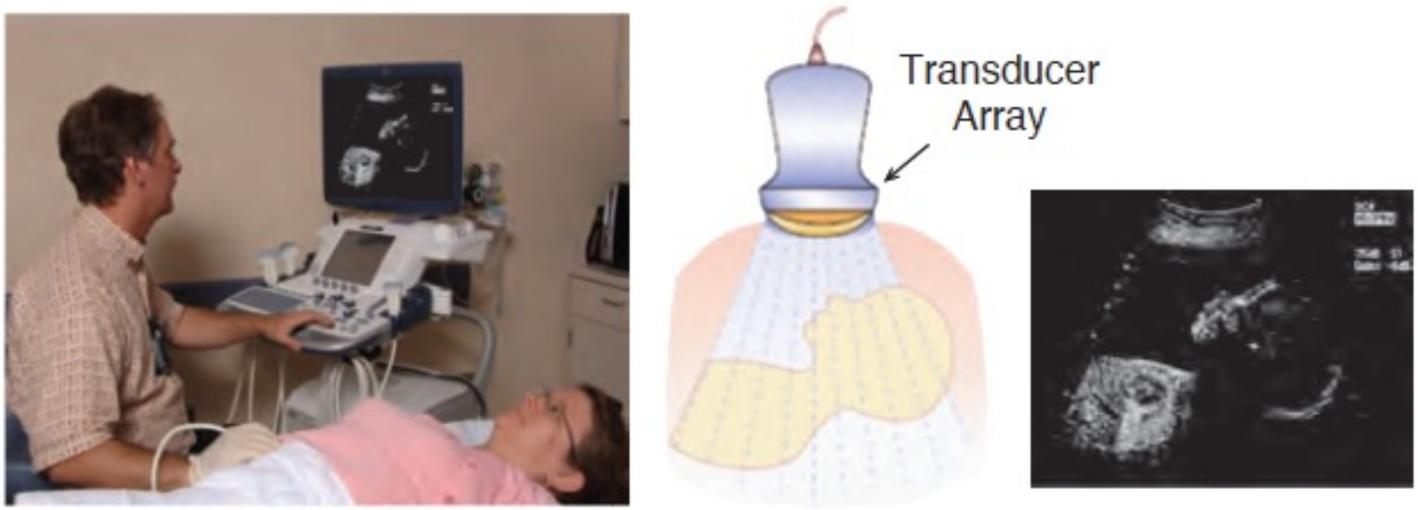
Ultrasonography (US)

- A small, spatially localized pulse of ultrasound is transmitted into the patient
- US reflected echoes are produced and detected



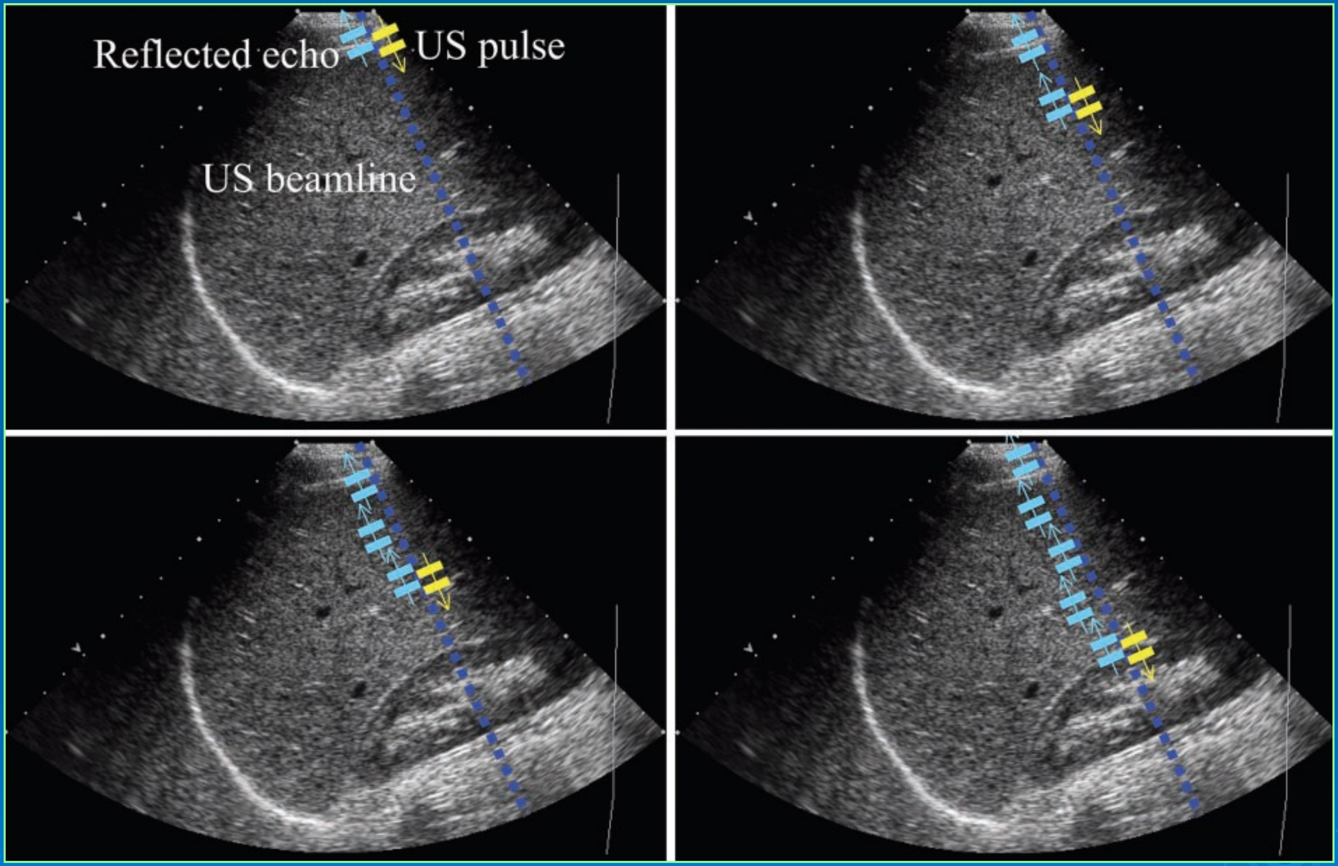
Interrogate body with acoustic "pulses" generated by transducer array

Acquire and record echoes arising from tissue interfaces

Construct "acoustic image" of tissues

Ultrasonography (US)

- US
  - Anatomical image
- Doppler US
  - Flow and functional imaging



"US is a relatively inexpensive, portable, safe, and real-time modality, all of which make it one of the most widely used imaging modalities in medicine."

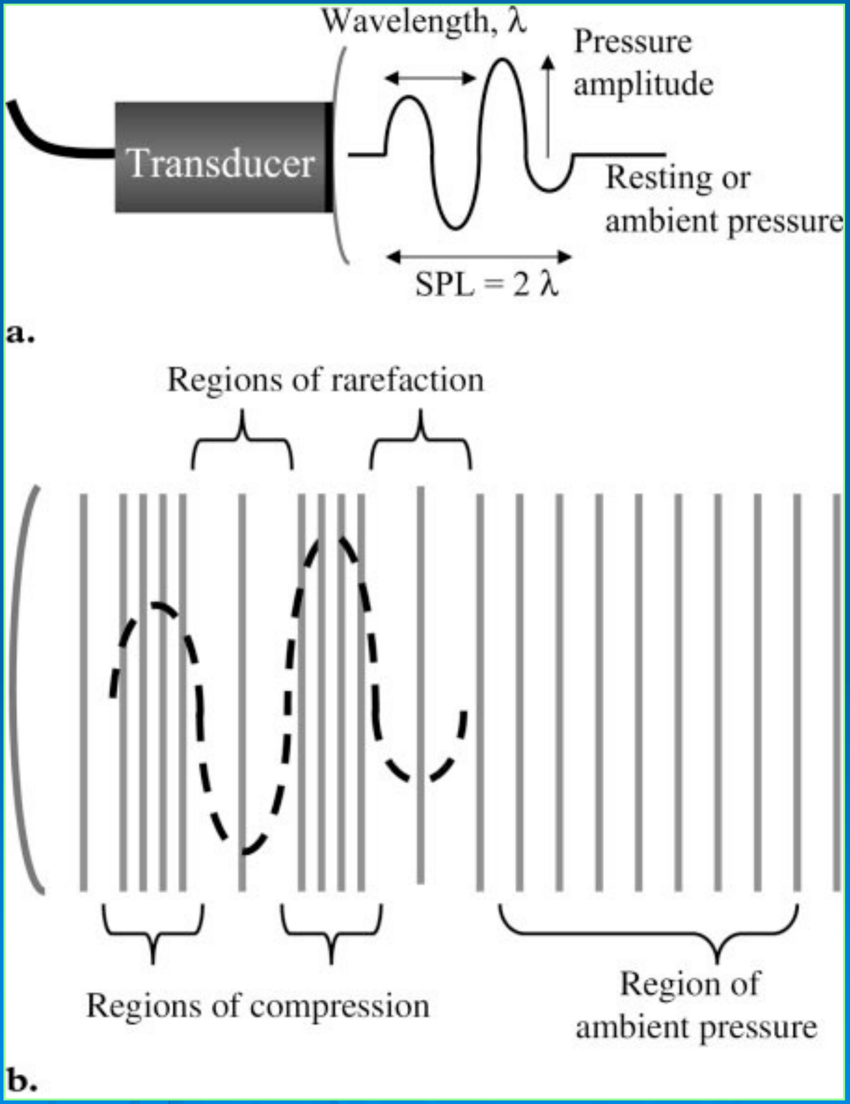
N.J. Hangiandreou RadioGraphics 2003; 23:1019–1033

Basic US Physics

Sound

- mechanical waves
  - Propagation in fluid
  - Longitudinal waves

- Audible acoustic spectrum:
  15 Hz – 20 kHz
- Ultrasound : ν>20 kHz
- Infrasound: ν<15 Hz

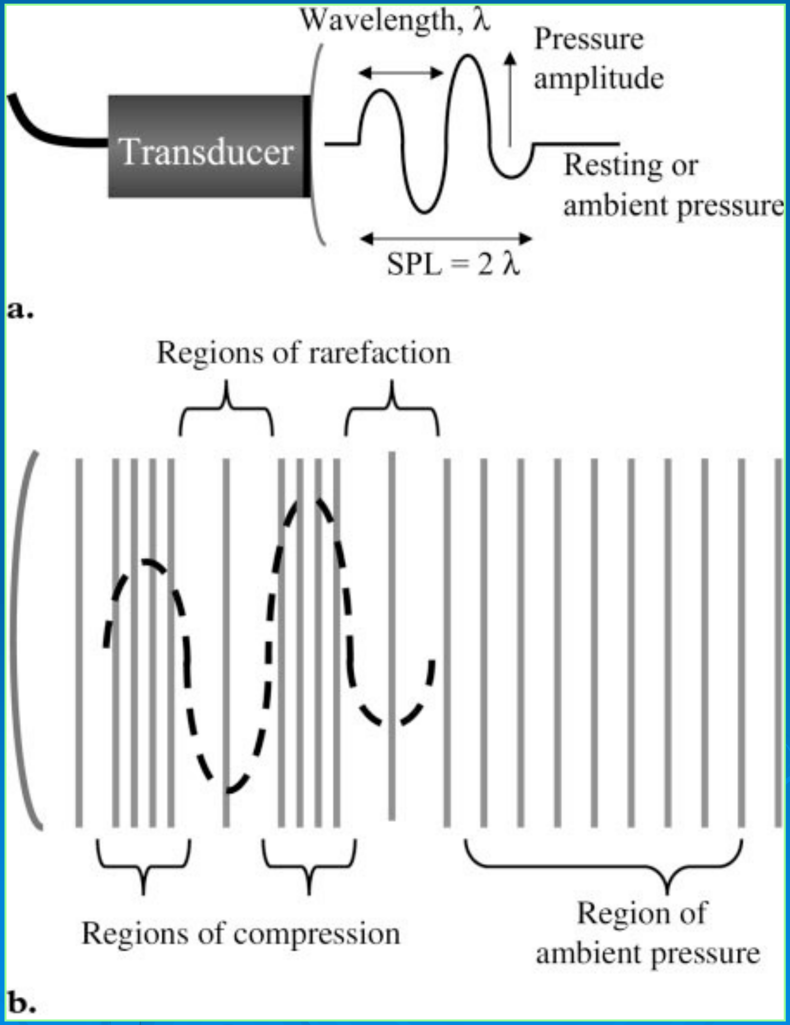


SPL= spatial pulse length

Basic US Physics

ultrasound:

- Propagation into the soft tissues
  - Strongly reflected from the bone
- Undergo reflection and refraction
  - Similar to optics
- $\lambda = c/\nu$ wavelength
  - c speed of sound
- λ << organs to be investigated

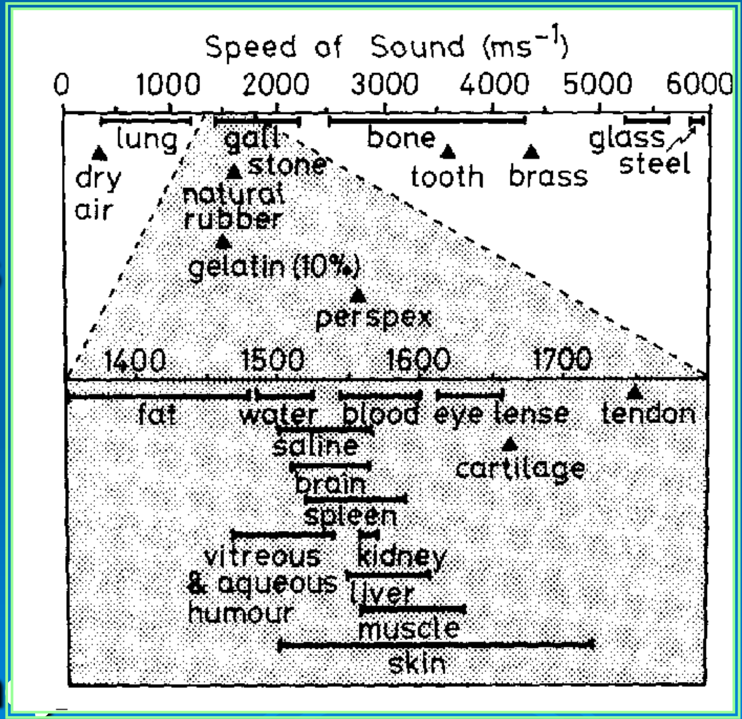


Medical US

- Frequency range of ~2–15 MHz
  - Selected by the radiologist during the exam
- Longitudinal wave speed

$$c=(B/\rho)^{1/2}$$

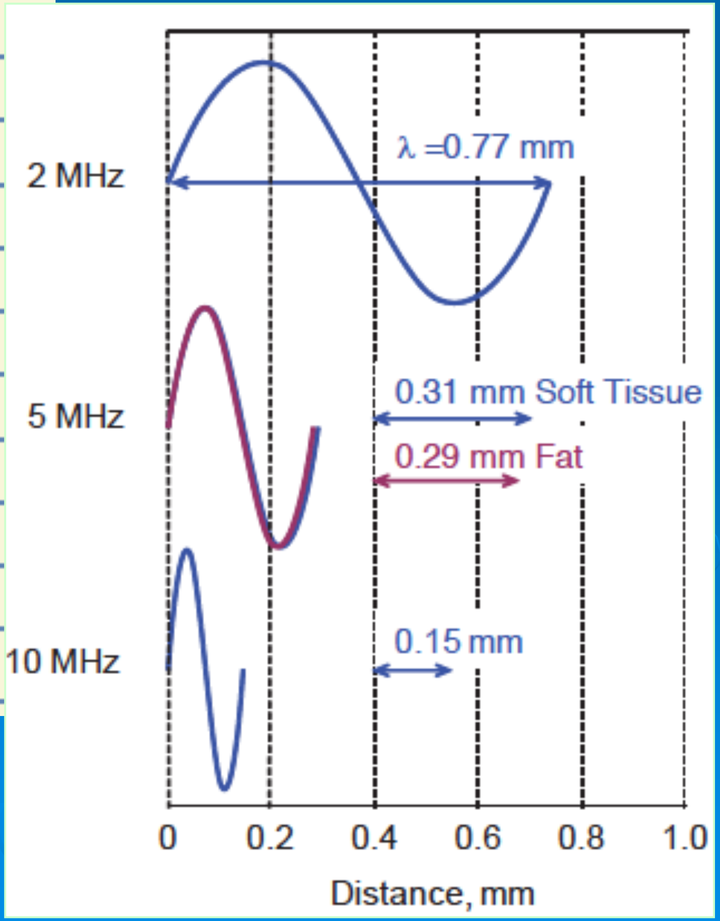
  - B adiabatic bulk elastic modulus
  - ρ mean density
  - $c=1540$ m/s $\pm$ 6% soft tissue
    - $c=1.54$ mm/μs
    - Experimental data
    - Not a strong function of the frequen
  - In bones c ~ 3500 m/s
  - In air c=330 m/s



Medical US

TABLE 14-1 DENSITY AND SPEED OF SOUND IN TISSUES AND MATERIALS FOR MEDICAL ULTRASOUND

MATERIAL	DENSITY (kg/m³)	c (m/s)	c (mm/μs)
Air	1.2	330	0.33
Lung	300	600	0.60
Fat	924	1,450	1.45
Water	1,000	1,480	1.48
"Soft Tissue"	1,050	1,540	1.54
Kidney	1,041	1,565	1.57
Blood	1,058	1,560	1.56
Liver	1,061	1,555	1.55
Muscle	1,068	1,600	1.60
Skull bone	1,912	4,080	4.08
PZT	7,500	4,000	4.00



Basic US properties

- **Wavelength $\lambda = c/\nu$**
  - $\nu = 1$ MHz $\lambda = 1.5\ 10^{-3}$ m = 1.5 mm
  - $\nu = 10$ MHz $\lambda = 1.5\ 10^{-4}$ m = 0.15 mm
  - $\nu = 15$ MHz $\lambda = 1.0\ 10^{-5}$ m = 0.1 mm
- **λ is dependent on the medium**
  - ν is unaffected by changes in sound speed
- **The spatial resolution of the US image depend on the wavelength**
- **The attenuation of the US beam energy depend on the frequency**

Exercise

A US pulse is 10^{-6} s long in time

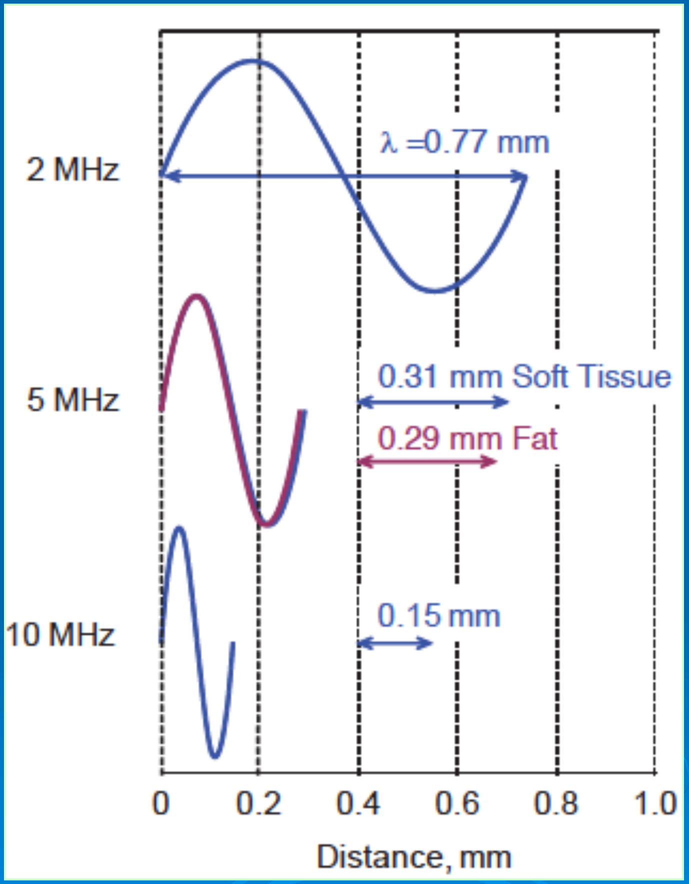
The applied frequency is $\nu = 12$ MHz

How long is the wavelenght?

How long is the pulse in space ?

Results in mm

c=1500 m/s

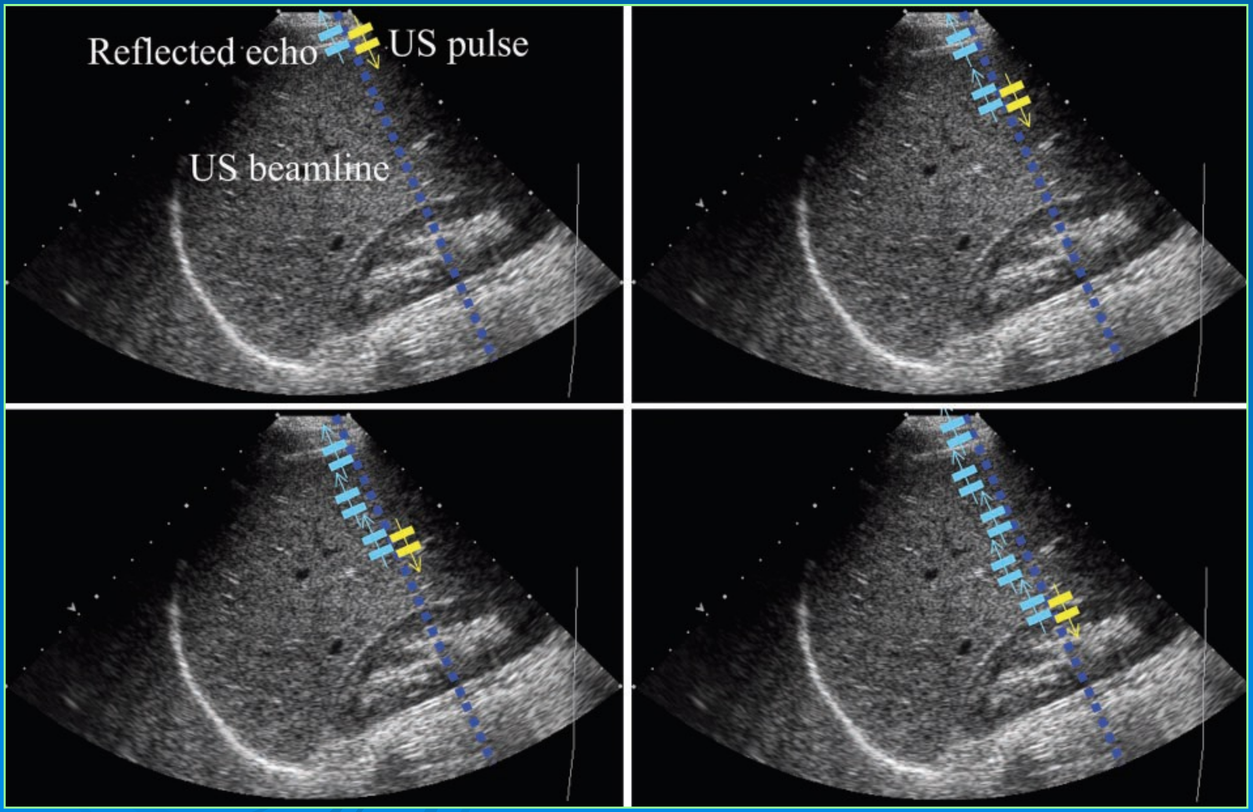


https://www.youtube.com/watch?time_continue=1&v=cI7ULKNhVcw

Basic US properties

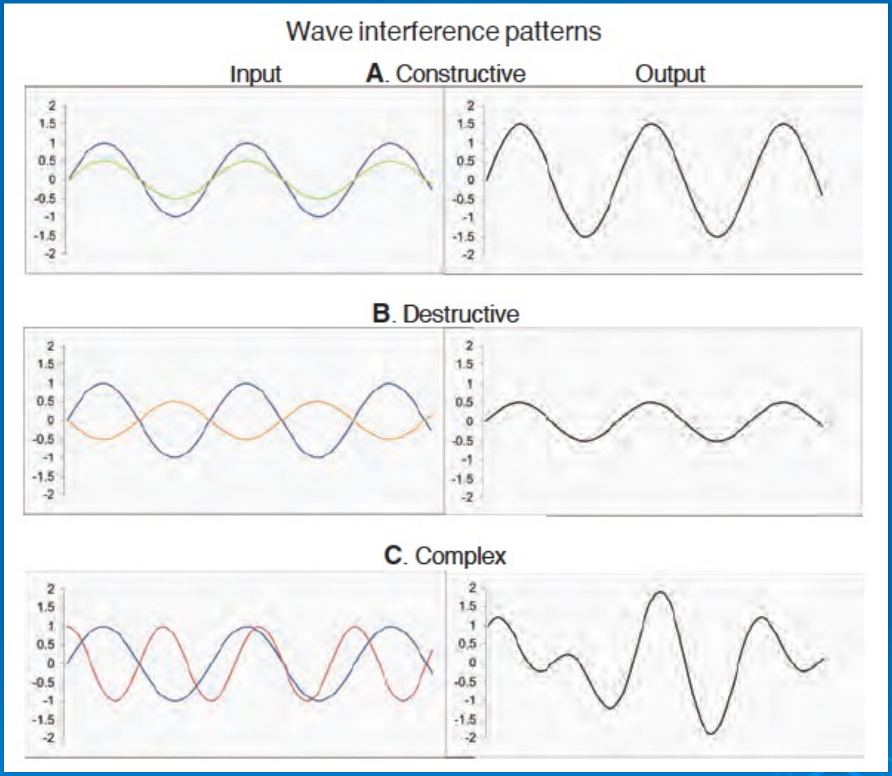
- US imaging is based on the time recording of the echo signal
- If the distance between 2 interfaces is 7.5 cm: the Δt of the 2 echo signals is 10^{-4} s

- c=1500 m/s, d=15 cm, $\Delta t=10^{-4}$ s
- echo resolution is feasible
- US pulses are ~ 10^{-6} s long
- ~2 mm spatial resolution in axial direction



Basic US properties

- Modern US equipment consists of multiple sound transmitters aligned in an array
  - creating sound beams independently
- Interaction of two or more separate US beams in a medium can result in constructive and/or destructive wave interference



Acoustic parameters

Pressure amplitude

- the peak maximum or peak minimum value from the average pressure on the medium in the absence of a sound wave
- SI unit: N/m^2 = Pascal (Pa)
- Diagnostic US beams typically deliver peak pressure levels that exceed 10 times the earth's atmospheric pressure
  - The average atmospheric pressure on earth at sea level ~100000 Pa

Acoustic parameters

Intensity

- the amount of power (energy per unit time) per unit area
- $I \propto P^2$
- Medical diagnostic US intensity levels are described in units of milliwatts/cm^2
  - The absolute intensity level depends upon the method of ultrasound production
- Relative intensity and pressure levels are described as a logarithmic ratio, the decibel (dB)

$$\text{relative intensity (dB)} = 10 \log \left(\frac{I_2}{I_1} \right)$$

Acoustic parameters

In diagnostic US

- the ratio of the intensity of the incident pulse to that of the returning echo spans a range of one million times
  - or more
  - The logarithm function compresses the large and expands the small ratios into a more manageable number range

$$\text{relative intensity (dB)} = 10 \log \left(\frac{I_2}{I_1} \right)$$

TABLE 14-2 INTENSITY RATIO AND CORRESPONDING DECIBEL VALUES

INTENSITY RATIO		DECIBELS (*dB*)
I_2/I_1	LOG (I_2/I_1)	
1	0	0
2	0.3	3
10	1	10
100	2	20
10,000	4	40
1,000,000	6	60
0.5	–0.3	–3
0.01	–2	–20
0.0001	–4	–40
0.000001	–6	–60

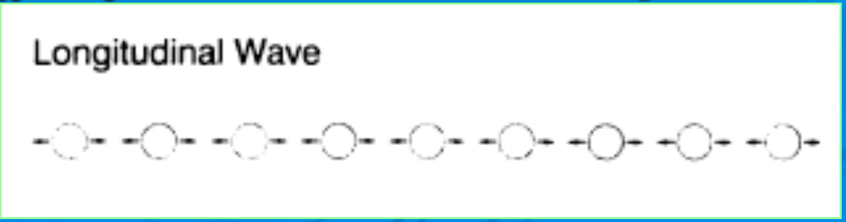
Acoustic field parameters

- sound intensity
  - Ideal conditions

$$I = \frac{c}{2}(\rho \upsilon^2 + \frac{p^2}{\rho c^2})$$

$$= \frac{1}{2}(Z\upsilon^2 + \frac{p^2}{Z})$$

  - W/m^2
  - c sound speed in the medium
  - υ molecules velocity
  - ρ density
  - p oscillating incremental pressure
    - Energy density
  - $Z = \rho c = (\rho B)^{1/2}$ Acoustic impedance



Acoustic parameters: a numerical example

if I = 2 10^5 W/m^2 and p=0:

- $\upsilon_{max}=(2\,I\,/\,c\rho)^{1/2} = 0.5$ m/s
  - $\rho = 10^3$ kg/m^3, c = 1500 m/s

Oscillation $\nu=10^6$ Hz:

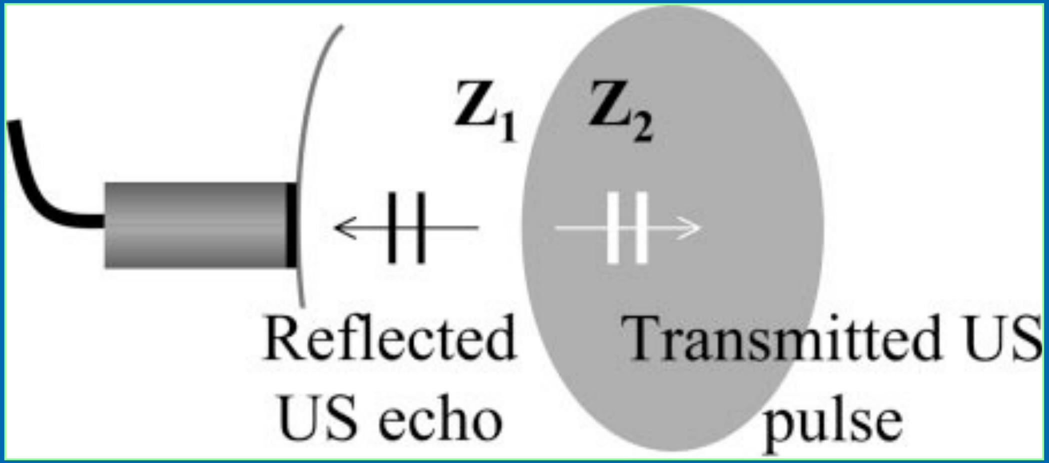
- $\Delta x_{max} = 8\ 10^{-8}$ m
  - Amplitude of the oscillation
- $P_{max}= 8$ atm
  - Local pressure
- $a_{max}= 3\ 10^5$ g

In medical imaging short pulse (1-2 μs)

Average I ~ 5mW, I peak 5 W

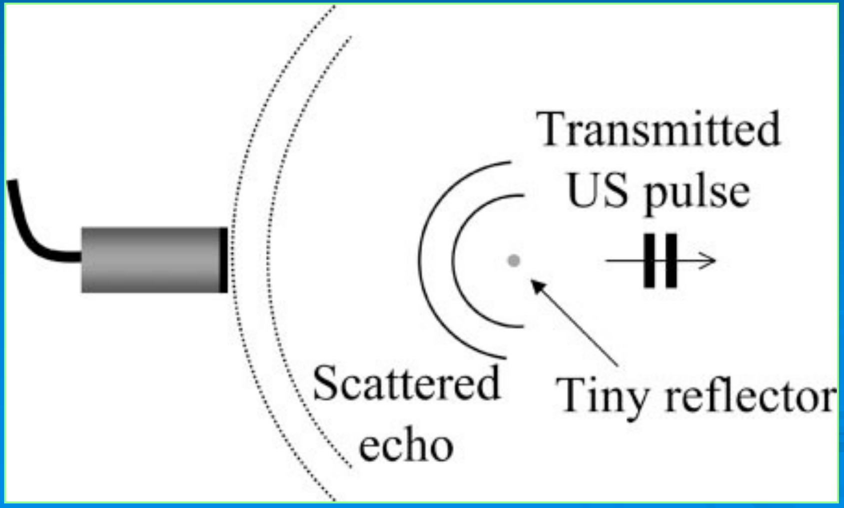
Interactions of US with Matter

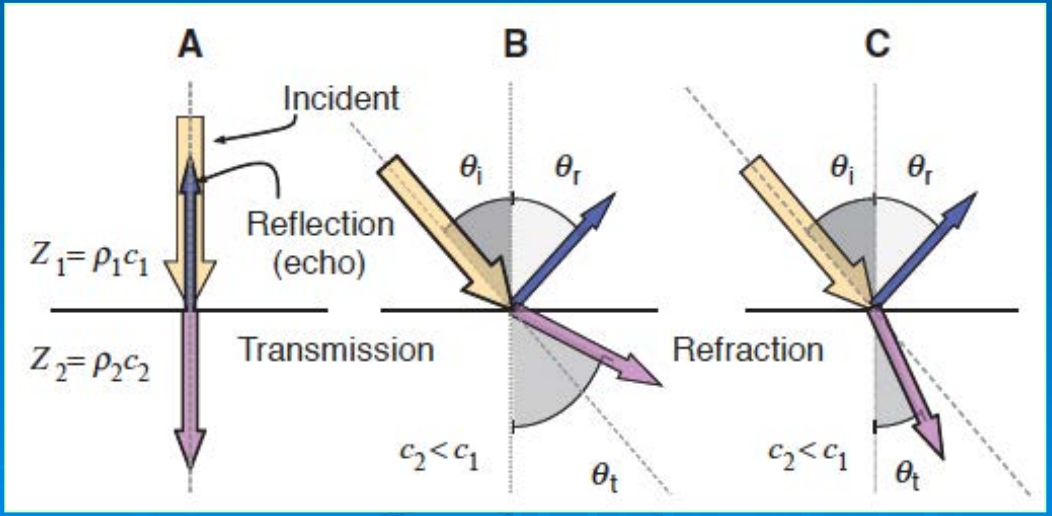
US interactions are determined by the acoustic properties of matter



Interactions include

- reflection
- refraction
- scattering
- absorption





Acoustic impedance

✓ $Z = \rho c = (B\rho)^{1/2}$

- SI unit: $kg/(m^2 s)$
- 1 Rayls = 1 $kg/(m^2 s)$

TABLE 14-3 ACOUSTIC IMPEDANCE, $Z = \rho c$, FOR AIR, WATER, AND SELECTED TISSUES

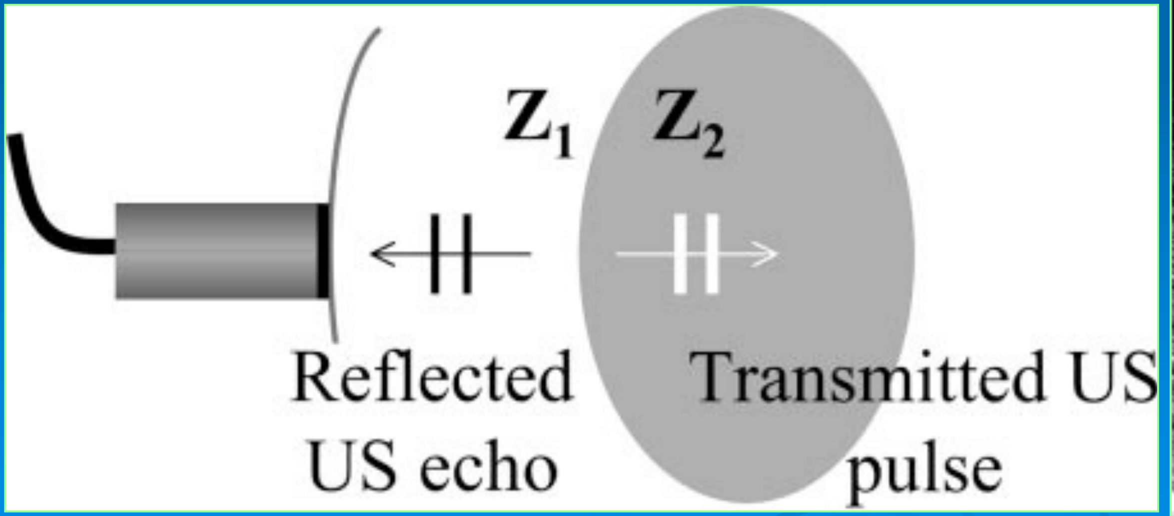
TISSUE	Z (RAYLS)
Air	0.0004×10^6
Lung	0.18×10^6
Fat	1.34×10^6
Water	1.48×10^6
Kidney	1.63×10^6
Blood	1.65×10^6
Liver	1.65×10^6
Muscle	1.71×10^6
Skull bone	7.8×10^6

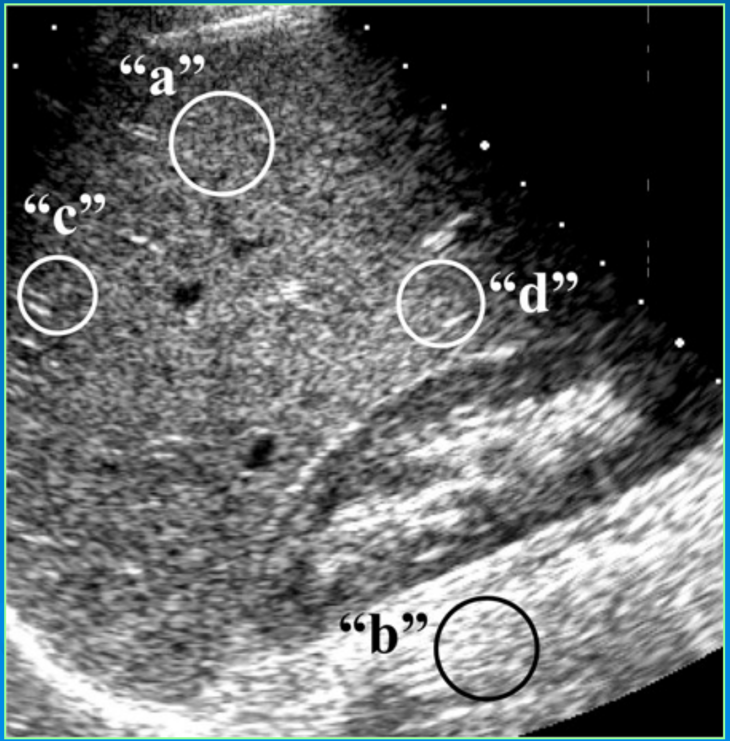
Acoustic impedance

- Acoustic impedance can be likened to the stiffness and flexibility of a compressible medium
- When tissues with different compressibility are connected together, the energy transfer from one to another depends on stiffness
- Minor differences in stiffness or compressibility allow the continued propagation of energy
  - little reflection at the interface

Acoustic impedance

- Acoustic impedance gives rise to differences in transmission and reflection of ultrasound energy
- producing an image using pulse-echo techniques



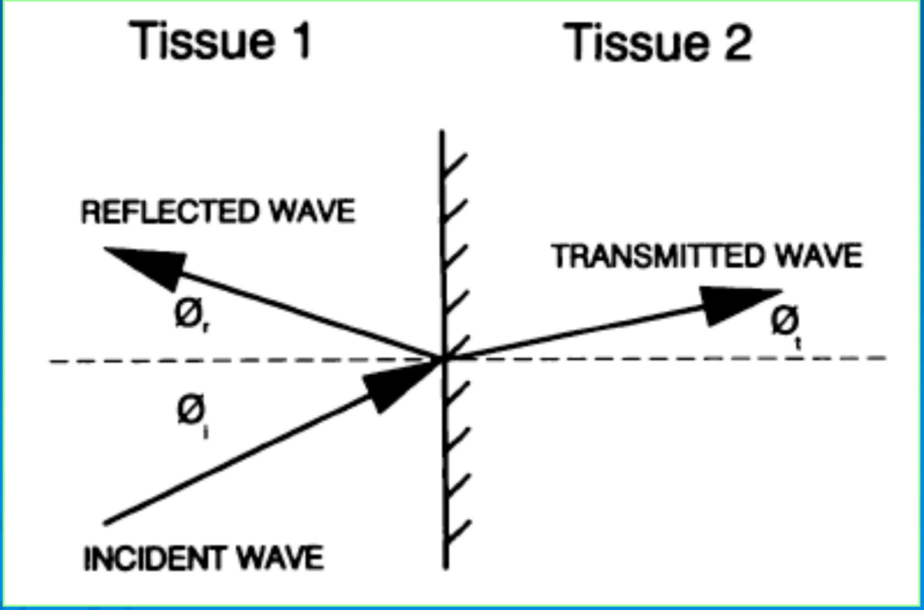


Refraction and Reflection

- Similar to optics
  - Valid if $d \gg \lambda$
- Snell's law: $\sin\theta_i / \sin\theta_t = c_1/c_2$
- R_I intensity reflection coefficient
  - $R_I = I_{refl}/I_{inc}$

$$R_I = \left(\frac{Z_1 \cos\theta_i - Z_2 \cos\theta_t}{Z_1 \cos\theta_i + Z_2 \cos\theta} \right)^2$$

Ziskin M.C.
Radiographic 1993; 13: 705-709

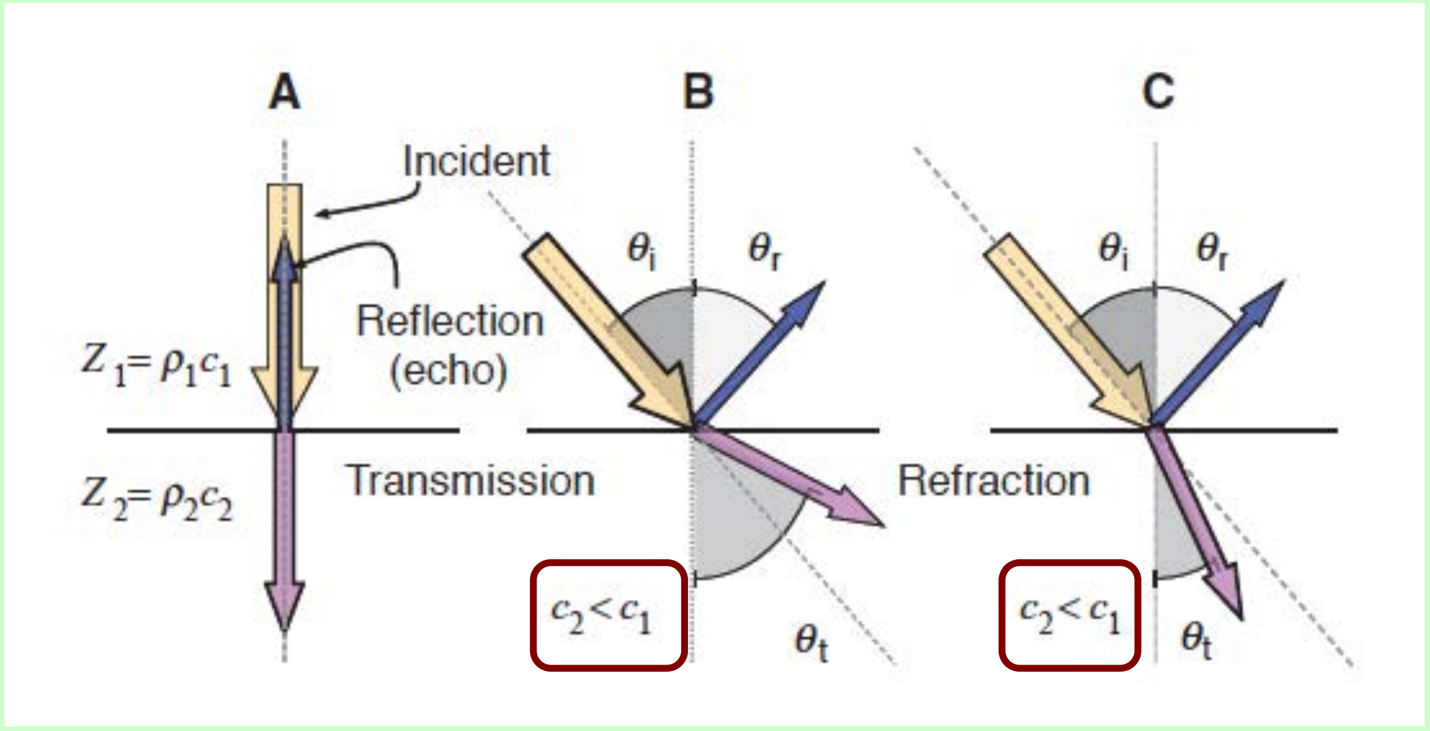


Reflection

- For normal (or 90°) incidence
- $R_I = I_{refl}/I_{inc}$

$$R_I = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2}\right)^2$$

  - $R_I = 0$?
  - R_I max?



Reflection

TABLE 14-4 PRESSURE AND REFLECTION COEFFICIENTS FOR VARIOUS INTERFACES

TISSUE INTERFACE	PRESSURE REFLECTION	INTENSITY REFLECTION
Liver-Kidney	–0.006	0.00003
Liver-Fat	–0.10	0.011
Fat-Muscle	0.12	0.015
Muscle-Bone	0.64	0.41
Muscle-Lung	–0.81	0.65
Muscle-Air	–0.99	0.99

$$R_I = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2} \right)^2$$

- The relatively strong echoes are generated by a muscle-fat interface (R_I=1.5%)
- A liver-kidney interface generates weaker echoes (R_I=0.03%)

Refraction

- Refraction describes the change in direction of the transmitted US energy
  - with nonperpendicular incidence
- Snell's law

$$\frac{\sin\theta_i}{\sin\theta_t} = \frac{c_1}{c_2}$$

  - If $c_1=c_2$ no refraction
    - c speed of sound in the tissue
- $T_I = 1 - R_I$
  - the fraction of the incident intensity transmitted

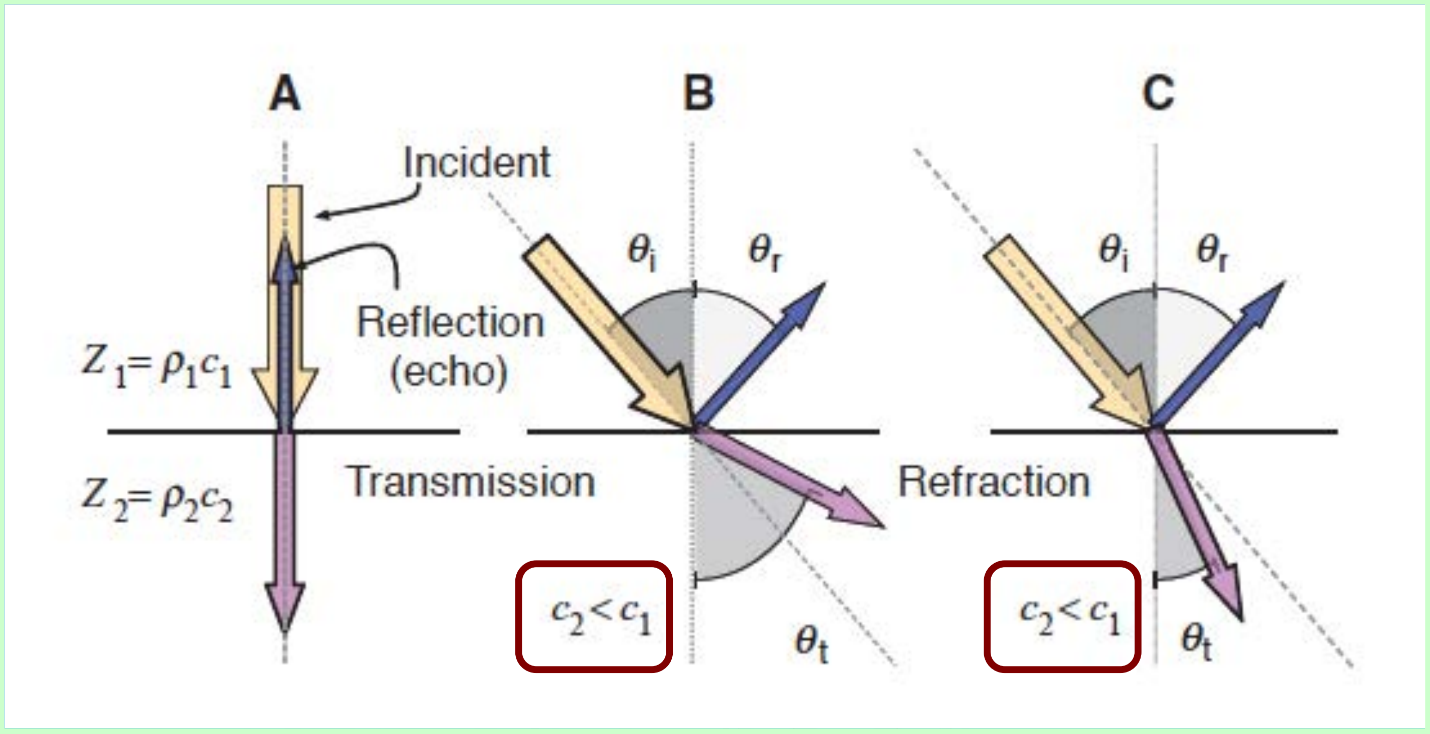
Refraction

- Snell's law

$$\frac{\sin\theta_i}{\sin\theta_t} = \frac{c_1}{c_2}$$

  - If $c_1=c_2$ no refraction

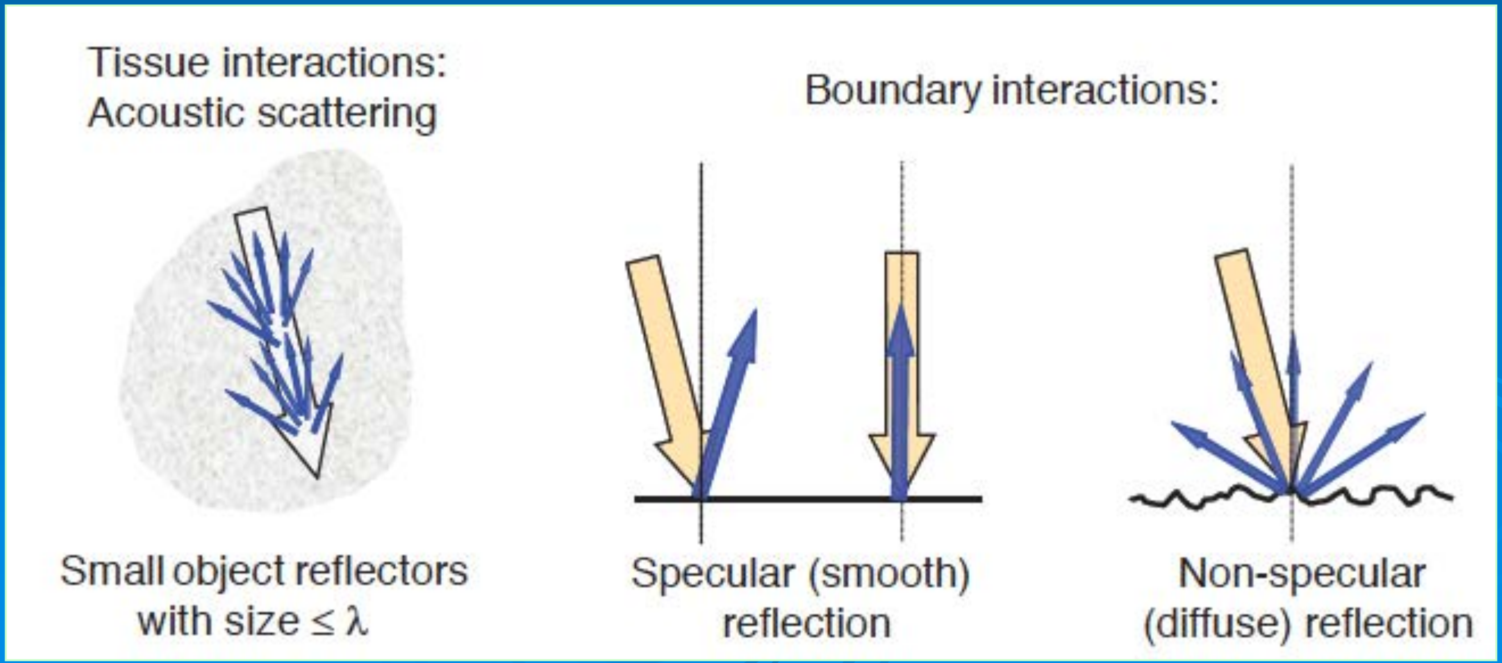
- $T_I = 1 - R_I$



Scattering

Acoustic scattering arises from objects and interfaces that are :

- **about the size of the wavelength**
- **a rough or nonspecular reflector surface**



US attenuation

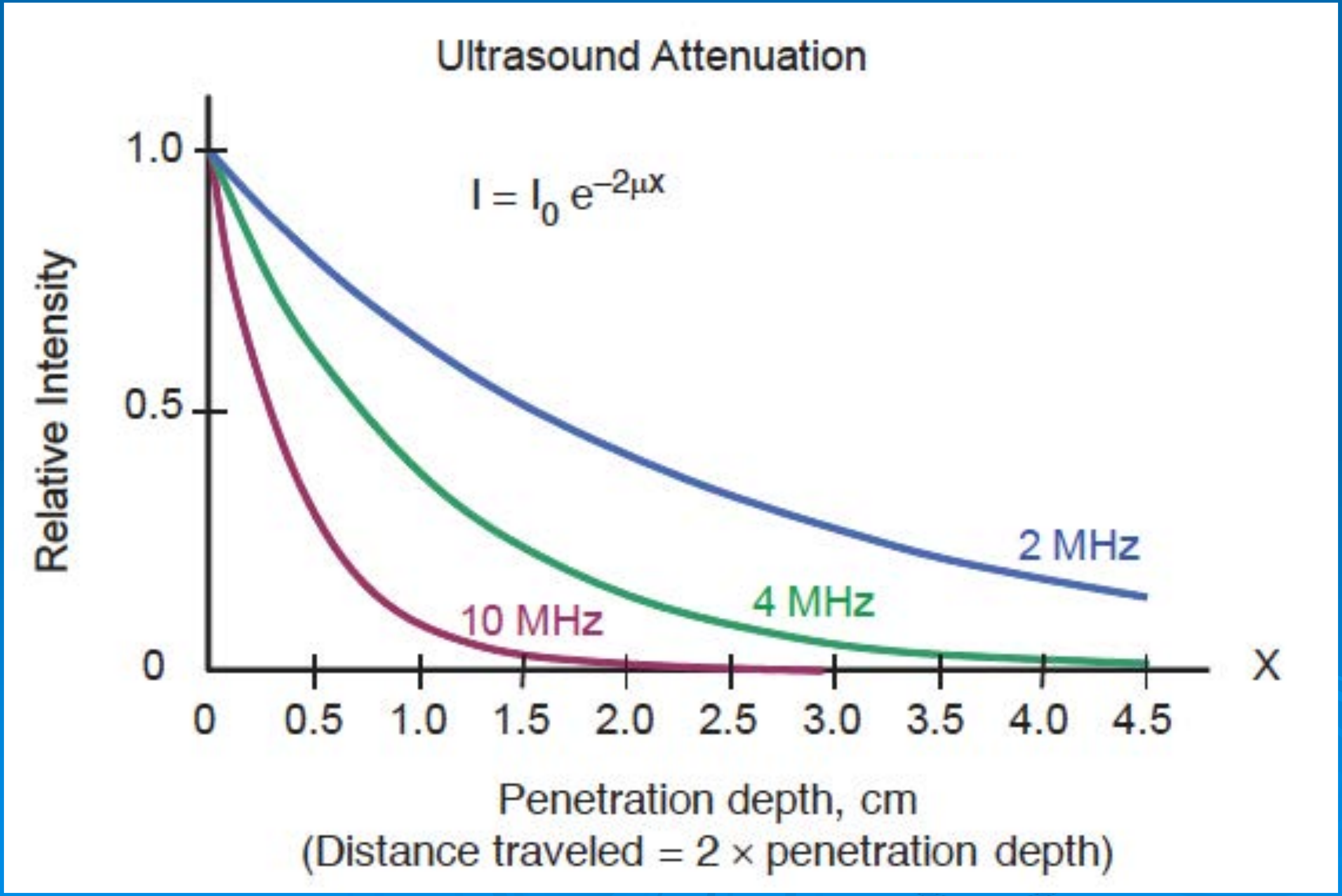
$$I(x)=I_0\, e^{-\mu x}$$

The loss of energy with distance traveled is due to:

- Scattering
- Tissue absorption of the incident beam
  - Absorbed acoustic energy is converted to heat

Attenuation and frequency

$\mu = 0.5$ (dB/cm)/MHz



Half intensity depth

In soft tissue

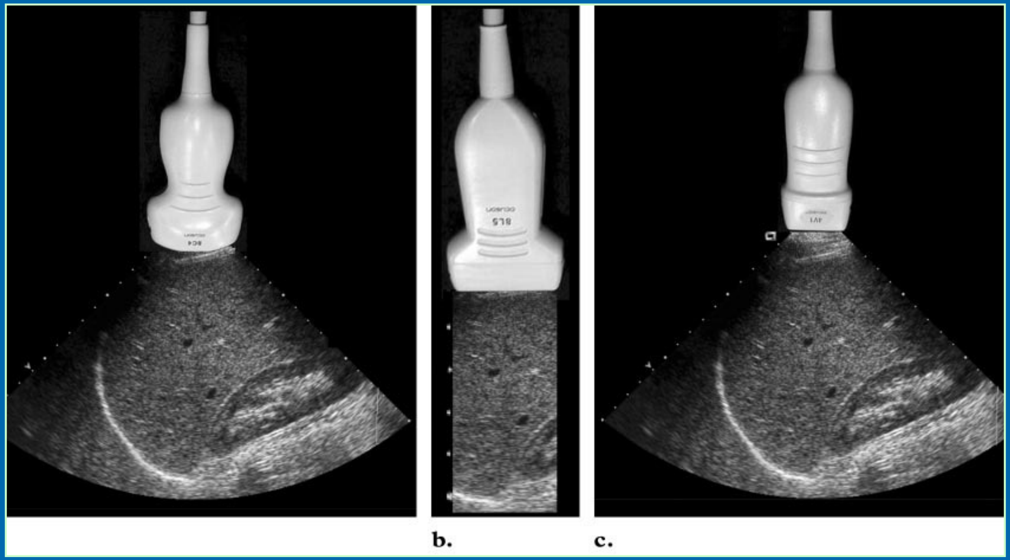
ν (MHz)	λ(mm)	$L_{1/2}$,(cm)
1	1.54	6.0
5	0.31	1.2
10	0.154	0.6

Ratio between $L_{1/2}$ and λ constrant ~ 40

Table 2
Attenuation Coefficients and Half-Value Layers of Various Materials

Material	Attenuation (dB/cm/MHz)	Half-Value Layer (cm)
Water	0.0022	1,360.0
Blood	0.15	20.0
Soft tissue	0.75	4.0
Air	7.50	0.4
Bone	15.00	0.2

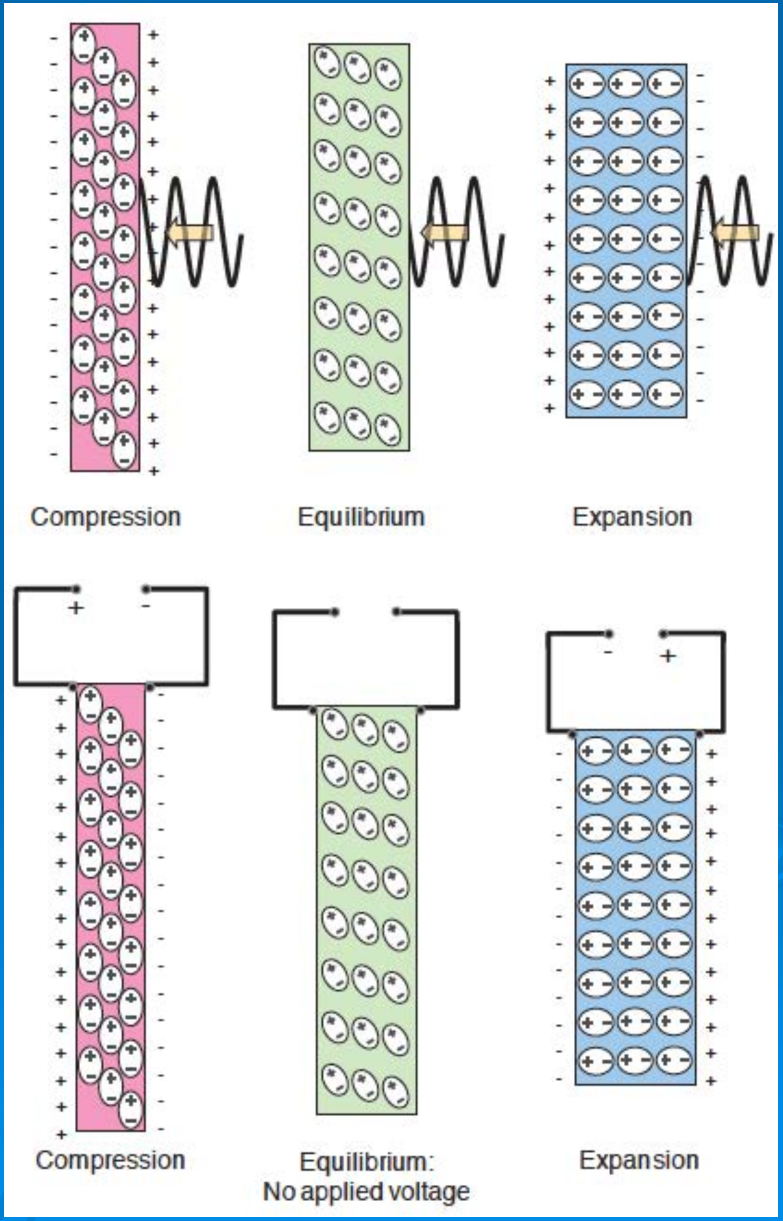
US transducers



- US pulse are produced and the echoes are detected with a transducer
- The ceramic element converts:
  - electrical energy into mechanical energy to produce US
  - mechanical energy into electrical energy for echoes detection

Piezoelectric Materials

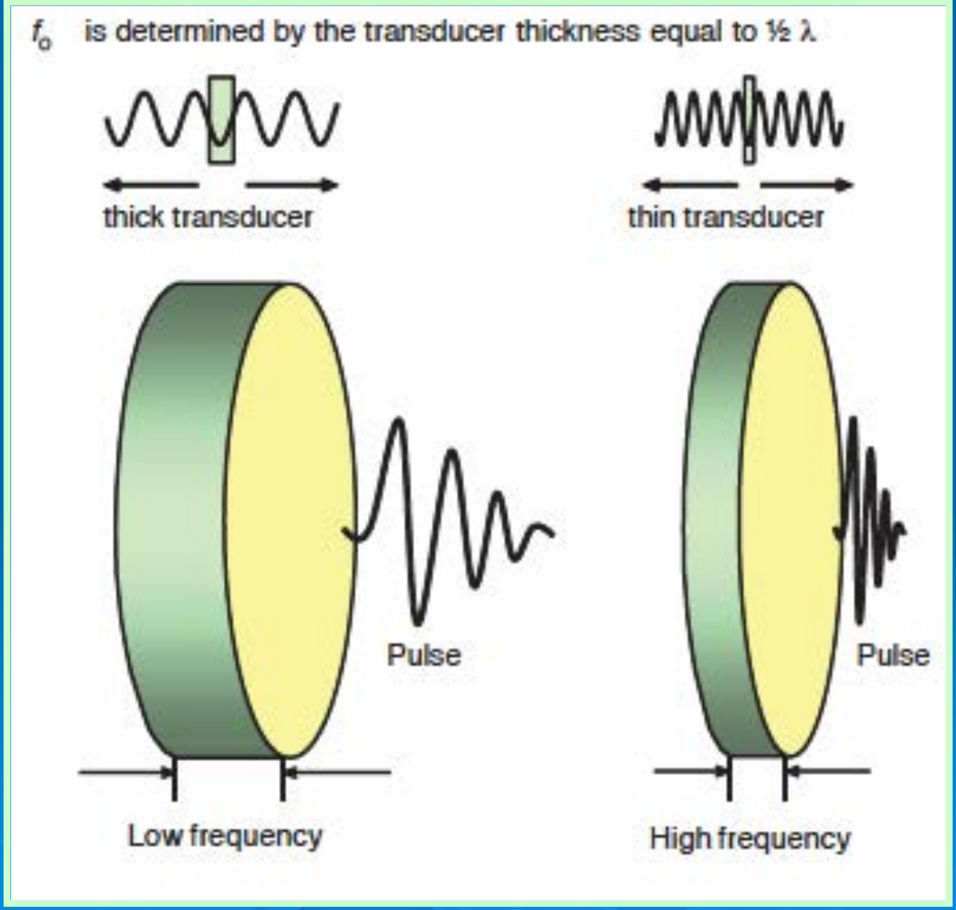
- Electrical energy is converted into mechanical energy by deformation of the crystal structure
  - well-defined molecular arrangement of electrical dipoles
- Mechanical pressure applied to its surface creates electrical energy
  - voltage



Resonance Transducers

Resonance transducers operate in a "resonance" mode

- a voltage (~ 150 V) of short duration (~1 μs) is applied
- the piezoelectric material to initially contract and then subsequently vibrate at a natural resonance frequency

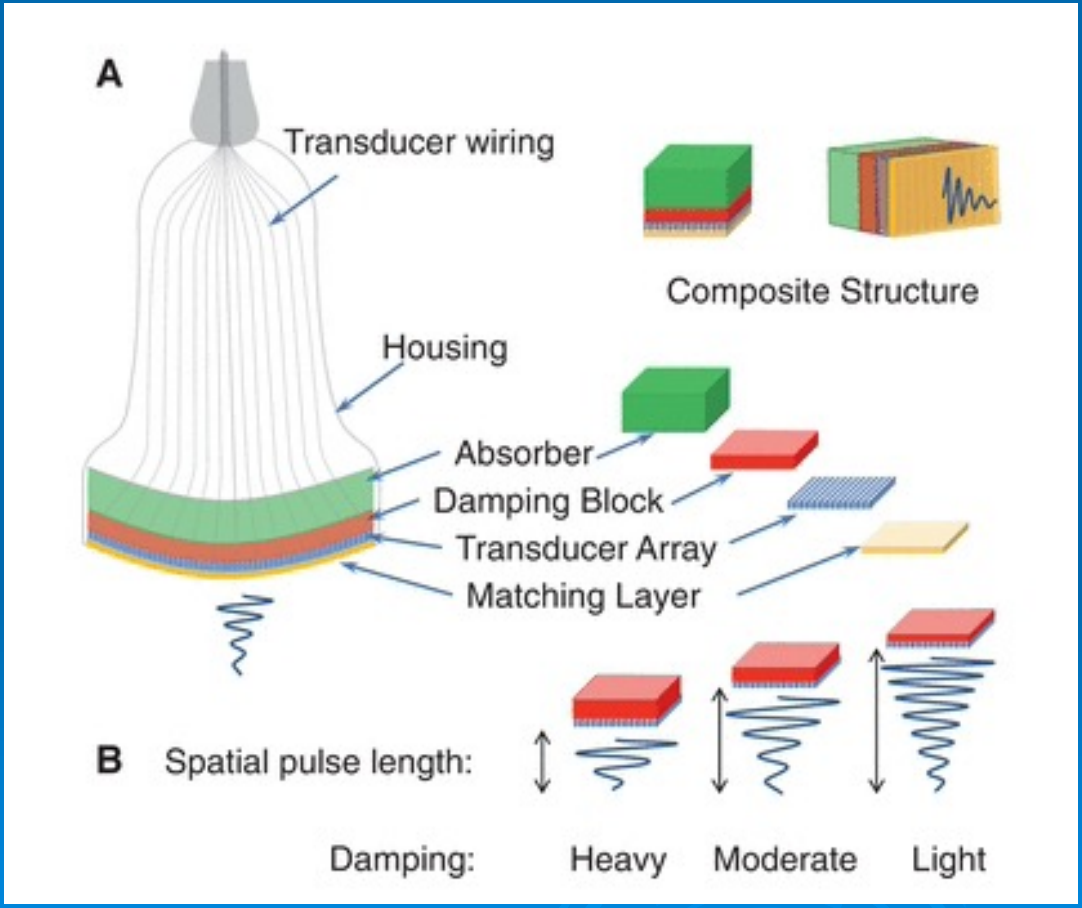


The Essential Physics of Medical Imaging
Bushberg et al, 2012 Lippincott Williams & Wilkins

US transducers

Major components:

- matching layer
- the piezoelectric material
  - up to 100 individual elements
- damping block
- acoustic absorber
- tuning coil
- insulating cover
- sensor electrodes



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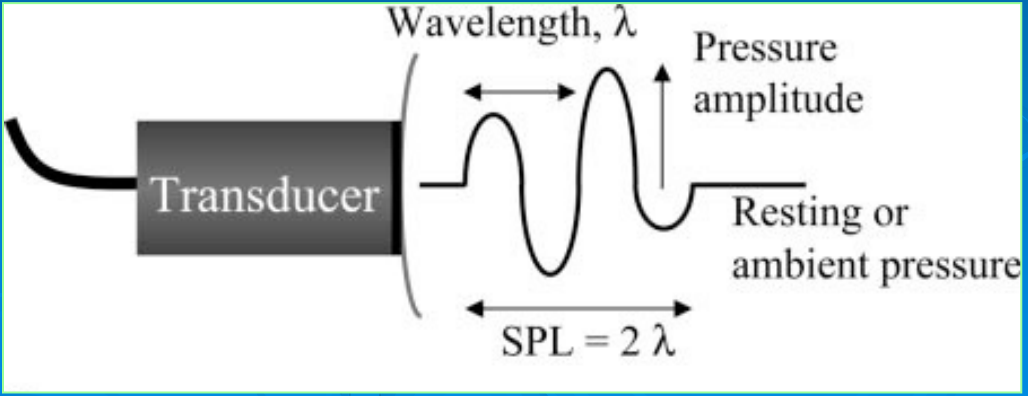
Damping Block

- The damping block dampens the transducer vibration to create an US pulse with a short spatial pulse length (SPL)
  - necessary to preserve detail along the beam axis
    - axial resolution



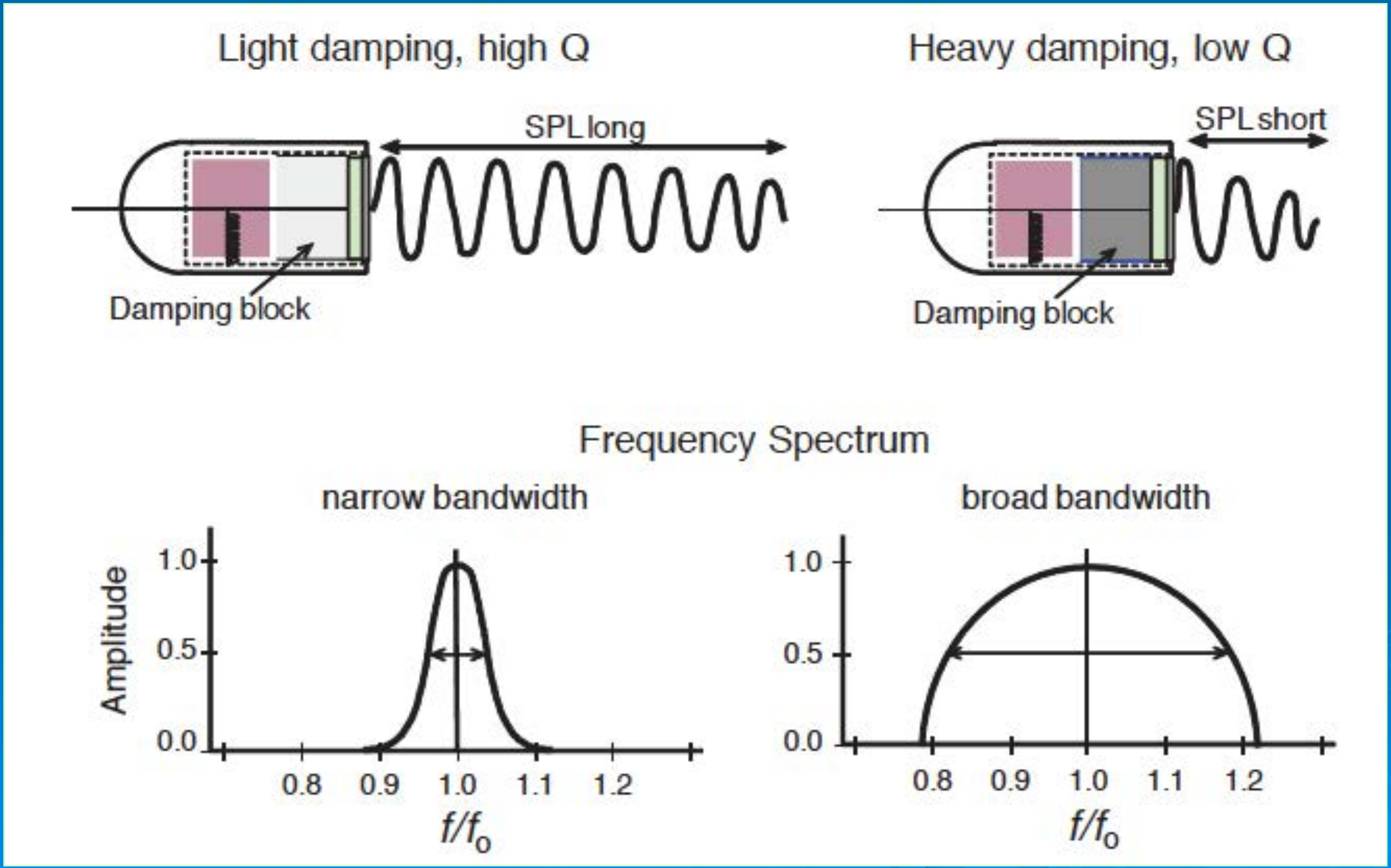
Damping Block

- Dampening the vibration introduces a broadband frequency spectrum

$$Q = \frac{f_0}{\text{bandwidth}}$$

- A "high Q" transducer has a narrow bandwidth and a corresponding long SPL
  - very little damping
- A "low Q" transducer has a wide bandwidth and short SPL

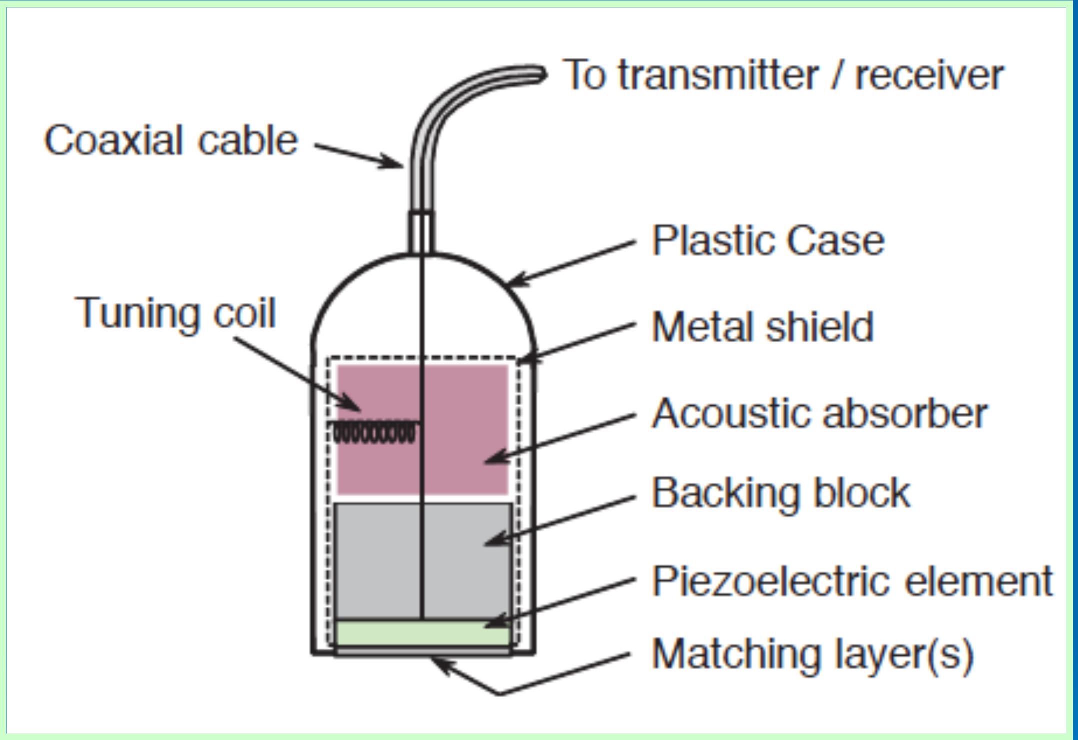
Damping and Q factor



low Q does not imply poor quality in the signal

Matching layer

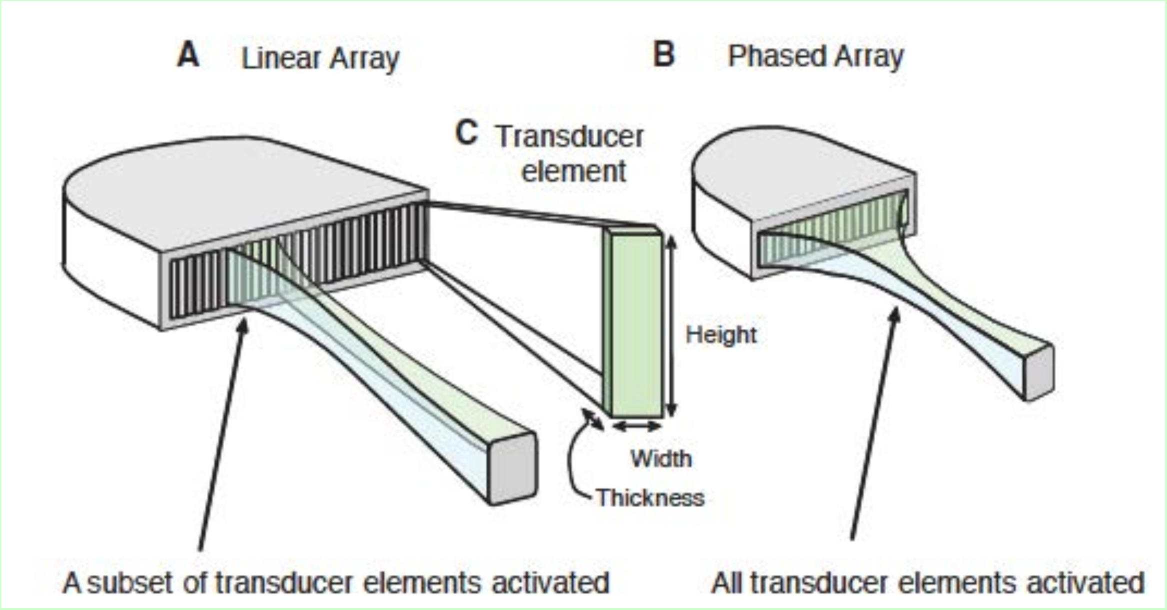
- The matching layer minimizes the acoustic impedance differences between the transducer and the patient
  - layers of materials with acoustic impedances that are intermediate to soft tissue and the transducer material



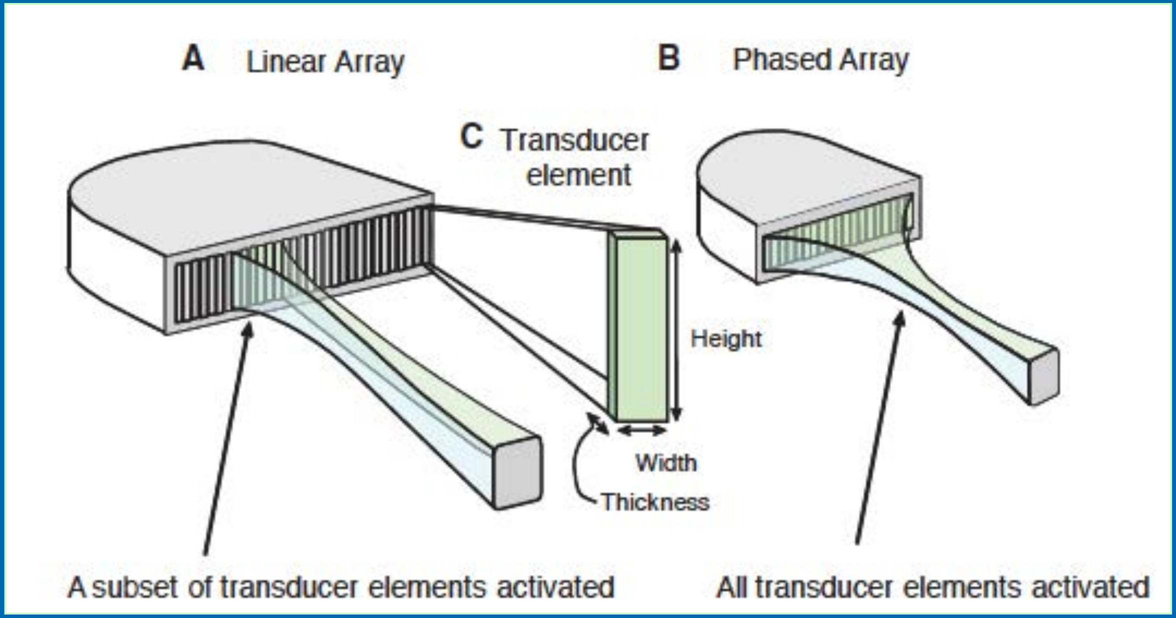
In addition acoustic coupling gel is used between the transducer and the patient to eliminate air pockets

Transducer Arrays

- Transducer arrays have with many individual rectangular piezoelectric elements arranged in linear or curvilinear arrays
  - 128 to 512 individual rectangular elements
  - Each element width typically less than 1/2 λ and height of several mm



Transducer Arrays



A phased-array transducer is usually comprised of 64 to 128 individual elements in a smaller package than a linear array transducer

- activation/receive modes
  - linear
    - Sequential
    - Firing a subset of the total number of transducer elements as a group (about 20 elements)
  - phased
    - A beam is produced from all of the transducer elements fired with fractional time delays in order to steer and focus the beam

Ultrasound Beam Properties

The US beam propagates as a longitudinal wave and exhibits 2 distinct beam patterns

- a slightly converging beam out to a distance determined by the geometry and frequency of the transducer
  - the near field
- a diverging beam beyond that point
  - the far field

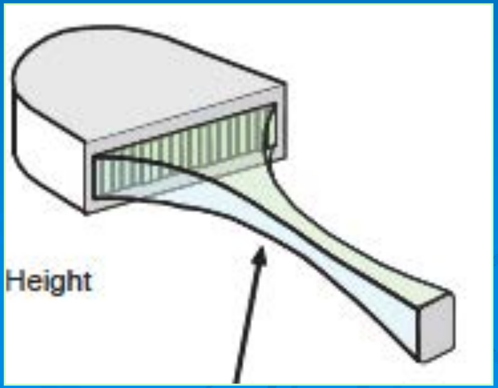
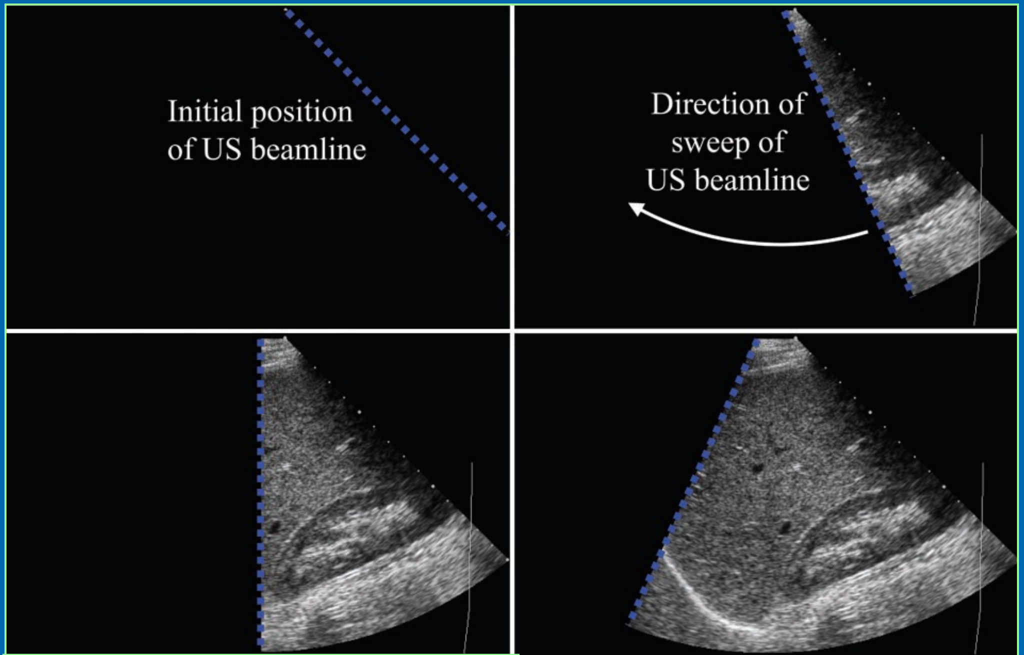
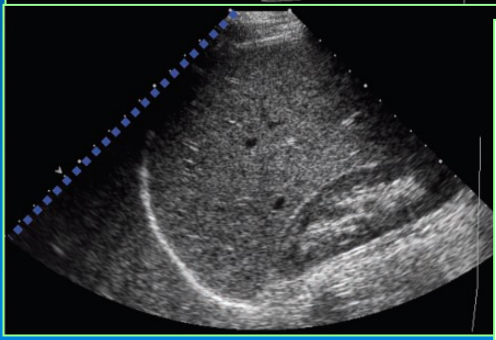
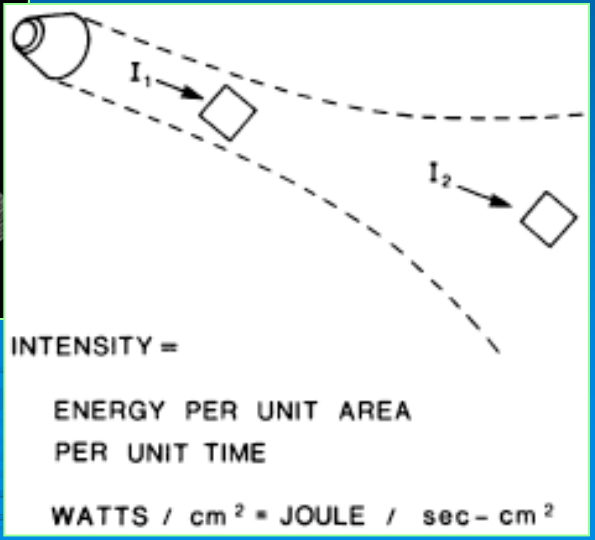


Image formation

- ✓ Collection of the echoes over time and recording of the echo amplitudes
- ✓ Repetition of the process hundreds of times with a small incremental change in the direction of the pulse interrogates a volume
    - from which a gray-scale tomographic image can be synthesized



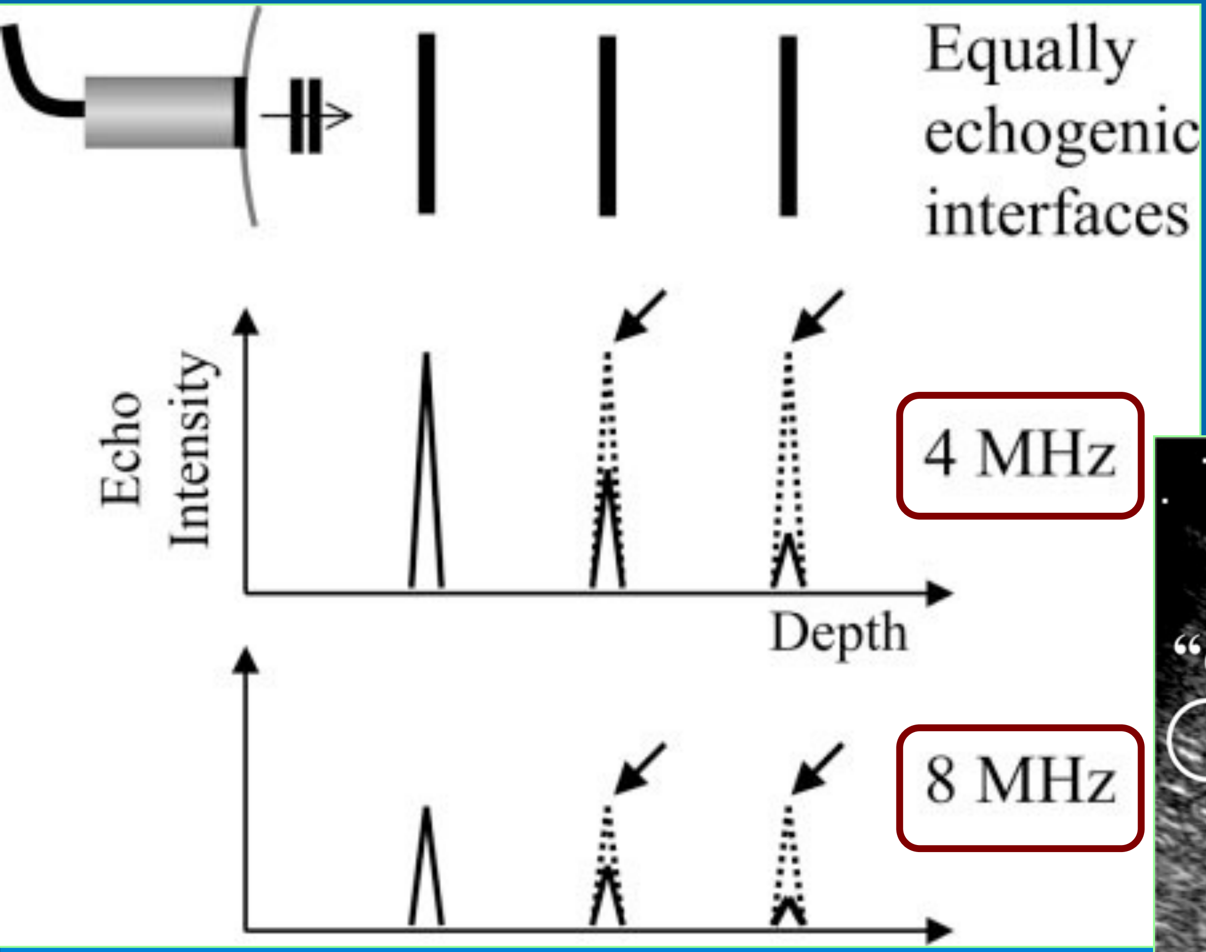


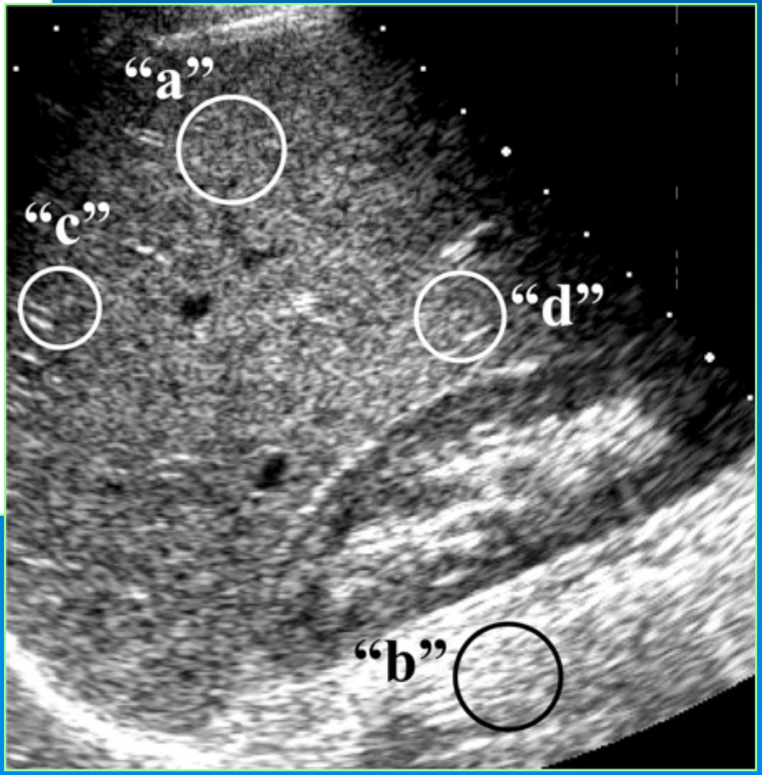


https://www.youtube.com/watch?time_continue=9&v=jxGUId2lBaA

Image formation

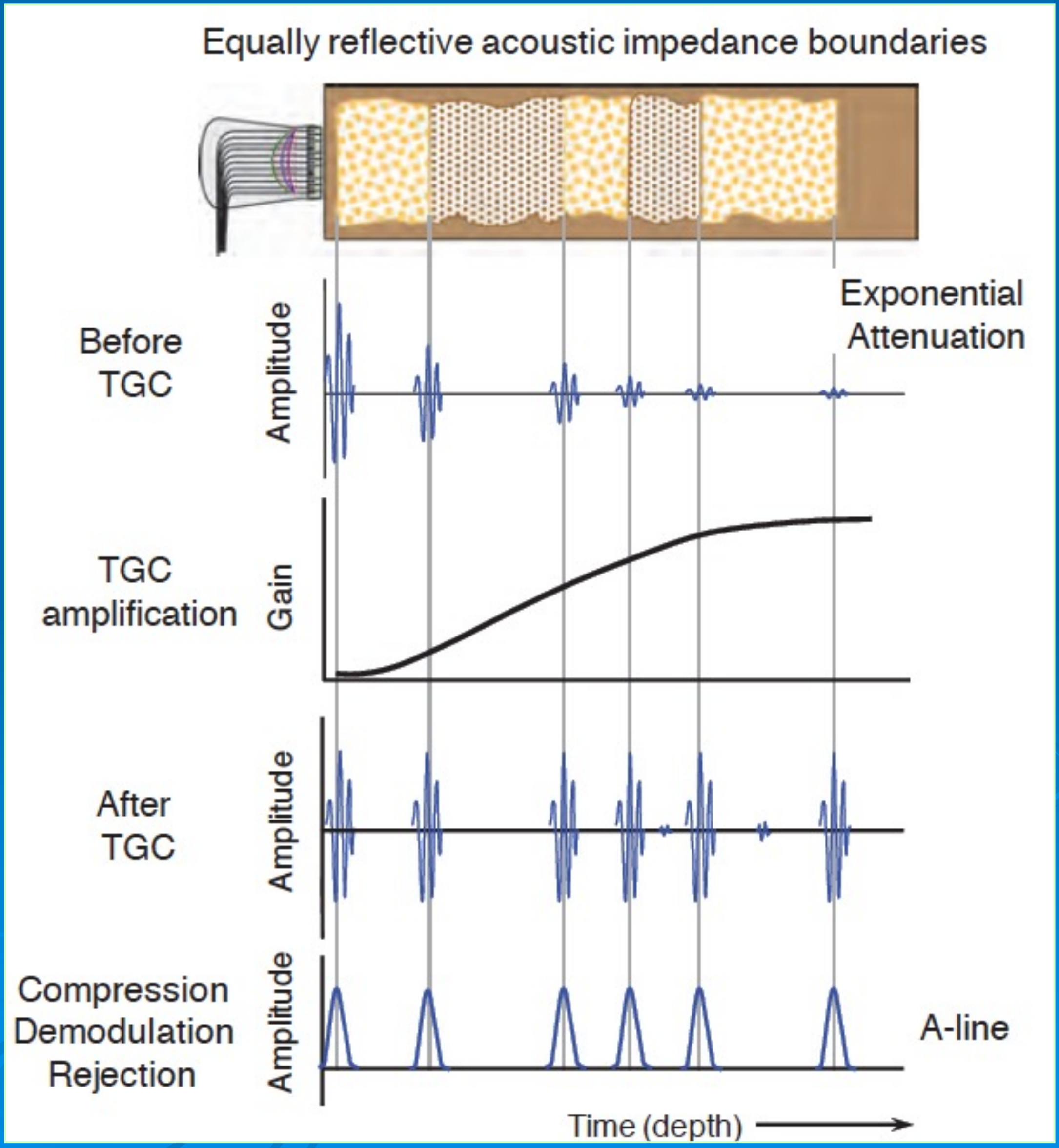
Diagram shows the effects of attenuation on echo intensity from three equally reflective structures



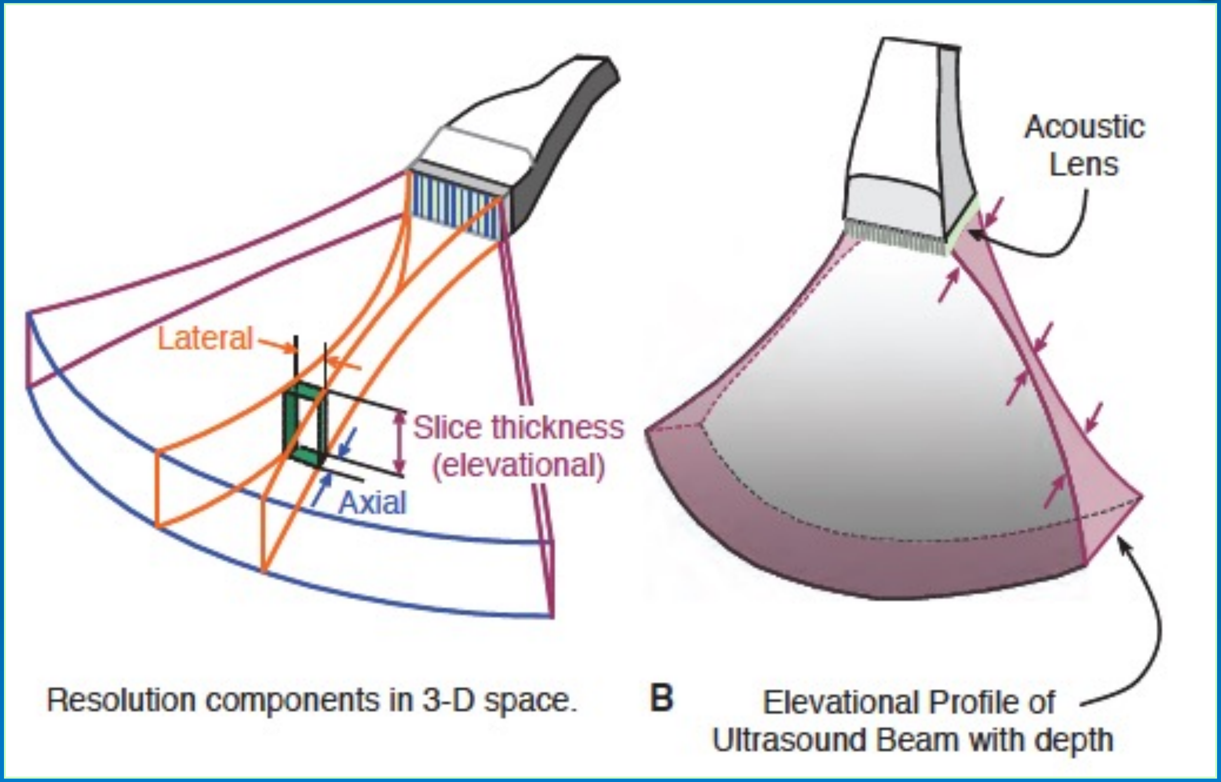


Time gain correction

TGC amplifies the acquired signals with respect to time after the initial pulse by operator adjustments



US image plane
the volume of the acoustic pulse

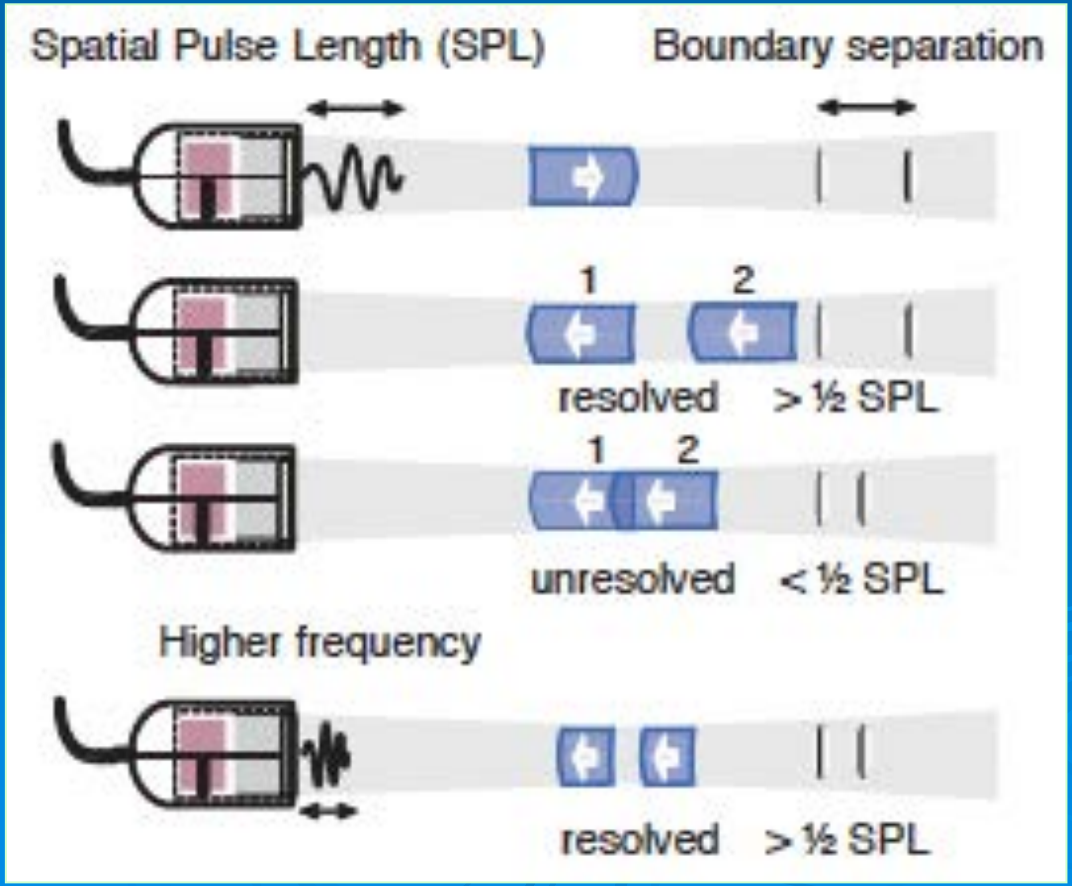


- The axial, lateral, and elevational dimensions determine the minimal volume element
- Each dimension has an effect on the resolvability of objects in the image

The Essential Physics of Medical Imaging
Bushberg et al, 2012 Lippincott Williams & Wilkins

US axial resolution

The spatial pulse length (SPL) in the axial direction is equal to the number of cycles in the pulse multiplied by the US wavelength

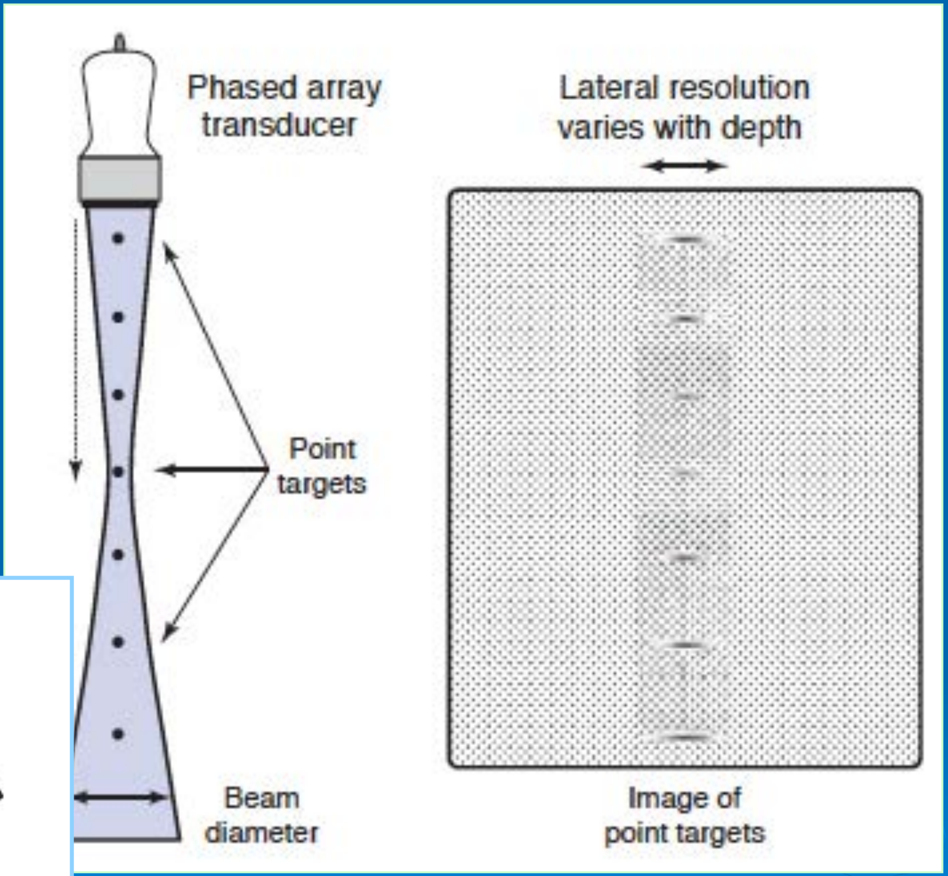


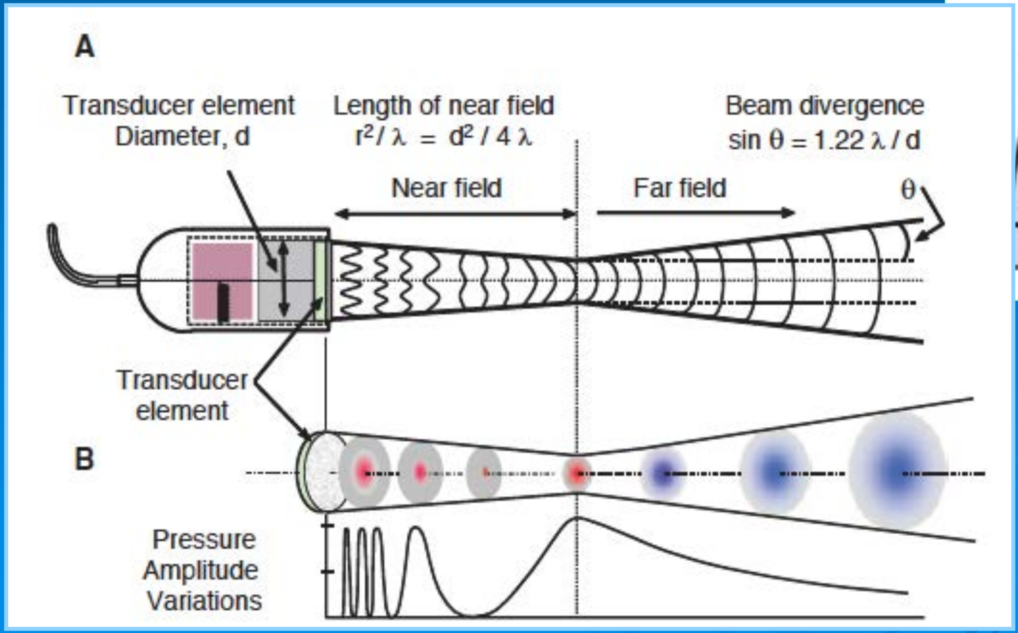
US axial resolution

- Short SPL produce US images with the greatest sharpness in the axial direction (axial resolution)
  - Short pulses are produced by electrically exciting the piezoelectric elements for about 1 μs or less
- SPL remains constant as the pulse propagates to greater depths
- Higher frequencies increase the spatial resolution
  - SPL decreases

US lateral resolution

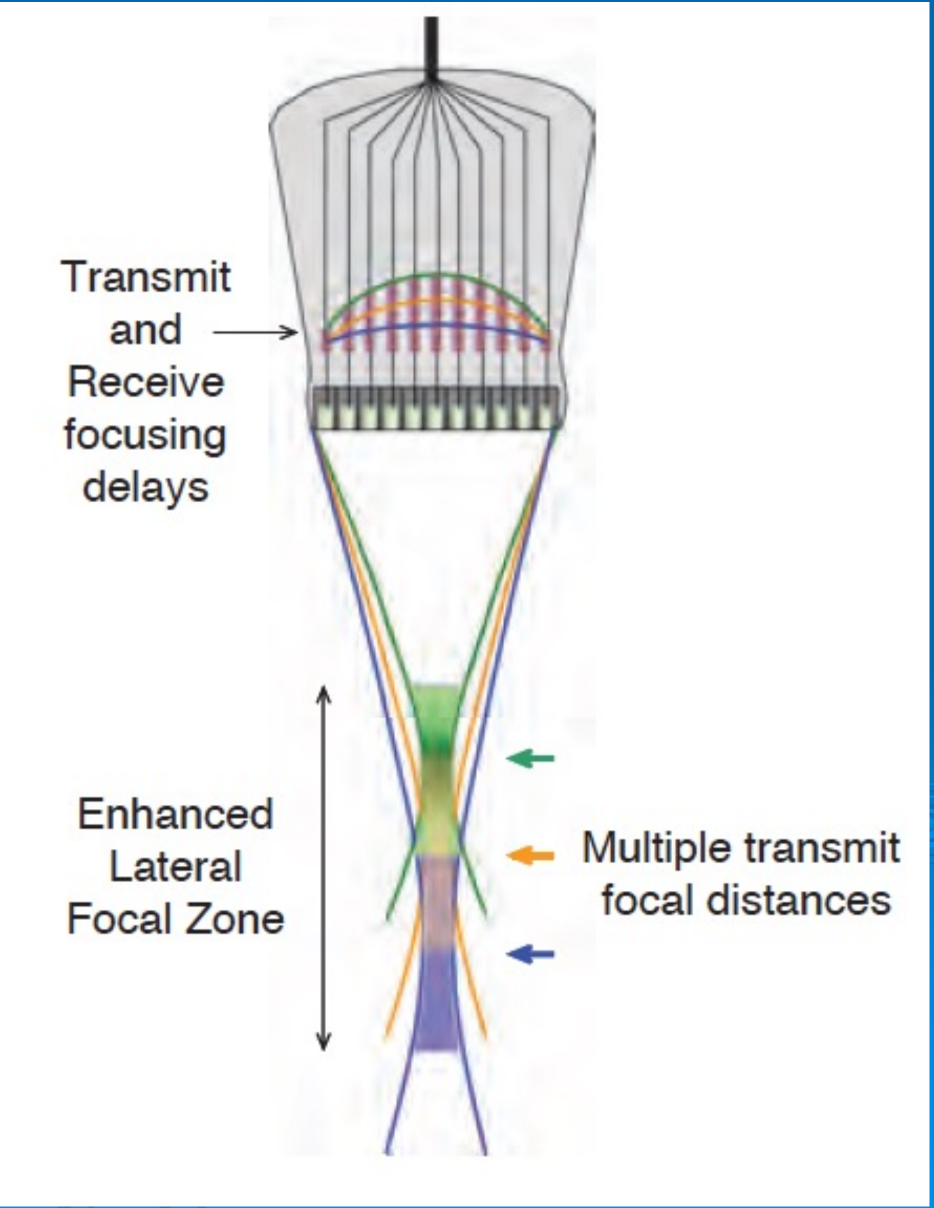
- US pulses that are narrow in the lateral direction produce images with greatest sharpness in that direction
  - lateral resolution





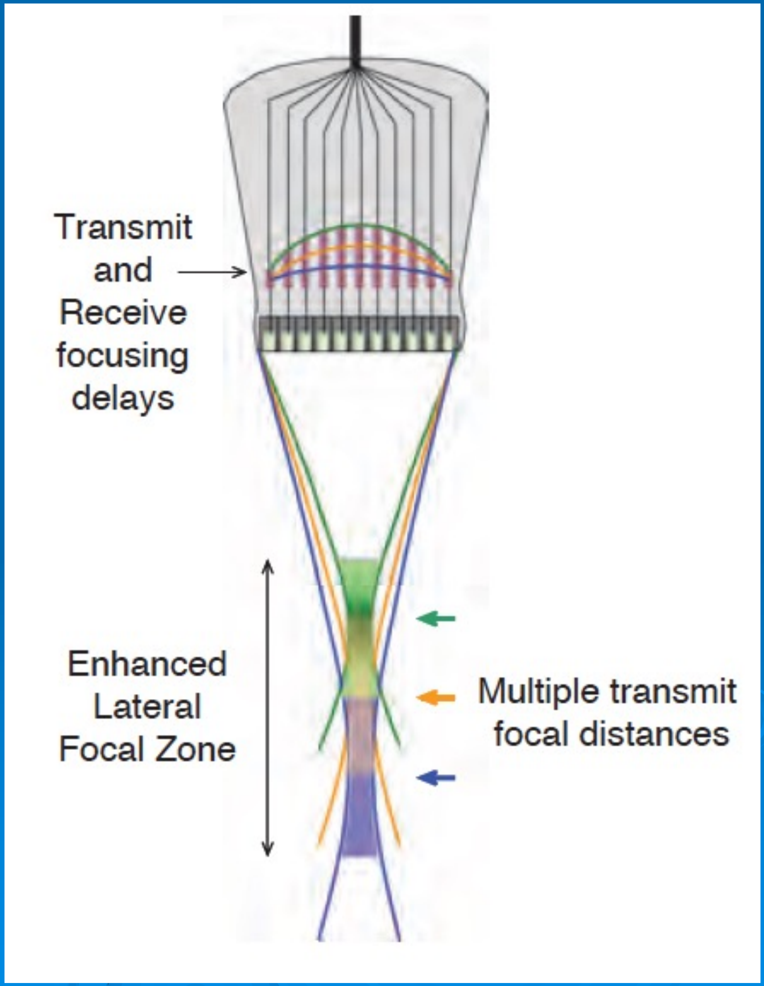
US lateral resolution

- **Phased-array transducers have multiple user selectable <u>transmit</u> and <u>receive</u> focal zones**
  - **Each focal zone requires the excitation of the entire array for a given focal distance**



US lateral resolution

- Good lateral resolution over an extended depth is achieved, but <u>the image frame rate is reduced</u>
  - Subsequent processing meshes the independently acquired data to enhance the lateral focal zone over a greater distance

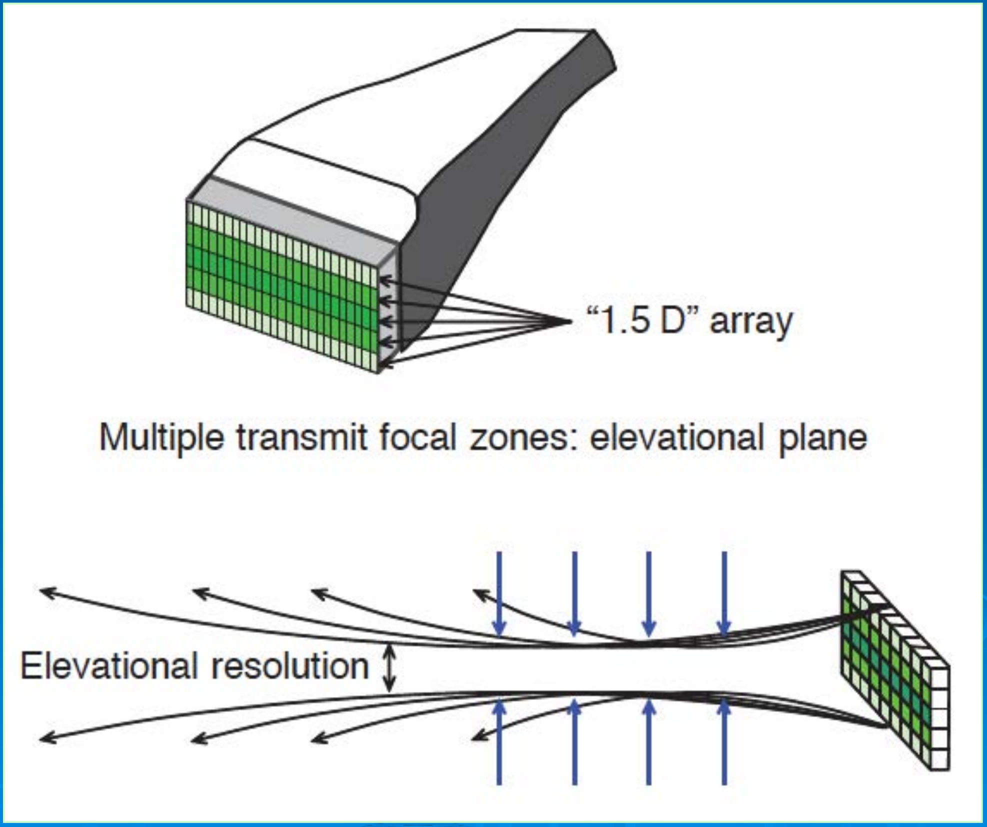


Elevational resolution

- The elevational dimension is perpendicular to the image plane
  - Slice thickness plays a significant part in image resolution
- Elevational resolution is dependent on the transducer element height
  - typically, the weakest measure of resolution for array transducers

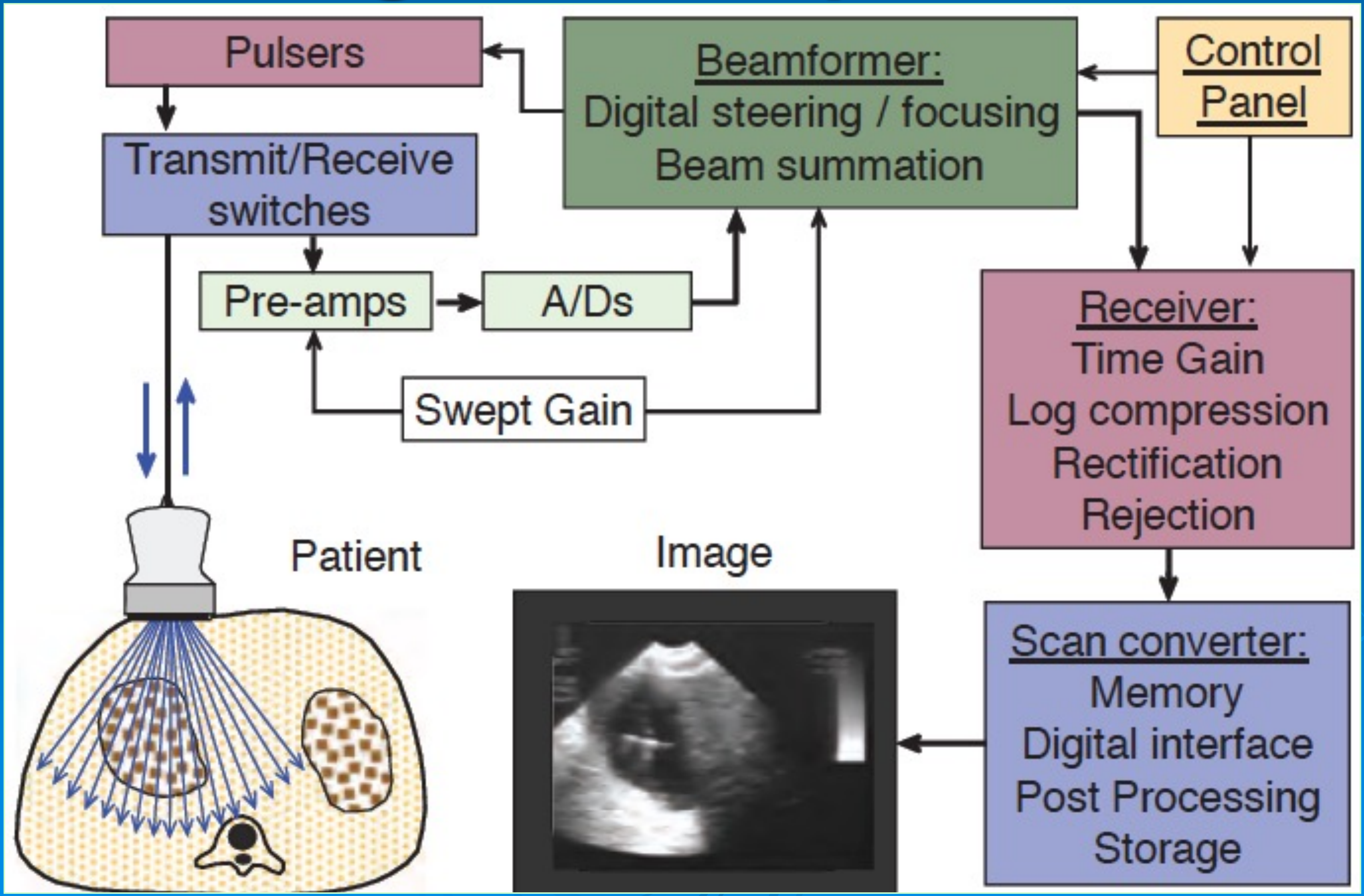
Elevational resolution

- Elevational resolution with multiple transmit focusing zones is achieved with "1.5D" transducer arrays to reduce the slice-thickness profile over an extended depth
- Five to seven discrete arrays replace the single array
  - Phase delay timing provides focusing in the elevational plane



An US scanner

Image data acquisition



Beam former

- The beam former generates the electronic delays for individual transducer elements in an array to achieve transmit and receive focusing
  - A beam former controls integrated circuits that provide for each of the transducer elements in the array
    - transmit/receive switches
    - digital-to-analog and analog-to-digital converters (ADCs)
    - preamplification and time gain compensation (TGC) circuitry

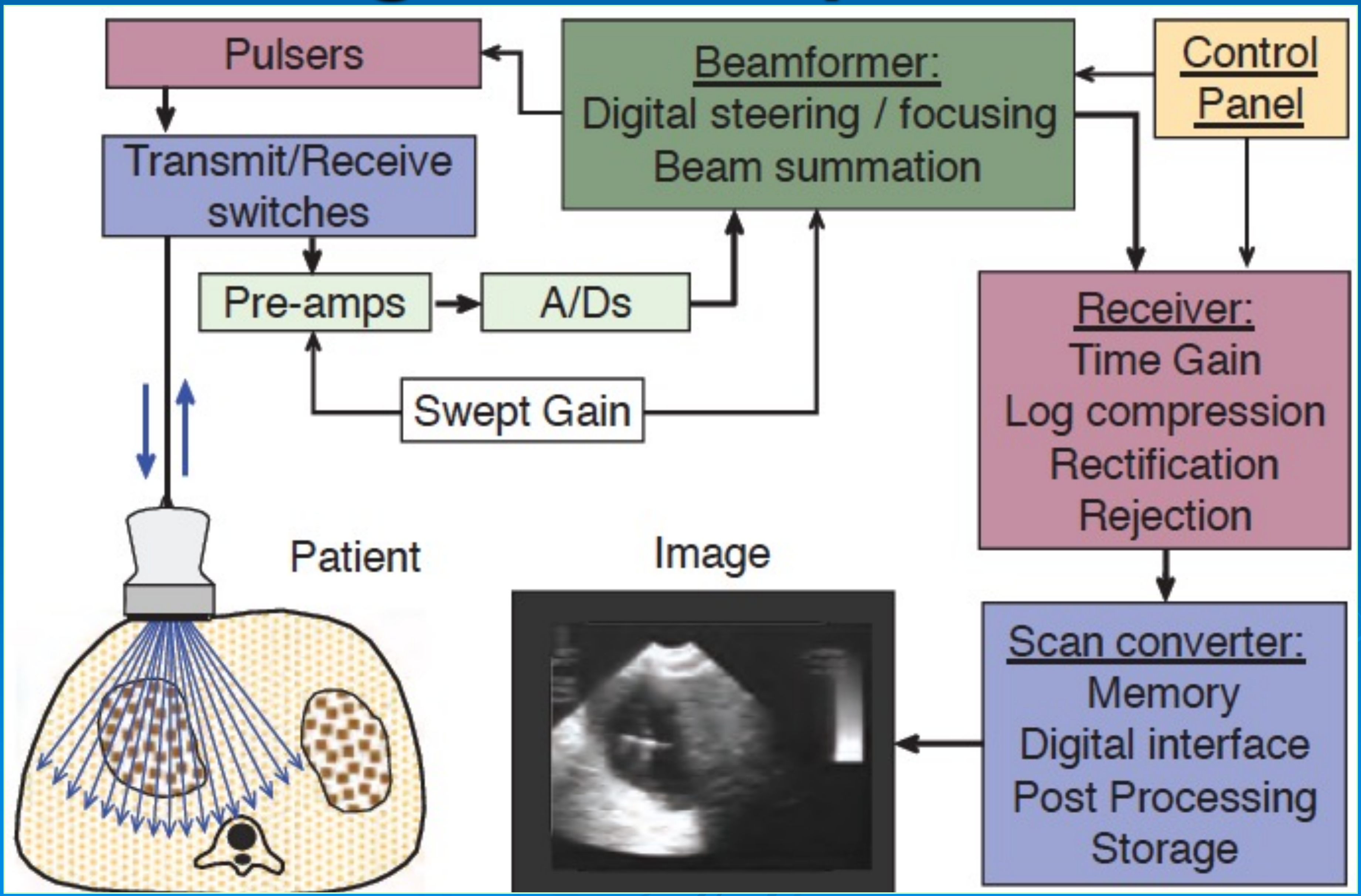
The transmitter (or the pulser)

- The pulser provides the electrical voltage for exciting the piezoelectric transducer elements and controls the output transmit power by adjustment of the applied voltage
  - An increase in transmit amplitude creates higher intensity sound and improves echo detection from weaker reflectors
    - A direct consequence is higher signal-to-noise ratio in the images but also higher power deposition to the patient

Transmit/Receive Switch

- **The transmit/receive switch, synchronized with the pulser, isolates the high voltage associated with pulsing (~150 V) from the sensitive amplification stages during receive mode**
  - **with induced voltages ranging from approximately 1 V to 2 μV from the returning echoes.**
  - **over a period up to about 1 ms**

Image data acquisition



Pulse-Echo Operation

In the pulse-echo mode, the ultrasound beam is intermittently transmitted

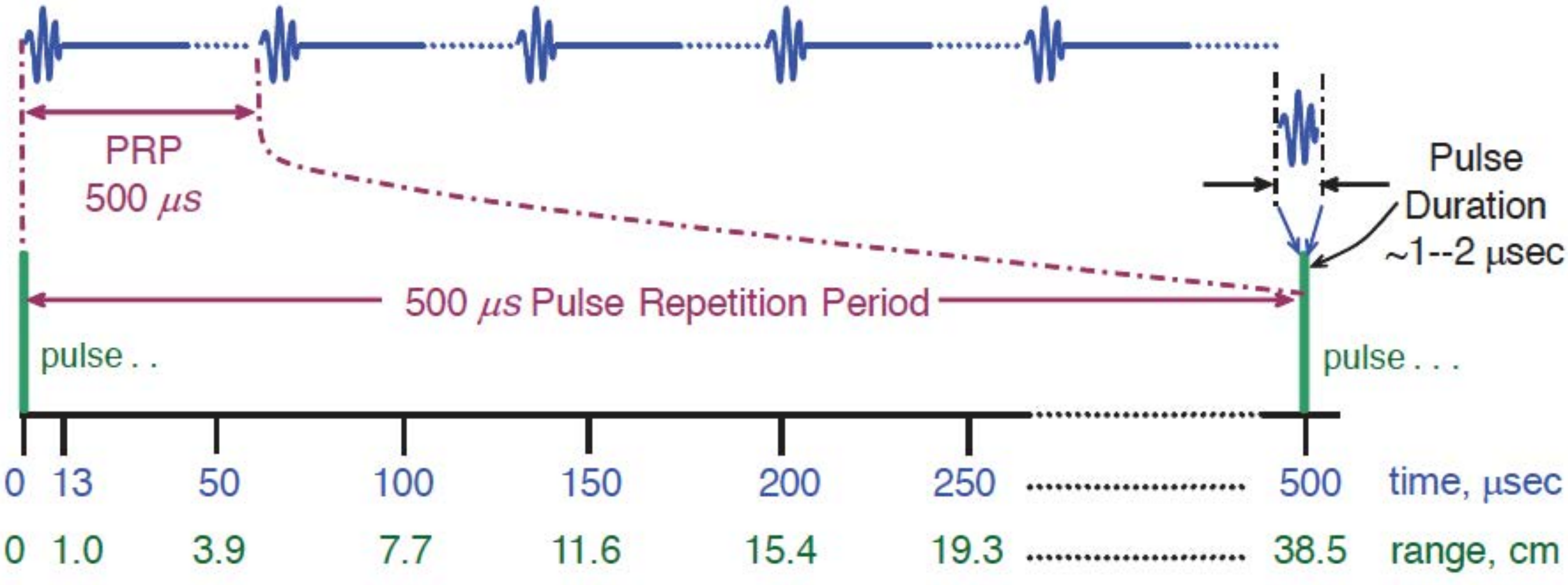
- a majority of the time occupied by listening for echoes
- One pulse-echo sequence produces one amplitude- modulated (A-line) of image data.

$$\text{Time } (\mu s) = \frac{2D(cm)}{c\,(cm/\mu s)} = \frac{2D(cm)}{0.154\,cm/\mu s} = 13\mu s/cm \times D(cm)$$

$$\text{Distance } (cm) = \frac{c\,(cm/\mu s) \times \text{Time } (\mu s)}{2} = 0.077 \times \text{Time}(\mu s)$$

the constant 2 represents the round-trip distance

Pulse-Echo Timing



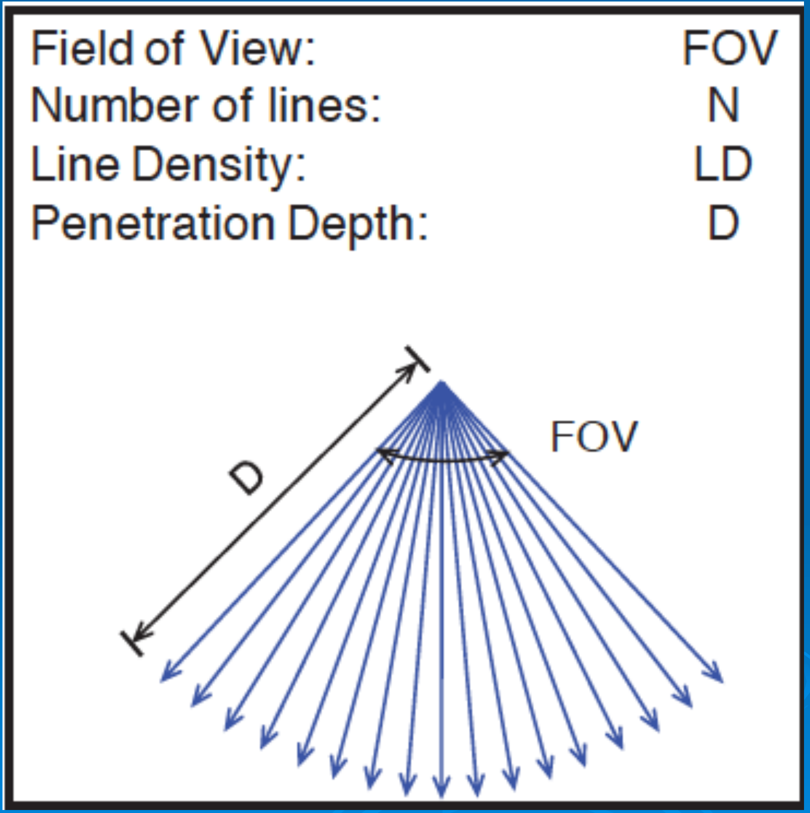
$$PRF = \frac{1}{PRP} = \frac{1}{500\,\mu s} = \frac{1}{500 \times 10^{-6}\,s} = \frac{2000}{s} = 2\,\text{kHz}$$

The maximum PRF is determined by the time required for echoes from the most distant structures to reach the transducer

Real-Time Ultrasound Imaging

The 2D image is created from a number of A-lines, N (100 or more)

- A larger number of lines will produce a higher quality image
- The finite time for pulse-echo propagation places an upper limit on N and impacts the temporal resolution

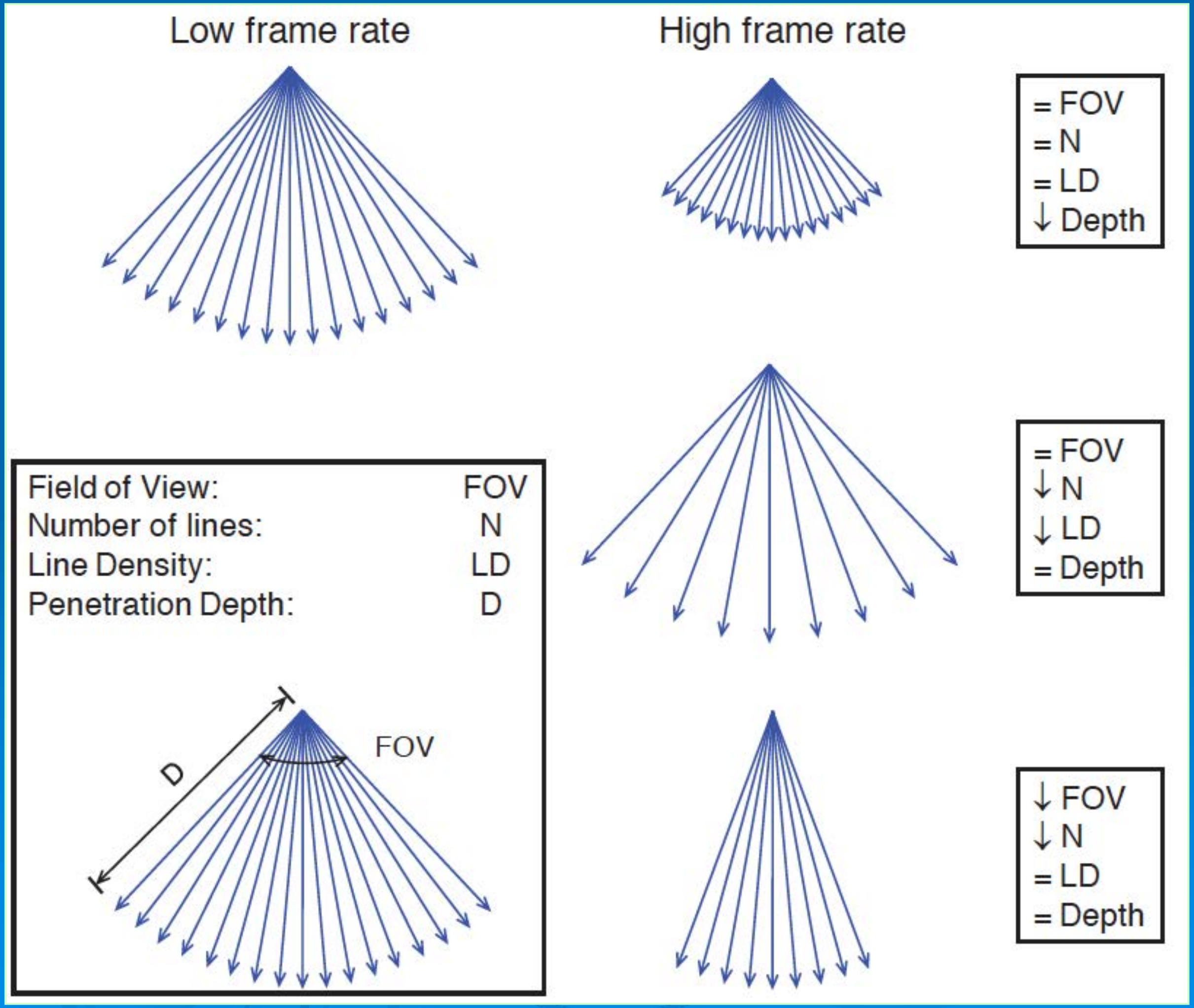


Frame Rate, FOV, Depth, Spatial Sampling Trade-Offs

Acquisition time for each line:

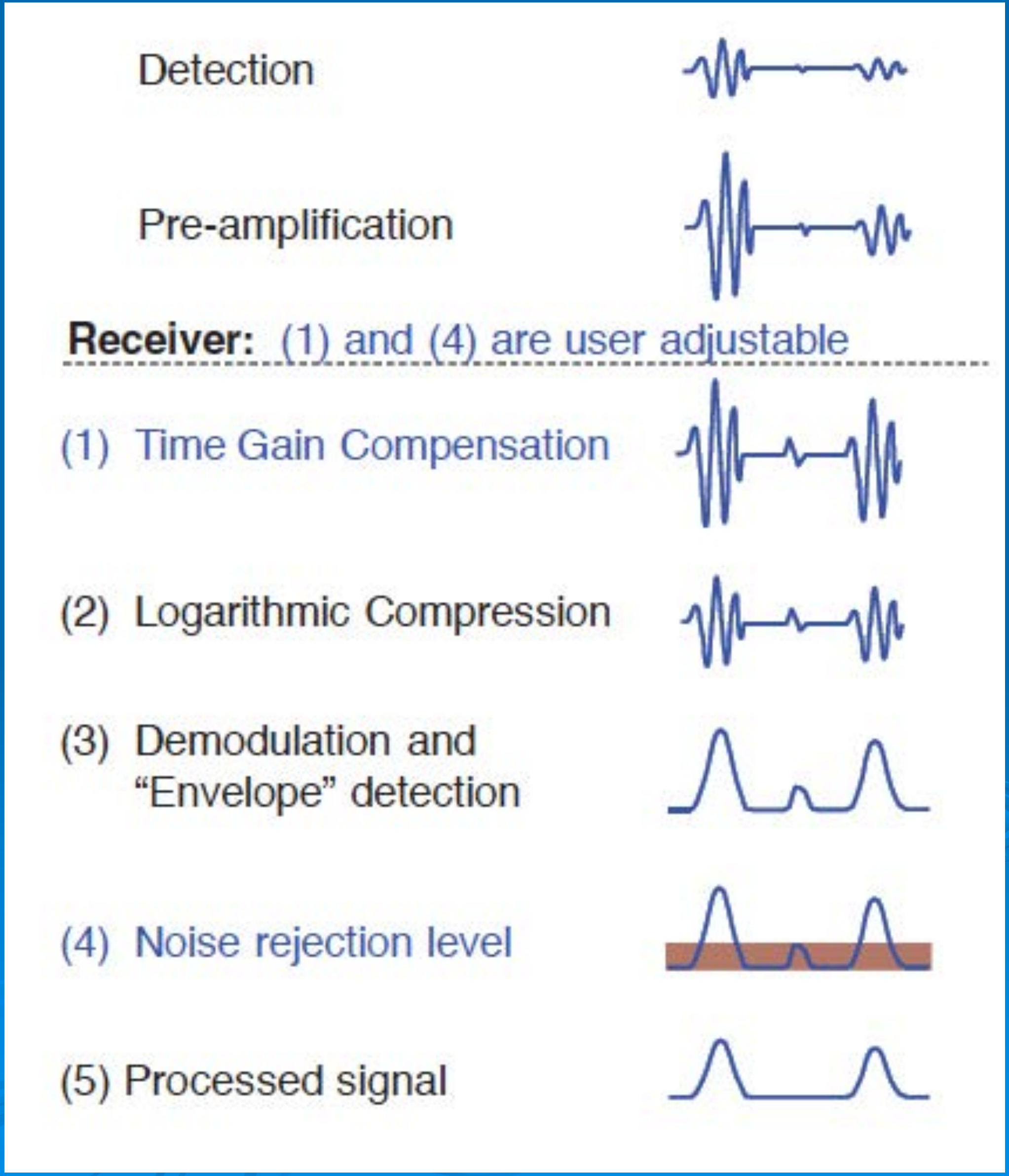
T_{line}=13μs/cm x D (cm)

It is required for the echo data to be unambiguously collected from a depth D



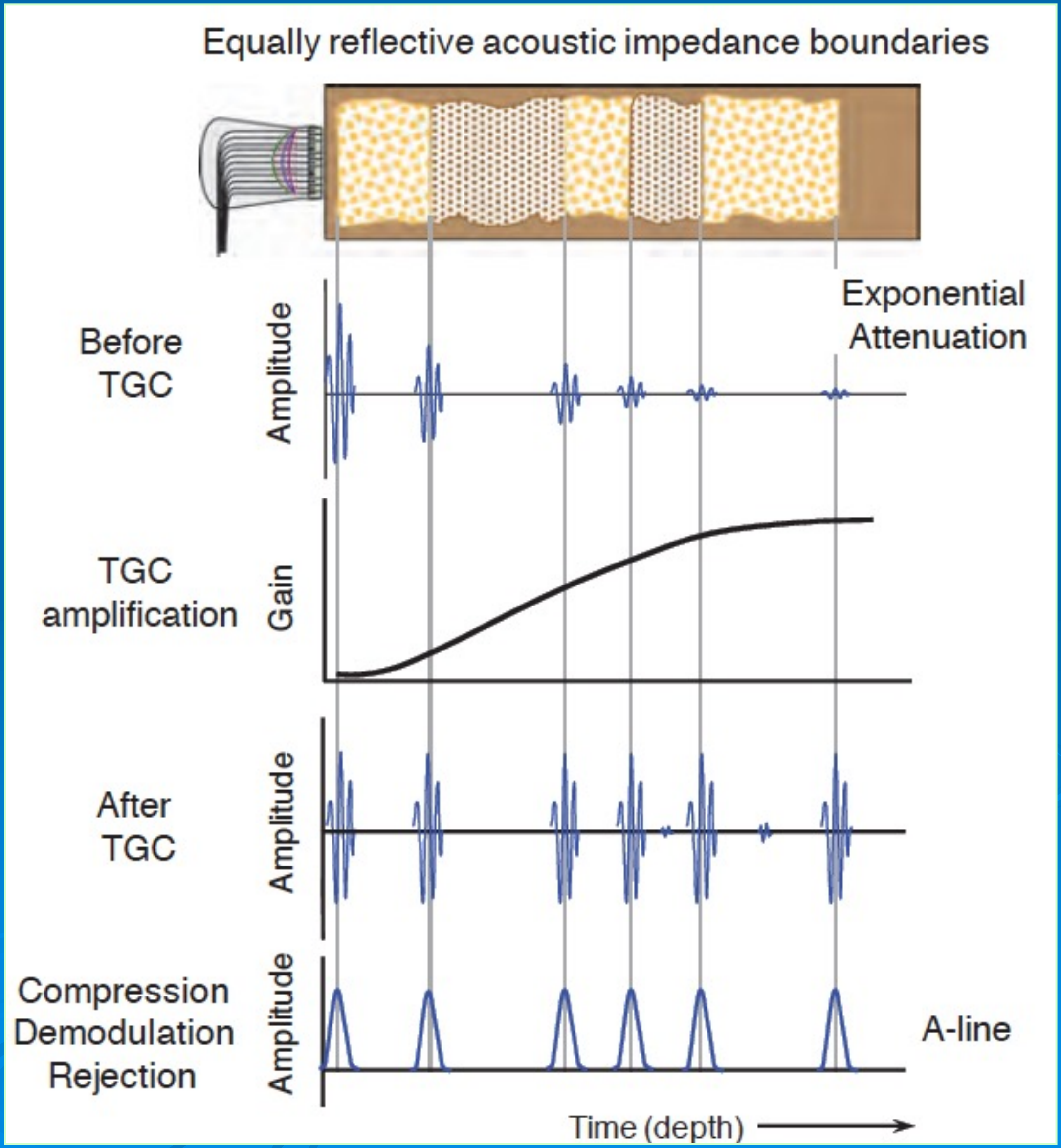
US signal processing

The user adjusts the Time Gain Compensation (TGC) and the noise rejection level

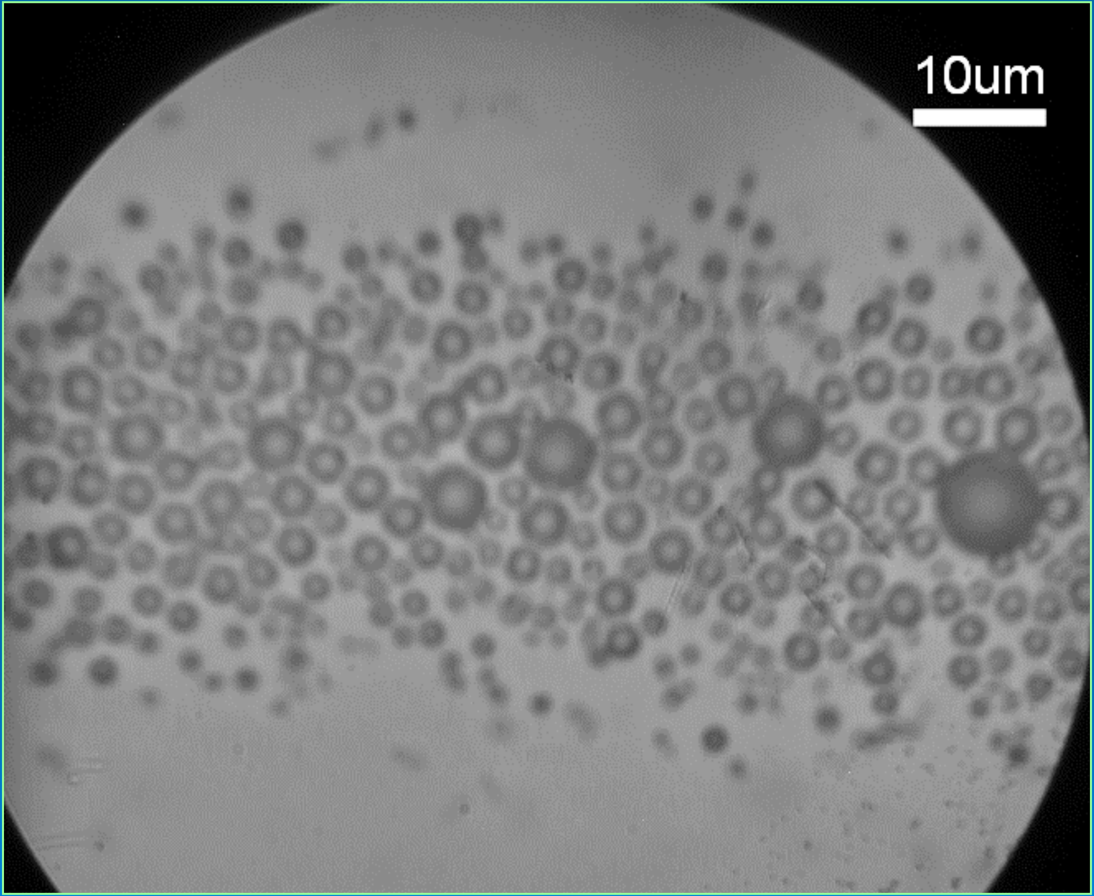


Time gain correction

TGC amplifies the acquired signals with respect to time after the initial pulse by operator adjustments



Microbubble for US



- The US contrast agents are based on microbubbles that are made to oscillate and burst due to cavitation

SonoVue® microbubbles in a cellulose capillary tube. Image at x100 magnification

http://ubimon.doc.ic.ac.uk/isc/public/HPosters06_151-200/paper158.pdf

Microbubbles for US

- US contrast agents are small gas bubbles encapsulated by a stabilizing shell, with a typical diameter on the order of microns
- Injected intravenously
- Bubbles remain within the blood pool and circulate in a manner similar to red blood cells

Table 1. Physical characteristics and dosage recommendations for ultrasound contrast agents used in imaging and drug delivery studies.

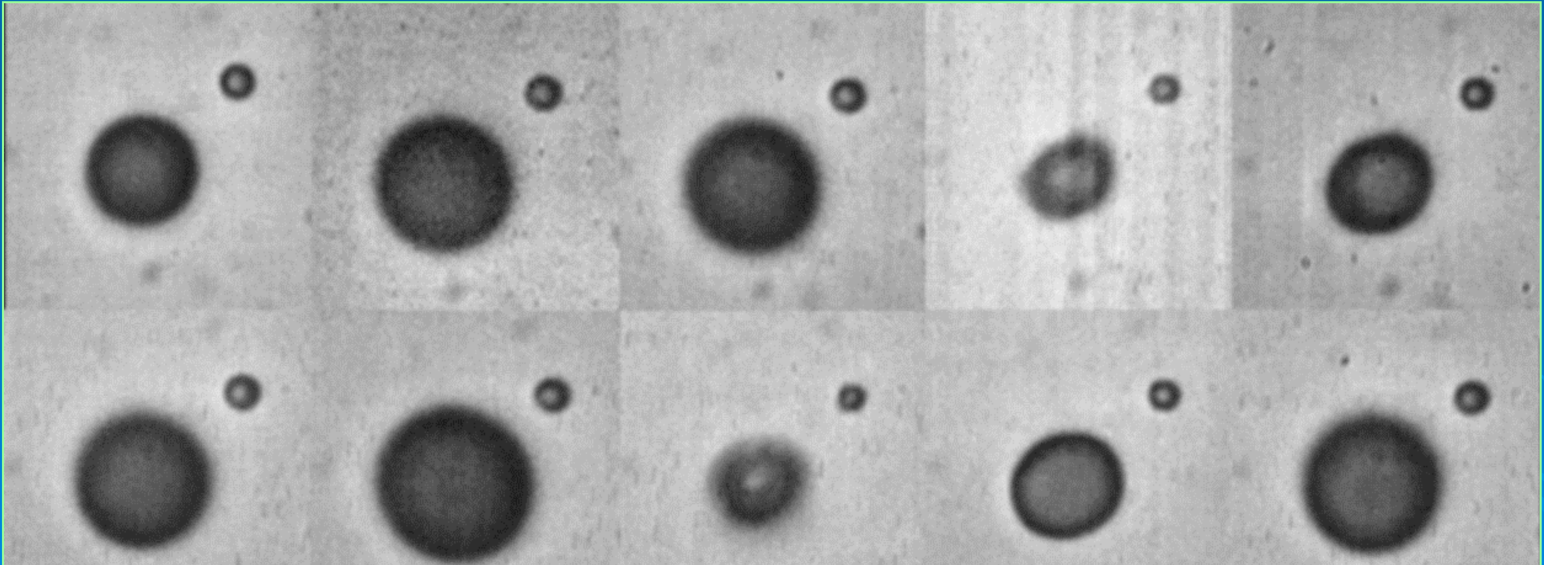
Formulation	Shell	Gas	Concentration (mL^{-1})	Mean diameter (μm)	Recommended dose[a]	Reference
Optison	Albumin	C_3F_8	$5.0–8.0 \times 10^8$	3.0–4.5	0.5 mL	(Optison-Prescribing-Information)
Definity	Lipid	C_3F_8	1.2×10^{10}	1.1–3.3	$10\ \mu L\ kg^{-1}$	(Definity-Prescribing-Information)
PESDA	Dextrose albumin	C_4F_{10}	6.5×10^8	2.5–4.9	$2.5–10\ (\mu L\ kg^{-1})$	(Porter *et al* 1996)

[a] Bolus intravenous injection into the peripheral vein.

Qin et al, Phys Medi Biol (2009) 54:27-57

Microbolle per US

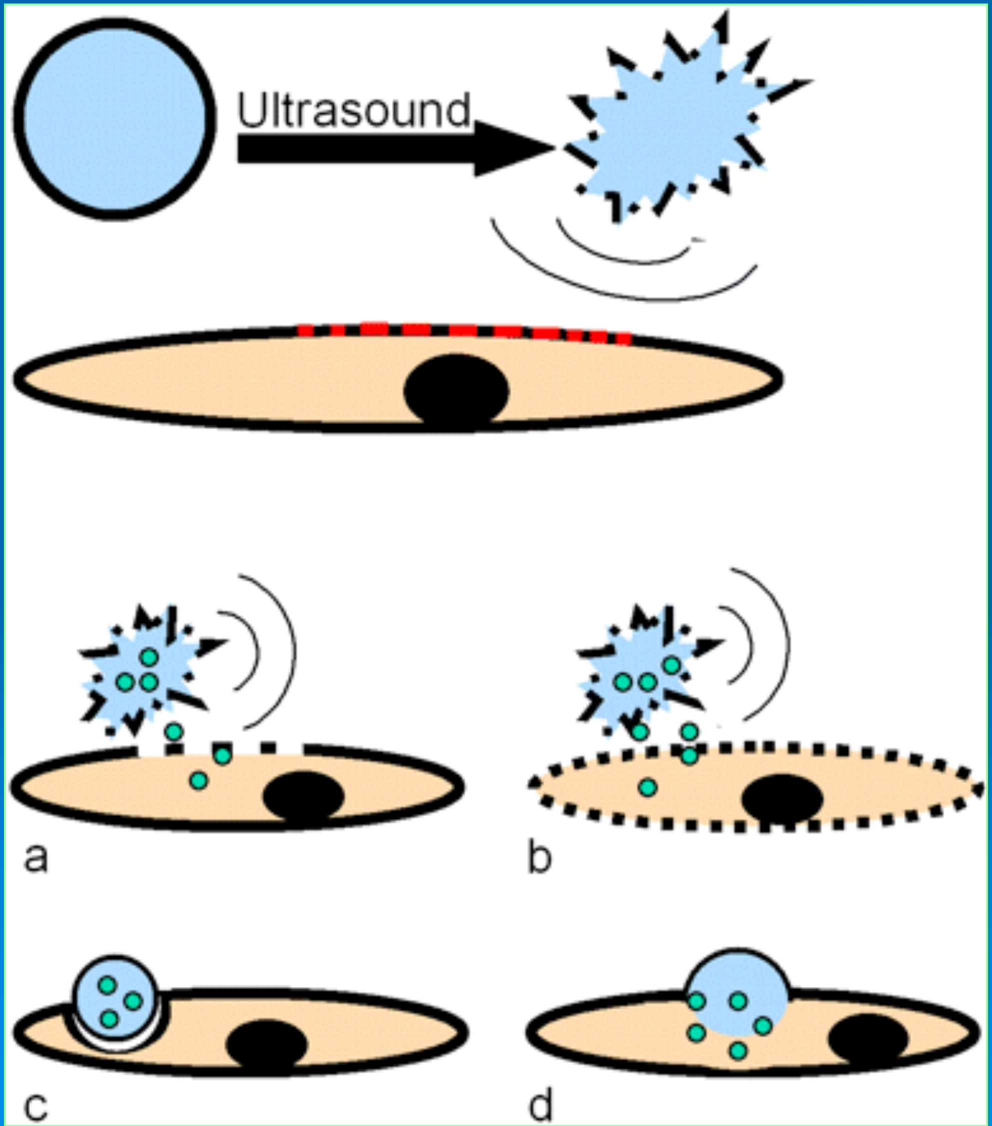
- US pulses are typically applied with a frequency near the resonance frequency of the gas bubble
- The bubbles increase and decrease in diameter, producing strong echoes from regions of perfused tissue



http://ubimon.doc.ic.ac.uk/isc/public/HPosters06_151-200/paper158.pdf

- High speed photography of two microbubbles oscillating in a 500 kHz 2 cycle ultrasound pulse. Image exposure time 500 ns.

Microbubbles and drug delivery



drug delivery through microbubbles: microbubbles containing the active substance are destroyed by US when they arrive at destination