A lifetime of specialist care

Royal Brompton & Harefield NHS
NHS Foundation Trust

The Physics of Ultrasound

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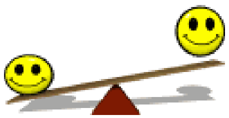
Paediatricians with Expertise in Cardiology Special Interest Group

Transducer choice: resolution vs penetration

The key is to balance both the need for image strength and image resolution



Use the highest frequency transducer available which gives enough image strength

Attenuation

Ultrasound waves attenuate (i.e. lose energy) due to:
- absorption (heat)
- reflection and scattering (energy redirected)
- diffraction (energy redirected)

Measured in decibels (dB) where each 3dB loss is a 50% reduction in intensity.

Attenuation coefficient in soft tissue = 1dB/cm/MHz

Attenuation: absorption

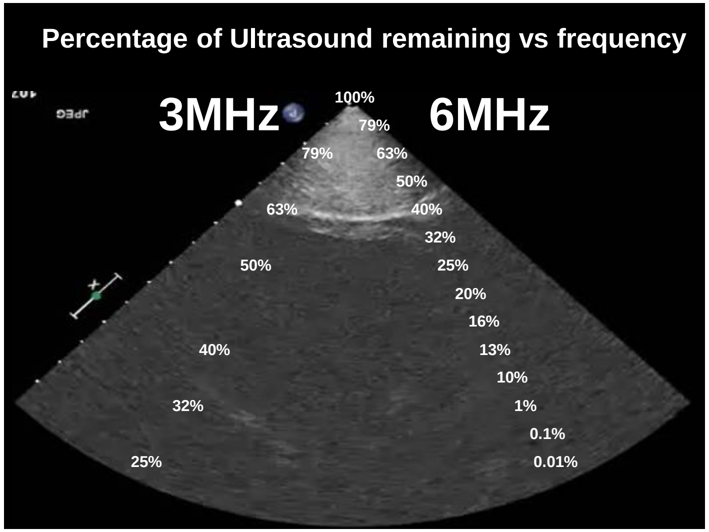
Ultrasound energy dissipates within a media due to energy absorbed as heat.

A higher frequency ultrasound wave causes more molecular motion and loses more energy to absorption (loss to heat).

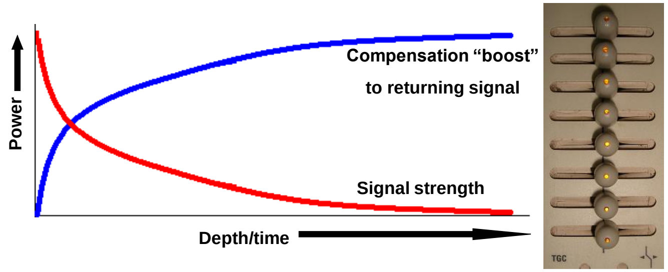
Therefore at any given depth a higher frequency ultrasound wave will be weaker.

Attenuation coefficient in soft tissue = 1dB/cm/MHz

double the frequency, double the rate of absorption

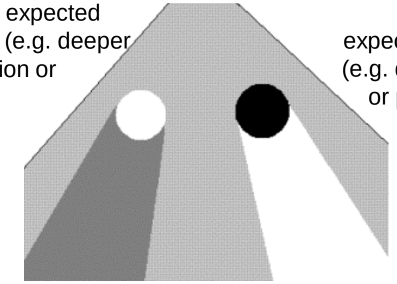


Time-gain compensation



Attenuation artifacts

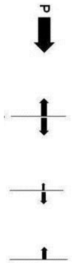
Acoustic shadow from higher than expected attenuation (e.g. deeper to calcification or prosthesis)



Acoustic enhancement from lower than expected attenuation (e.g. deeper to a cyst or pericardial fluid)

Attenuation assumed to be 1dB/cm/MHz

Attenuation: reflection/scattering

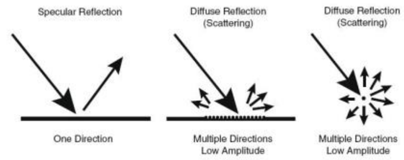


Large (>λ) smooth surface such as cardiac wall/valve are called specular reflectors and cause 'mirror-like' reflection.

The strength of the reflected beam is related to the difference in acoustic impedance (Z). The remaining energy deeper to the reflector will be weaker due to energy being redirected back/away.

Small (usually <1λ) rough/irregular surfaces which reflect in multiple directions are called scatterers.

Higher *f* lower λ → increased backscatter.



Attenuation: reflection

The strength of the reflected beam is related to the difference in acoustic impedance (Z).

Percentage reflected = $[(Z_2 - Z_1)/(Z_2 + Z_1)]^2 \times 100\%$

Material	Acoustic Impedance (Z)
Air	0.0004
Lung	0.26
Soft-tissue (avg)	1.63
Bone	7.8

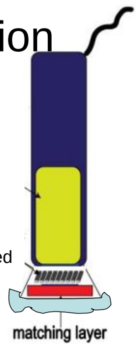
% reflected at an air/soft tissue interface?

??

% reflected at an bone/soft tissue interface?

??

Maximising transmission



matching layer

thin layer between the piezoelectric elements and the skin

"accoustic matching"

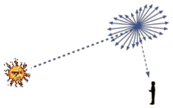
reduces reflection → less attenuation and more energy transmitted

matching layer + gel

Additional intermediate "accoustic matching"

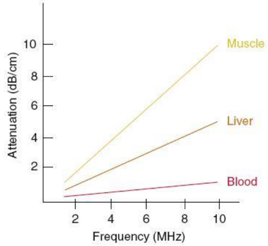
reduces reflection → less attenuation and more energy transmitted

Ultrasound and red blood cells: Rayleigh scattering



Very small (<<λ) reflectors which reflect ultrasound energy concentrically, e.g. RBC (7.5µm).

(recall yesterday, we worked out the λ of a 6MHz signal was 0.25mm (Q9))



Amplitude of reflected signal is less in RBC's.

Attenuation is less in RBC's.

Spontaneous contrast due to RBC aggregation.

Reverberation artefact

Assumption: ultrasound beam is reflected only once.

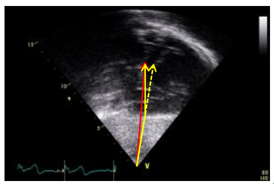
Reverberation artefact occurs when echoes bounce between two highly reflective interfaces resulting in depth perception errors.

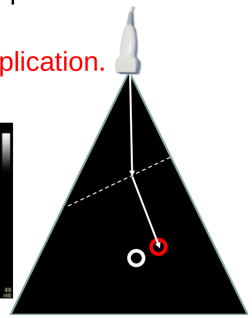
Refraction artifact

Assumption: ultrasound beam travels in a straight line

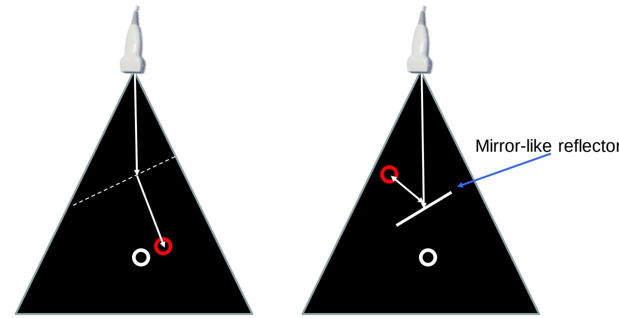
Refraction occurs when the ultrasound beam strikes an interface at an angle and where the speed of sound is different (according to Snells Law).

Results in improper placement or duplication.





Refraction vs mirror image artifacts



= actual object, usually displayed in correct position

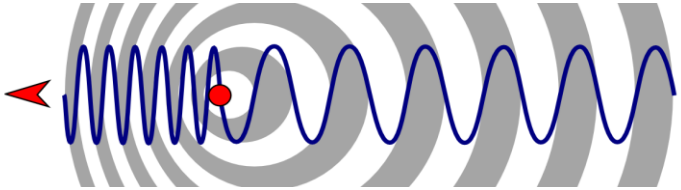
= object duplication as displayed

The Doppler Effect

When ultrasound interacts with a moving object (i.e. RBC's) the reflected frequency changes.

If the RBC's are traveling towards the transducer
→ the ultrasound wave is "squashed" ↓λ and ↑f
→ positive Doppler shift, i.e. red or above the line

If RBC's are traveling away: "stretched" →↑λ and ↓f



The Doppler Equation (!!!)

Blood flow velocity (m/s)

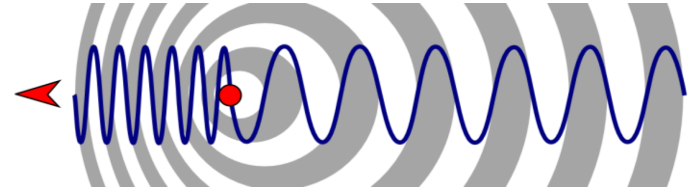
Transmitted frequency (Hz)

Incident angle

Doppler Shift (Hz)

$$\pm \Delta f = \frac{2 f_t V \cos\theta}{c}$$

Assumed speed of sound in soft tissue (i.e. 1540m/s)



The Doppler Equation

Assumed speed of sound in soft tissue (i.e. 1540m/s)

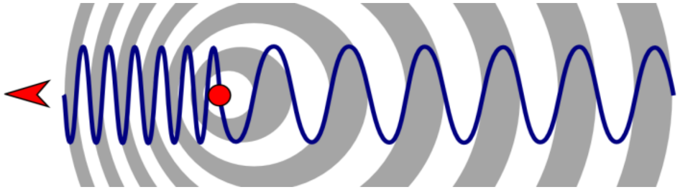
Doppler Shift (Hz)

Blood flow velocity (m/s)

$$V = \frac{c(\Delta f)}{2\, f_t \cos\theta}$$

Transmitted frequency (Hz)

Incident angle



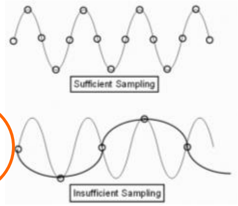
The Nyquist Limit (Aliasing)

The maximum Doppler shift (Δf_{max}) able to be displayed without aliasing.

Determined by the sampling rate (PRF).

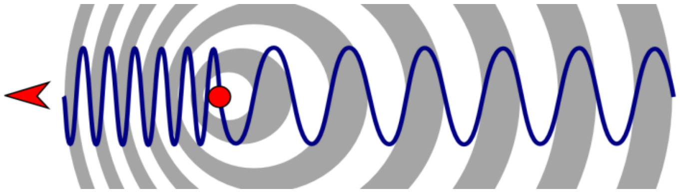
Nyquist Limit: $\Delta f_{max} = \frac{PRF}{2}$

Recall that: PRFmax = $\frac{c}{2D}$ $\Delta f_{max} = \frac{c}{2x2D}$



The Doppler Equation

$$V_{max} = \frac{c\,(\Delta f_{max})}{2\, f_t \cos\theta} = \frac{c}{2\, f_t \cos\theta} \times \frac{c}{2x2D}$$



Determining the Nyquist Limit

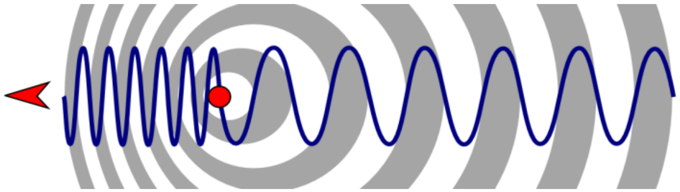
Assumed speed of sound in soft tissue (i.e. 1540m/s)

$$V_{max} = \frac{c^2}{8\, f_t \cos\theta D}$$

Transmitted frequency (Hz)

Incident angle

Depth



Determining the Nyquist Limit

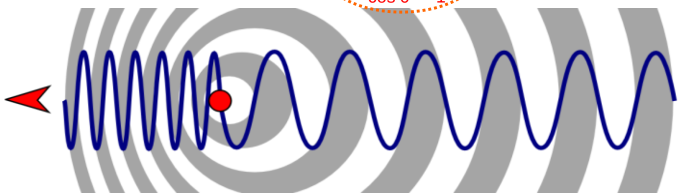
Angle (θ)	cos θ	% error
0	1.00	0
10	0.98	2
20	0.94	6
30	0.87	13
40	0.77	23
50	0.64	36
60	0.50	50

Assumed speed of sound in soft tissue

$$V_{max} = \frac{1540^2}{8\, f_t \cos\theta D}$$

Incident angle assumed to be 0º

cos 0º = 1



Determining the Nyquist Limit

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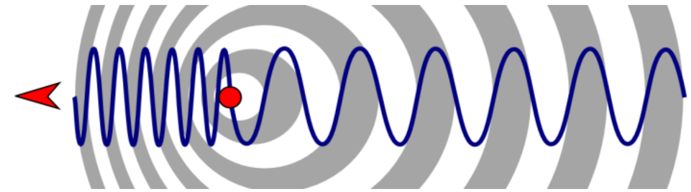
50	0.64	36
60	0.50	50

$$\uparrow V_{max} = \frac{1540^2}{8\, f_t \cos\theta D}$$

↑ **Incident angle**

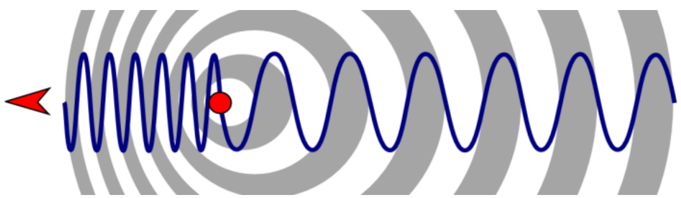


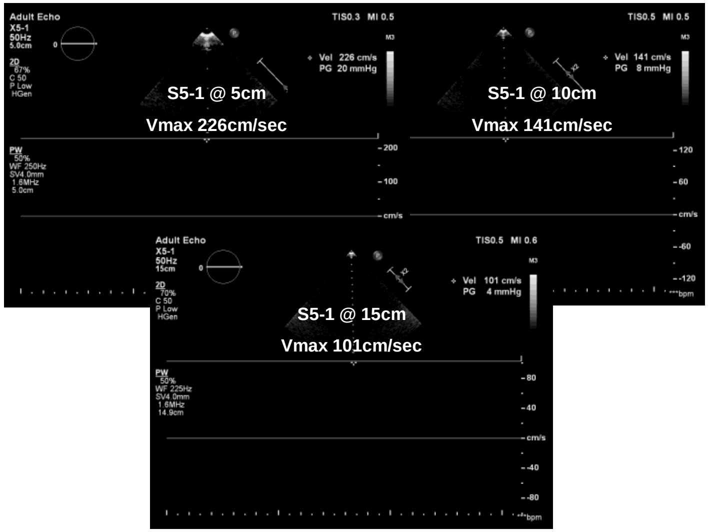
Determining the Nyquist Limit

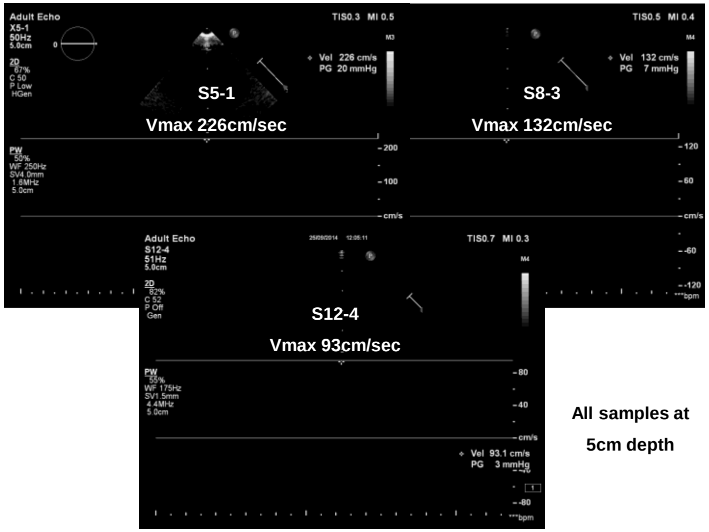
$$\downarrow V_{max} = \frac{1540^2}{8\, f_t \cos\theta D}$$

↑ **Transmitted frequency (Hz)**

Depth ↑





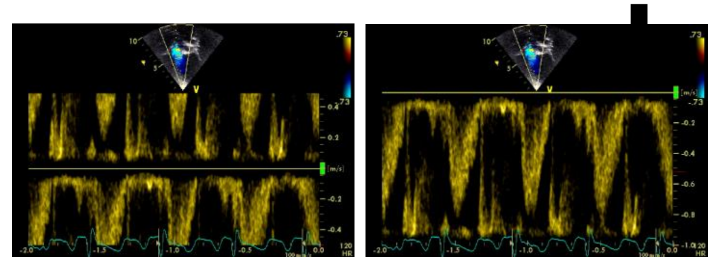


All samples at

5cm depth

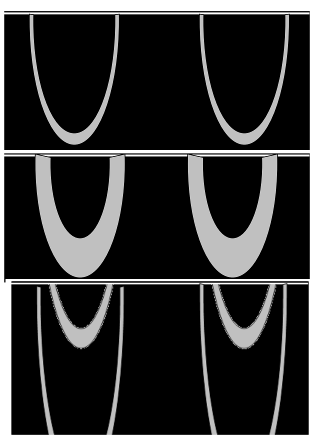
PW Doppler

PW Doppler works by selectively listening



Digital “cut and paste”

PW Doppler



On reflection the ultrasound wave consists of a spectrum (hence spectral Doppler) of frequencies which are digitally subtracted → Doppler frequency shifts (*f* difference is in the audible range → “sound”). From this complex wave, the process of fast fourier transformation separates each individual frequency and its amplitude and then plots this information on the spectral Doppler graph.

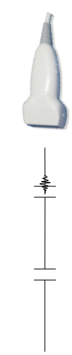
PW Doppler will display less velocities at any single point in time vs. CW Doppler → narrow spectral envelope

When there is a large variation in velocities at any point in time the spectral envelope will be broader. This may indicate acceleration or that your gate size is too large.

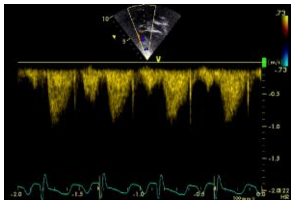
When the Nyquist Limit is exceeded then you will get aliasing which manifests itself as a ‘wrap-around’ signal. To counteract this, you either use high PRF PW or CW.

High PRF

- Transducer sends out an additional pulse before the original pulse has returned.
- In effect it doubles the PRF and therefore doubles the Nyquist limit.
- The disadvantage is that the exact origin of the Doppler shift is not known.
- Potential for range ambiguity artifact (“depth confusion”)



CW Doppler



CW Doppler works by listening all the time

Continuous transmission and reception of ultrasound.

No maximum velocity but….

No range resolution.

Colour flow Doppler

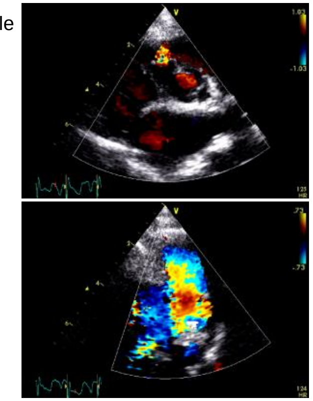
Effectively a multi-sampled PW from multiple sites (100-400) superimposed on a 2D image→ low FR!!!

Each area sampled minimum of 3 times to calculate a Doppler frequency shift and estimate mean velocity.

Frame rate determined by:

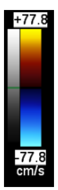
- **Sector size** ↓width/depth↑FR
- **Packet size:** The packet size is the number of pulses transmitted per line. ↓packet size ↑FR

Same limitations as PW Doppler (i.e. Nyquist limit), however as it is detecting mean velocity the Nyquist limit is lower → aliases earlier



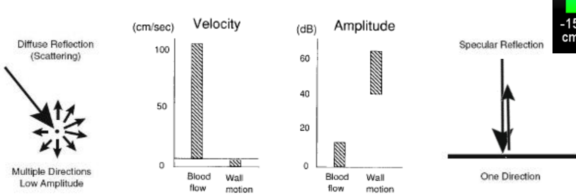
Colour and TDI

Filters are used to discriminate between myocardium and tissue in colour imaging:



Blood is a low amplitude scatterer (recall Rayleigh scattering) with relatively quick velocities.

Myocardium is a high amplitude spectral reflector with relatively slow velocities.



Biological effects of ultrasound

Thermal Effects of Ultrasound: amount of heat produced has to do with the intensity of ultrasound, the time of exposure, and the specific absorption characteristics of the tissue.

Thermal Index (TI): relative potential for temperature rise.

Non-thermal/biological effects of Ultrasound: rapid and potentially large changes in bubble size can occur → cavitation, may increase temperature and pressure within the bubble and thereby cause mechanical stress on surrounding tissues, precipitate fluid microjet formation, and generate free radicals.

Mechanical Index (MI): potential effects for cavitation, microstreaming and radiation force. Highest in PW Doppler**.**

Recommendation:

Minimise exposure time

ALARA: As Low As Reasonably Achieveable.