Ultrasound Physics & Hardware

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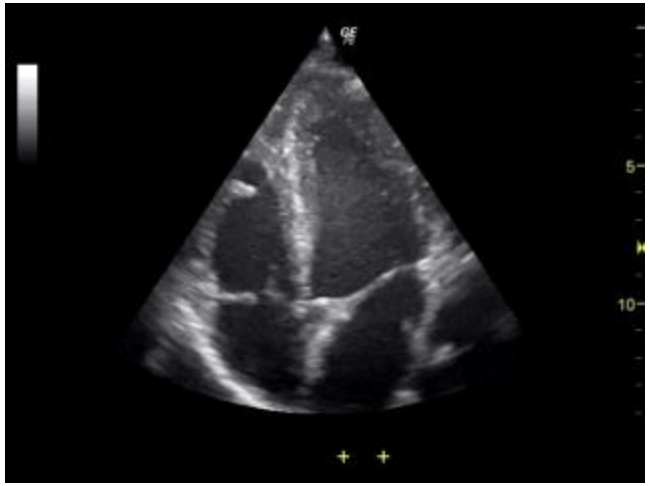
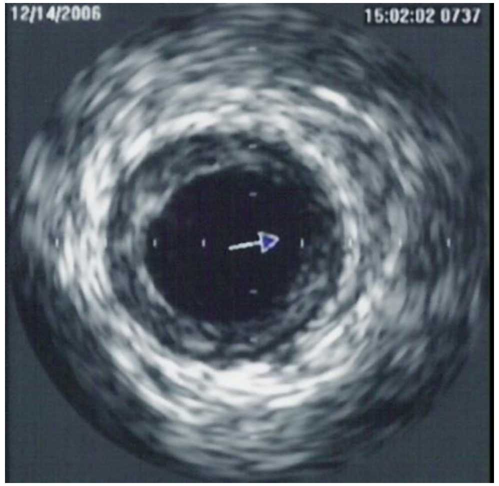
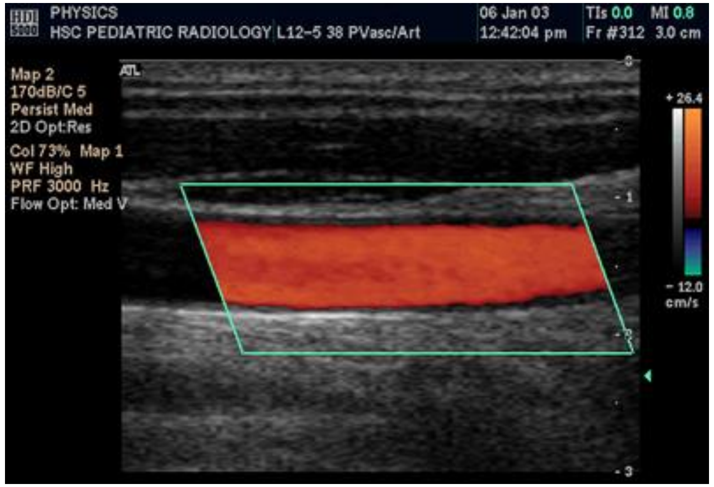
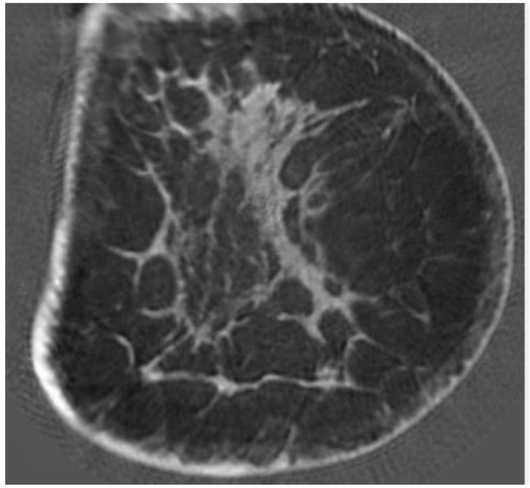
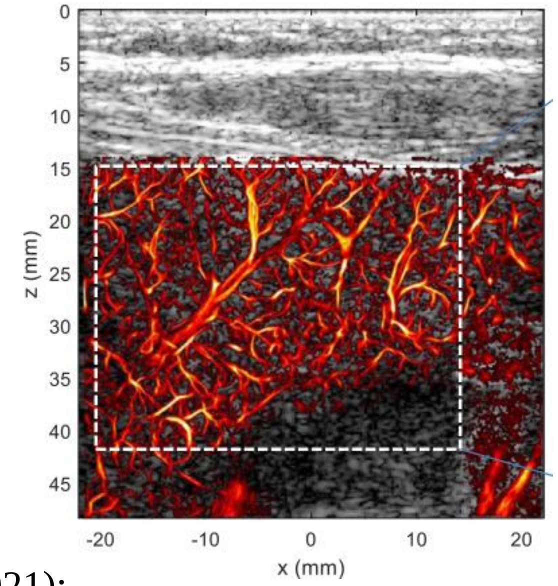
Ultrasound imaging







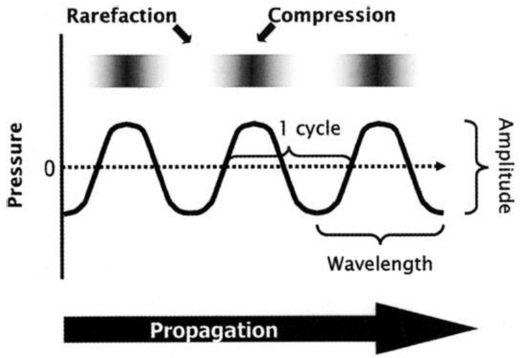




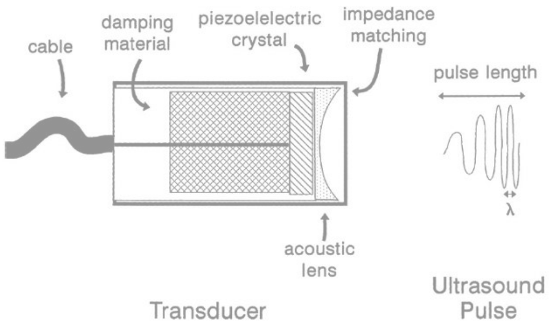
Picture courtesy: Wikipedia; Huang et al., PMB (2021);
Wiskin et al., IEEE TUFFC (2017)

Learning outcomes

- After these lectures, you should be able to:
  - Explain how ultrasound interacts with tissue
  - Understand where ultrasound imaging contrast comes from
  - Describe how ultrasound signals are generated and detected
  - Explain how anatomical ultrasound images are formed
  - Compare the different clinical approaches to performing ultrasound imaging and discuss emerging new applications of ultrasound
  - Describe the origin of the hazards that arise from ultrasound imaging and how they can be mitigated
  - Explain the key governing legislation around ultrasound safety
  - Describe common approaches to quality assurance / control

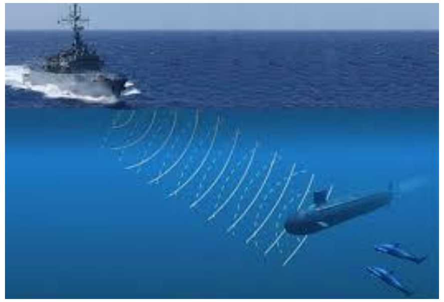


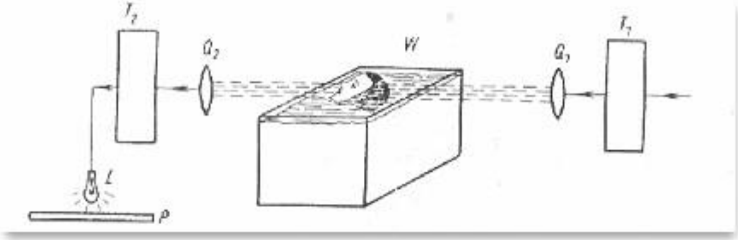
Ultrasound Physics

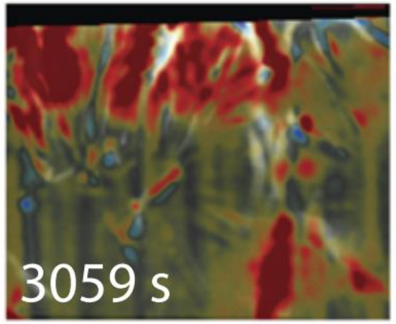


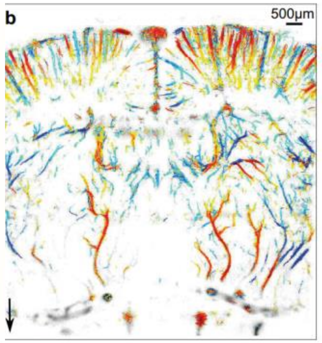
Ultrasound Hardware

Some historical context









1794
Spallazani discovered 'non-audible' sound

1877
Pierre Curie discovered piezo-electric effect

1912
Destruction caused by U-boats in WWI provides drive for development of SONAR

1917
Langevin produced ultrasound device using piezoelectrics

1942
Dussik investigates ultrasound transmission of the brain

1950s
Pulsed ultrasound developed at multiple institutions enabling 'B Mode' imaging

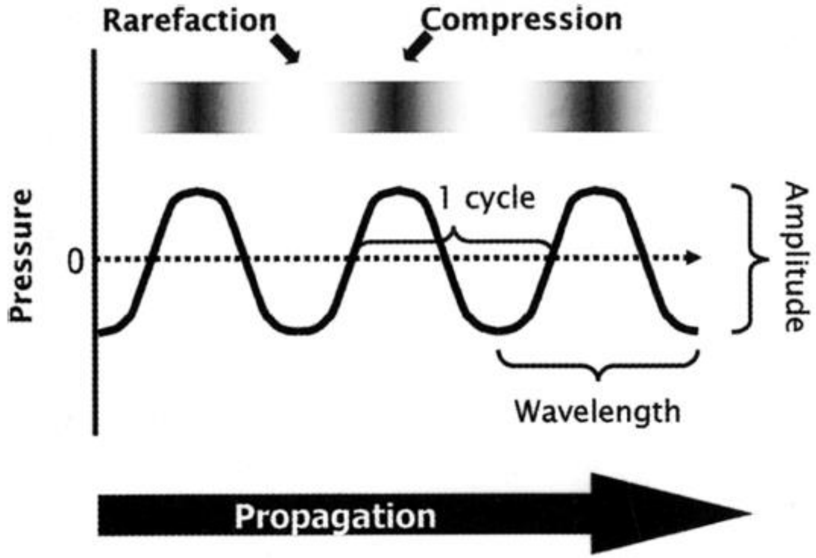
1980s
Real time ultrasound possible

1990s
3D and 4D ultrasound emerge

2000s
Elastography
Ultrafast ultrasound imaging

2010s
Functional ultrasound imaging
Ultrasound localization microscopy

Ultrasound refers to mechanical waves with a frequency greater than 20 kHz



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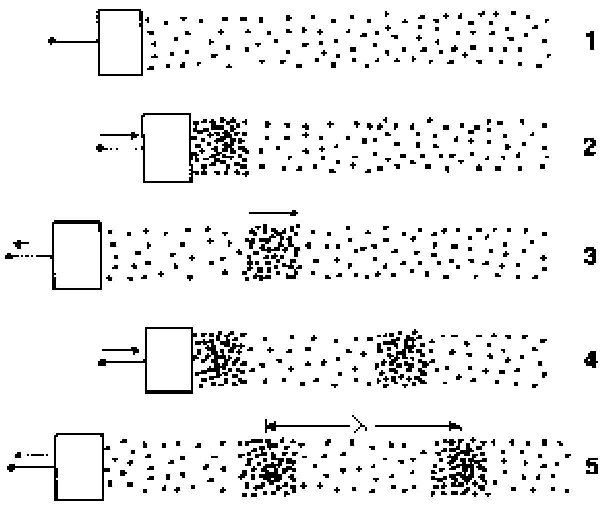
Speed of sound: speed at which the wave propagates, units of metres per second (ms^{-1})

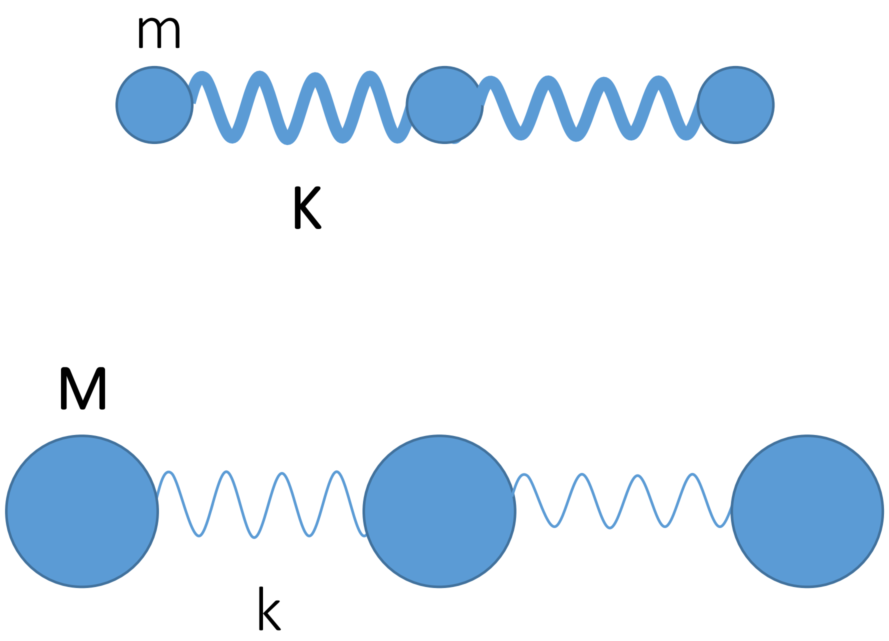
Wavelength: distance between successive compressions, units of metres (m)

$$c = f\lambda$$

Frequency: number of compressions passing a stationary observer per second, units of Hertz ($1Hz = 1s^{-1}$)

The speed of sound is determined by the properties of the medium: density





The quantitative relationship to the speed of sound is given by their ratio

$$c = \sqrt{\frac{K_a}{\rho}}$$

Adiabatic elastic bulk modulus

Density

Tissues generally differ more in stiffness than in density, so although bone is much denser than muscle, it has a higher speed of sound because it is much stiffer

The propagation of ultrasound in a medium is determined by the acoustic impedance

The specific acoustic impedance of a plane wave is:

$$Z_{sp} = \frac{p}{\dot{u}}$$

p = acoustic pressure

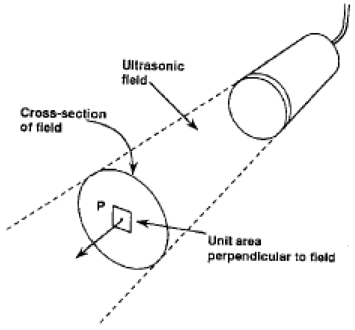
$\dot{u}$ = particle velocity

For perfect plane wave conditions, the characteristic acoustic impedance of the medium is equal to Z_{sp}:

$$Z_{sp} = Z_0 = \rho_0 c_0$$

ρ_0 = density of the medium

c_0 = speed of sound in the medium



The (time averaged) product of pressure and particle velocity gives the intensity of the wave, or the energy flowing per unit time through unit area:

$$I = \frac{1}{2} Z_{sp} \dot{u}^2 = \frac{1}{2} \frac{p_0^2}{Z_{sp}}$$

Speed of sound through tissue depends on fat, collagen and water content

- An increase in water and fat content leads to a decrease in wave speed.
- An increase in collagen content leads to an increase in wave speed.
- Resolution is related to the wavelength.
    - A wavelength of 0.8 mm and wave speed of 1540 ms^{-1} corresponds to a frequency of 2 MHz.

Acoustic properties vary tremendously between different biological tissues

Material	ρ Density (kg m^{-3})	c Speed (m s^{-1})	Z Impedance (Mrayl)
Perspex	1180	2680	3.16
Air	1.2	330	0.004
Bone	1912	4080	7.8
Water	1000	1480	1.48
Lung	400	650	0.26
Fat	952	1459	1.38
Soft Tissue	1060	1540	1.63

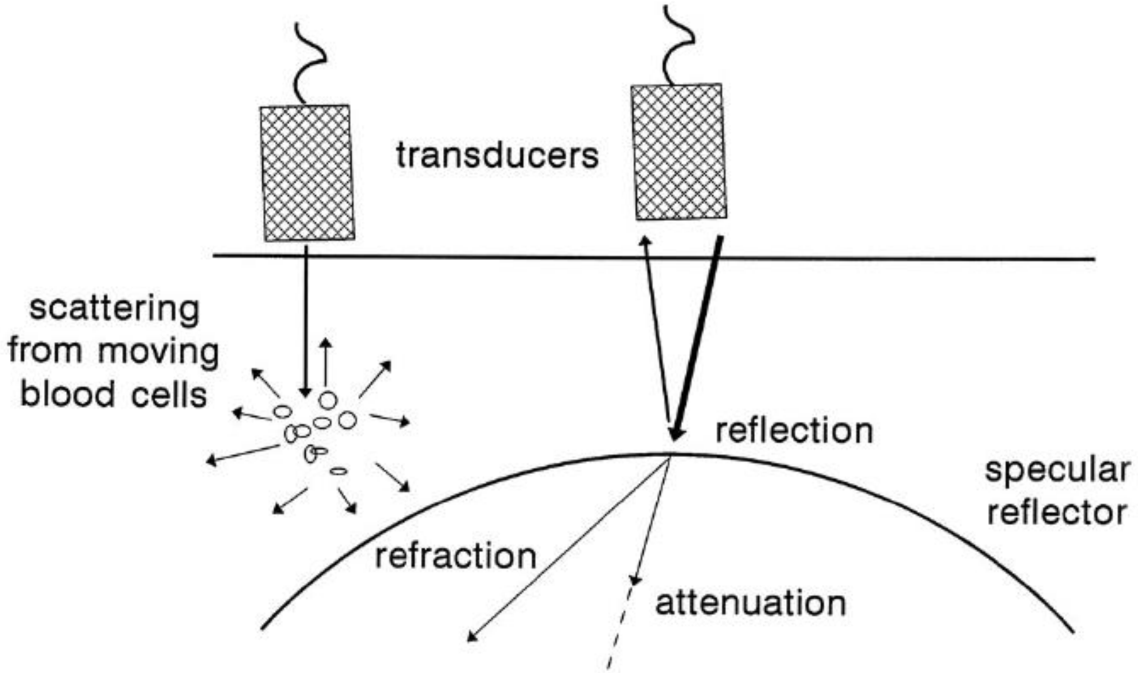
The acoustic mismatch, or reflection coefficient, is the key source of contrast in medical ultrasound

Interface	Reflected intensity (%)
Fat / muscle	1.1
Bone / muscle	41.0
Soft tissue / lung	52.5
Soft tissue / air	99.9
Soft tissue / water	0.2

Water based gel is used to remove air interfaces between transducer and skin

Ultrasound can undergo a range of interactions in soft tissue

- Reflection
- Scattering
- Refraction
- Absorption



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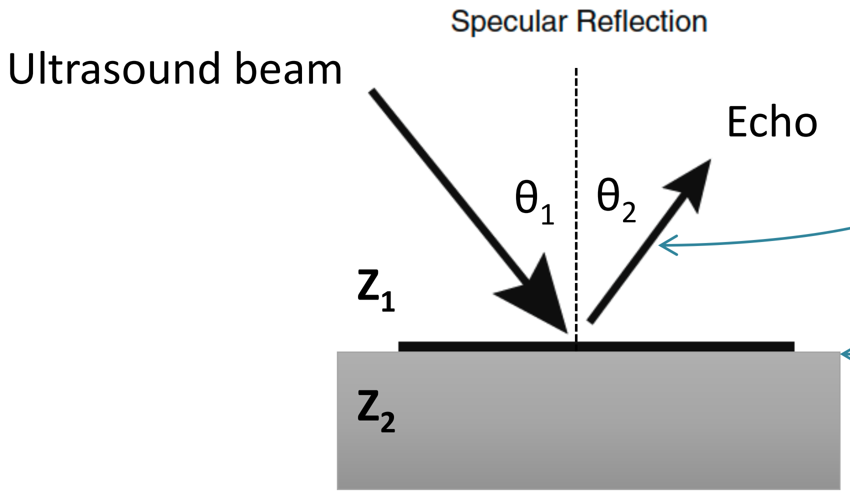
Reflection of ultrasound occurs at boundaries of media with different acoustic impedances

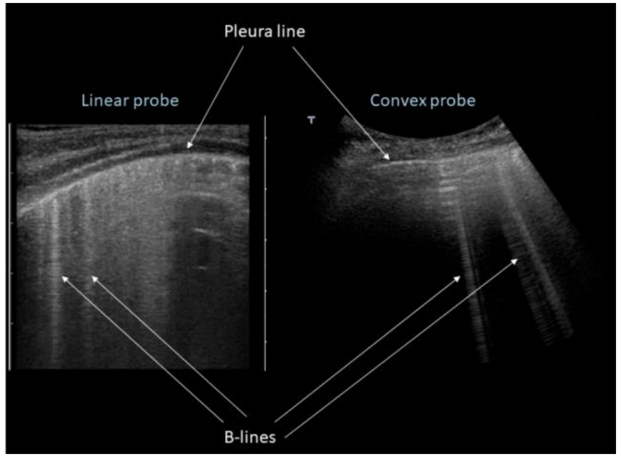


$$q_1 = q_2$$

Reflection coefficient for normal incidence: $\dfrac{I_r}{I_i} = \dfrac{(Z_1 - Z_2)^2}{(Z_1 + Z_2)^2}$

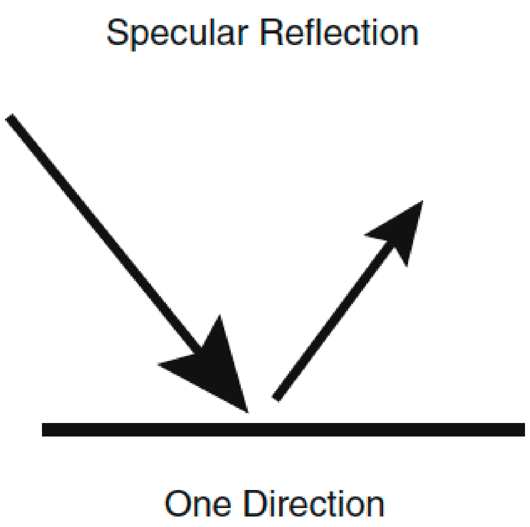
Reflections can be good and bad!

Interface	Reflected Intensity (%)
Fat/kidney	0.6
Fat/muscle	1.1
Bone/muscle	41.0
Soft tissue/air	99.9
Soft tissue/lung	52.5
Soft tissue/PZT	79.8
PZT/air	99.99



- Gel is used to remove air interfaces between the transducer and the skin.
- It is difficult to image the lung and behind bones, and impossible to image across bowel gas.
- There are not many interfaces in the body that are large and smooth on the scale of ultrasound wavelength (1mm or less). Examples include the diaphragm/liver, bladder wall and some large blood vessels.

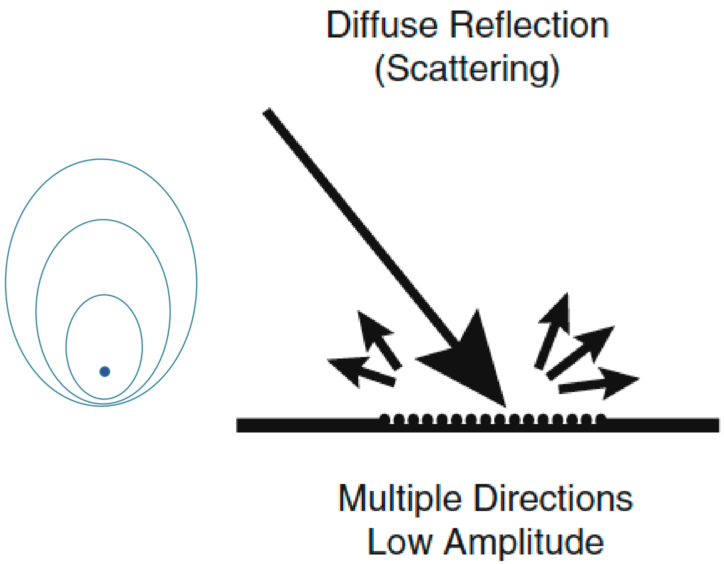
Specular reflections of ultrasound are analogous to looking into a mirror



Scattering condition: $\boxed{s >> \lambda}$

Scattering strength: Low

Scattering can arise from rough or irregular surfaces

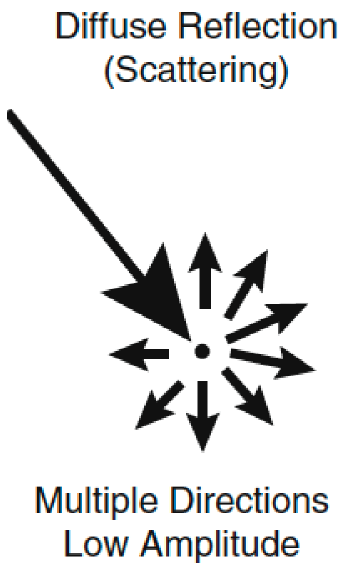


Scattering condition: $\boxed{s \approx \lambda}$

Scattering strength: Moderate

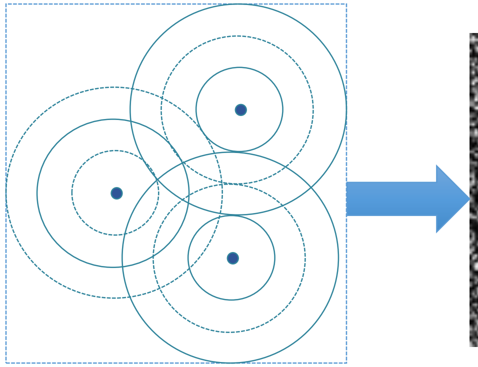
At 2.5 MHz, the signal from red blood cells is 1/1000 that of a fat/muscle specular reflection

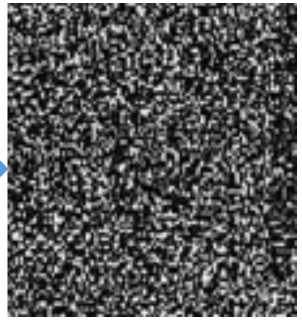
Scattering can also arise from objects smaller than the ultrasound wavelength



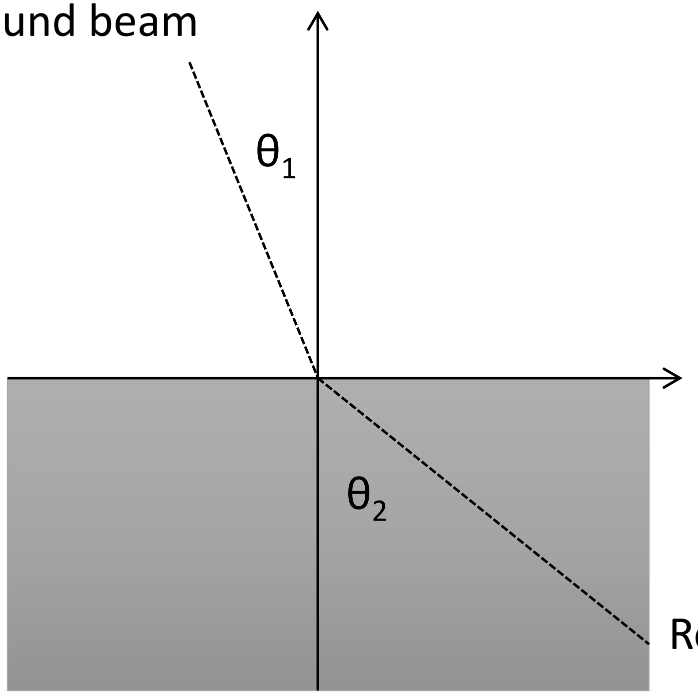
Scattering condition: $\boxed{s \ll \lambda}$

Scattering strength: High





Refraction occurs when ultrasound is incident on a medium with a different speed of sound



$$\boxed{\frac{\sin(q_1)}{\sin(q_2)} = \frac{c_1}{c_2}}$$

If $c_1 > c_2$, beam bends toward normal

If $c_1 < c_2$ and θ_1 is large may get total internal reflection

Interface	c_1/c_2	Angle of incidence	Angle of refraction
Fat / muscle	1.09	5	4.6
Bone / fat	2.81	5	1.8

Absorption occurs when mechanical energy of the ultrasound beam is converted to heat energy

- Absorption in tissues is strong, accounts for 80 – 90 % of all energy loss by an ultrasound beam
- Depends on:
  - Frequency
  - Viscosity of the medium
  - Relaxation time of the medium
- Relaxation:
  - at low frequencies the particles move easily with the passing pressure wave and return to equilibrium before the next disturbance so all energy is transmitted
  - at higher frequencies, particles are unable to keep up so do not pass all energy

Attenuation describes the loss of intensity as ultrasound passes through the tissue

- Attenuation includes both scattering and absorption

$$I = I_0 e^{-afl}$$

- Where *I* is intensity, *a* ~ 0.5 dB cm^{-1} MHz^{-1} in soft tissue, *f* is frequency (MHz) and l is thickness of tissue (cm)
- Analogy to X-ray HVT:

Material	HVT (cm) @ 2MHz	HVT (cm) @ 5MHz
Air	0.06	0.01
Bone	0.1	0.04
Liver	1.5	0.5
Blood	8.5	3.0
Water	344	54

To avoid the exponential in attenuation calculations, the decibel scale is used

$$\frac{I}{I_0} = e^{-afl}$$

I/I_0	dB
1,000,000	60
100	20
10	10
2	3
1	0
0.01	-20

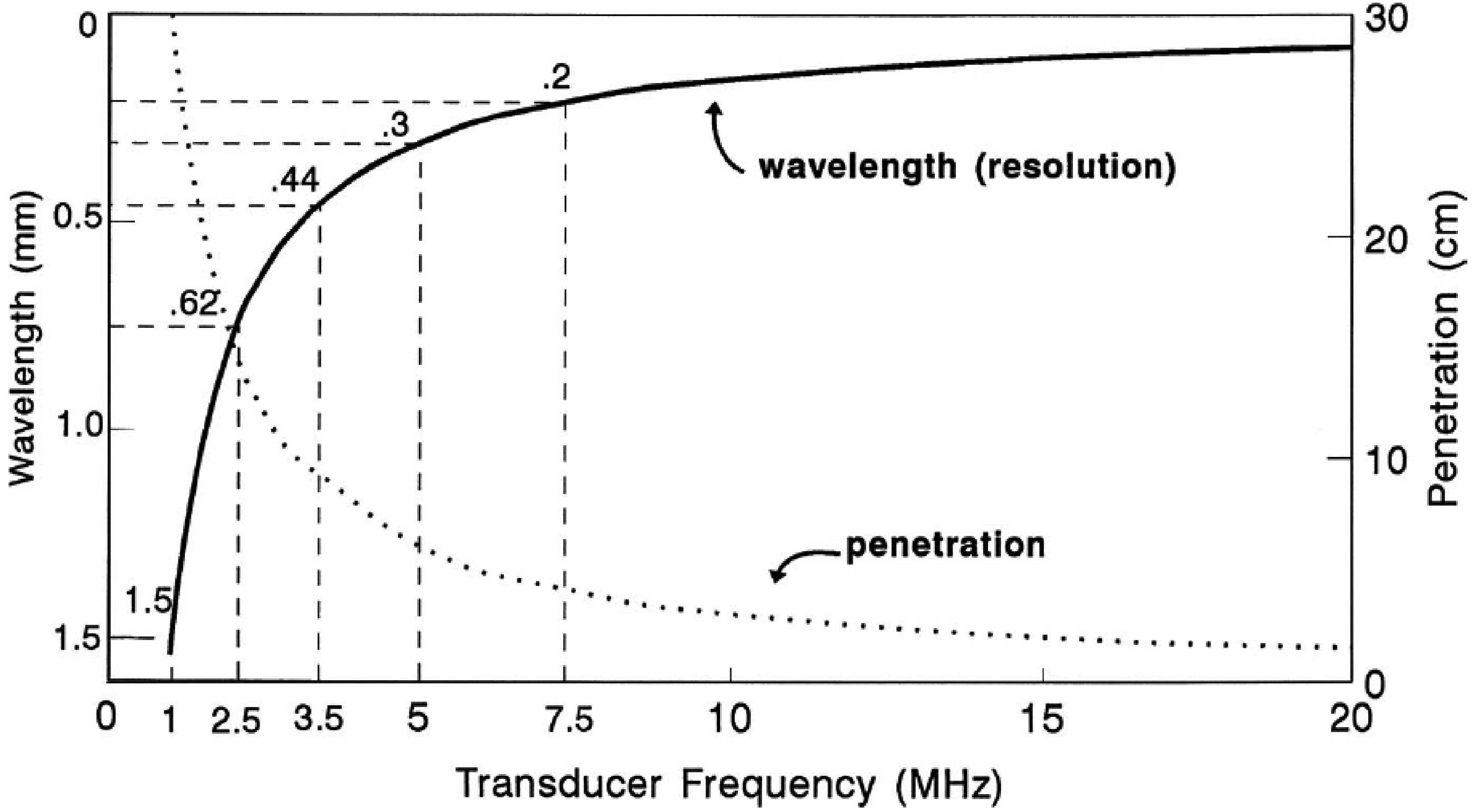
Echo pressure amplitudes vary by a factor of 10^5 or greater, so a logarithmic scale helps:

$$\text{Intensity ratio (dB)} = 10\log_{10}\left(\frac{I}{I_0}\right)$$

$$\text{Amplitude ratio (dB)} = 20\log_{10}\left(\frac{A}{A_0}\right)$$

(Factor of 2, since intensity is proportional to square of amplitude)

Ultrasound imaging thus requires a trade off between imaging resolution and penetration depth



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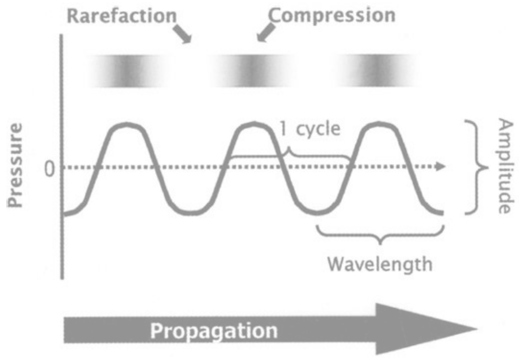
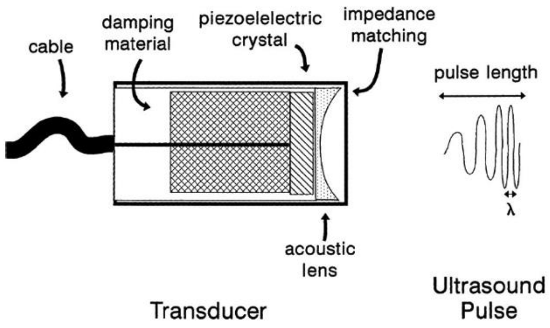
Summary 1: Ultrasound Physics

- Impedance mismatch causes acoustic reflections
- Ultrasound can undergo reflection, refraction, absorption and scattering in tissue
  - Depends on the angle of incidence, size of the object relative to the ultrasound wavelength, acoustic impedance
- Resolution must be traded against penetration depth because
  - High frequency ultrasound provides better spatial resolution
  - but high frequency ultrasound is strongly attenuated in tissue



Ultrasound Physics



Ultrasound Hardware

Ultrasound imaging is based on the 'pulse-echo' principle

- The distance of a reflecting object can be established by the return time of a short pulse if the speed of the pulse is known
- For a measured time t and known speed of sound c, the distance in the pulse-echo technique is given by d:

$$d = \frac{ct}{2}$$

Ultrasound imaging is based on the 'pulse-echo' principle

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- The maximum pulse repetition frequency is therefore

$$\mathrm{PRF} = \frac{c}{2d}$$

- The frame rate, or number of images produced per second is then dependent on the number of scan lines needed to make up the B-mode image:

$$\text{Frame rate} = \frac{c}{2dn}$$

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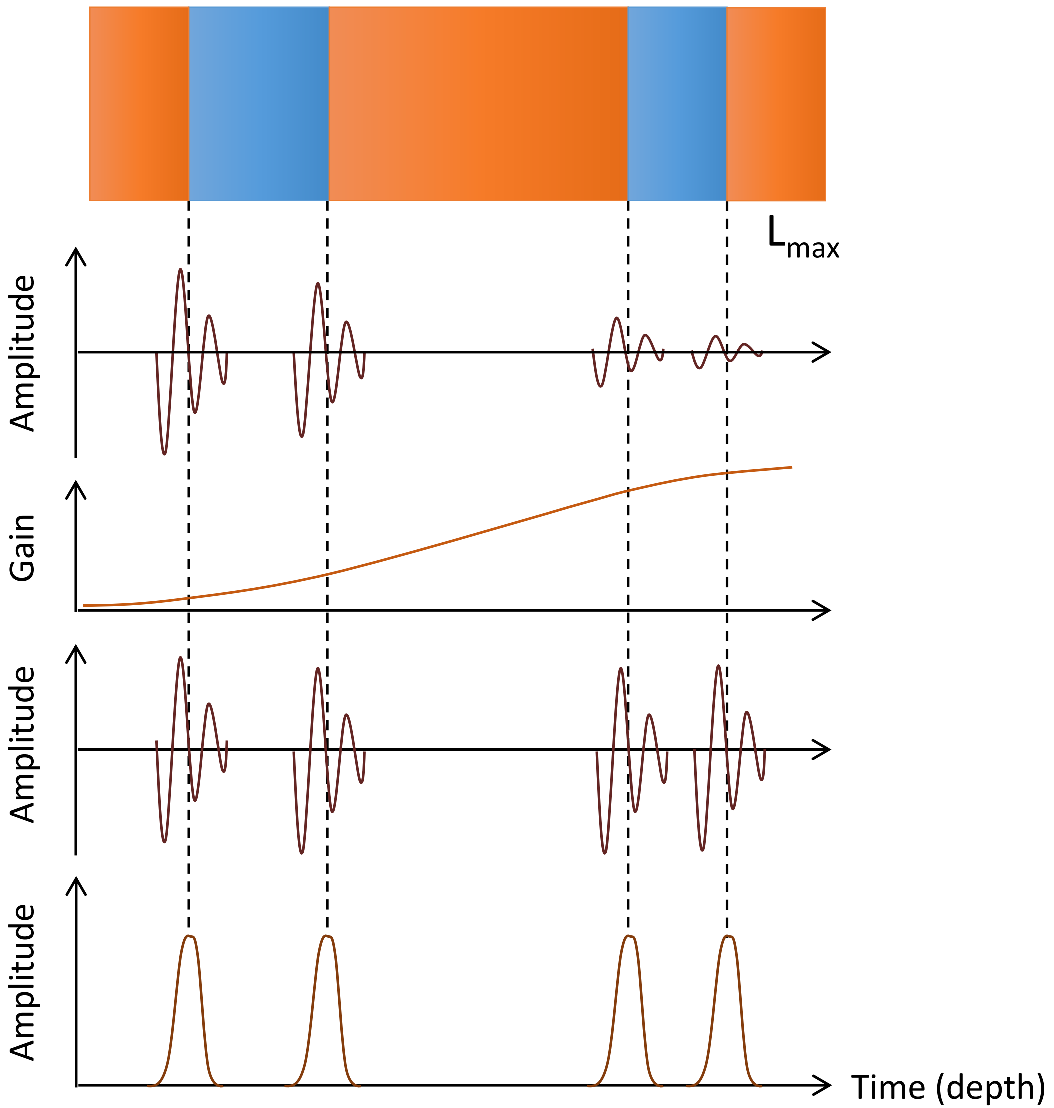
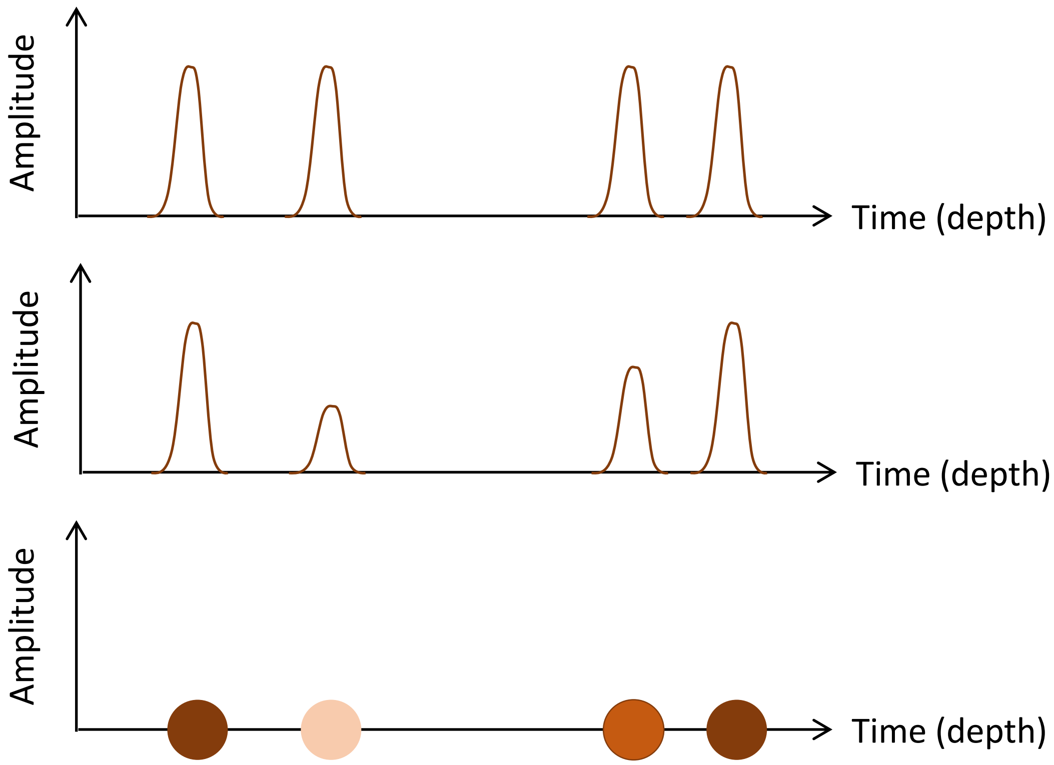
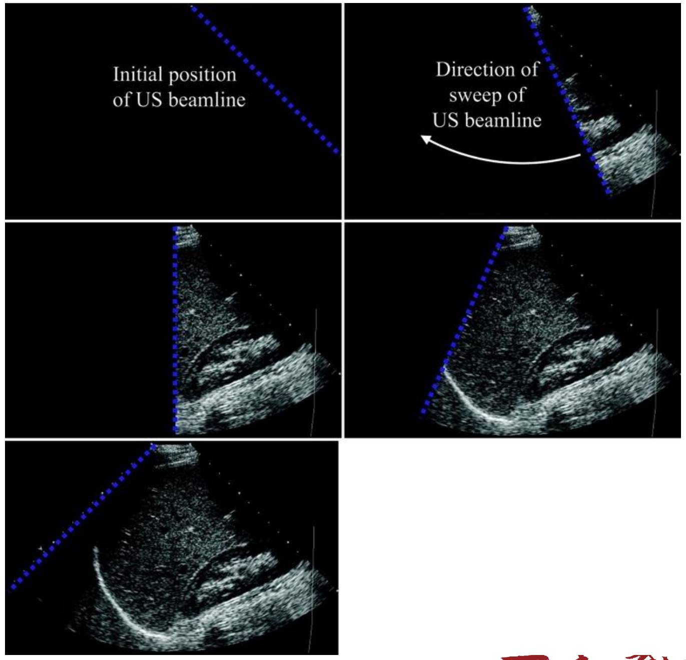
Amplitude (A) mode ultrasound displays the ultrasound echoes along one beam, or ‘A Line’



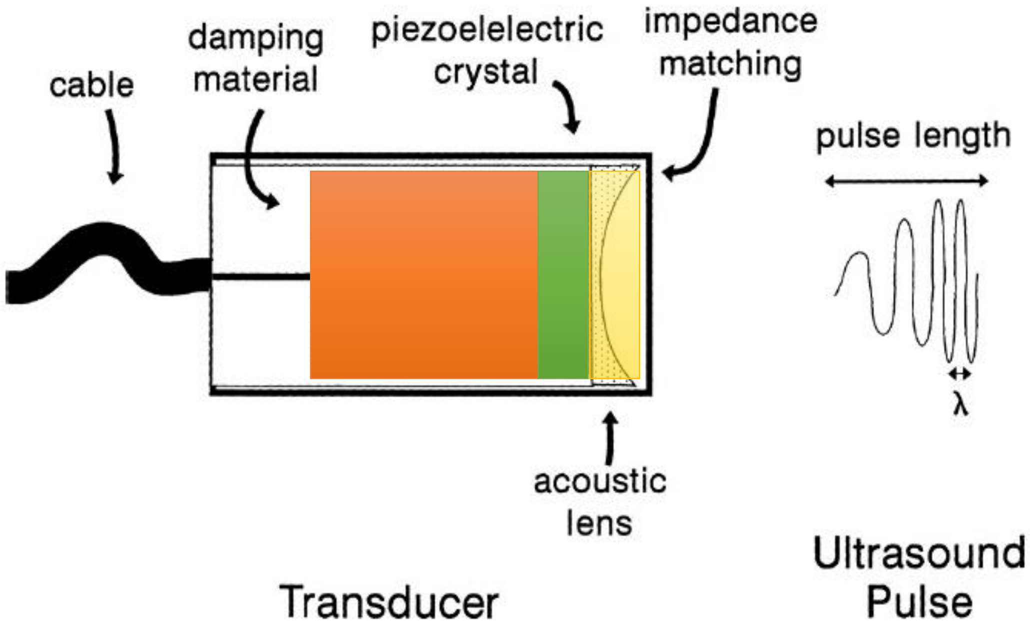
Brightness (B) mode uses each individual echo strength to build up a 2D image

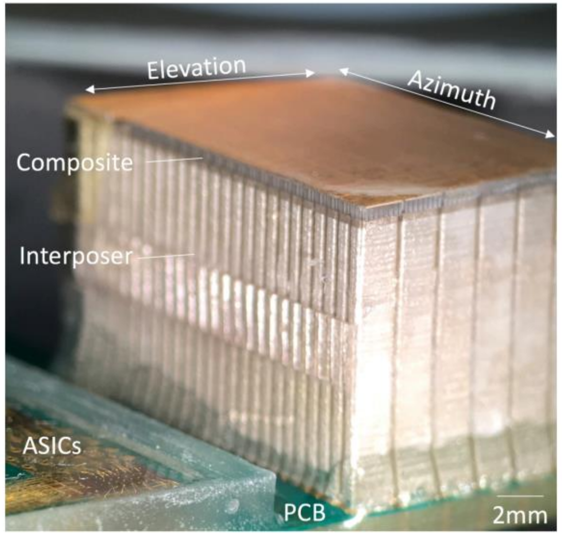


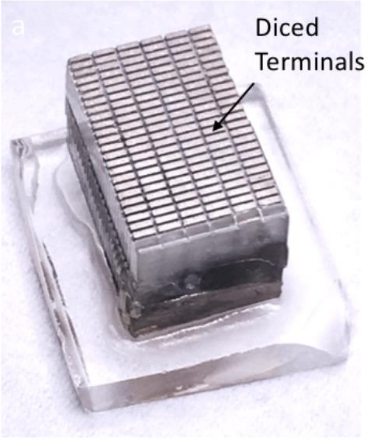


More reflective structures appear brighter

An ultrasound transducer is composed of three main parts







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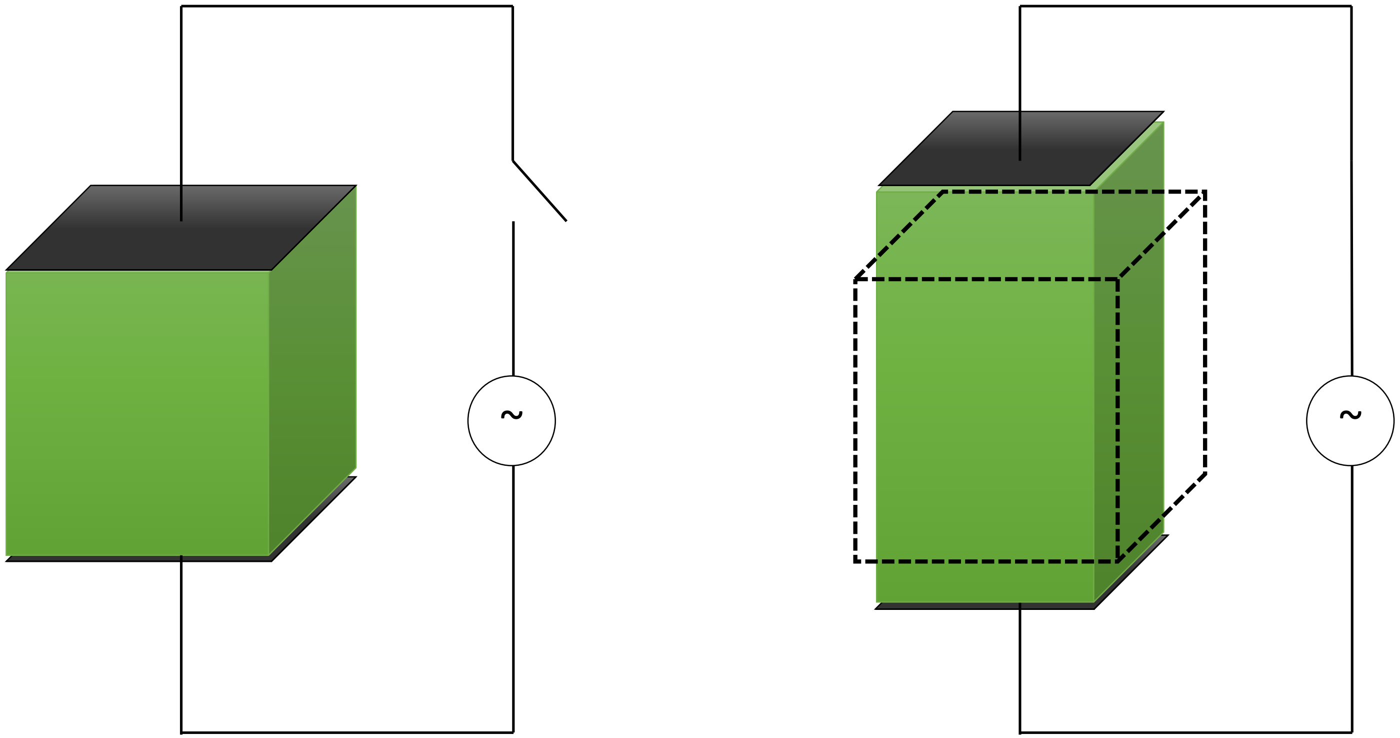
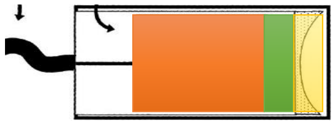
Piezoelectric elements both generate and detect ultrasound waves

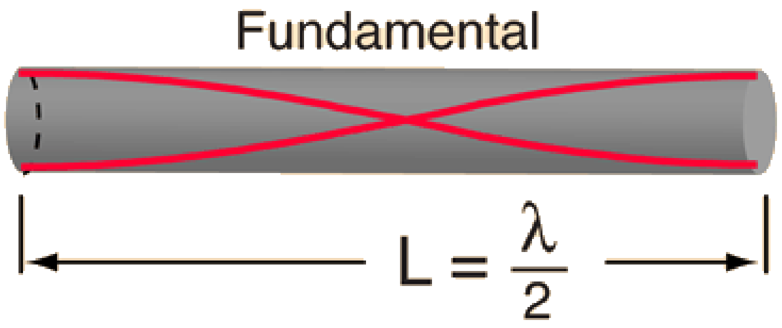


The application of a short (~ 1 μs) pulse of high voltage (~ 150 V) causes PZT contraction and subsequent vibration at a natural resonant frequency

The crystal thickness (l) determines the ultrasound frequency (f) produced



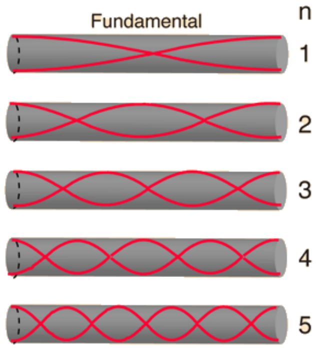
The time for the wave to make a return trip between the faces of the crystal is one period, T (units of seconds):



$$T = \frac{2l}{c_{PZT}}$$

$$f_0 = \frac{1}{T} = \frac{c_{PZT}}{2l}$$

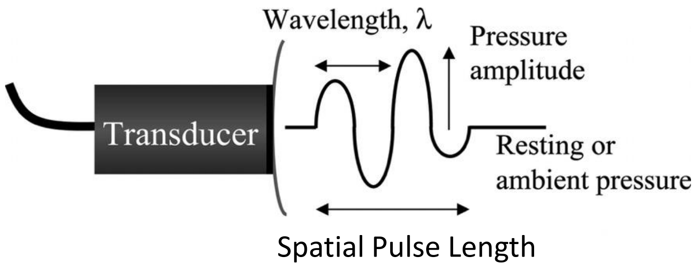
The fundamental mode (maximum pressure) occurs when $l = \frac{\lambda}{2}$



The pulse duration determines ultrasound axial imaging resolution

- Axial resolution is determined by the speed of sound (c) and the pulse duration (τ):

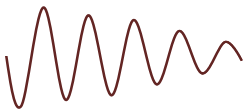
$$d_{a\,min} = \frac{\tau c}{2}$$

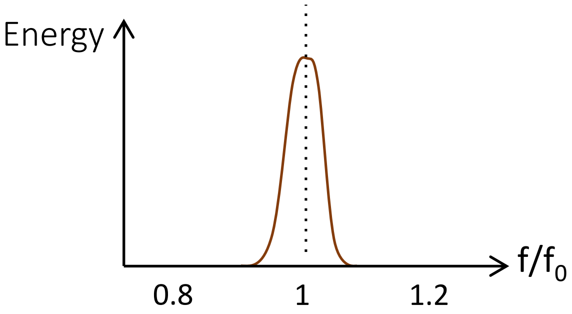


- Increasing the ultrasound frequency means that each pulse can be made even shorter in time, hence the axial spatial pulse length (τc) is smaller, giving better resolution
- The lateral extent of the disturbance must be narrower than the distance between the features to be resolved (more later)

The pulse frequency bandwidth is an important consideration in transducer design

Long spatial pulse length
Poor axial resolution
Narrow frequency bandwidth

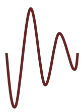


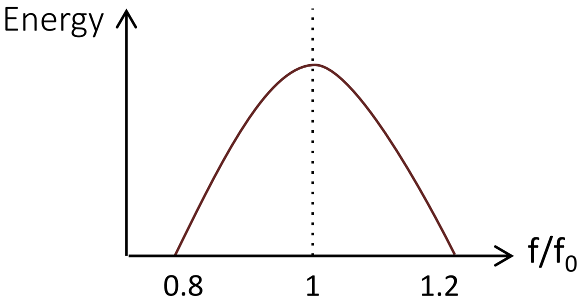


"Ringing"

Pulse bandwidth Δf
~ 1/pulse duration

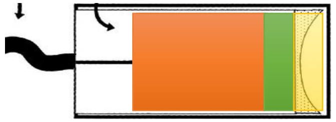
Short spatial pulse length
Good axial resolution
Wide frequency bandwidth

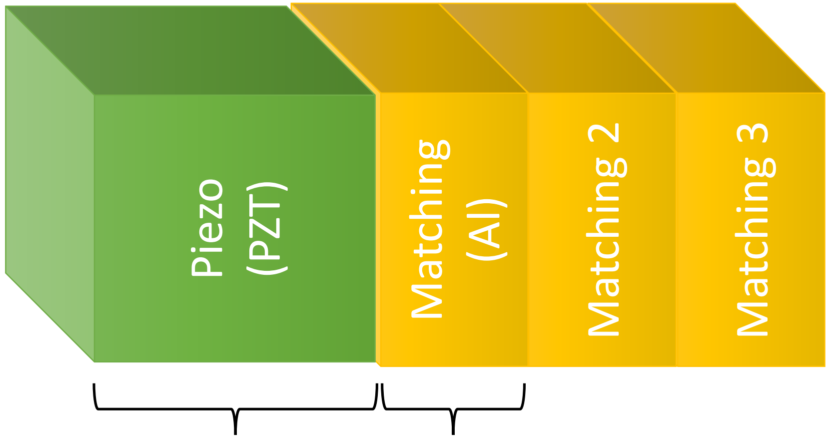




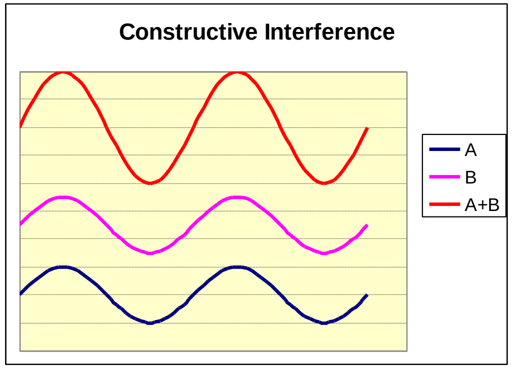
Damped
High sensitivity
Optimal axial resolution

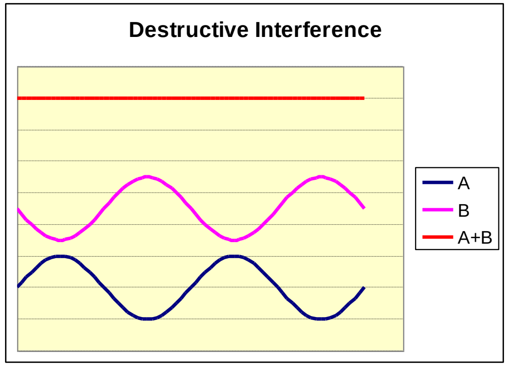
Impedance matching determines the intensity of ultrasound emitted





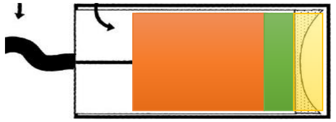
$L = \frac{l}{2}$ $\quad$ $L = \frac{l}{4}$

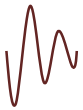


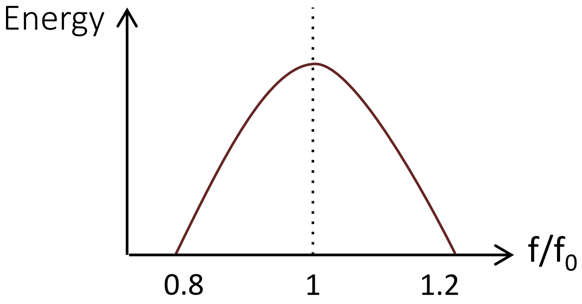


The material should have an acoustic impedance of: $Z_{match} = \sqrt{Z_{PZT} + Z_{tissue}}$

Damping determines the bandwidth of ultrasound emitted





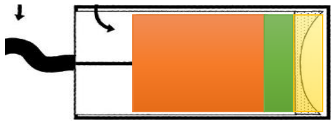


Damped

High sensitivity

Optimal axial resolution

Q factor describes how damped an oscillator is

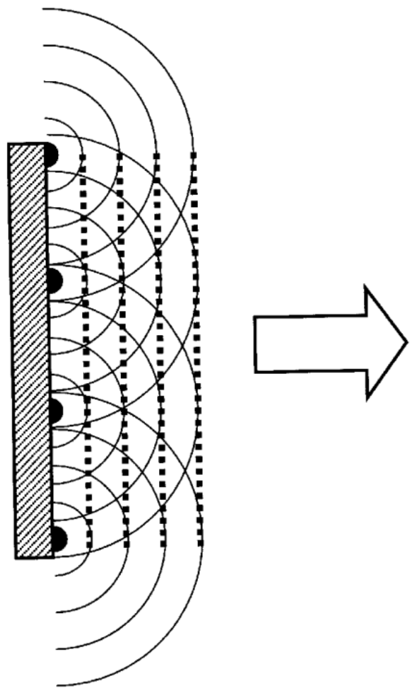


$$Q = \frac{f_0}{\Delta f} \quad \text{or} \quad Q = \frac{\text{energy stored per cycle}}{\text{energy lost per cycle}}$$

High Q transducer is lightly damped, so good for continuous wave ultrasound

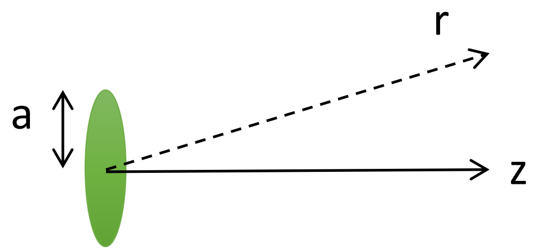
Low Q transducer is highly damped, so good for pulse echo imaging ultrasound

The simplest ultrasound case is a continuous wave created by a circular disk of PZT



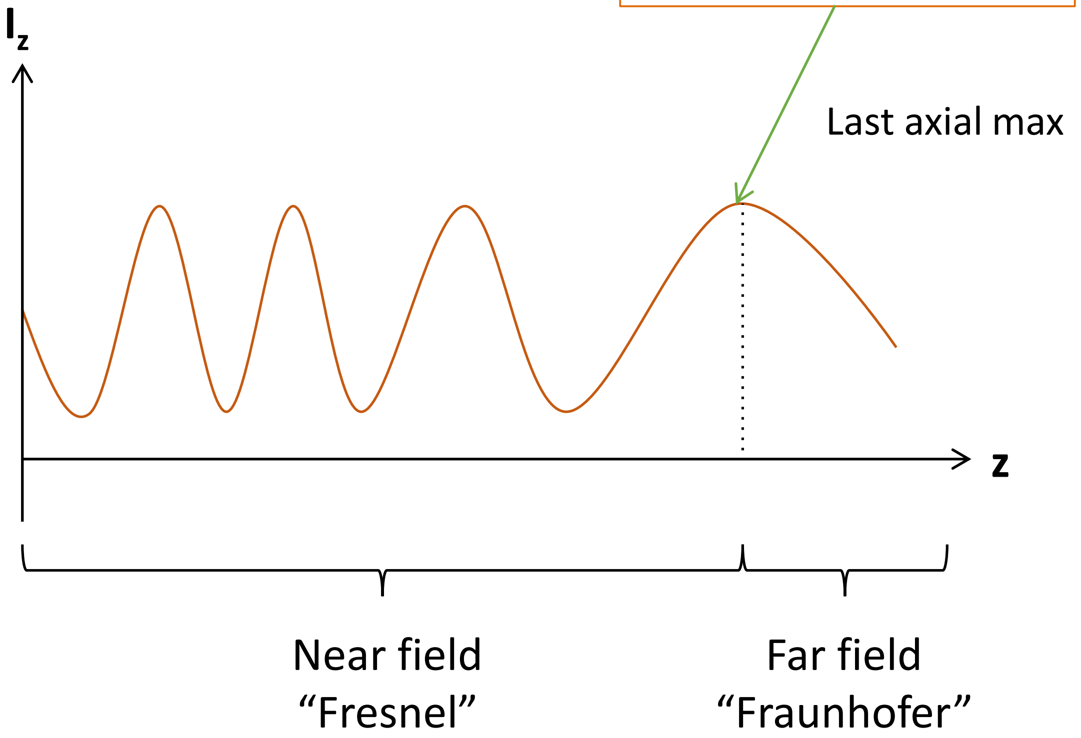
- The pressure (and intensity) field can be calculated using Huygen's principle for superposition of wavelets.
- Every point on the transducer surface is considered to emit a spherical wave.
- The resulting pressure field is found by summing all the waves, taking into account the phase of each contribution.
- The mathematical integral is difficult to solve and is typically treated numerically.

Considering the axial behaviour along the z axis normal to the centre of the disk:

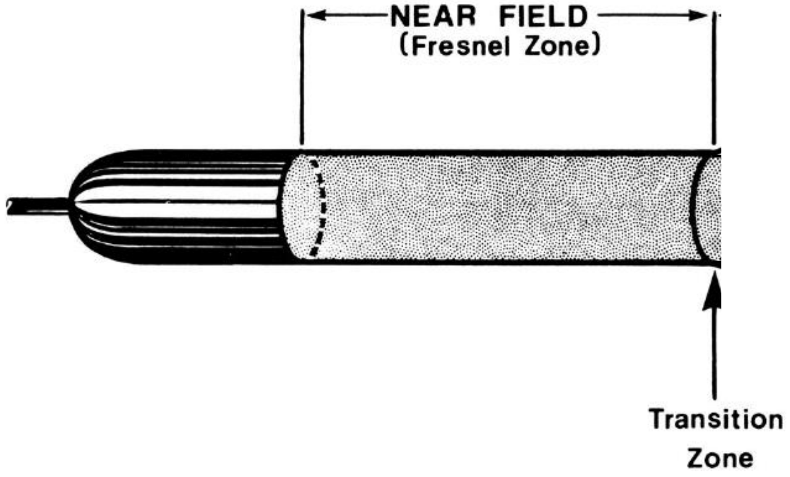


$$\frac{I_z}{I_0} = \sin^2\left[\frac{\pi}{\lambda}\left(\sqrt{a^2+z^2} - z\right)\right]$$

$$a^2 \gg \lambda^2 \Rightarrow z'_{\max} = \frac{a^2}{\lambda}$$



In the far field regime, the cylindrical ultrasound beam diverges



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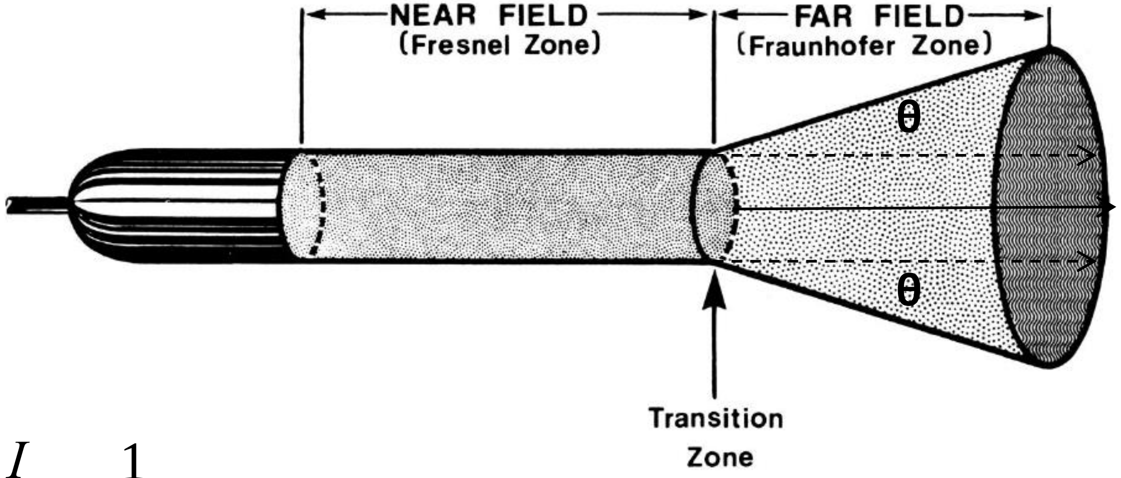

Lateral behaviour

On axis: $\frac{I_z}{I_0} \mu \frac{1}{z^2}$

Off axis: $\frac{I_z}{I_0} \mu \frac{2J_1(ka \sin q)}{ka \sin q}$ where J1 is a Bessel function of the first kind

The central lobe is confined to a region defined by: $\sin q = \frac{3.83}{ka} = \frac{0.61/}{a}$

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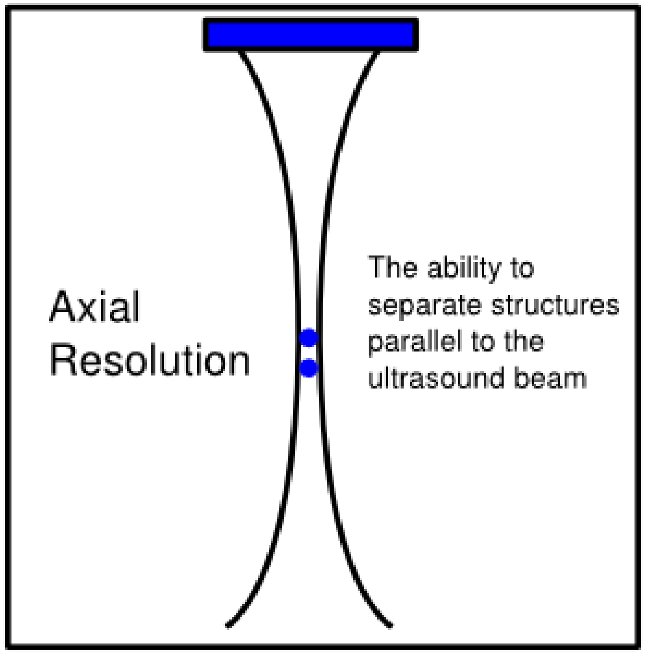
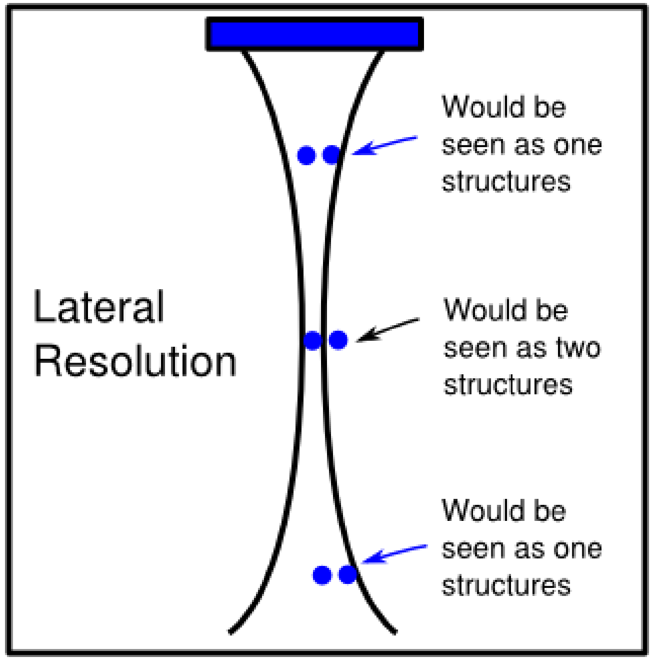
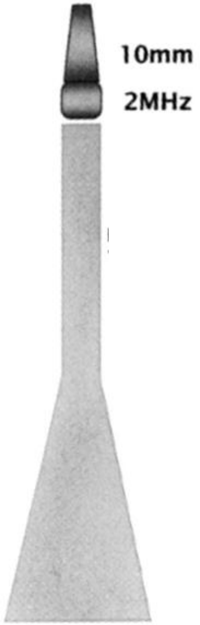
Lateral resolution is determined by beam divergence

- To optimise resolution the cross section of the beam should be narrow and the fresnel zone as long as possible, but is generally poorer than axial resolution
- This can be achieved by increasing centre frequency or physical size of PZT disk





The ultrasound beam can therefore be shaped by adjusting transducer geometry



$$z'_{\max} = \frac{a^2}{\lambda}$$

$$a = 1\text{cm},\ f = 2\ \text{MHz},\ c = 1540\ \text{ms}^{-1} \Rightarrow \lambda = 0.77\text{mm}$$

$$z'_{\max} = 13\text{cm},\ \theta = 2.7^\circ$$

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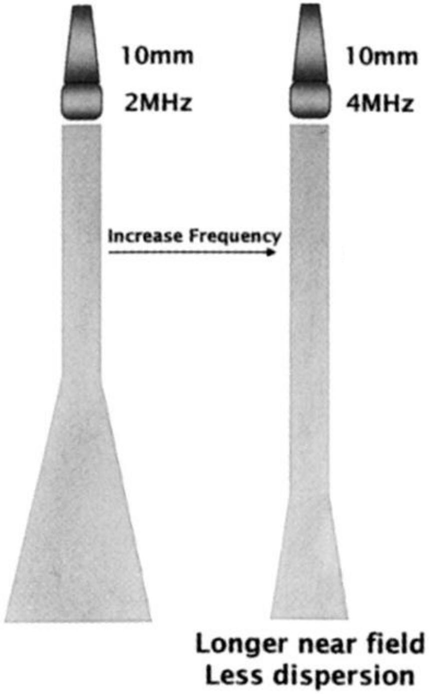

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$$\mathrm{a = 1cm,\ f = 4\ MHz,\ c = 1540\ ms^{-1}} \Rightarrow \lambda = 0.39\mathrm{mm}$$

$$z'_{\max} = 25.6\mathrm{cm},\ \theta = 1.4^{\circ}$$

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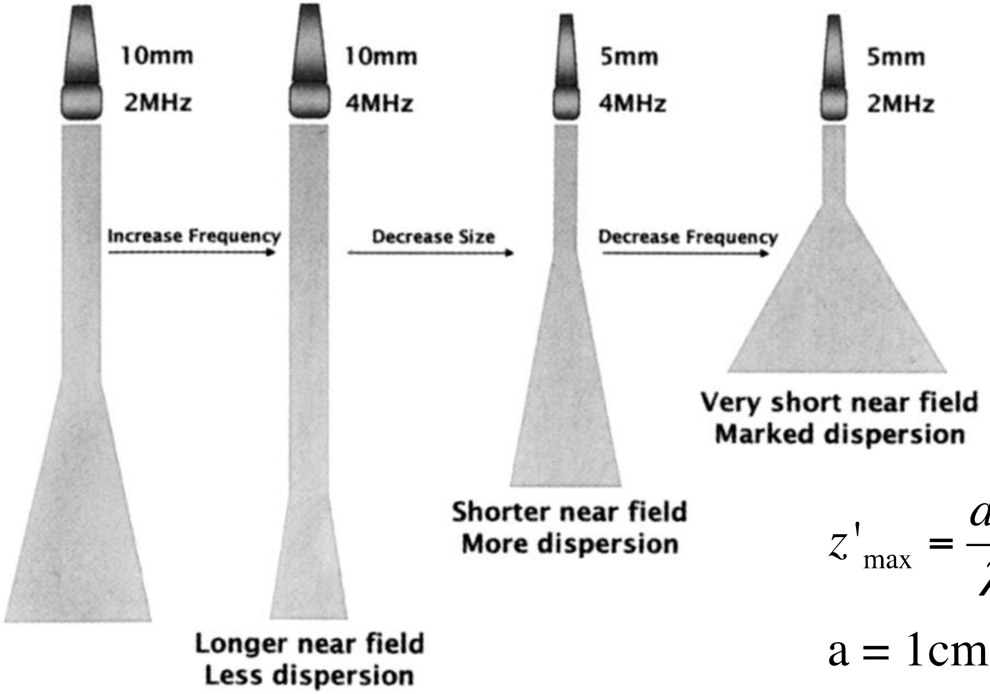

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$$z'_{\max} = 25.6\mathrm{cm},\ \theta = 1.4^{\circ}$$

$$\mathrm{a} = 0.5\mathrm{cm},\ \mathrm{f} = 2\ \mathrm{MHz},\ \mathrm{c} = 1540\ \mathrm{ms}^{-1}$$

$$z'_{\max} = 3.2\mathrm{cm},\ \theta = 5.4^{\circ}$$

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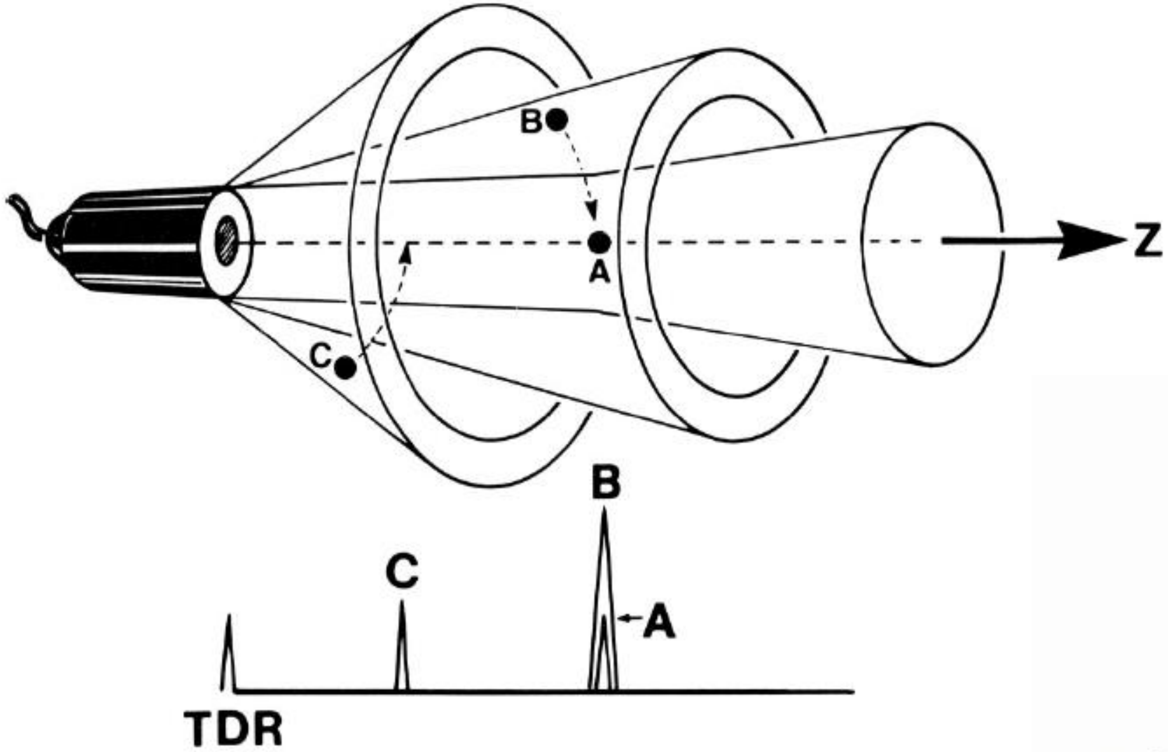
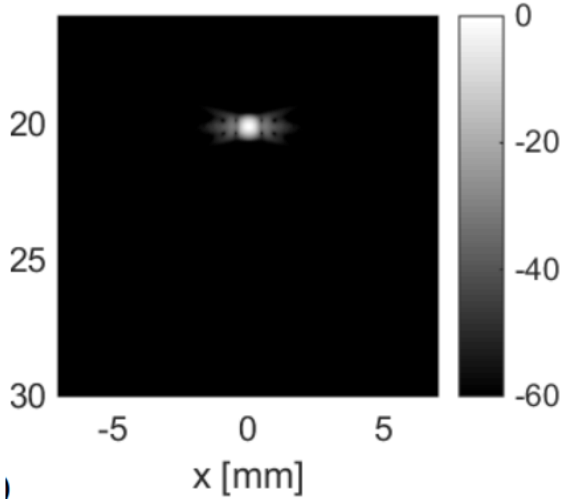
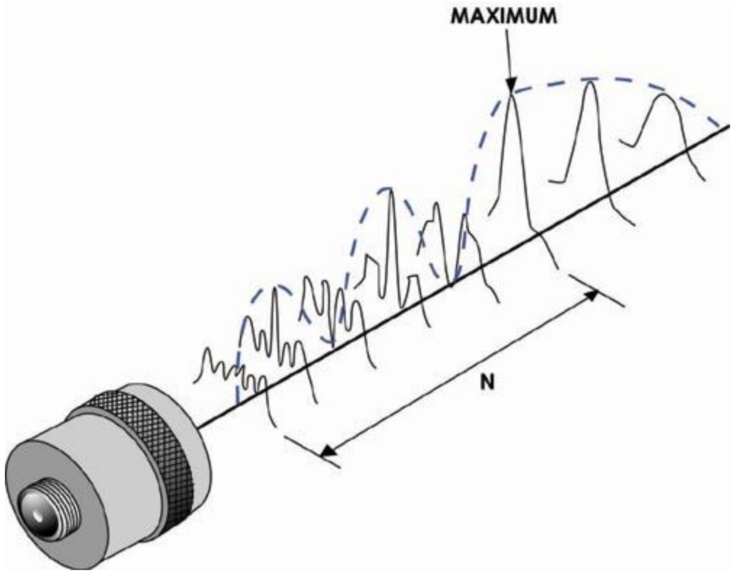
In addition to the inherent far field divergence, in reality no beam is a perfect cylinder







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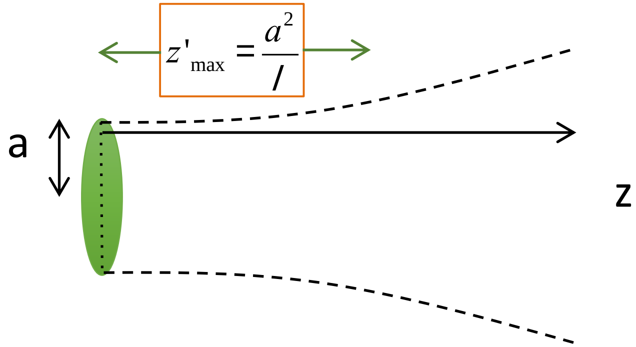
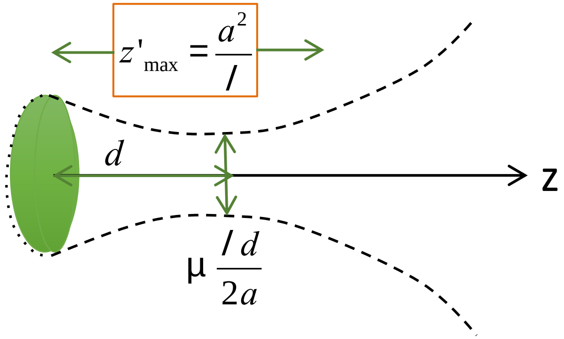
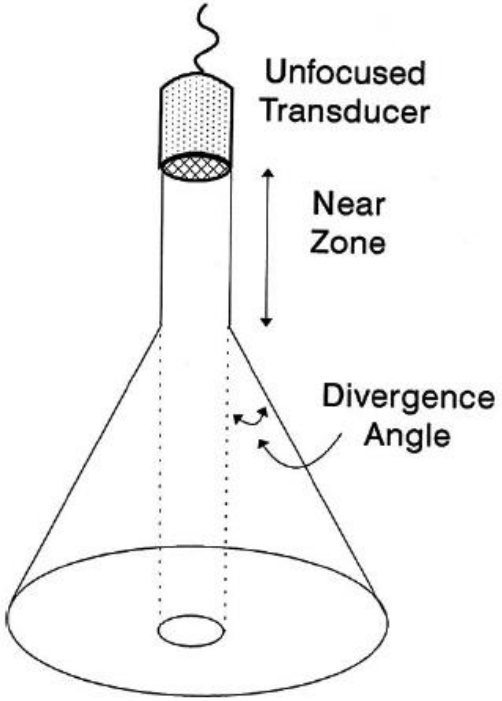
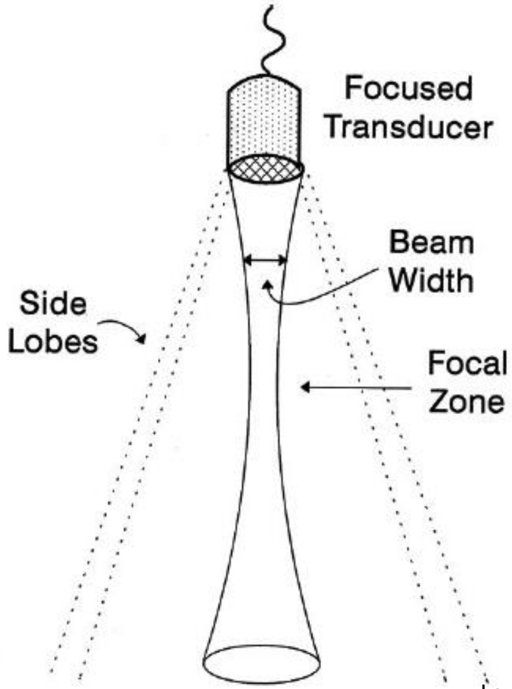
http://www.ndt-ed.org/EducationResources/CommunityCollege/Ultrasonics/EquipmentTrans/radiatedfields.htm

Using a curved transducer or acoustic lens allows for focusing of the ultrasound beam









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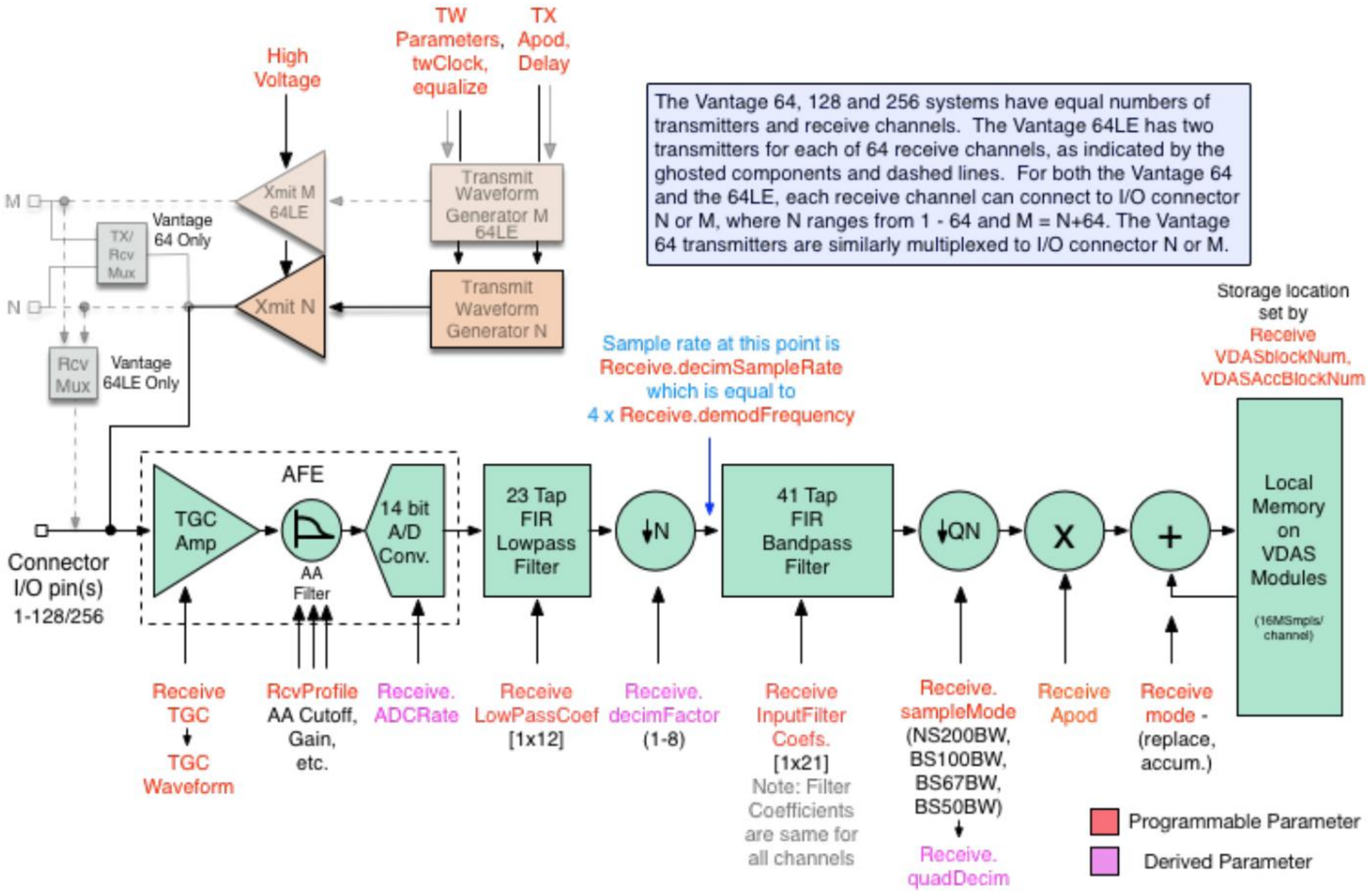
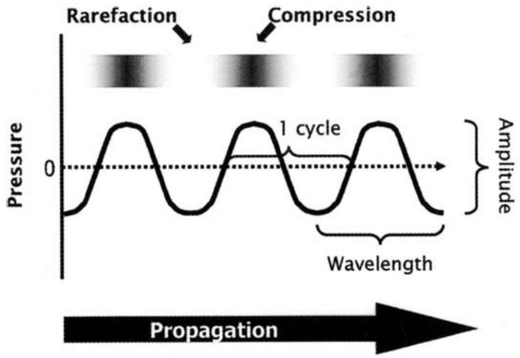
■ Signal path on an open ultrasound platform

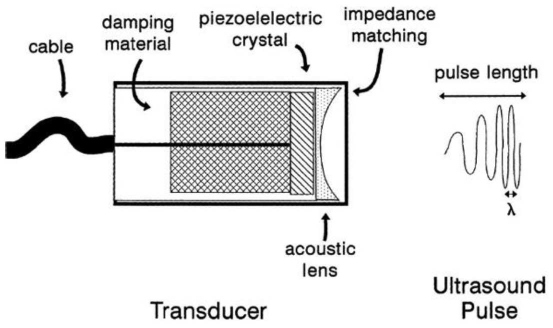


Summary 2: Ultrasound hardware

- The pulse echo approach is used to form a brightness (B) mode ultrasound image
- Transducers are composed of:
  - A piezoelectric element to generate and detect acoustic waves
  - Matching elements to maximise coupling of acoustic waves to the piezoelectric element
  - Backing material for damping to create a short pulse length and improve axial resolution
- The finite transducer size results in a far field divergence of the beam and addition of side lobe imperfections
  - Focusing can partially compensate for this
- …for next time: transducers are commonly combined into arrays for imaging



Ultrasound Physics



Ultrasound Hardware