure 7. Doppler shift. Usual angle of insonation is 60 degrees

Doppler is used to measure and display blood flow. Several versions are available:-

Colour Flow Doppler (CFD) converts the frequency shift to velocity, which displayed on a colour scale. Interrogation of an area around the nerve is important to reduce the possibility of inadvertent vessel penetration and intravascular injection. This is particularly important in the axilla and supraclavicular regions.

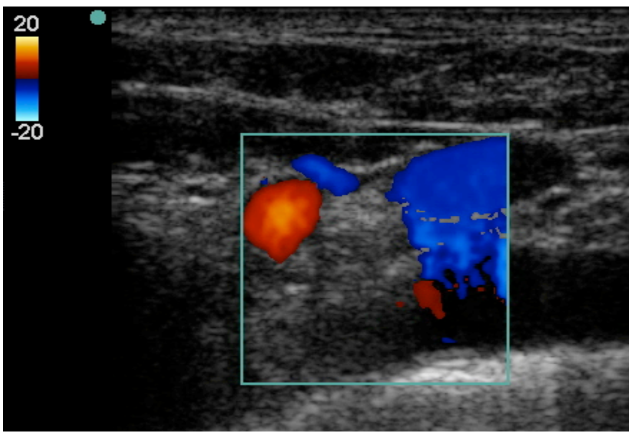

Figure 8. Colour flow doppler. Colour is applied according to a user selectable velocity scale (top left). No colour on 2D image outside the box.

Continuous Doppler – Samples all velocities along a user selected line.

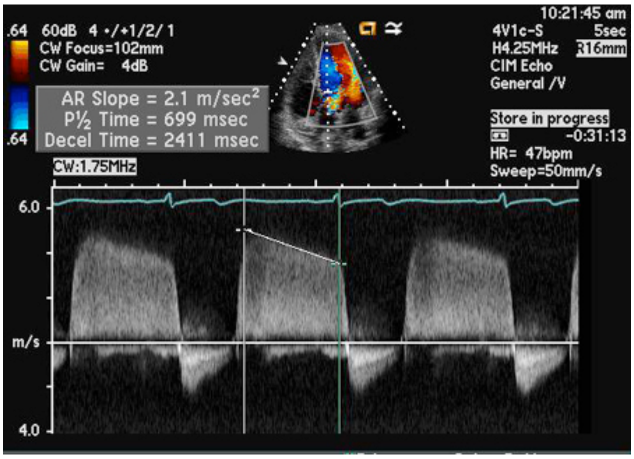

Figure 9. Continuous Doppler. All velocities along the cursor line are shown on a scale with time on the x axis. The ECG is displayed for timing with the cardiac cycle.

Pulsed wave Doppler – Samples a small area defined on a 2D image. A box is placed over the area of interest on the cursor line. The ultrasound machine then only looks at the echoes returning from this small gate by measurement of the returning frequency after a specific time delay.

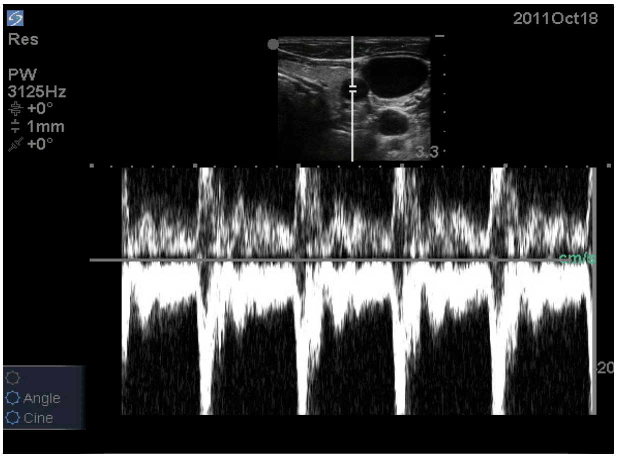

Figure 10. Pulse wave doppler of Carotid artery

Ultrasound artefacts

Definition: Any part of an image that does not represent the anatomical structures present in the area being examined.

Ultrasound relies on physical principles to create that image and each part of the process of image production may be accompanied by errors. The US image is created by pulse formation, propagation, attenuation, reflection and echo detection, only some of which are user dependent. An understanding of the way in which ultrasound images are generated is important if one is become expert in its use. Artefacts are the result of a failure of Ultrasound to conform to the physical assumptions made in interpreting returning echoes.

Physical assumptions are made in Ultrasound imaging that in certain situations lead to the production of artefacts:-

Echoes originate from the main beam
Echoes return after a single reflection
Ultrasound energy is uniformly attenuated
Ultrasound velocity is constant
Sound always travels in a straight path
The depth of an object is determined by time

Table 3. Physical assumptions in Ultrasound imaging

Artefacts due to beam characteristics

The processing computer in the US machine believes that returned echoes are the result of only one beam of minimal width, and of defined shape. Unfortunately this is not true. The beam can be divided into three zones, the near field or Fresnel zone, a focus zone, and the far field or Fraunhofer zone. The near field depth is dependent on frequency (l) and transducer radius (R).

Near field length = $R^2 / 4 \times l$

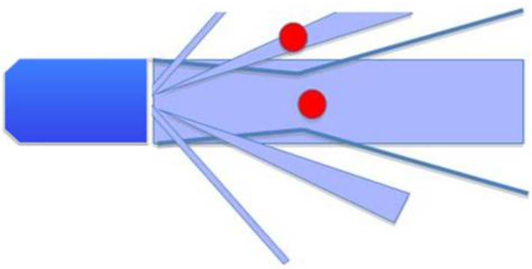

Figure 11. Side lobe artefact. Echo from a strong reflector in the side lobe is positioned as though it was within the main beam.

Higher frequency transducers have a greater Fresnel zone depth. Beam divergence occurs in the far field and is controlled by a focusing lens in the probe or electronic focusing. Some machines have a user selectable focus and it is therefore important to set this at the target depth. Where there is no independent control of focus, the focus is automatically set for the depth at the centre of the screen. The main beam is assumed to be the only source of echoes, but crystals often possess multiple side lobes and grating lobes. A reflection from one of these will be placed as though it originated from the main beam. This is known as a side lobe artefact. These are usually weak echoes only seen within areas of low echogenicity such as blood vessels, bladder or cysts. As these lobes are of lower energy, changing the insonation angle or reducing gain will eliminate them.

Beam divergence affects the lateral resolution of the system, so that two objects may be appear as one. The divergence angle is frequency dependent and given by the formula:

Divergence angle (degrees) = 70 l/D where D is the diameter of the transducer.

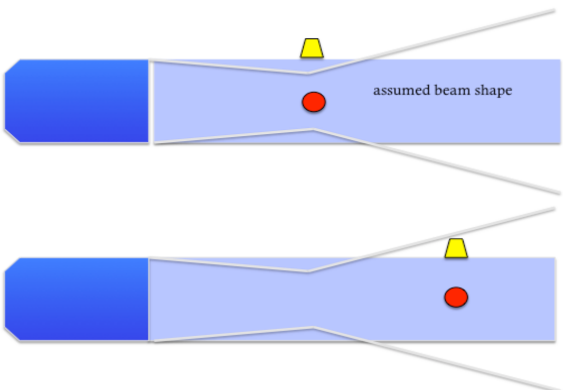

Figure 12. Beam width (lateral resolution) artefact. Grey area represents assumed beam. Note two reflectors the same width apart are combined as one in the far field, but only the red is seen at the focus zone.

Axial resolution is the ability to distinguish two points above and below each other in the US beam. It is a equal to half the spatial pulse length; Spatial pulse length is the product of the number of cycles in a pulse or ultrasound and the wavelength. Most pulses are of two or three cycles in length. Therefore the ability to distinguish two objects in depth is improved with increasing frequency.

Temporal resolution is of more concern in Echocardiography and is the ability to distinguish moving structures in time. Higher frame rates are possible with reduced depth and narrow frame width. See Ng and Swanevelder [2].

The discussion above makes it clear that what we see on the monitor is very dependent on a variety of tissue and sound properties, and explains why changing frequency changes the screen representation of tissue. Low resolution images will therefore suffer from loss of information – a form of drop out artefact.

Attenuation artefacts

The Transmission of sound through tissue is a function of acoustic impedance with air being notably poor at conducting sound energy. It is therefore important to eliminate air in the system. A probe artefact may occur due to inadequate coupling of the US probe to skin and is easily remedied with the use of acoustic gel or fluid. Acoustic gel should be sterile; 5% dextrose is a useful coupling agent particularly if nerve stimulation is used [3].

As sound passes through the through the tissues, energy losses occur due to heat (80%), reflection and refraction as previously described. As the sound energy falls during passage through tissues the amplitude of returning echoes becomes smaller. In order to compensate for this, the machine progressively increases the amplification of reflected energy in an attempt to normalize the grey scale on the monitor. This is known as Time-Gain compensation (TGC). On many portable machines this is automatic or there are controls for near, far and overall gain. More versatile machines have a series of sliders so the user can adjust each layer independently. Failure to adjust these correctly leads to artefacts due to over-gain or under-gain in the image. Inadequate gain (Low energy signal) will result in a poor quality image with none or only very strong reflectors being visible. This is easily remedied by slowly increasing gain until a good image is revealed. As the sound energy levels are increased, excessive gain (high energy signal) results in increased “noise” from specular and scattering reflectors resulting in a white out. Imaging in areas with high ambient light levels often results in over-gain problems, and mirror artefacts, which can be eliminated in lower light levels. *Always use the minimum amount of gain* to achieve the desired image. Whilst there are few defined health risks from Ultrasound [4], we should use as little power as is reasonable.

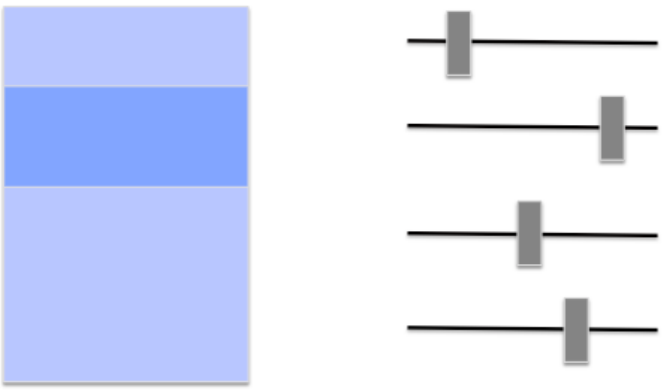

Figure 13. Banding of image due to inappropriate Time-Gain Compensation settings. There is over-gain in the second slice.

Post-cystic enhancement or distal enhancement occurs when the sound passes through a material with low attenuation. As the sound reaches the next reflector its energy is greater than expected for tissue due to the TGC applied in post processing. This results in a stripe of increased echogenicity beyond the material. This usually but not always indicates fluid and is commonly seen deep to water containing structures such as cysts, bladder or blood vessels. In regional anaesthesia practice this may be misinterpreted as a nerve particularly deep to the axillary artery in infraclavicular block.

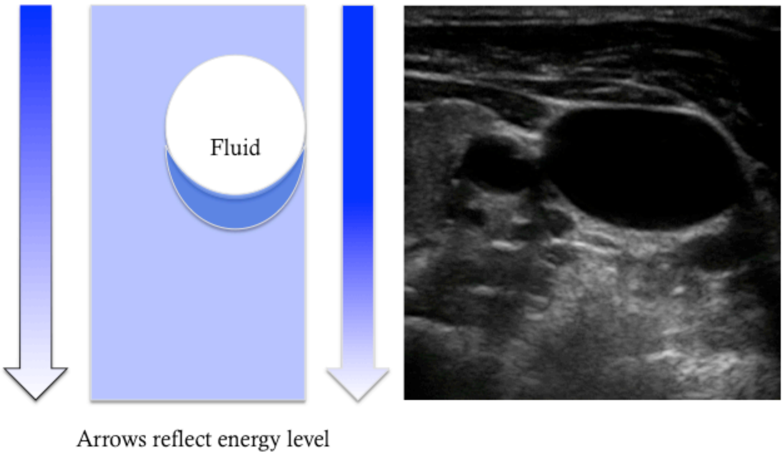

Figure 14. Distal enhancement. Note increased brightness deep to vessels.

Propagation artefacts

Acoustic Shadowing artefact occurs when the sound energy is reflected, rather than transmitted through tissue. This occurs with bone and air leading to dropout of information deep to the object. Probe repositioning may assist in viewing behind the reflector. It is used as an indicator of rib position in the supraclavicular fossa and assists in spinal imaging to determine the correct angle for needle placement [5].

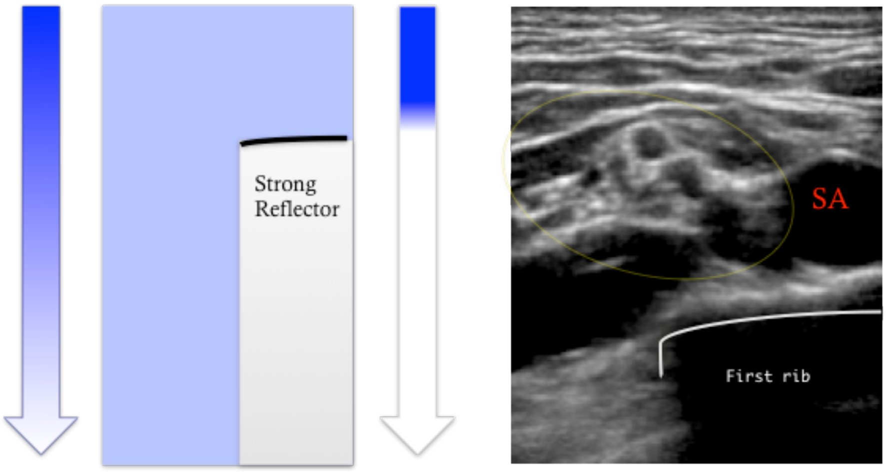

Figure 15 Acoustic shadowing. Strong reflector prevents Sound transmission deep to it creating a shadow

Reverberation Artefact

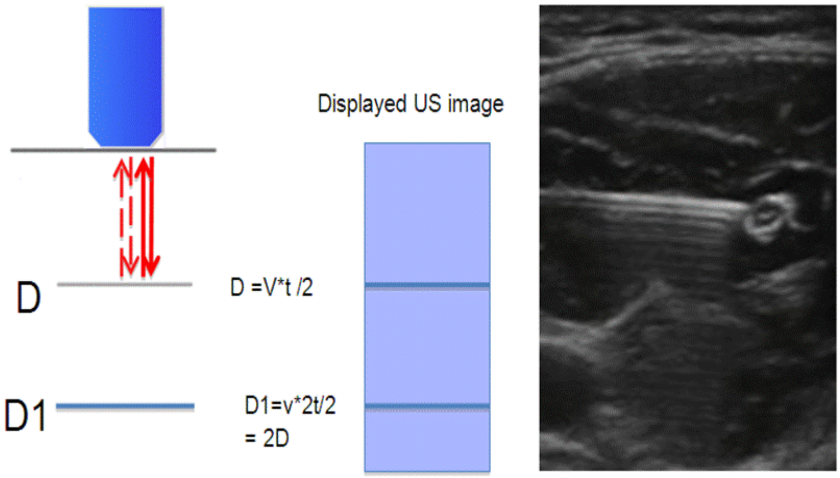

Figure 16. Reverberation. Reflection between two strong reflectors lead to multiple returning echoes at recurring time intervals which are placed in the image at a constant periodicity as a series of parallel lines. The US image to the right shows the reverberation artefact of a needle during block performance.

US assumes that an echo returns once to the transducer and the time interval denotes its depth. When there are two highly reflective surfaces the echo may travel back and forward between the two multiple times resulting in a series of parallel lines at fixed intervals. This may occur between probe and reflector or between two reflectors. Needles imaged perfectly in long axis produce a pronounced reverberation artefact, a useful sign to confirm accuracy of needle position under the probe.

The comet tail artefact occurs when reverberation occurs between two closely spaced reflective surfaces. Reverberation between these produces a thin bright artefact, which appears deep to the structures and gets progressively narrower with depth as the energy dissipates with each reflection. The close spacing means it is impossible to see each individual reflection. This artefact is a useful sign in lung imaging, its presence confirming pleural apposition ruling out a pneumothorax. The "lung point" is the position at which the lung sliding sign disappears and can be used to map out the extent of a pneumothorax. Lung rockets are brighter lines created by reflection between adjacent air bubbles and are seen when there is increased interstitial lung water in consolidated lung. They reach the edge of the screen.

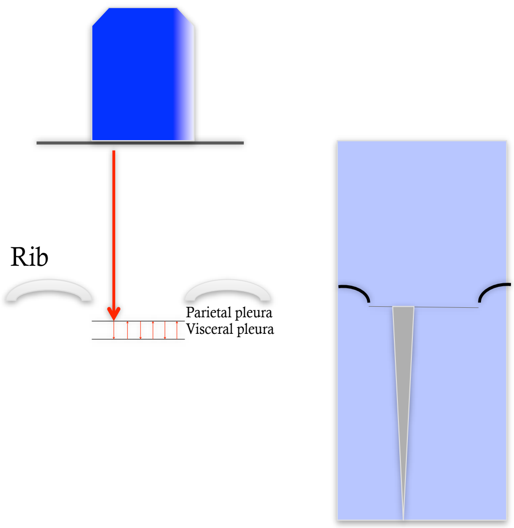

Figure 17. Comet tail artefact. Reverberation in a thin layer produces a bright reflection of narrow width.

See: http://www.criticalecho.com/content/tutorial-9-lung-ultrasound

Lung rockets or B lines are produced by reverberation in a cluster of four bubbles. They indicate thickening of the interlobular septa of the lung, and produce multiple bright lines, which reach the depth of the image as shown below. These are always pathological.

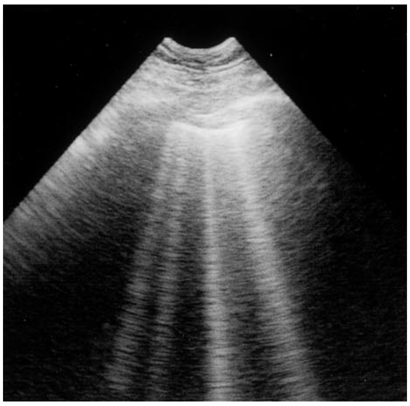

Figure 18. Lung rockets

Mirror artefacts occur when there is a strong reflector (e.g. Bone) deep to a weaker reflector (e.g. fascial layer). The image of the weak reflector is then displayed deep to the strong reflector due to the later timing of the returning echo. This is more likely to occur at higher gain settings and the artefact should disappear with probe angulation and reduced gain.

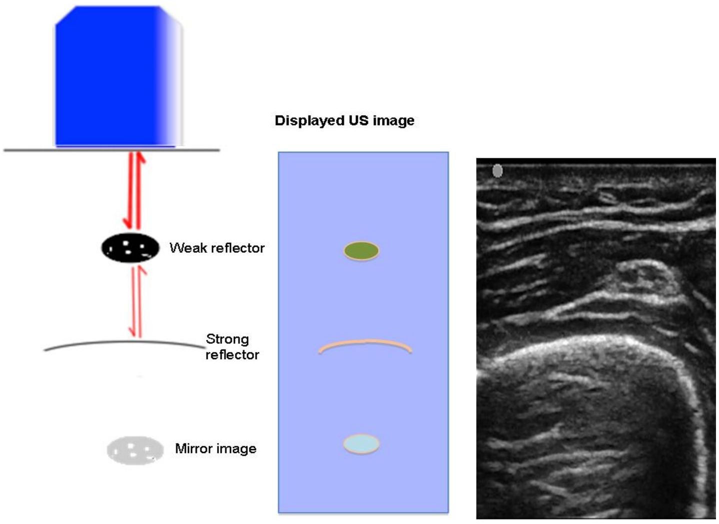

Figure 19. Mirror artefact. Reflected wave produces mirror image deep to strong reflector. US image on right shows fascial planes mirrored in humerus.

Anisotropy

Specular reflection obeys the laws of physics where the angle of reflection is equal to the angle of incidence. The strongest echo and therefore image occurs when the reflector is at 90 degrees to the sound waves. This property is very important in nerve imaging where the nerve can disappear from view with small deviations in angle. This directional dependence is termed Anisotropy and is well seen for many peripheral nerves. When searching for a nerve, you need to be aware of the path it takes and angle the probe to find the nerve. For example the popliteal sciatic nerve follows the line of the femur and anterior thigh so that the probe should be held perpendicular to that direction. This means the probe appears to be pointing caudally as it rests in the popliteal fossa.

Velocity artefacts

The speed of sound transmission varies between tissues dependent on their density and elastic properties (see Table.1). The US machine places echoes dependent on a fixed velocity of $1540 m.sec^{-1}$. If sound is propagated at greater velocity it will appear in the image more superficial than it is, conversely a slower velocity will result in a deeper image. This is termed a speed displacement artefact but is probably of little significance in regional anaesthesia. Where sound encounters two dissimilar tissues at an oblique angle to the beam then the transmitted beam is refracted towards the slower medium. The degree of refraction is given by Snell's Law:-

Sin q *r* / sin q *i* = c2 / c1

where: q *r* = refraction angle; q *i* = incident angle; c2 and c1 are the wave velocities in the two media

The refraction makes it possible for objects outside the assumed main beam to appear as though they are within the beam. Changing the beam angle will change the refraction angle and assists in recognition of this artefact.

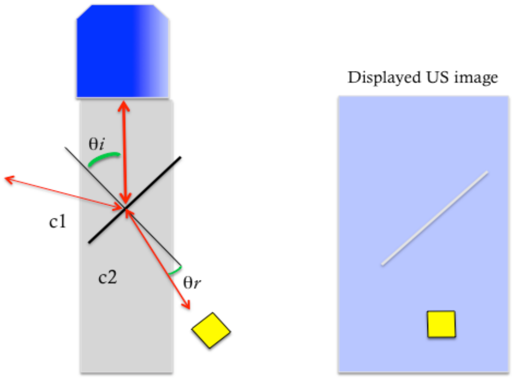

Figure 20. Refraction artefact- as a result of refraction an object may be placed as though it is within the beam

This artefact is readily also seen as Edge shadowing distal to a curvilinear surface. A common example is the shadowing deep to the edges of blood vessels.

Figure 21. Edge artefact and distal enhancement around the femoral artery A.
Bold arrow shows Femoral Nerve. Small arrows point to Fascia Iliaca

Andrew Gray described the bayonet artefact [6] where he noted an apparent bend in a needle placed through the Axillary artery. In this instance the speed through blood (water) is faster than through tissue so the returning echo is received earlier and placed nearer the surface.

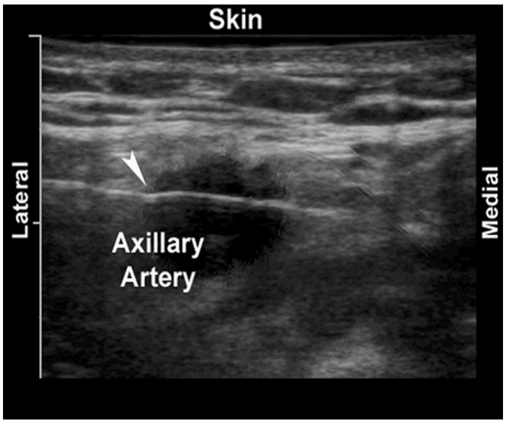

Figure 22. Bayonet artefact

Doppler artefacts

Interrogating the vessel at 90 degrees to flow (i.e. probe perpendicular to vessel) will not produce a frequency shift and therefore no colour flow indication since there is no movement of reflector (red cells) towards or away from the transducer. It is therefore vital that the probe is angulated in several directions to search for vessels in hypoechoic areas.