Royal Brompton & Harefield NHS Trust
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BACK TO BASICS

General Principles of Cardiac Ultrasound

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Physics of Ultrasound

TYPES OF WAVES

Mechanical Waves

Characteristics
Requires a medium (matter) to travel

How they move
- Longitudinal wave
- Compressional wave
- Transverse wave

Examples
- Sound waves
- Seismic waves
- Ropes and Springs
- Water waves

Both

Transmit Energy

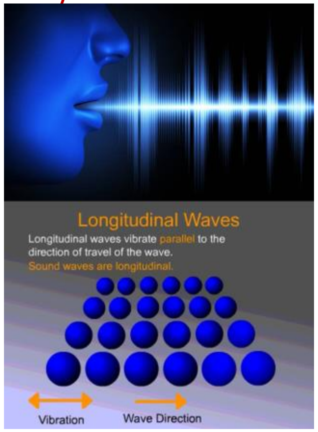
Do NOT transmit matter

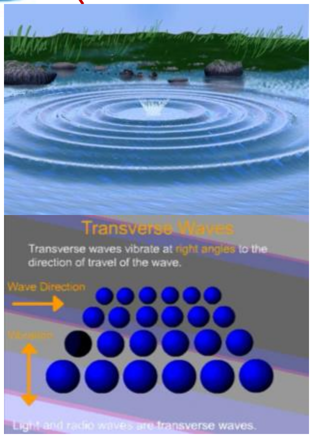
Electromagnetic Waves

Characteristics
Can travel through the vacuum of empty space

How they Move
- Transverse wave

Example
- Light





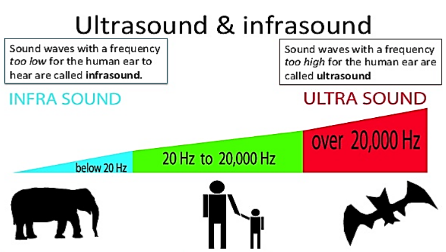
Sound Waves

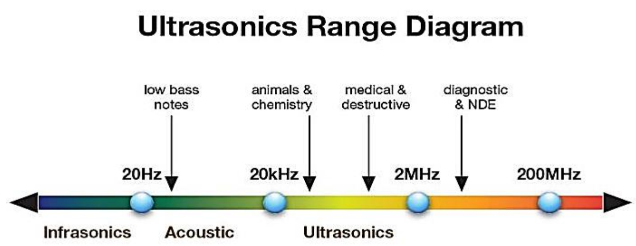
SOUND WAVES: mechanical vibrations that induce alternate compression and rarefaction of physical medium through which they pass

TRANSVERSE WAVES
(S waves)

LONGITUDINAL WAVES
(P waves)

ULTRASOUND WAVES: longitudinal sound waves higher than audible frequencies





INFRASOUND	*< 20Hz*
AUDIBLE SOUND	*20 - 20,000Hz*
ULTRASOUND	*> 20,000 Hz*

CARDIAC IMAGING ULTRASOUND: 1-20MHz

Transthoracic	1–8 MHz
Transoesophageal	3–10 MHz
Intracardiac	3–10 MHz
Epicardial	4–12 MHz
Intracoronary	10–20 MHz

**Currently used frequency range for cardiac imaging applications (including paediatric)*

Physical Properties of Sound wave

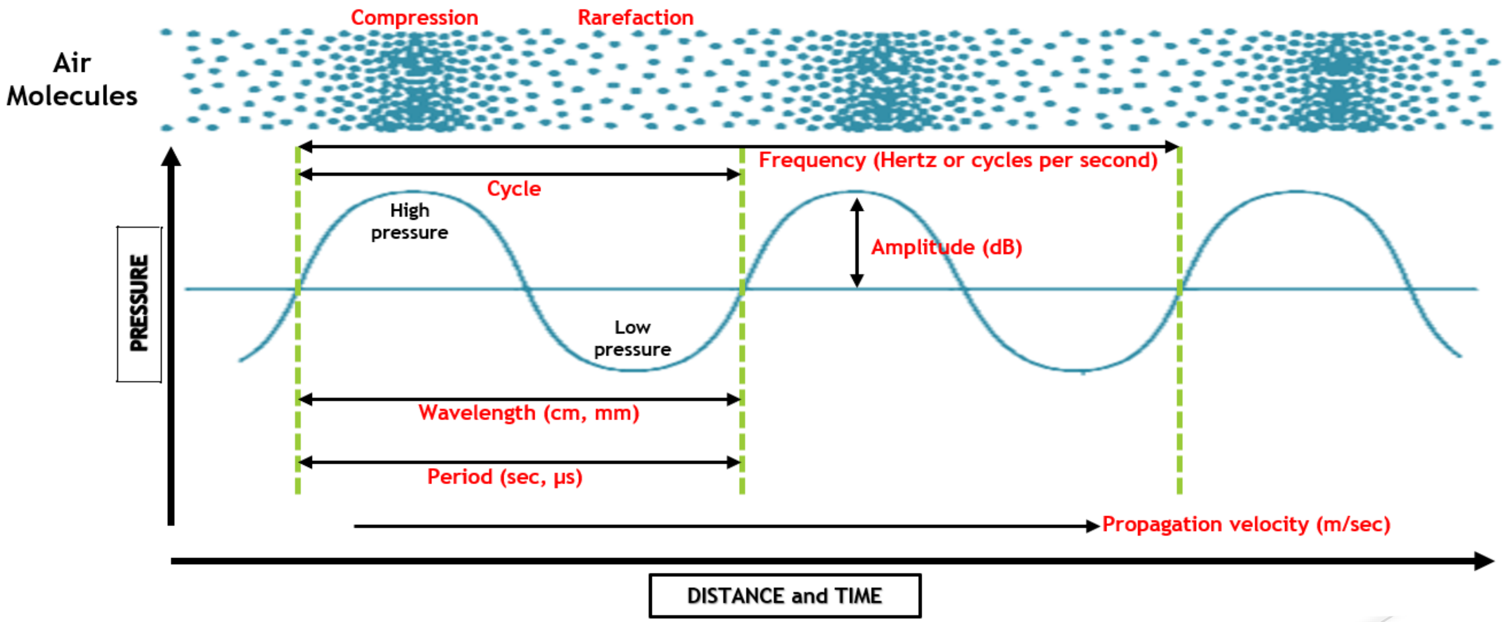
VELOCITY (m/sec): speed of sound propagation through a medium ($v = \lambda * f$)

FREQUENCY (Hz): number of cycles per second

Cycle: combination of one compression and rarefaction

WAVELENGTH (mm): distance between two similar adjacent points on a wave

AMPLITUDE (dB): strength of the signal



PERIOD (sec): length of time it takes for the completion of a single cycle

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WAVE PARAMETERS

1. Propagation Velocity
2. Frequency
3. Wavelength
4. Period
5. Amplitude
6. Power
7. Intensity

ACOUSTIC VARIABLES

1. Pressure
2. Density
3. Particle Displacement
4. Temperature

Wave Parameters

PROPAGATION VELOCITY

Medium	M/sec
Air	330
Fat	1460
WATER (20° C)	1480
Ave. Soft Tissue	**1540**
WATER (50° C)	1540
Liver	1559
Blood	1570
Muscle	1580
Bone	3500
PZT (Crystal)	4000

-unit of Measurement: *m/sec*, *cm/sec*
-speed with a given direction (and temperature)
-depends on DENSITY or STIFFNESS of medium
-is constant for human soft tissue: 1540m/sec

DENSITY (ρ) - mass per unit volume, difficult to get "vibrating"
STIFFNESS (Β) - or BULK MODULUS (measure of stiffness of tissue), not compressible so sound travels at high velocities

↑ DENSITY ∝ ↓ VELOCITY
↓ DENSITY ∝ ↑ VELOCITY

↑ STIFFNESS ∝ ↑ VELOCITY
↓ STIFFNESS ∝ ↓ VELOCITY

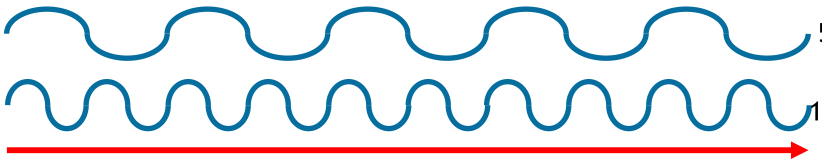
FORMULA:

$$\text{Velocity} = \frac{\sqrt{stiffness}}{density}$$

$$V = \sqrt{\beta/\rho}$$

FREQUENCY

-unit of Measurement: *HERTZ = 1 cycle per second*
-determined by the source (function of transducer)
-constant within the given medium

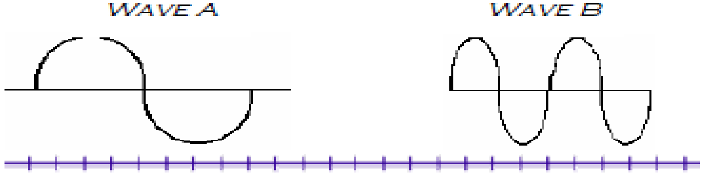


KILOHERTZ = 1,000 Hz
MEGAHERTZ = 1,000,000 Hz

5MHz Transducer = 5,000,000 Hertz or CPS

WAVELENGTH

-unit of Measurement: *mm*, *µm*
-describes the spatial dimension of the a wave
-directly affects image quality
-not operator adjustable



↑ FREQUENCY ∝ ↓ WAVELENGTH
↓ FREQUENCY ∝ ↑ WAVELENGTH

Remember....

FORMULA:
(λ=V/F)

Exercises

EXAMPLE 1: Calculate the wavelength for 2.25MHZ

λ =v/f

=1,540m/sec ÷ 2,250,000cycles/sec

=0.00068m x 1,000 mm/m

=0.68mm

V (constant) = 1,540m/sec

EXAMPLE 2: Calculate the wavelength for 3.5MHZ

λ =v/f

=1,540m/sec ÷ 3,500,000cycles/sec

=0.00044m x 1,000 mm/m

=0.44mm

EXAMPLE 3: Calculate the wavelength for 5.0MHZ

λ =v/f

=1.540m/sec ÷ 5,000,000cycles/sec

=0.00031m x 1,000 mm/m

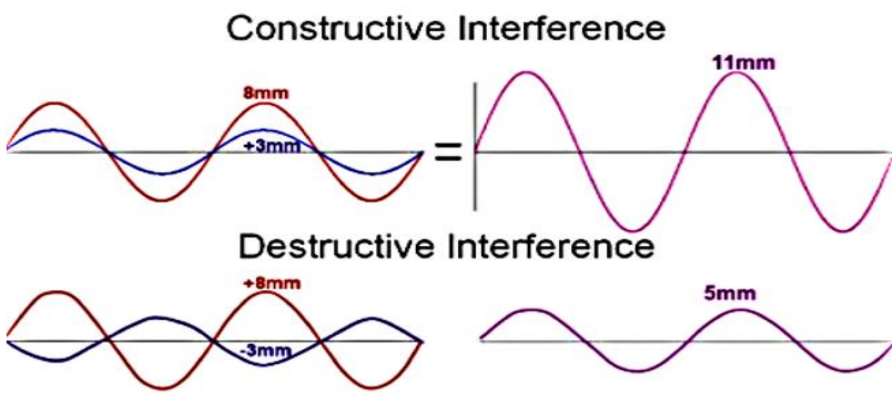
=0.31mm

↑ FREQUENCY = ↓ WAVELENGTH

Wave Parameters

Wave Phase -two waves of different frequencies are produced at the same time and combine to create a new wave

-important in understanding the formation of sound beams, electronic beam and focusing methods, Doppler instrumentation designs and matching layer transducer designs



Both scenarios assume that the waves are 100% in phase and 100% out of phase.

HIGHER FREQUENCIES = SHORTER WAVELENGTHS
SHORTER WAVELENGTHS = BETTER AXIAL RESOLUTION
BETTER AXIAL RESOLUTION = MORE DIAGNOSTIC INFORMATION

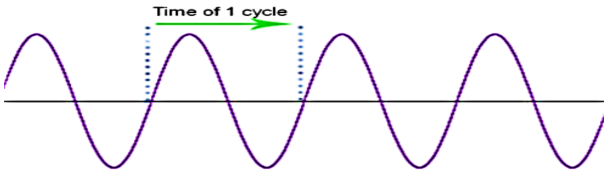
BUT

HIGHER FREQUENCIES = LESSER PENETRATION

↑FREQUENCY = ↓WAVELENGTH = ↑AXIAL RESOLUTION = ↑DIAGNOSTIC INFORMATION = ↓PENETRATION

Wave Parameters

PERIOD
- -unit of Measurement: ***seconds*, *ms*, *µs***
- **-the time it takes for one cycle to occur**
- **-not operator adjustable**



FORMULA:
PERIOD = 1/ frequency

↑FREQUENCY = ↓WAVELENGTH = ↓PERIOD

Exercises

EXAMPLE 1: Determine the period of a single waveform emitted by a 3.5MHz transducer

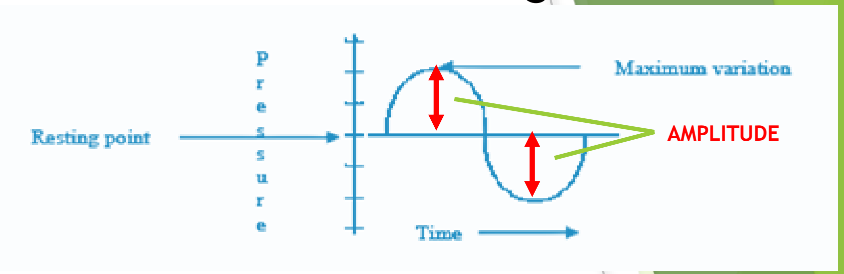
Period =1/frequency
=1/3,500,000cycles/sec
=0.000000285sec/cycle
=0.285µs

EXAMPLE 2: Determine the period of a single waveform emitted by a 5.0MHz transducer

=1/5,000,000cycles
=0.0000002sec/cycle
=0.20µs

Wave Parameters

AMPLITUDE
-unit of Measurement: *Varies (cm or mm, grams/cc3, mmHg)*
-height of the compression or depth of the rarefaction
-determined by the sound source, decreases as it travels through the tissue (attenuation)
-directly proportional to POWER
-operator adjustable



POWER
-unit of Measurement: *Watts or mWatts*
-rate energy transmitted into substance or the rate work is performed
-determine by the sound source
-operator adjustable

↑POWER = ↑AMPLITUDE

FORMULA:
$POWER_{watts} \propto (AMPLITUDE)^2_{mm}$

INTENSITY
-unit of Measurement: *Watts/cm² or mWatts/ cm²*
-measure of ultrasound energy concentration present in human soft tissue
-rate energy travels through a substance
-measurement can be over a given area (spatial) or period of time (temporal)
-operator adjustable

↑INTENSITY = ↓AREA = ↑ENERGY

FORMULA:
$INTENSITY_{watts/cm2} = \frac{POWER_{watts}}{AREA_{cm2}}$

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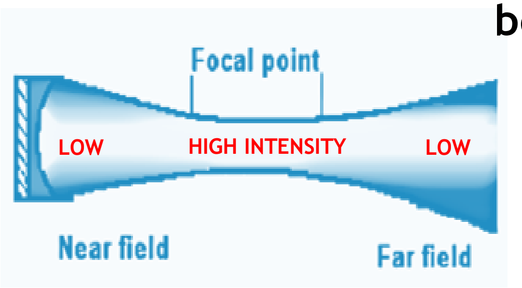
Wave Parameters

SPACIAL INTENSITY -exact measurement of energy dispersed over a given area
-energy is highest at the narrowest point along the central beam, than in the beginning and end



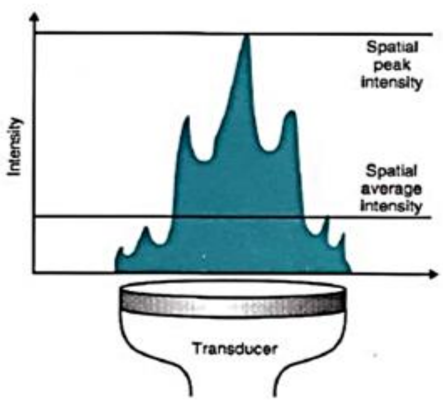
FORMULA:

$$\text{SPATIAL INTENSITY}_{watts/cm^2} = \frac{\text{POWER}_{watts}}{\text{AREA}_{cm^2}}$$

SPACIAL PEAK - *measured along the central beam at the narrowest point*

SPACIAL AVERAGE - *average intensity in the sound beam and is usually measured at the transducer face*

BEAM UNIFORMITY RATIO/COEFFICIENT (SP/SA) - *ratio between spatial peak and spatial average intensities*



FORMULA:

$$\text{SP/SA FACTOR} = \frac{\text{SPACIAL PEAK}_{watts/cm2}}{\text{SPACIAL AVERAGE}_{watts/cm2}}$$

NOTE*: *SP/SA FACTOR ≥1

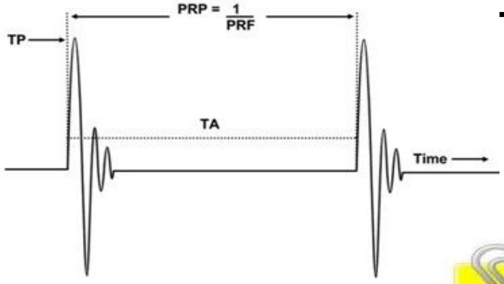
EXAMPLE 1: If the SPACIAL PEAK = 2mW/cm^2; SPACIAL AVERAGE = 1mW/cm^2

$$\text{SP/SA FACTOR} = \frac{2mW/cm^2}{1mW/cm^2}$$

= 2:1

Wave Parameters

TEMPORAL INTENSITY



-exact measurement of energy dispersed over a given time
-energy is less in the beginning and end of a pulse, than in the middle

TEMPORAL PEAK - the point in time when intensity reaches its maximum
TEMPORAL AVERAGE - average of all occurring pulses, only "ON" time is included
DUTY FACTOR- proportion of time that sound energy is actually produced, "ON" time

FORMULA:
TEMPORAL AVE.= TEMPORAL PEAK x DUTY FACTOR

NOTE: DUTY FACTOR values between 0 to 1

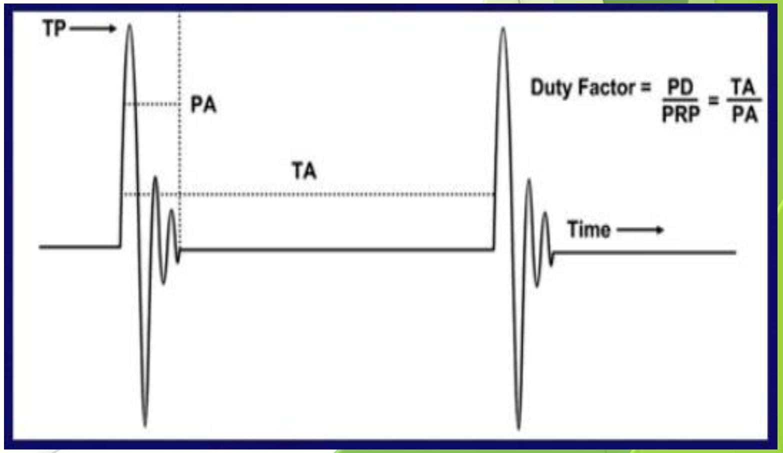
↑TEMPORAL AVERAGE = ↑TEMPORAL PEAK
= ↑DUTY FACTOR

DEFINITIONS:

SPATIAL: where in SPACE is intensity measured (depth in the body).
TEMPORAL: when in TIME is intensity measured.
PEAK: the MAXIMUM value
AVERAGE: the average, or mean, value

These definitions can be combined as follows in referring to the intensity of an ultrasound beam:

SPTP - spatial peak, temporal peak - HIGHEST VALUE
SATP - spatial average, temporal peak
SPTA - spatial peak, temporal average – MOST COMMONLY REFERENCED IN BIOEFFECTS
SATA - spatial average, temporal average - LOWEST VALUE
SPPA - spatial peak, pulse average - AVERAGE OVER DURATION OF *PULSE* ONLY.



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Acoustic Variables

PRESSURE -unit of Measurement: Newtons/m^2

- the ratio of a force acting on a surface of an object

FORMULA:

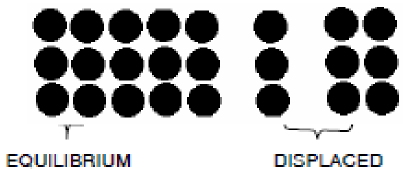
$$PRESSURE = \frac{FORCE}{AREA}$$

↑AREA = ↓PRESSURE

↑DENSITY = ↓COMPRESSIBILITY

PARTICLE DISPLACEMENT -unit of Measurement: units of distance

-distance that particles move from equilibrium positions



DENSITY -unit of Measurement: Kg/m^2, g/cm^2

-mass per unit volume

FORMULA:

$$DENSITY = \frac{MASS}{VOLUME}$$

TEMPERATURE-unit of Measurement: ºC, ºF

-measure of relative warmth or coolness of an object

-measures not the heat of the substance rather, the average kinetic energy of its molecules

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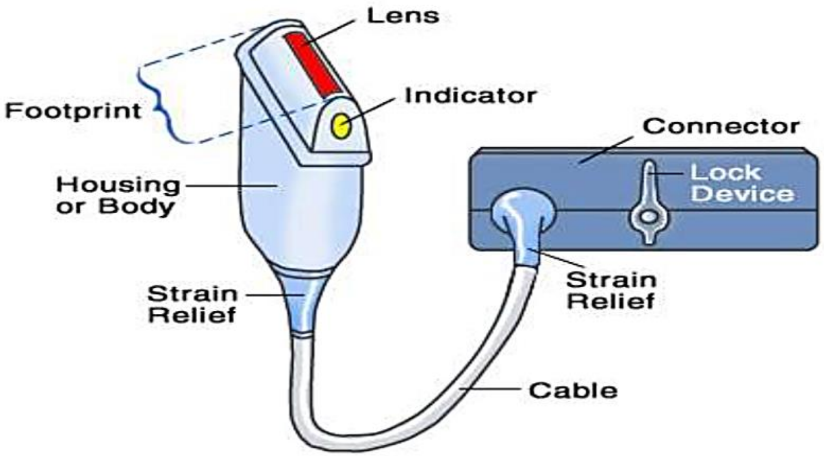
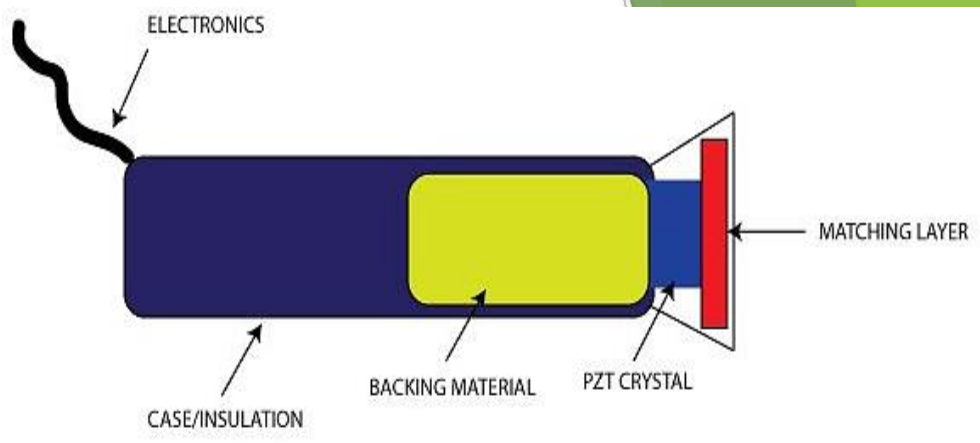
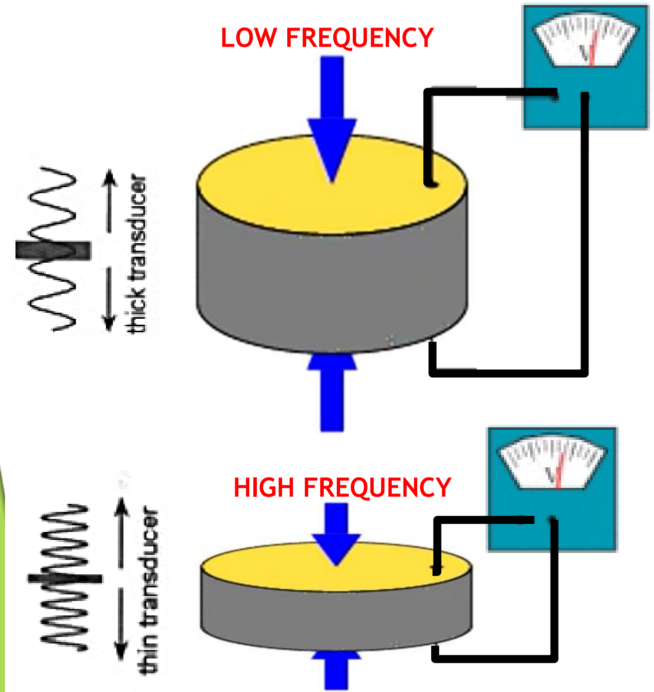
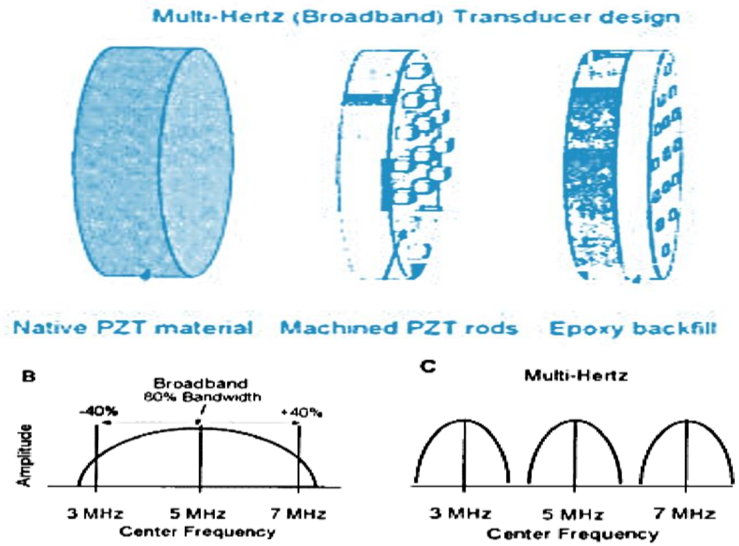
Transducers and Piezo-electric Effect



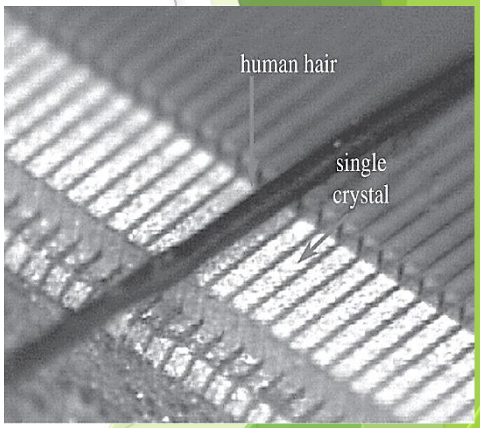




- VOLTAGE SPIKE OF 150V (SHORT DURATION)



- Large number of small "RODS" filled with epoxy resin to create a smooth surface
- SHORT SQUARE WAVE BURST OF 150V WITH 1 - 3 CYCLES

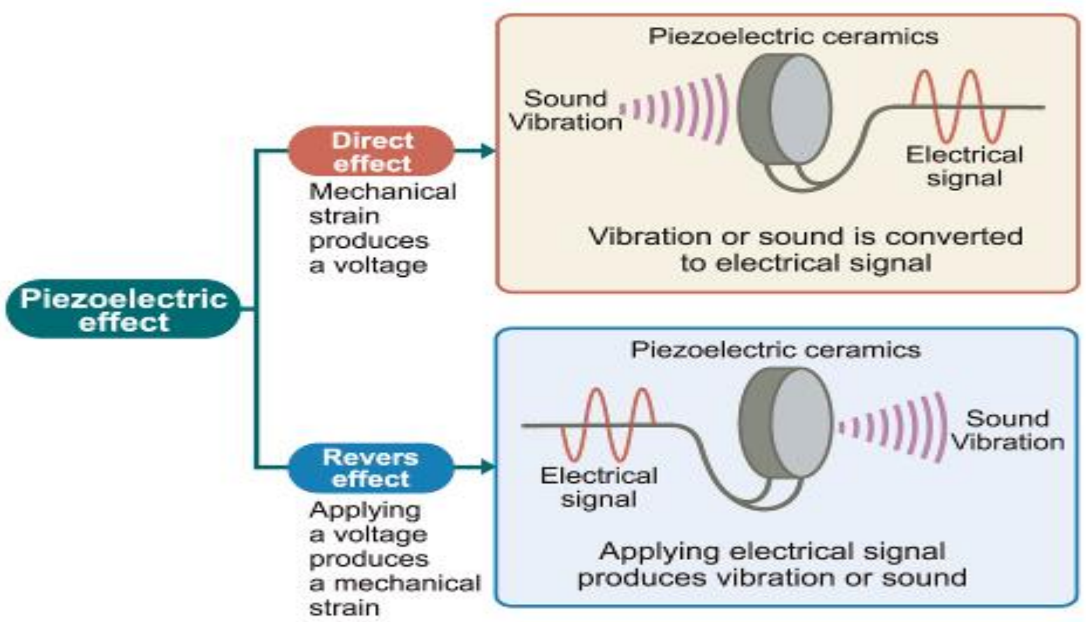


PIEZO ELECTRIC SINGLE CRYSTAL VERSUS HUMAN HAIR

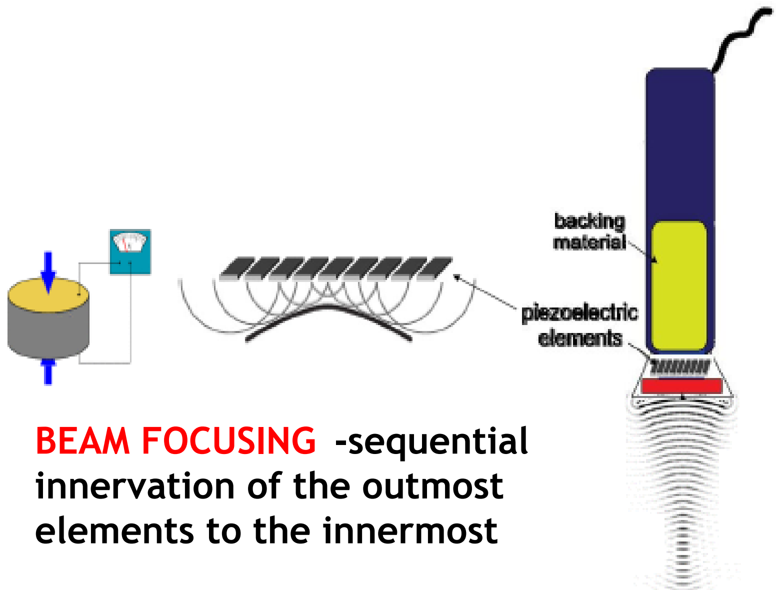
Transducers and Piezo-electric Effect

Piezoelectric Effect:

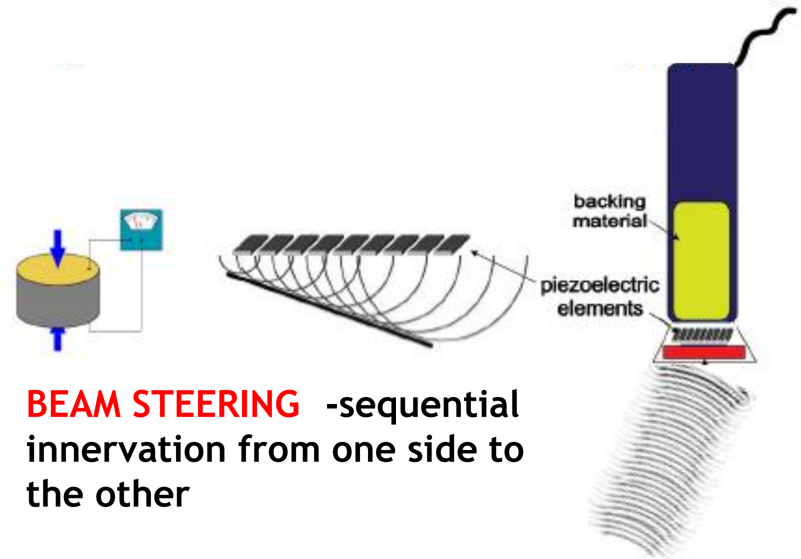
- The conversion of electrical energy into mechanical energy – transmission of the sound beam
- The conversion of mechanical energy in electrical energy – receiving the reflected beam information
- Electricity is applied to the piezoelectric material which vibrates (expands and contracts) to produce mechanical sound or pressure waves
- Returning sound waves cause mechanical vibrations (acoustic pressure) of the piezoelectric material that are converted into the electrical signal for the display



Transducers and Piezo-electric Effect



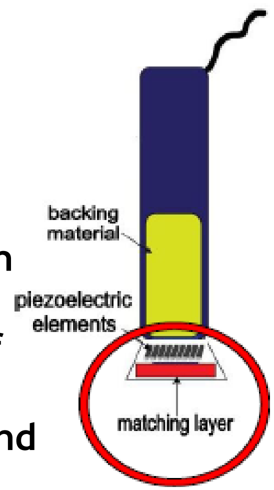
BEAM FOCUSING -sequential innervation of the outmost elements to the innermost



BEAM STEERING -sequential innervation from one side to the other

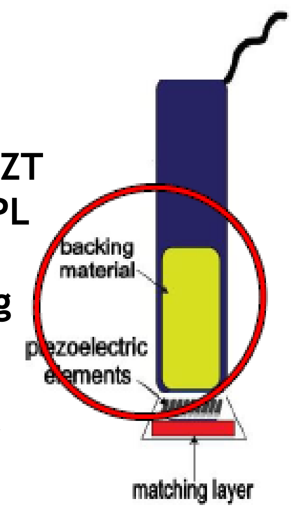
MATCHING LAYER

-thin layer on the surface of the probe (between the skin and piezoelectric elements
Thickness is generally ¼ the wavelength of the ultrasound produced
-purpose is to maximize transmission of sound from PZT to patient, reduce the amount of reflection at this interface and improve transmission to patient

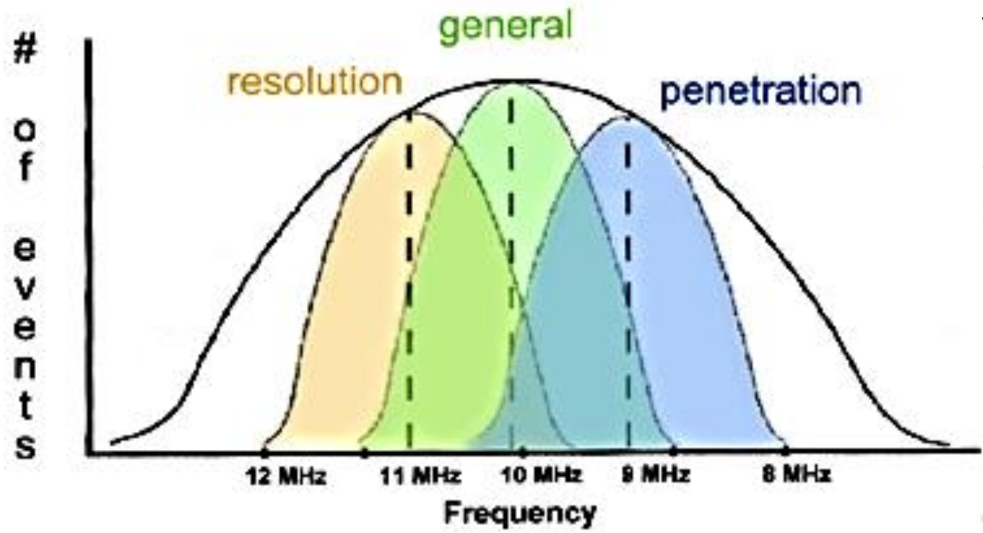


BACKING/DAMPING MATERIAL

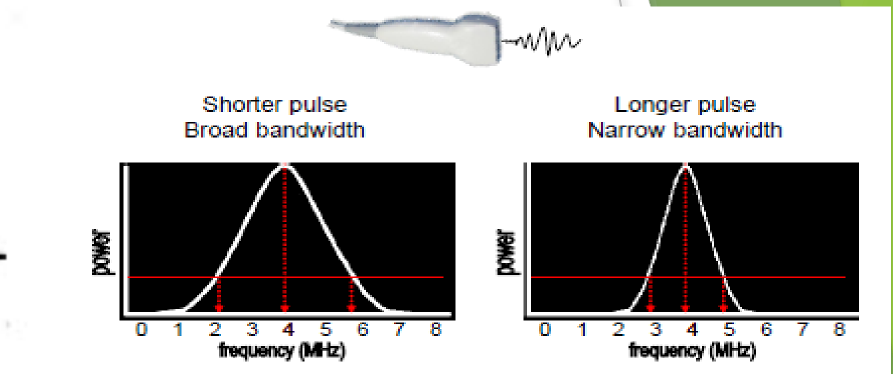
- reduces/damps "ringing" of PZT crystal, thereby shortening SPL and improves axial resolution
- widens bandwidth, decreasing quality factor
- diagnostic transducer: wide bandwidth, low quality factor



Transducers and Piezo-electric Effect



- A "**high Q**" transducer has a narrow bandwidth (i.e., very little damping) and a corresponding long spatial pulse length – organ imaging
- A "**low Q**" transducer has a wide bandwidth and short spatial pulse length – doppler



CENTER FREQUENCY *– frequency between the lower and upper cut-off frequencies*
BANDWIDTH *– width from the difference of the upper and lower cut-off frequencies*
Q FACTOR *– determines the purity of the sound and the length of time the sound persists*

FORMULA:

$$FREQUENCY_{center} = \sqrt{FREQUENCY_{low} \times FREQUENCY_{high}}$$

$$FREQUENCY_{bandwidth} = FREQUENCY_{high} - FREQUENCY_{low}$$

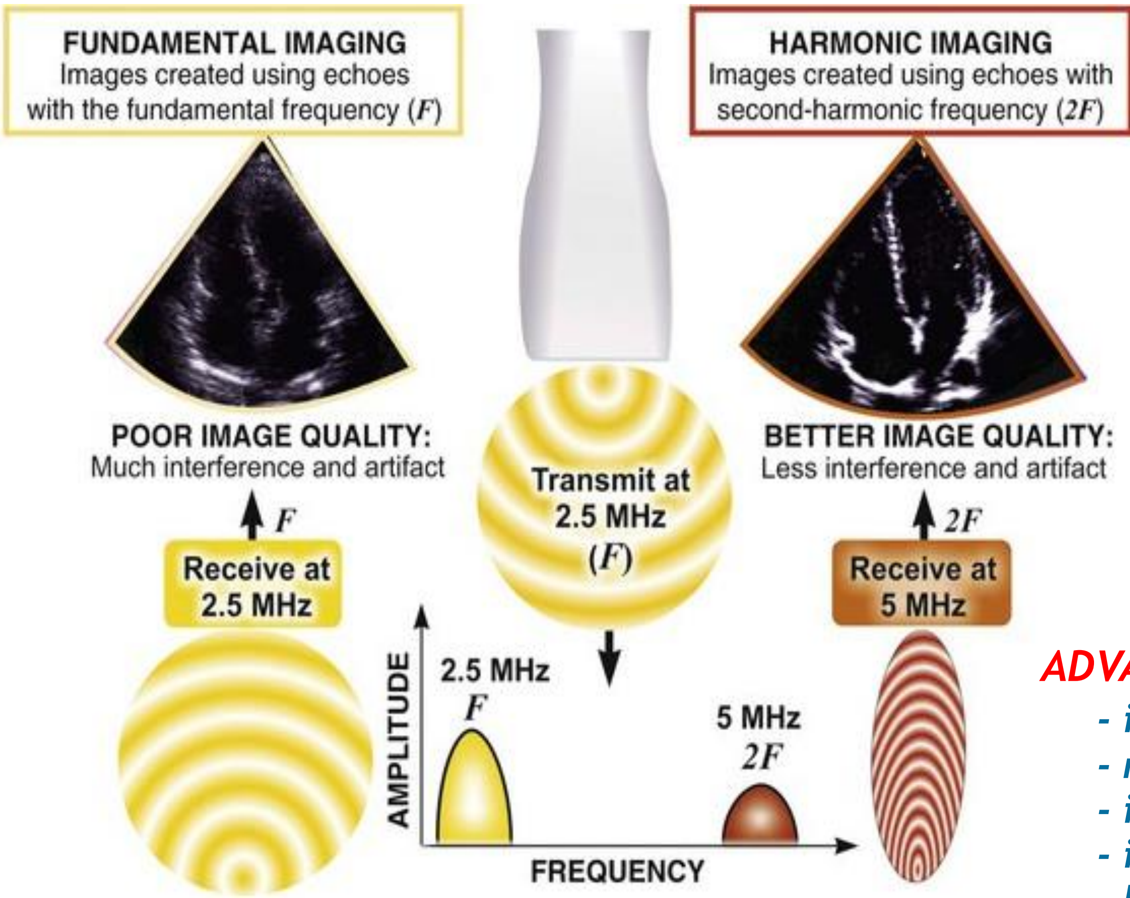
$$Q\ FACTOR = FREQUENCY_{center} / FREQUENCY_{bandwidth}$$

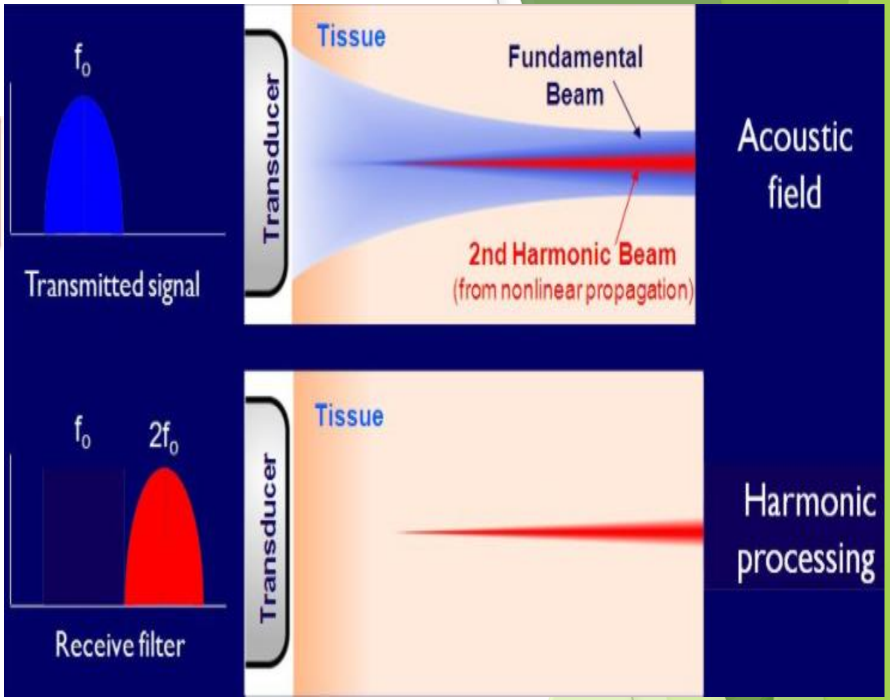
FREQUENCY = ↑ SPATIAL PULSE LENGTH
= ↑ BANDWIDTH
= ↑ AXIAL RESOLUTION
= ↓ Q FACTOR

- *Excitation allows the center frequency to be selected within the limits of the bandwidth*
- *Broad bandwidth permits reception of echoes within the wide range of frequencies, i.e. low frequency pulses received at higher frequency*

Transducers and Piezo-electric Effect

HARMONIC IMAGING -recently introduced technique that uses the ability of broadband transducers
-are formed by utilizing the harmonic signals that are generated by tissue and by filtering out the fundamental echo signals that are generated by the transmitted acoustic energy

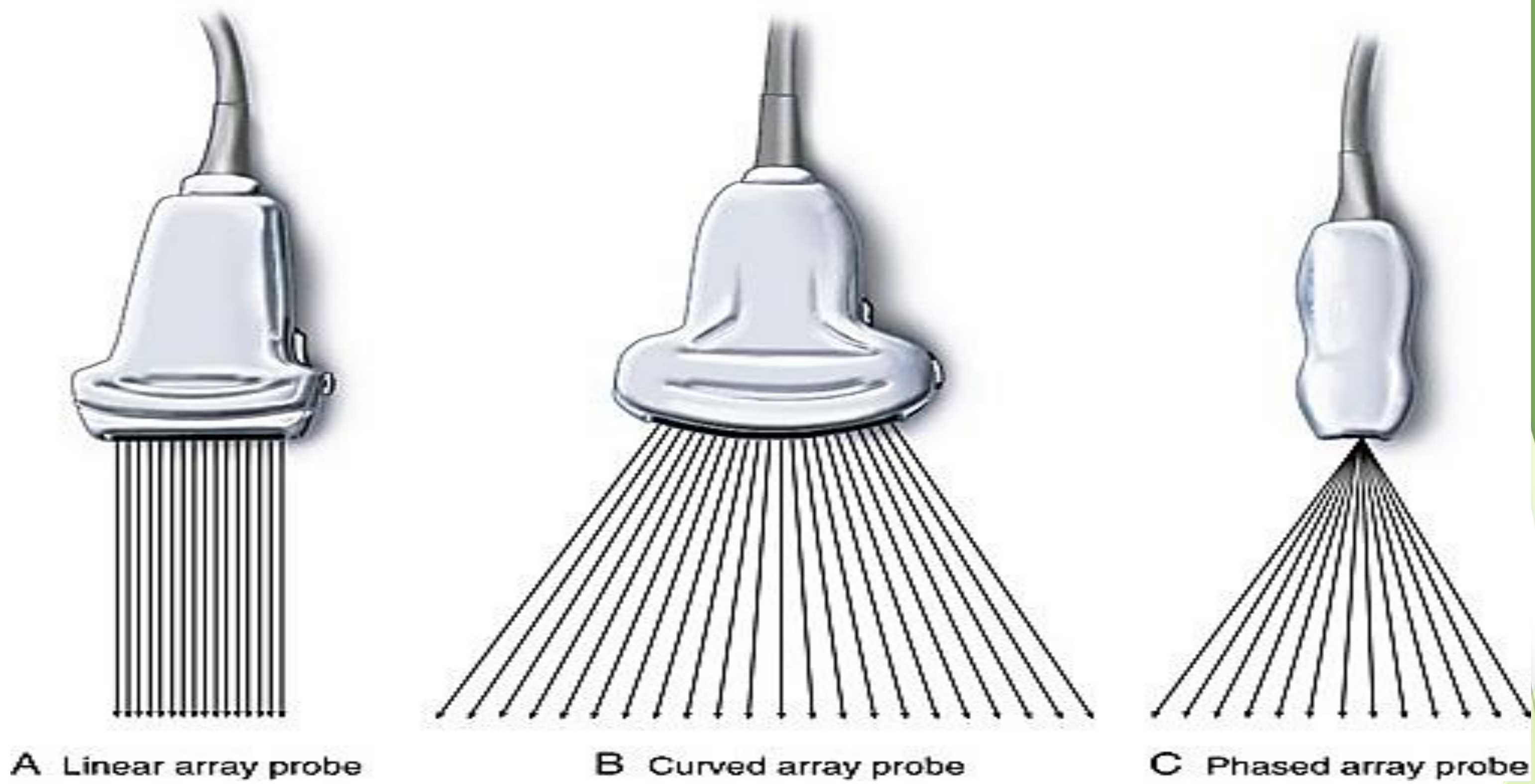




ADVANTAGES:

- *improve penetration and resolution*
- *reduce side lobe artifacts*
- *improve signal to noise ratio*
- *improve near field and far field image quality*
- *better clarity of structures and visualization of lesions*

Transducers Beam Geometry



-wide footprint and keep same field of view at deep part
-vascular application
-bandwidth usually 2.5MHz-12MHz

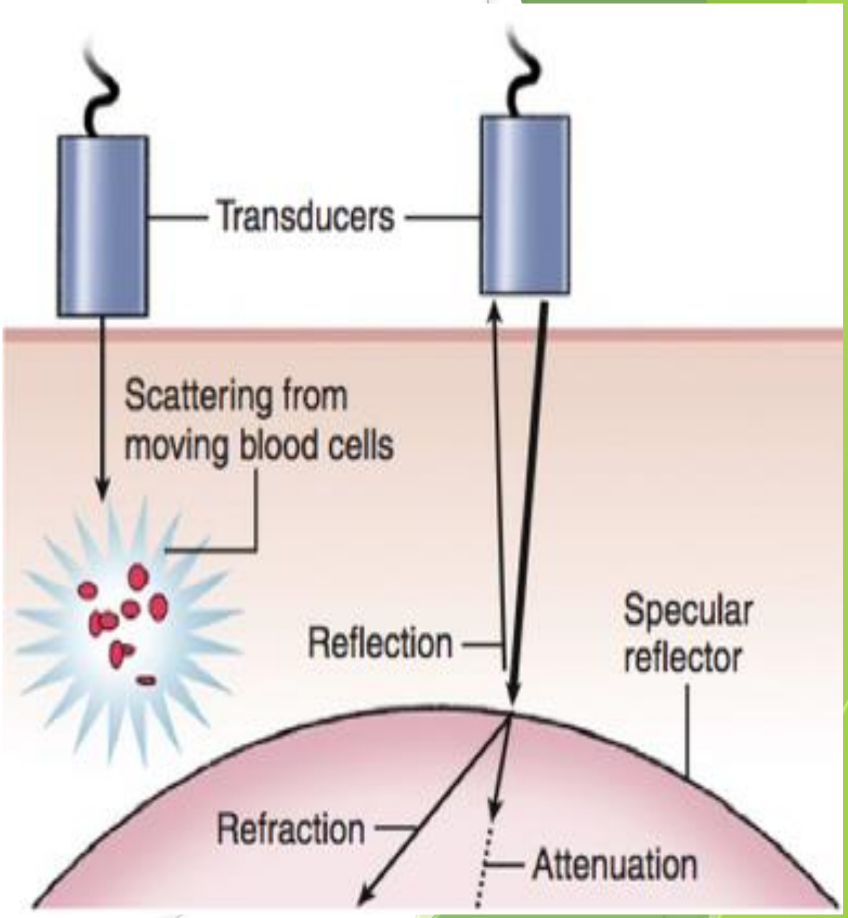
-wide footprint, field of view will be spreaded at deep part
-abdominal application
-bandwidth usually 2.5-7.5MHz

-small footprint, field of view will be spreaded widely at deep part
-cardiac application
-bandwidth usually 2-8MHz

Ultrasound-Tissue Interaction

SOUND
- travels the body tissues at certain speed
- heart tissue = 1540m/sec (Propagation velocity)
- interacts with human soft tissues in several predictable ways

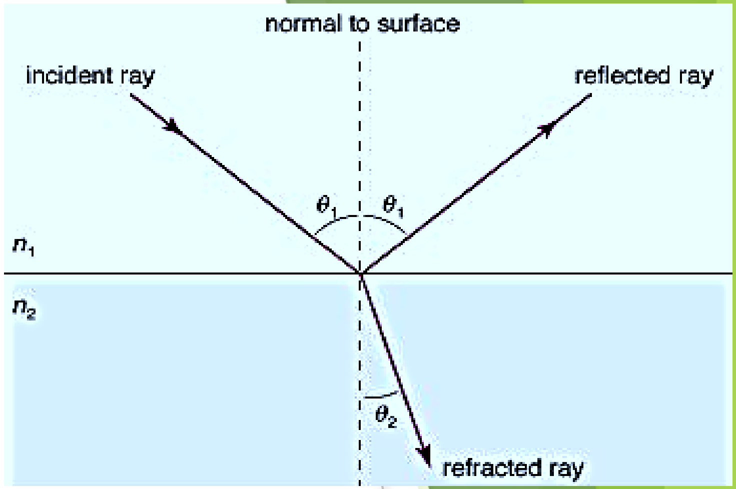
1. **SCATTERING** -hits small irregular objects and go in different direction, i.e. blood cells
2. **REFLECTION** -hits stationary objects and reflected back to the transducer and create a signal
3. **REFRACTION** -sound is bent as it goes through specular reflector, i.e. straw suspended in a glass of water
4. **ATTENUATION** -loss of energy or intensity through absorption as the sound travels through the tissue



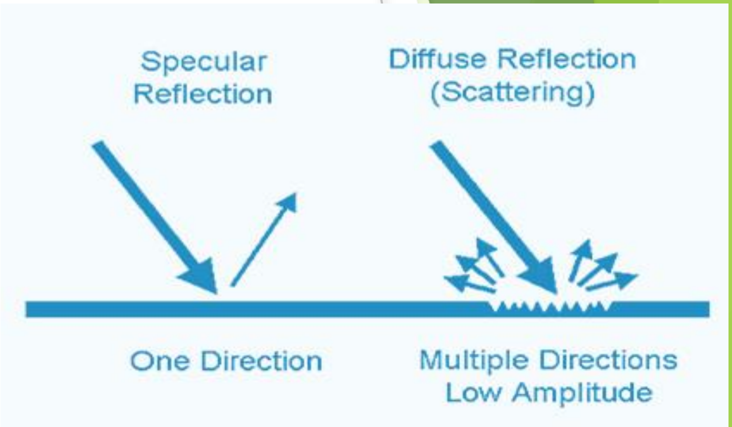
VISUALISATION OF STRUCTURES DEPEND ON HOW MUCH LIGHT IS REFLECTED AND TRANSMITTED IN ACOUSTIC INTERFACES, i.e. MYOCARDIUM, VALVES, ETC.

Ultrasound-Tissue Interaction

ANGLE OF INCIDENCE - *major determinant of reflection*
- *an ultrasound wave hitting a smooth mirror like interface at a 90 degree angle will result in a perpendicular reflection, less than 90 degrees will result in the wave being deflected away from the transducer at an angle equal to the angle of incidence but in the opposite direction (angle of reflection)*

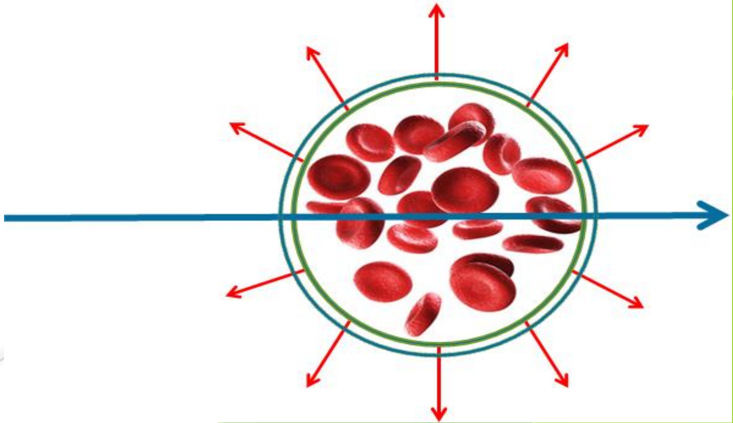


***Snell's Law** - *predicts the angle at which a light ray will bend, or refract, as it passes from one medium to another*



SPECULAR REFLECTORS - *relatively large objects, smooth walls*

RAYLEIGH SCATTERERS - *extremely small non specular reflector whose dimension are much less than that of the beams wavelength*
- *used in Tissue Doppler Speckle and Doppler imaging*



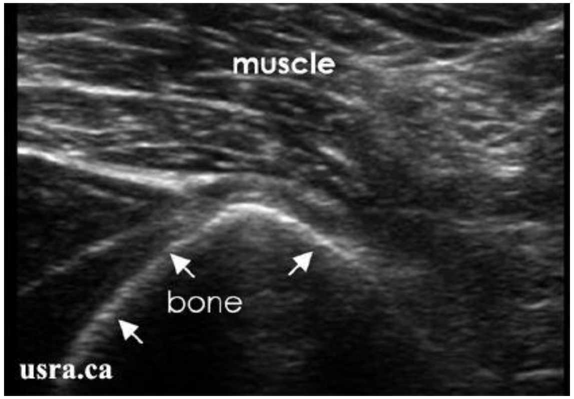
Ultrasound-Tissue Interaction

ACOUSTIC IMPEDANCE (Z) - *a physical property of tissue (important tissue property in imaging)*
- *product of the tissue's density and sound velocity within the tissue*
- *amplitude of returning echo is proportional to the difference in acoustic impedance between two tissues*
- *RAYL (g/cm2 x 10-5)*

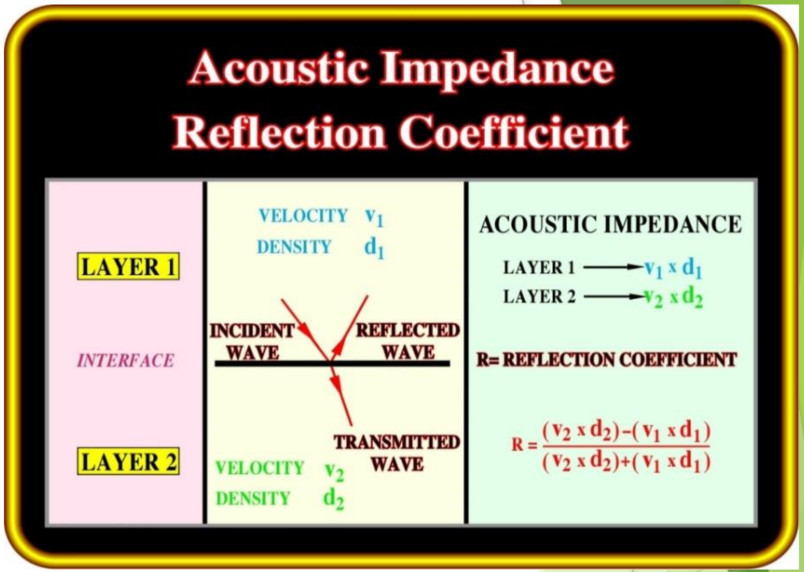
Remember....

FORMULA:
Z= DV

D = density of tissue (g/cm3)
V = propagation velocity (cm/sec)



The velocity of ultrasound in bone is 4080 m/s, in contrast to muscle where it is 1568 m/s. (high acoustic impedance of bone attenuates the energy carried in the ultrasound signal)



Reflection coefficient = ratio of intensity of reflected echo versus intensity of incident beam at the boundary

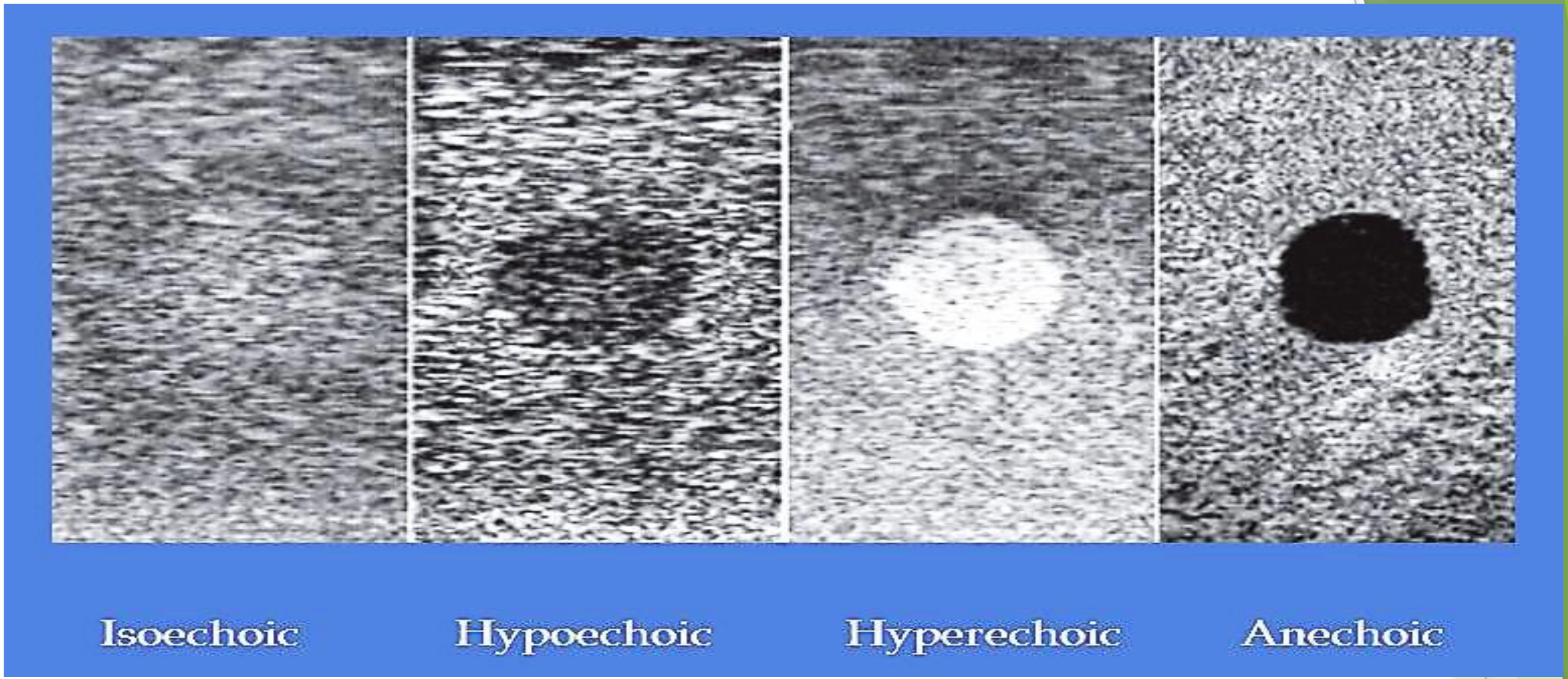
ACOUSTIC IMPEDANCE MISMATCH - *beam is reflected or absorbed when beam encounters two regions of very different acoustic impedances*
- *i.e. soft tissue – bone interface*

Remember....

FORMULA:

$$R = \left(\frac{z_2 - z_1}{z_2 + z_1}\right)^2$$

Ultrasound-Tissue Interaction



ECHOIC OR HYPERECHOIC (bright)
- *large reflection component*
- *waves returning to transducer*

ANECHOIC OR HYPOECHOIC (dark)
- *large attenuation component*
- *waves not returning to transducer*

MIXED ECHOGENICITY

Ultrasound-Tissue Interaction

4. ABSORPTION -removal of energy from the ultrasound beam
-eventual dissipation of energy as heat

5. DIFFRACTION -bending of the waves when encountering obstacles forming 2 beam patterns

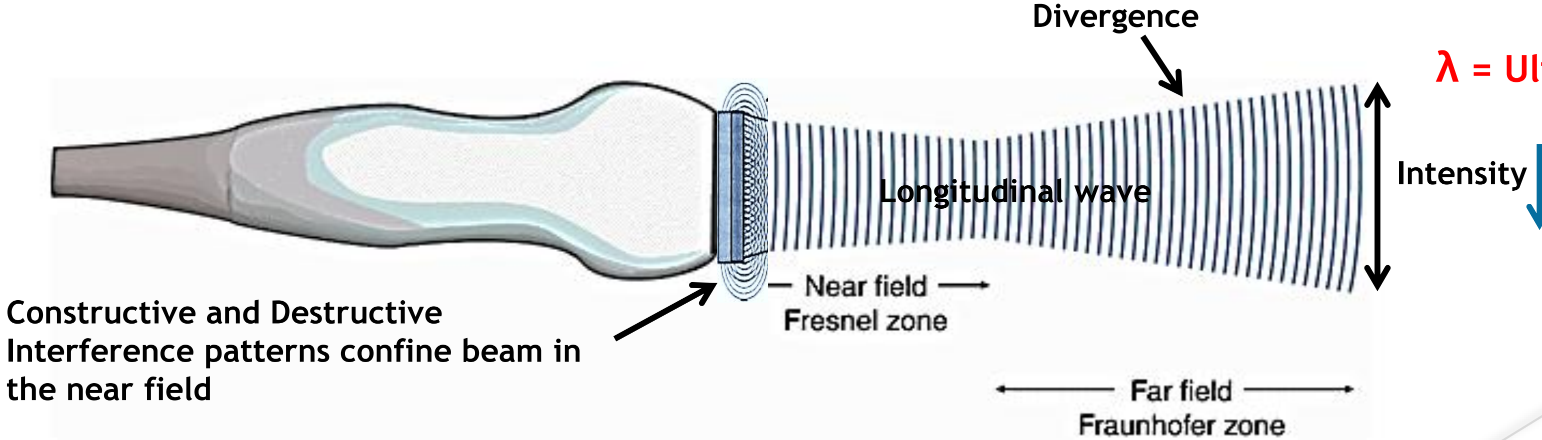
FRESNEL ZONE - *near field*
- *slightly converging beam out to a distance specified by the geometry and frequency of the transducer*
- *Fresnel (near-field) length is directly proportional to aperture of the transducer element and inversely proportional to transducer frequency*

FRAUNHOFER ZONE - *far field*
- *diverging beam beyond that point*

FORMULA:

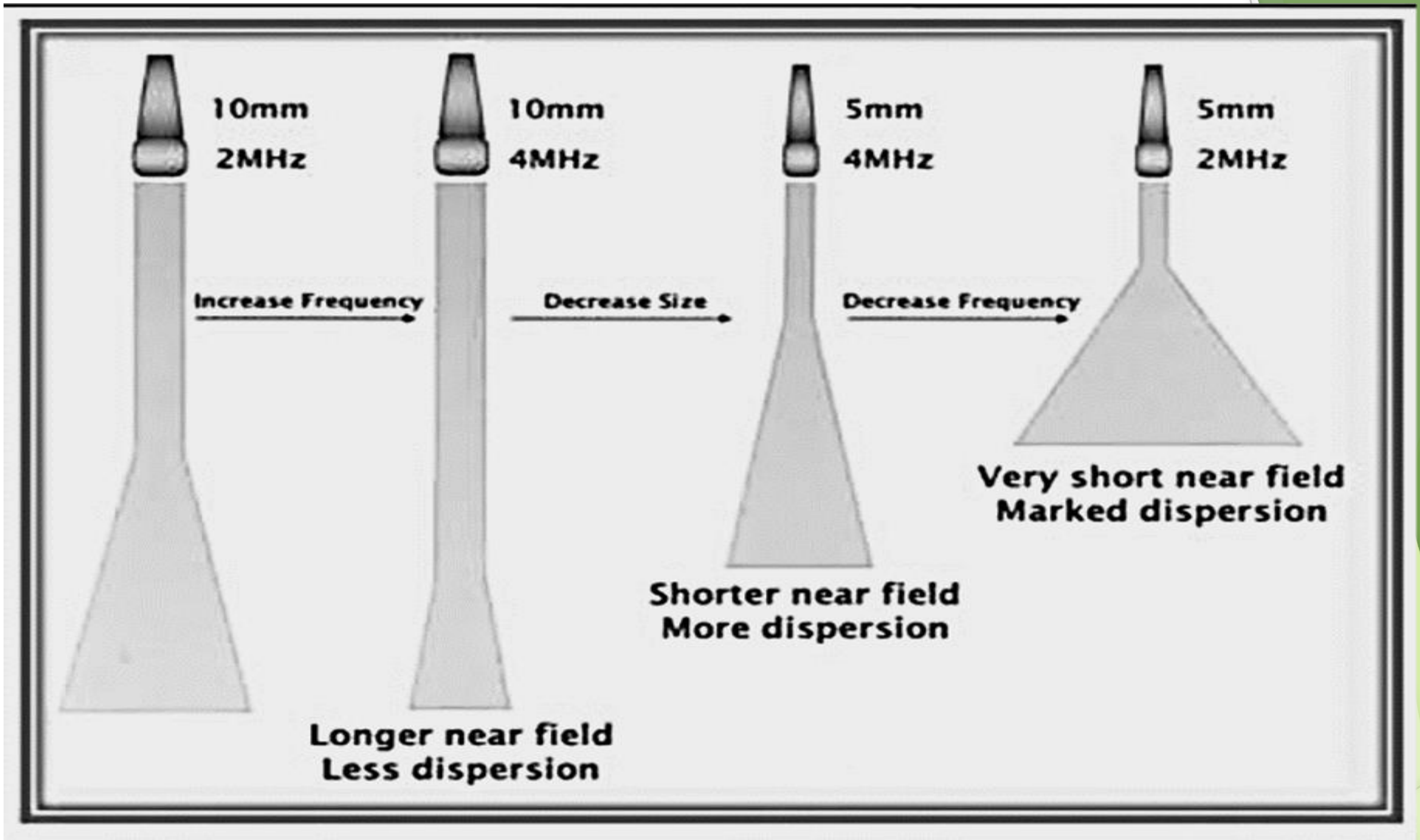
$$D_{fresnel} = \frac{D2}{4\lambda}$$

$D_{fresnel}$ = Fresnel length
D = Diameter or aperture of transducer
λ = Ultrasound wavelength



CHALLENGES IN LENGTHENING THE NEAR FIELD:
1. Transducer size limited by intercostal space
2. Higher wavelength results in greater attenuation
3. As Frequency increases, and Fresnel zone increases, so is greater absorption and side distortion
4. Far field decrease intensity and increase attenuation

Ultrasound-Tissue Interaction

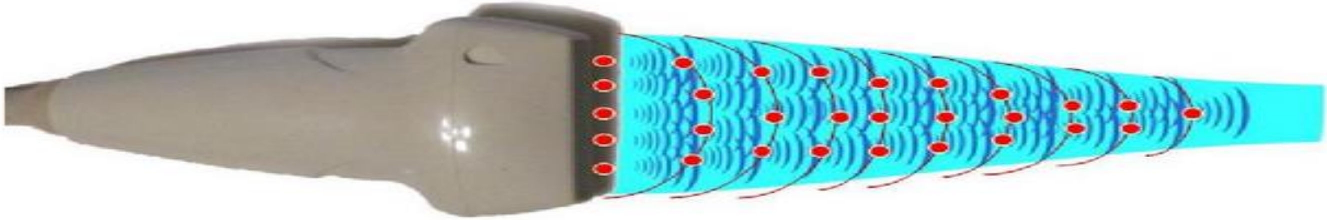


The length of the NEAR FIELD is determined by:

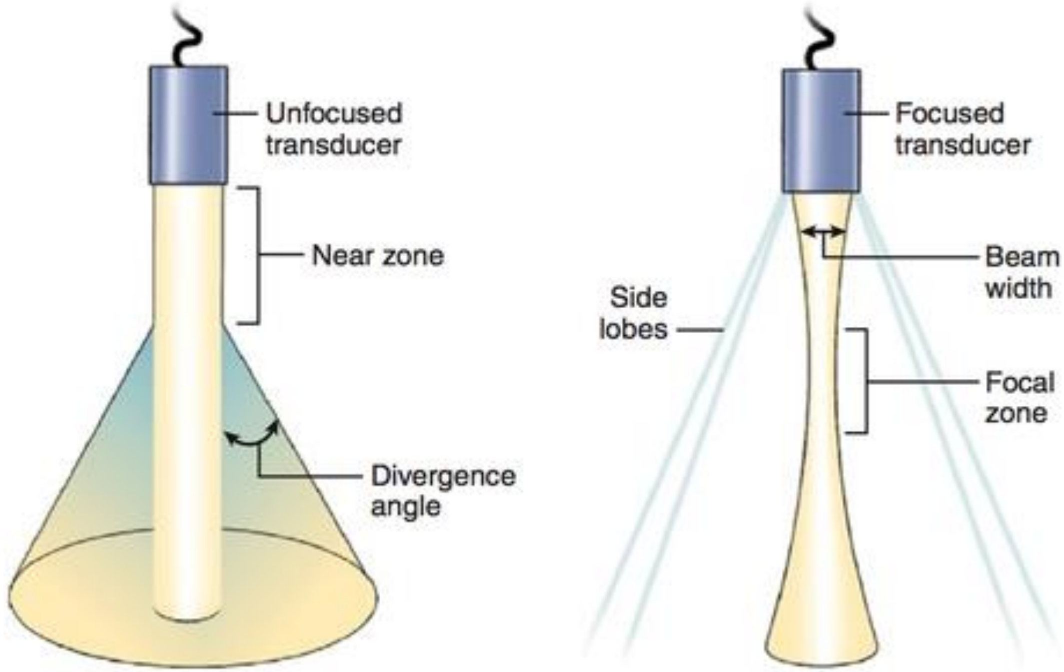
- ***RADIUS OF THE TRANSDUCER FACE (APERTURE)***
- ***WAVELENGTH***
- ***FREQUENCY OF THE TRANSMITTED ENERGY***

Ultrasound-Tissue Interaction

HUYGEN'S PRINCIPLE - sound waves produced by ultrasound transducers originate as numerous points on the surface of a piezoelectric element (each point is a source of small individual sound wavelets)



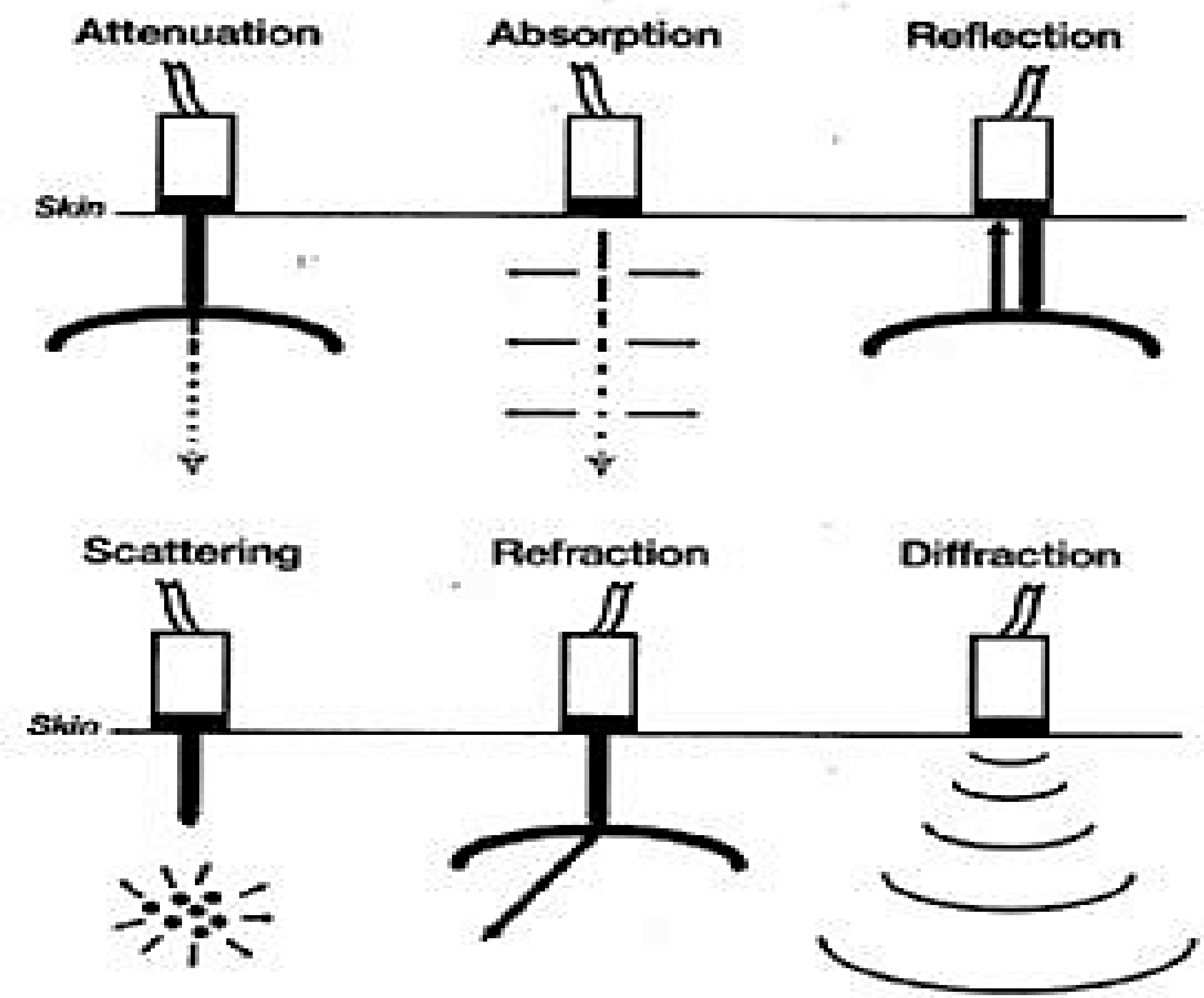
Comparison between an unfocused and focused ultrasound beam



- focused transducer uses a curved acoustic lens to decrease beam diameter at specified distance to transducer
- length of the near zone is a function of the frequency
- focal zone is found between the near and the far zone
- focused transducer on the right is a more accurate representation of ultrasound beam geometry caused by constructive and destructive interference from neighbouring sound waves
- side lobes contribute to blurring of the image

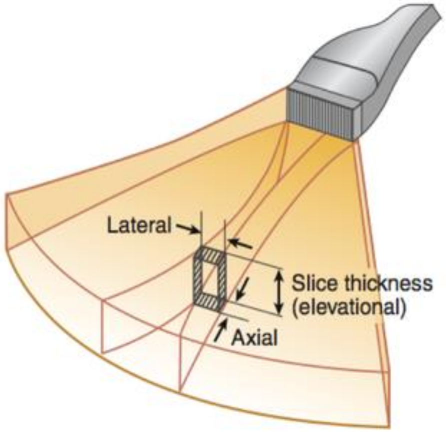
Ultrasound-Tissue Interaction

SUMMARY



RESOLUTION

1. SPATIAL -the ability to distinctively display two closely spaced reflectors in tissue



Axial - also known as linear, range, longitudinal or depth resolution
- is dependent on FREQUENCY/WAVELENGTH, DYNAMIC RANGE, PULSE LENGTH

Lateral - also known as azimuthal resolution
- is dependent on FOCAL DEPTH, BEAM WIDTH, GAIN, PROBE DIAMETER OR APERTURE WIDTH

Slice Thickness - also known as elevation resolution
- is dependent on the TRANSDUCER ELEMENT HEIGHT

2. TEMPORAL -ability to display in real time, events that are closely spaced in time
-depends on *FRAME RATE, IMAGE DEPTH, SECTOR WIDTH, SWEEP ANGLE, LINE DENSITY, PULSE REPITITION FREQUENCY*

3. CONTRAST -the ability to distinguish between different echo amplitudes of adjacent structures
-depends on various stages in imaging process including COMPRESSION, IMAGE MEMORY, the use of CONTRAST AGENTS and DYNAMIC RANGE

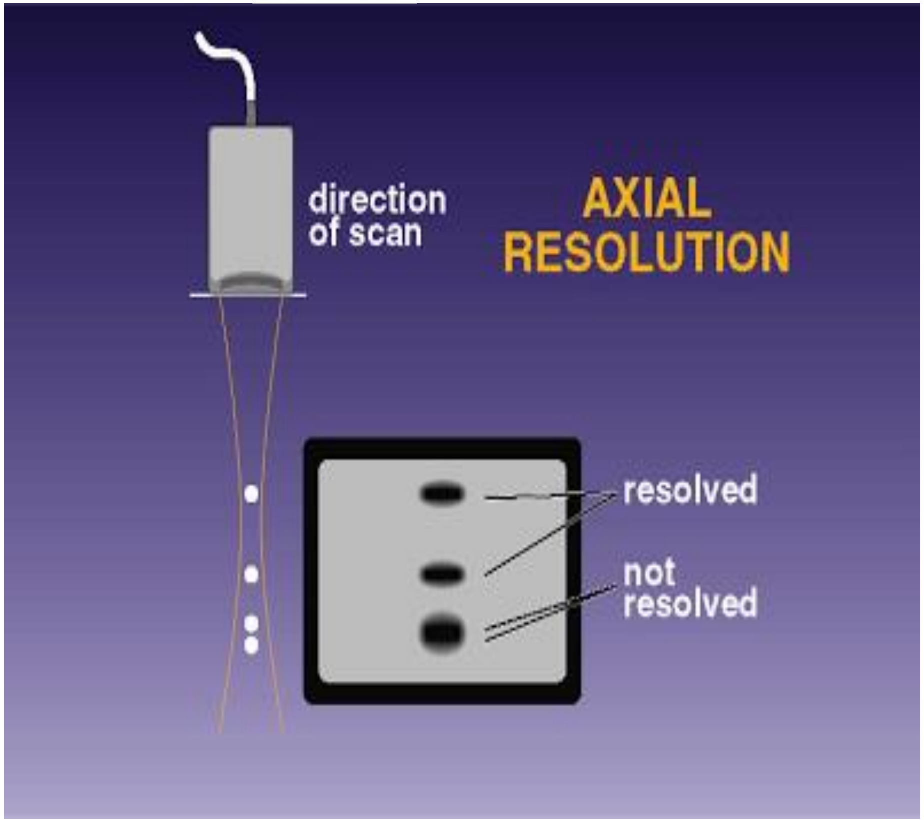
RESOLUTION

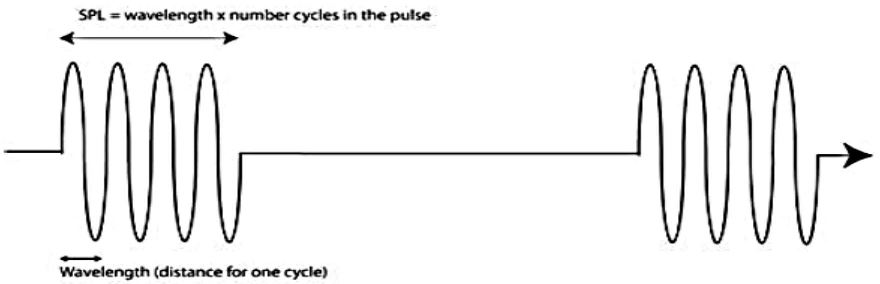
SPACIAL RESOLUTION

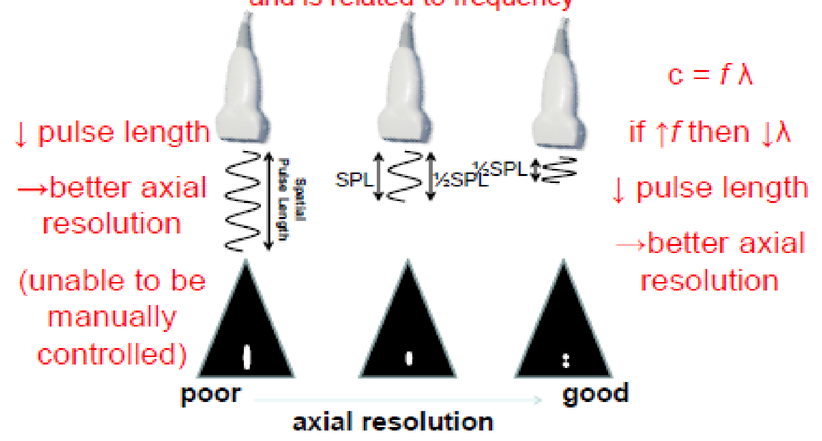
AXIAL - **ability to differentiate objects that are parallel the imaging beam axis, one-half spatial pulse length ($\frac{1}{2}$ SPL) to avoid overlap of returning echoes**

- **dependent on frequencies/wavelength**
- **if object is smaller than the wavelength, scattering occurs**

Remember.... SPATIAL PULSE LENGTH = WAVELENGTH x NO. OF CYCLES



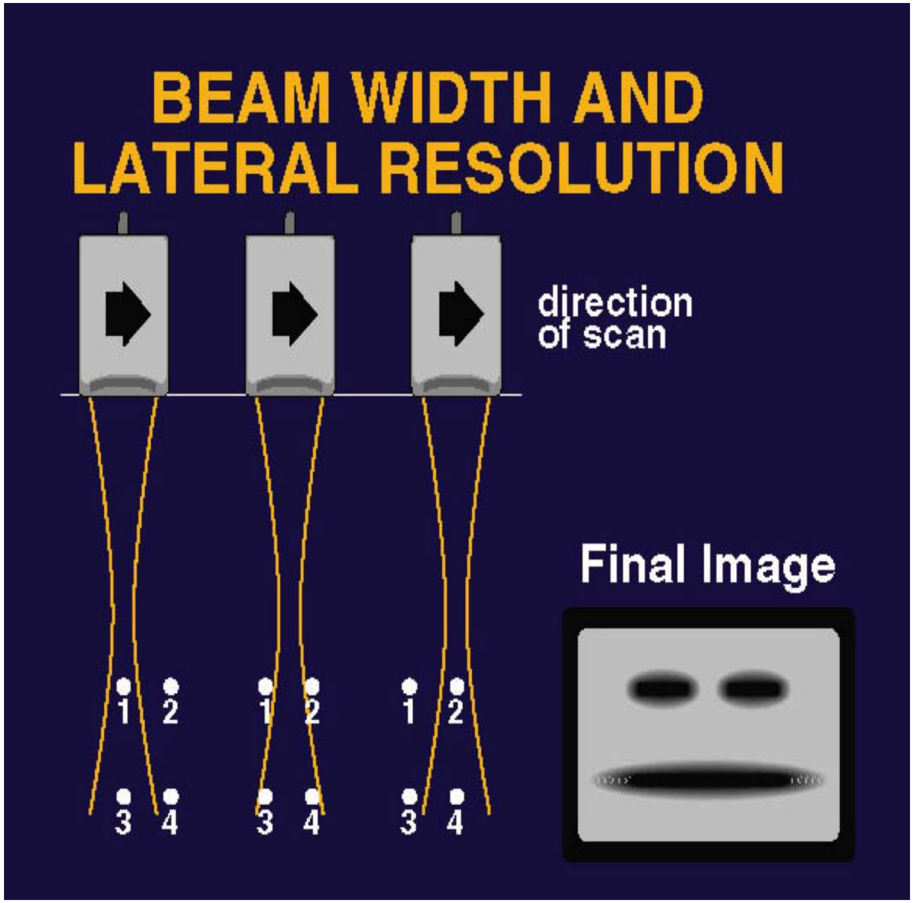


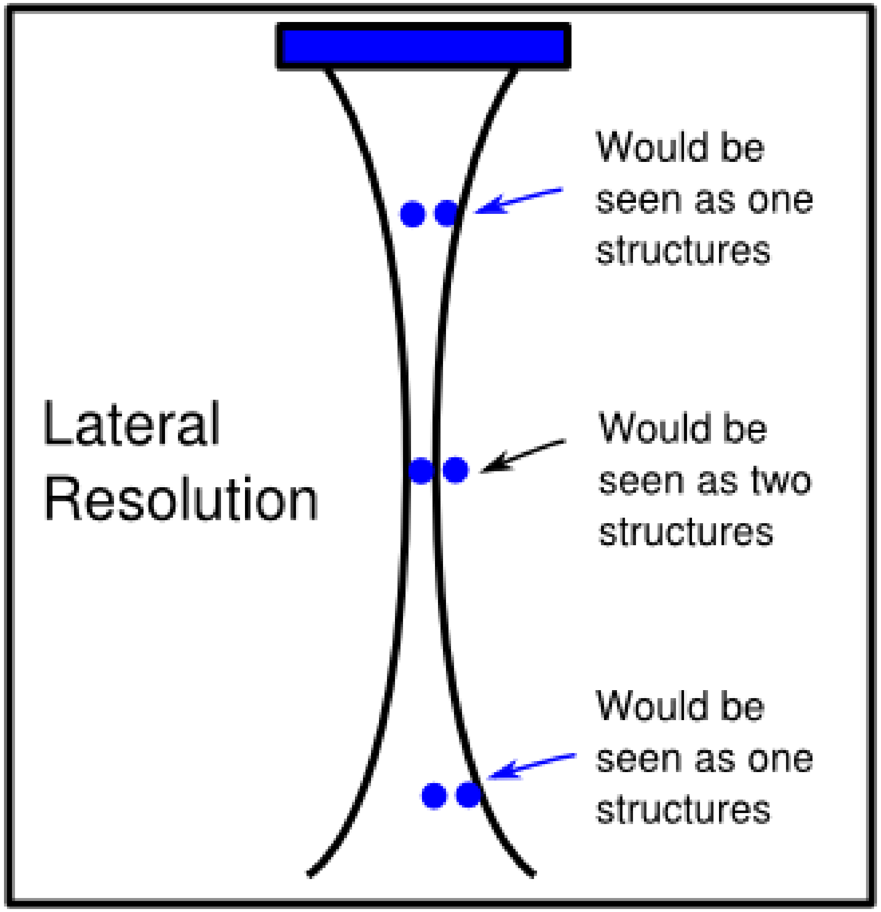


RB&HH SPECIALIST CARE

RESOLUTION

LATERAL - ability to differentiate objects that are located side to side or perpendicular to the beam axis
- dependent on focal depth
- effective beam diameter is approximately equal to half the transducer diameter

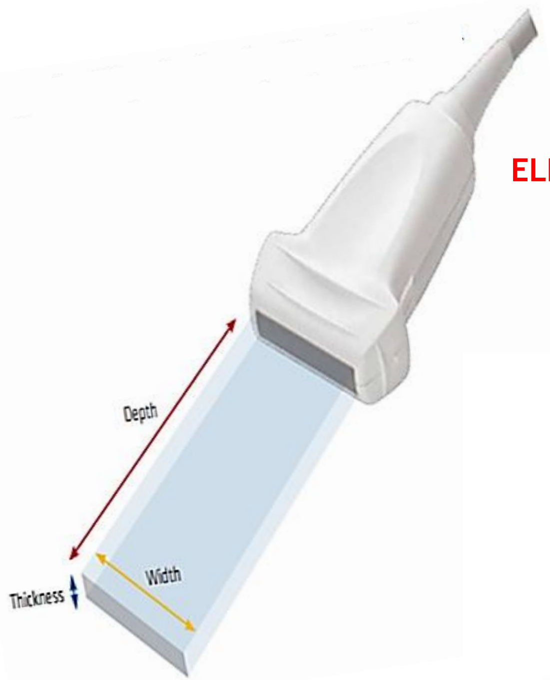




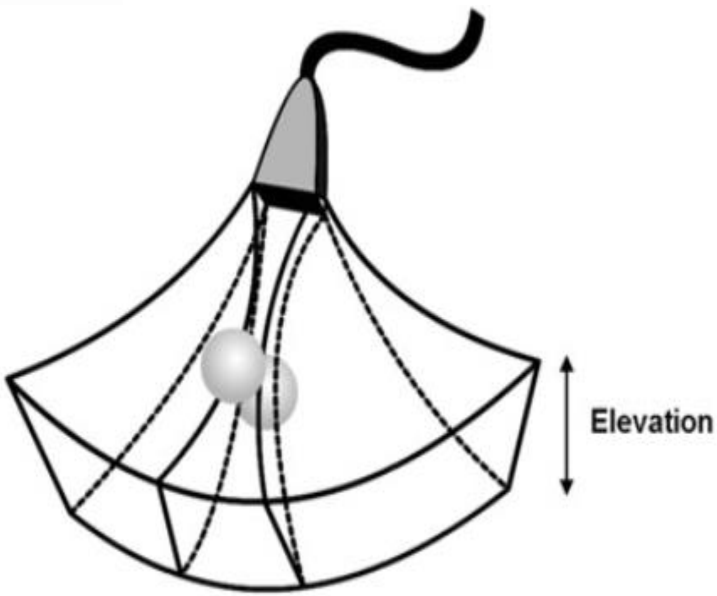
RB&HH SPECIALIST CARE

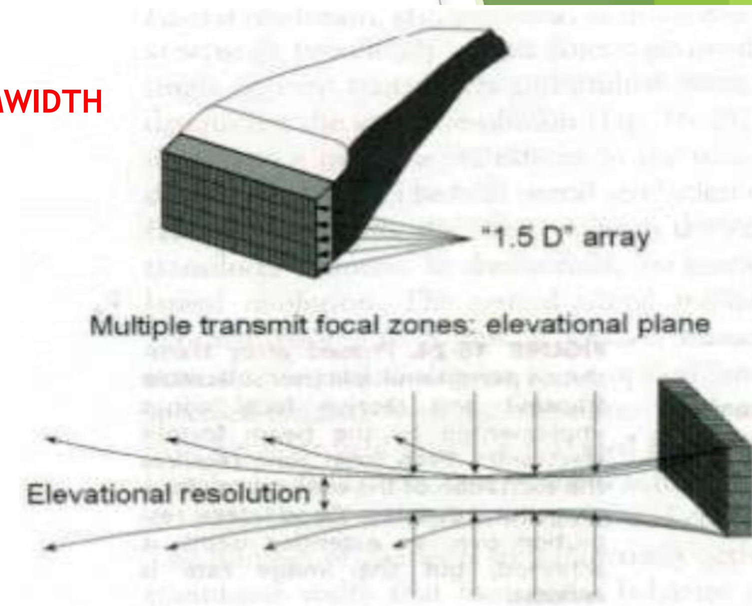
RESOLUTION

SLICE THICKNESS
- refers to thickness of the beam or elevation beamwidth
- plays a role in signal averaging of acoustic details in the regions close to the transducer and in the far field beyond the focal zone
- is dependent on the transducer element height or thickness



ELEVATION RESOLUTION = ELEVATION BEAMWIDTH

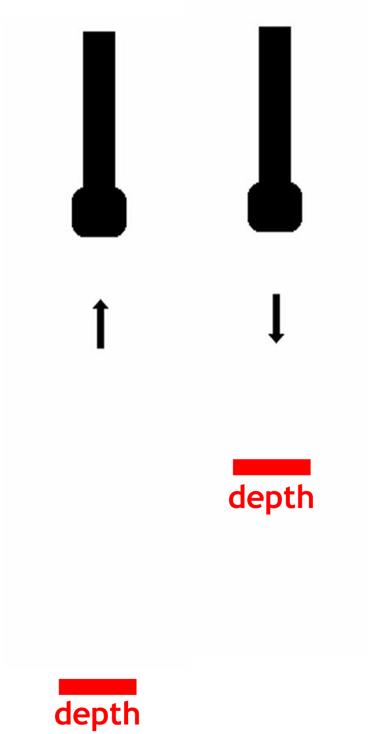


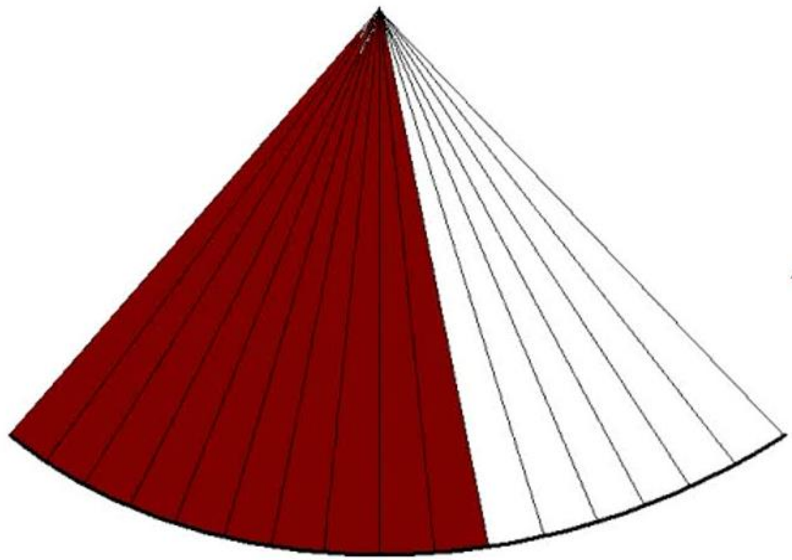


RB&HH SPECIALIST CARE

RESOLUTION

TEMPORAL RESOLUTION -ability to display in real time, events that are closely spaced in time
-the time from the beginning of one frame to the next
-represents the ability of the ultrasound system to distinguish between instantaneous events of rapidly moving structures
-depends on frame rate, image depth, sector width

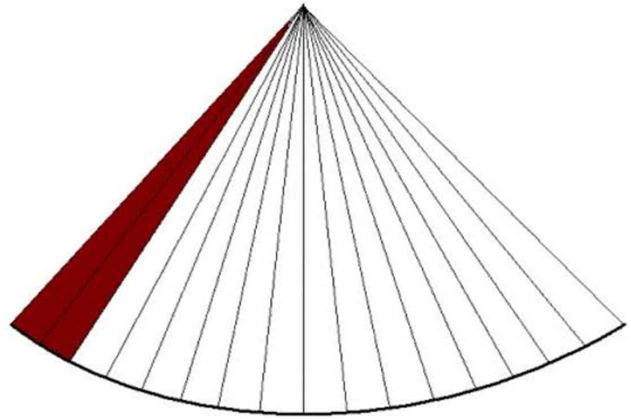




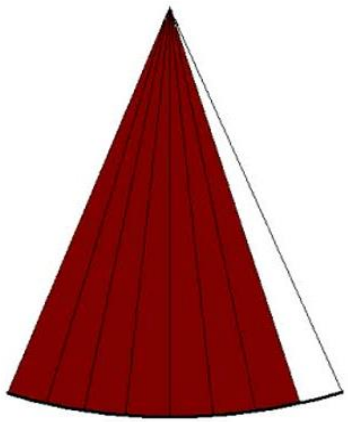
Increased Depth =
Lower Frame Rate

Decreased Depth =
Higher Frame Rate

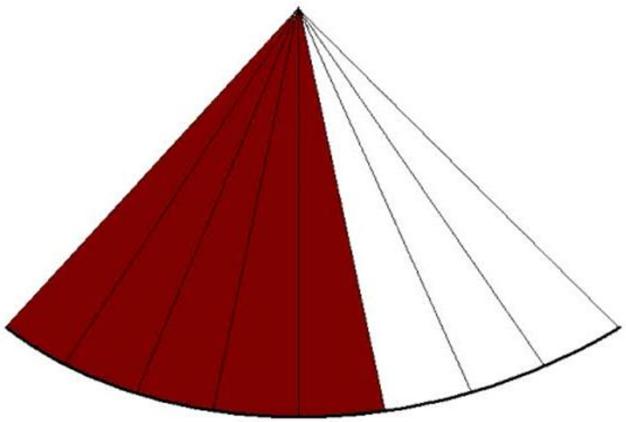
RESOLUTION



Sector width with a given depth and line density determines the frame rate

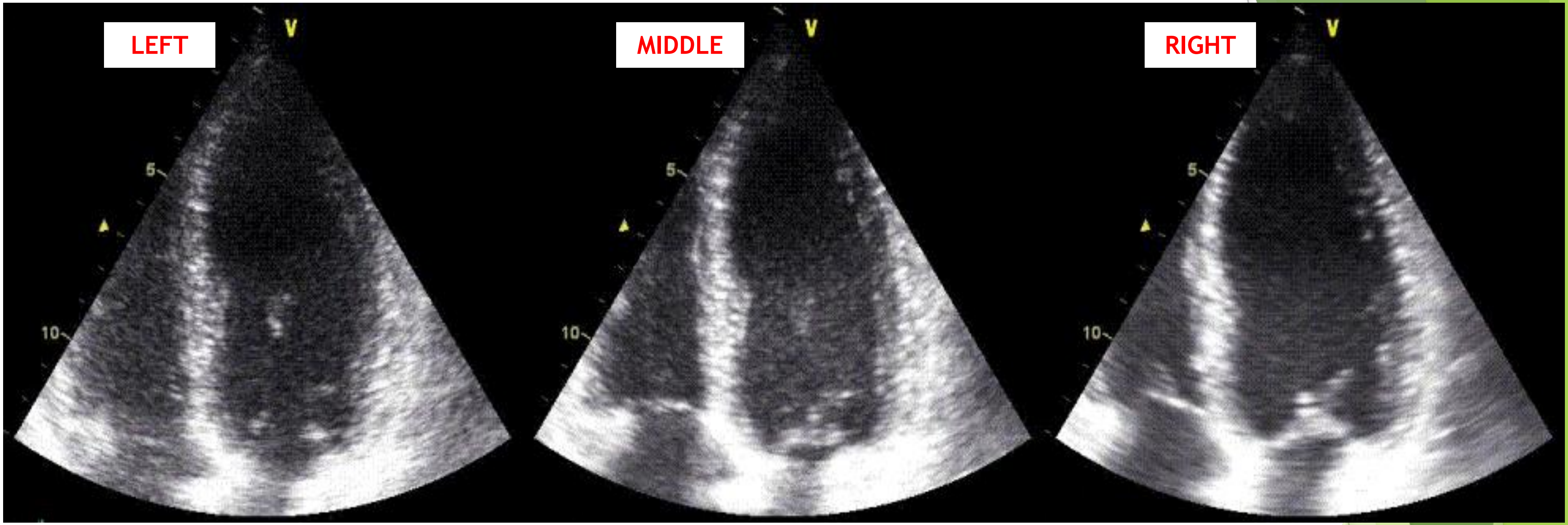


Sector width reduced, but maintaining line density = unchanged lateral resolution with higher frame rate



Reducing line density and maintaining sector width = decreased lateral resolution with high frame rate

RESOLUTION



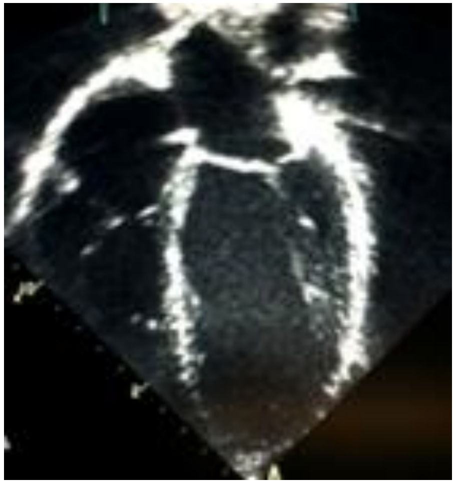
Same ventricle acquired at different Frame Rate (FR): 34 (left), 56 (middle), 112 (right). Frame Rate was increased by reducing line density, all other settings being equal. Right image: poor lateral resolution (lateral blurring/smearing). Left image: redundant and more grainy. Middle: appears the best quality. NOTE: IN DISTAL STRUCTURES (ATRIAL WALL AND MITRAL VALVE), PROMINENT SMEARING OCCURS (DUE TO DIVERGENCE) ALSO, DUE TO POOR LATERAL RESOLUTION, ENDOCARDIAL DEFINITION IS LOST

RB&HH SPECIALIST CARE

RESOLUTION

CONTRAST RESOLUTION -the ability to distinguish between different echo amplitudes of adjacent structures
-depends on various stages in imaging process, i.e. compression, dynamic range, image memory and contrast agents

COMPRESSION -occurs in the signal processor which reduces the dynamic range (ratio of the highest power to the lowest power)
- assigning stronger echo powers to maximum and weaker echo powers to zero
- high compression with a narrow dynamic range (e.g. 30 decibels) creates an image of high contrast, low compression with wide dynamic range (e.g. 60 decibels) displays an image of low contrast and with many shades of grey



LOW COMPRESSION

HIGH COMPRESSION

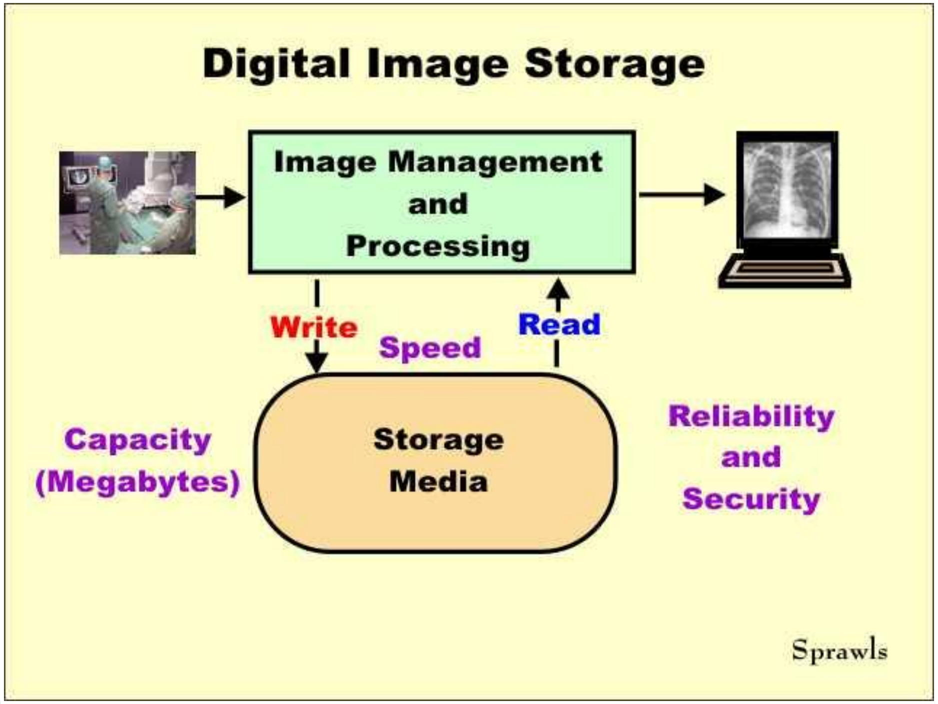
↑COMPRESSION = ↓DYNAMIC RANGE = ↑CONTRAST

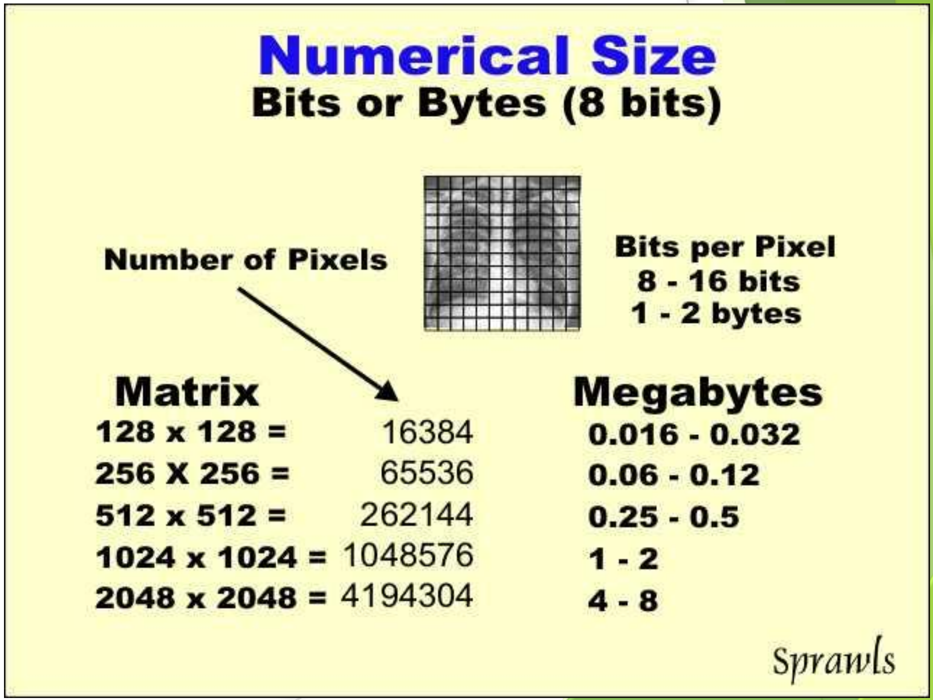
↓DYNAMIC RANGE = ↓LESS SHADES OF GREY
= ↓SPACIAL RESOLUTION

RESOLUTION

IMAGE MEMORY

- where storage of digitized information contained in the pulse waveforms occurs
- each part of the image memory called a pixel (picture element) must have as many layers of bits (binary digits) as possible to enable various shades of grey be visualise
- capacity (number of images that can be stored), speed (time required to write/record and read/retrieve images), reliability and security (to prevent loss of images)

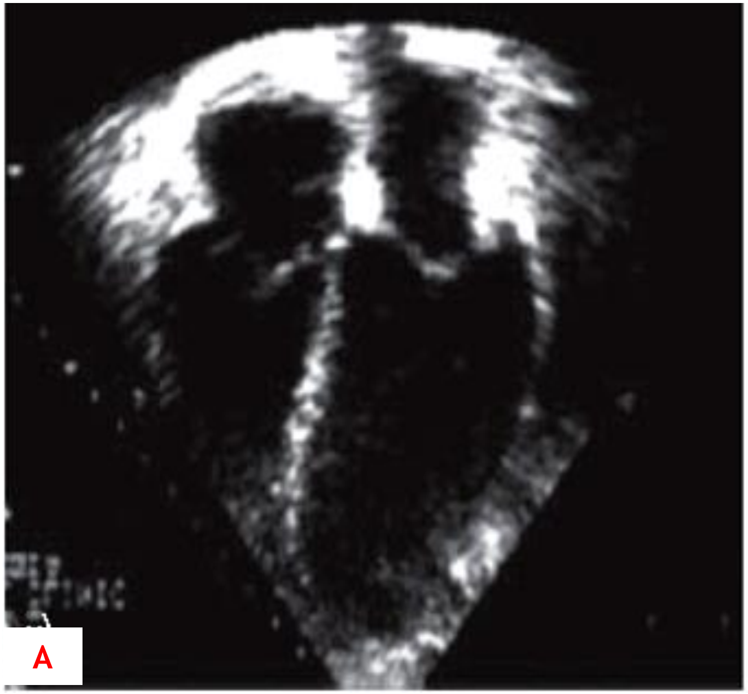




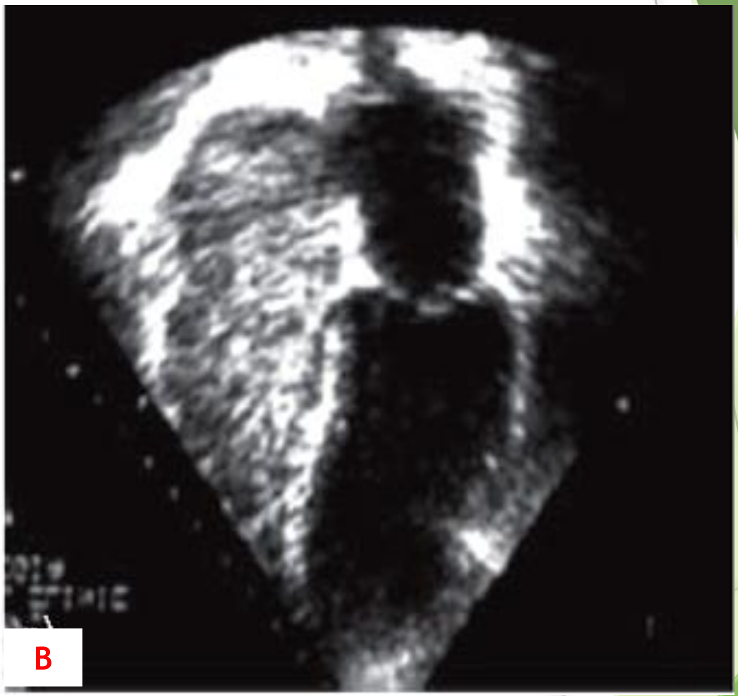
RESOLUTION

CONTRAST AGENTS -are used when conventional ultrasound imaging does not provide sufficient distinction between myocardial tissue and blood

-agitated saline as contrast agent



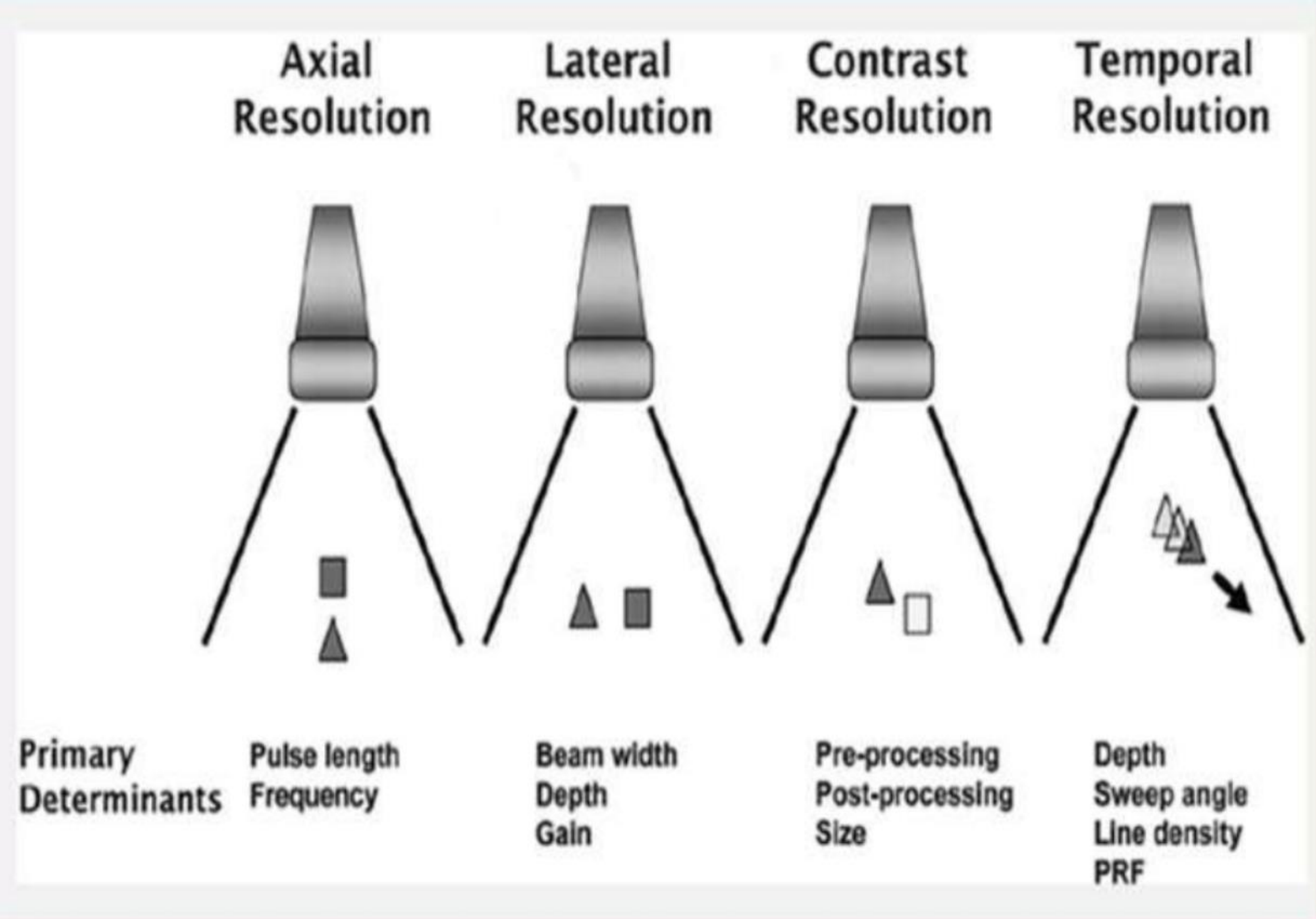
NORMAL HEART CHAMBERS



OPACIFICATION OF THE RIGHT HEART CHAMBERS

RESOLUTION

SUMMARY



RB&HH SPECIALIST CARE

IMAGE OPTIMIZATION

POWER -adjusts the amount of acoustic power transmitted by the transducer
-acoustic power = acoustic energy/time (ultrasound can produce heat)
-adjust power control to highest power level within thermal limits (mechanical index of approx. 0.3)

FOCUS -optimizes ultrasound intensity in near and far field
-affects lateral resolution (thinner beam = improved lateral resolution)
-enables focusing the ultrasound beam at selected distance by altering the sequences of electrical impulses sent to transducer elements by phased array transducers

FREQUENCY -denotes transmitted frequency which can be adjusted according to proximity of structures
-can be adjusted in broadband transducers (i.e. RES=highest freq. available, GEN=mid-range frequencies, PEN=lowest freq. range but with good penetration)
-for shallow structures, use higher freq., and deeper structures, lower freq.

SECTOR WIDTH -controls the angle of the sector displayed on the monitor
-wide sector size, lower frame rate
-fast moving structures, narrow sector with increased line density

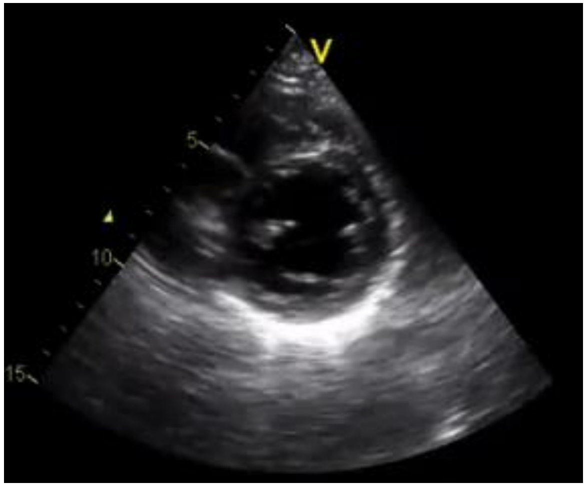
TILT -lateral orientation of the image sector, facilitates exploration of peripheral structures with better axial resolution

FREEZE -allows operator to stop the moving heart display (real time scanning or off-line) then select a single frame of interest in order to measure or acquire

RB&HH SPECIALIST CARE

IMAGE OPTIMIZATION

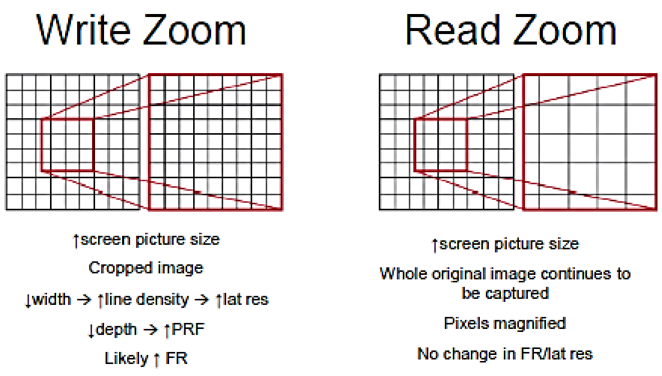
DEPTH -determines how "deeply" into the body one wishes to image
-is depicted along a scale at the side of the sector (in cm) and should always be selected
-influences spatial resolution, pulse repetition frequency (PRF), frame rate



ZOOM -magnify point of interest

WRITE ZOOM
-improves image quality
-only acquired live, re-scans selected smaller area of anatomy
-higher frame rate

READ ZOOM
-magnifies a selected area of pixels from both live or previously recorded image
-not capable of rendering any more structural detail because it simply enlarges the pixels
-frame rate is the same

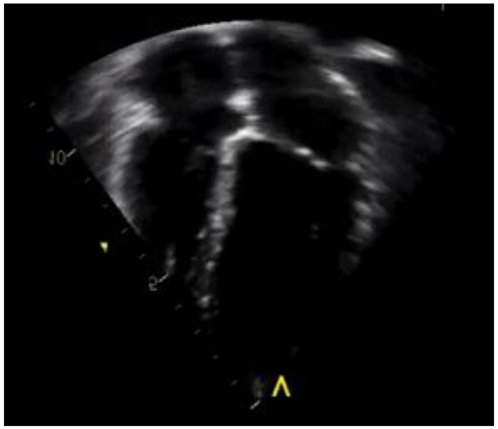


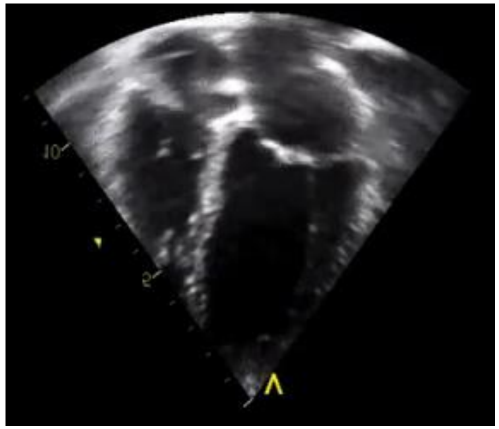
RB&HH SPECIALIST CARE

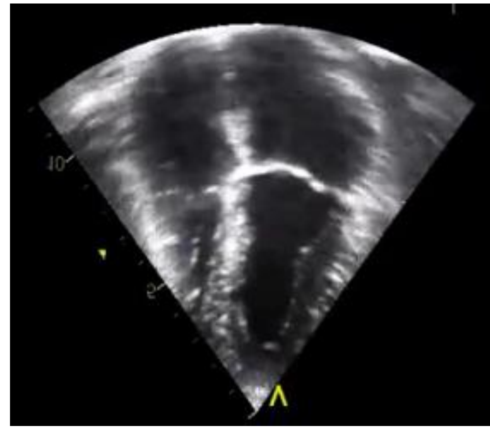
IMAGE OPTIMIZATION

GAIN

-adjust overall brightness by amplifying the return or receive echo signals
-compensates for attenuation
-excessive will result to adding "noise" resulting to the same maximum grayscale value on different shades of grey value reflectors
-if too low, tissue with low reflector intensity will not be seen





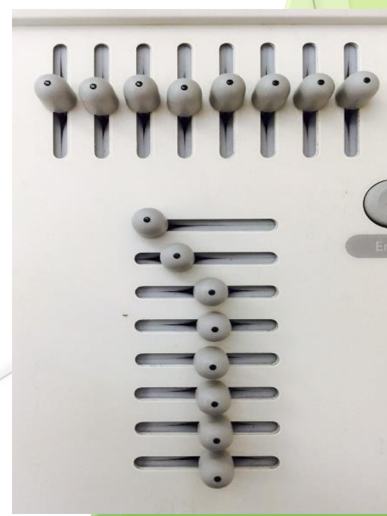


TGC

-TIME GAIN COMPENSION
-compensates differences in echo strength by adjusting amplification of returning signals from depth attenuation

LGC

-LATERAL GAIN COMPENSION
-compensates differences in echo strength by allowing higher amplification of the weaker lateral signal



RB&HH SPECIALIST CARE

IMAGE OPTIMIZATION

DYNAMIC RANGE -ratio between the largest and smallest signal
-displays in the monitor the range of compressed wide spectrum of signals detected as varying shades of grey
-adjusts overall number of shades of grey

GREY MAPS -similar effect on an image as changing dynamic range but different approach
-determines how dark or light each level of white/grey/black based upon the strength of the ultrasound signal

REJECT -eliminates greater number of low intensity signals termed as acoustic noise coming from refraction signals from within the body and electronic noise from the equipment itself or ventilators
-increase reject control eliminate random echoes from low intensity areas

PERSISTENCE -adjusts the updating and averaging of consecutive frames on the screen to reduce noise and too much speckle
-increase persistence will smooth out image but sacrifices crispness of moving structures, decrease will give grainier image
-higher persistence more desirable for slow moving structures, lower persistence for rapidly moving structures

EDGE ENHANCEMENT -improves border delineation enabling more accurate measurements and better visualisation of the endocardium for systolic function and regional wall motion assessment

IMAGE GENERATION

A-MODE - **AMPLITUDE**
- **visualized as spikes**

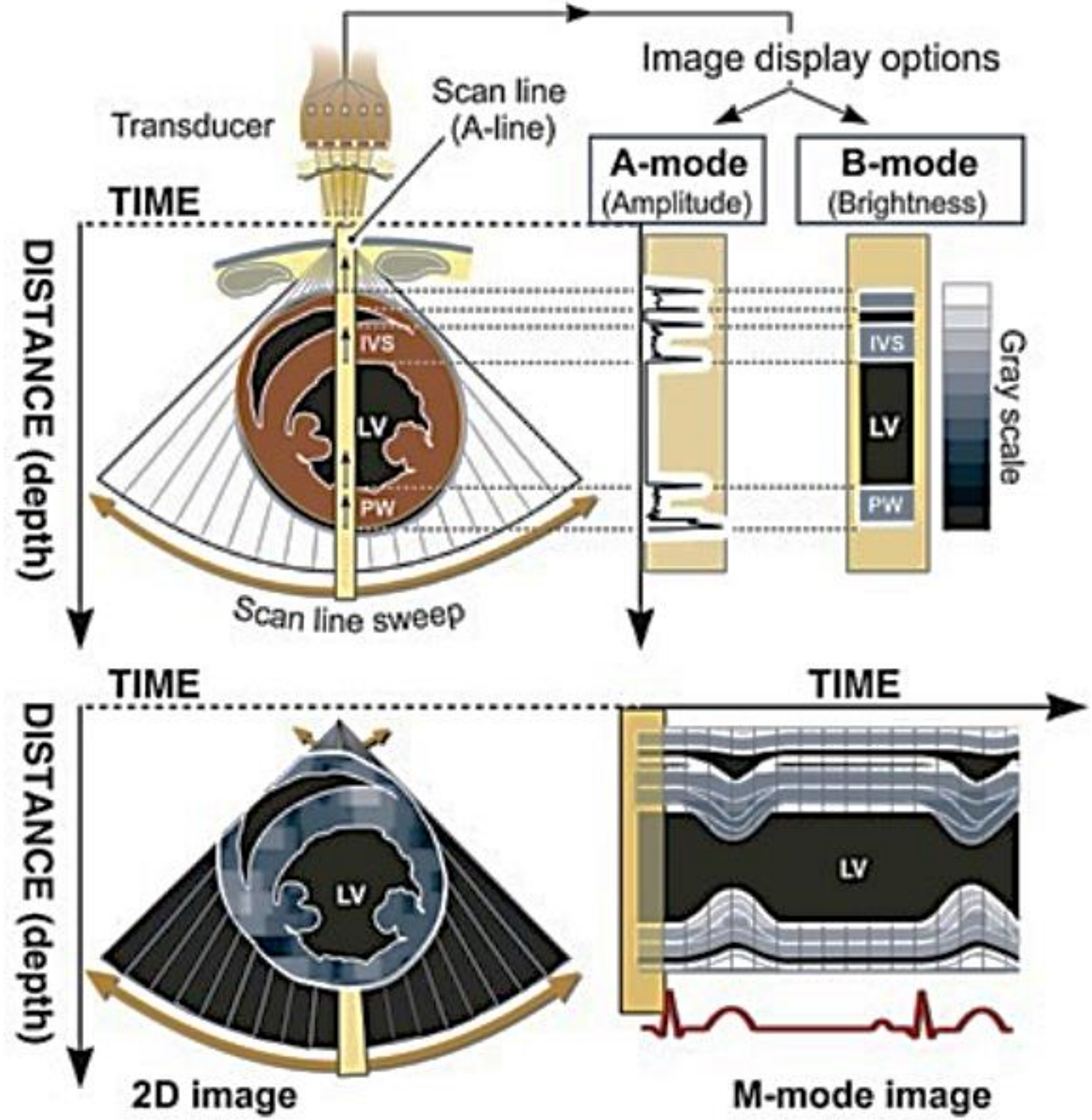
B-MODE - **BRIGHTNESS**
- **visualized as gray scale**

M-MODE - **MOTION**
- **brightness or gray scale over time**

B-MODE = 2D image over time (2D Real-time)
= 1D image over time (M-Mode)

M-MODE = high sampling rate of more than 2000x per second

2D REAL TIME = sampling rate of around 40-80 frames per second

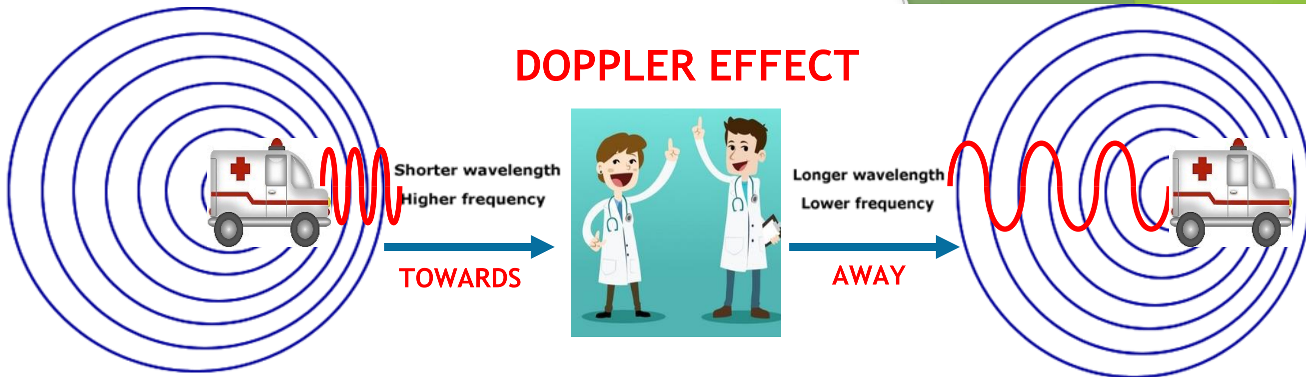


RB&HH SPECIALIST CARE

DOPPLER IMAGING PRINCIPLES

DOPPLER EFFECT

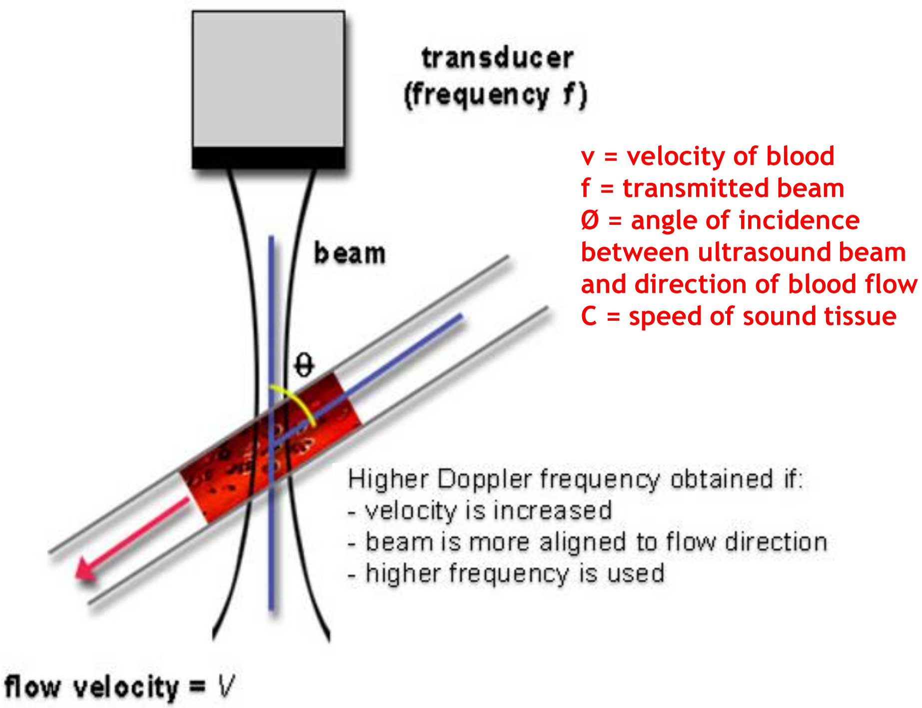
-apparent change in received frequency due to a relative motion between a sound source and sound receiver

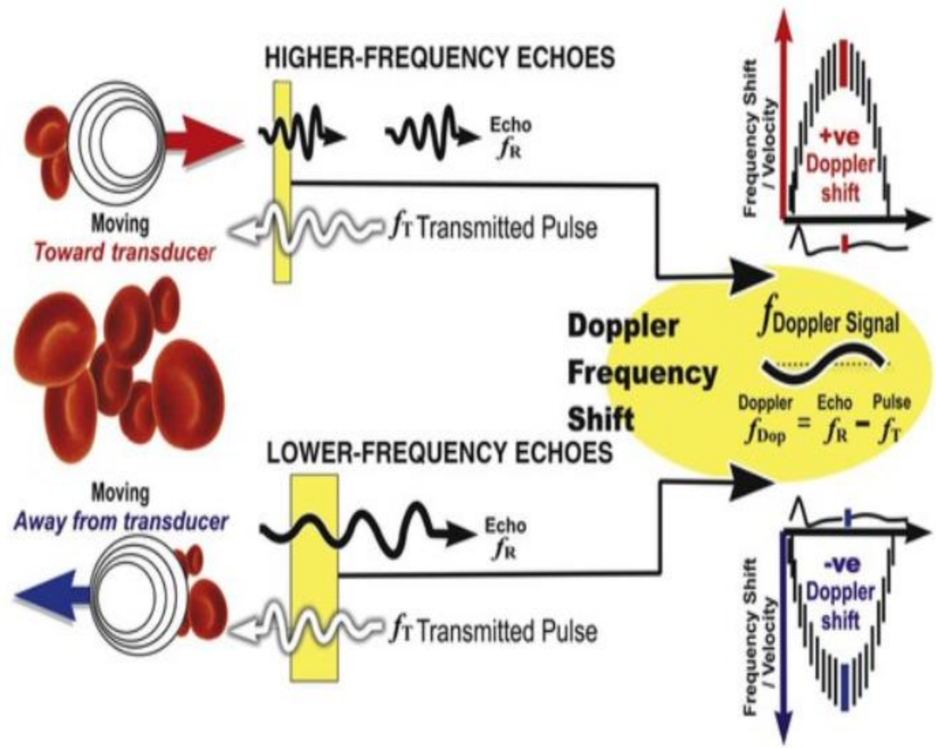


Remember....

FORMULA:

DOPPLER FREQUENCY SHIFT = $\frac{2vf \cos \emptyset}{C}$



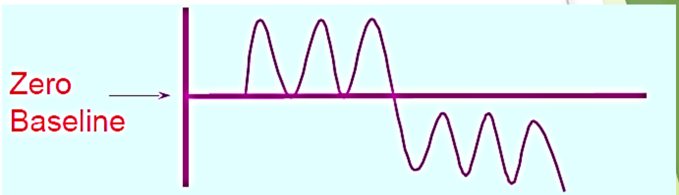


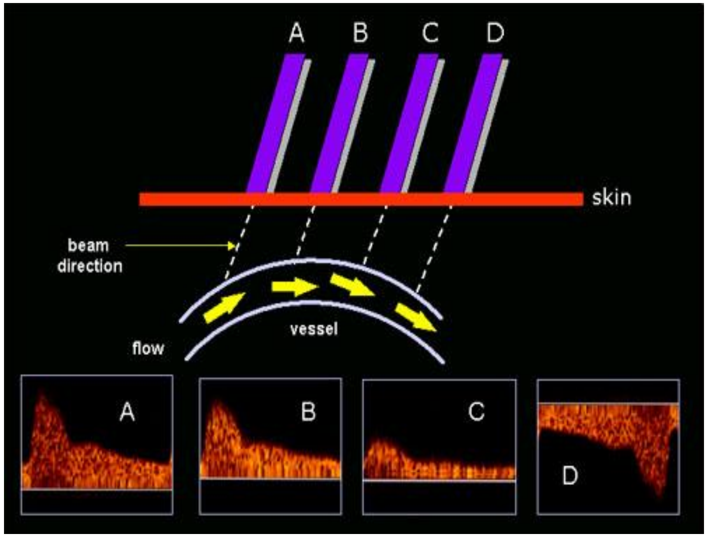
DOPPLER IMAGING PRINCIPLES

DOPPLER - measures the movement of the scatterers through the beam as a phase change in the received signal
- produces a graphical representation of flow (Spectral Waveform)
- resulting Doppler frequency can be used to measure velocity if the beam/flow angle is known
- alignment/parallel (≤60°) with flow is very important for optimal acquisition

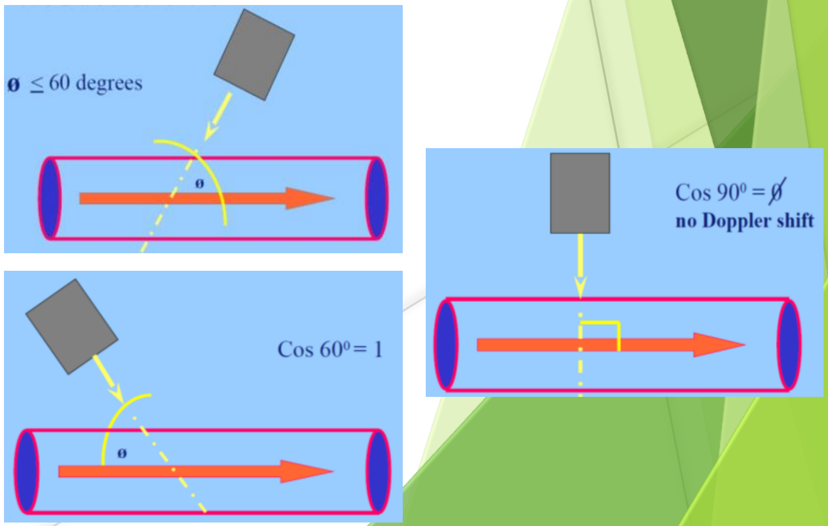
SPECTRAL WAVEFORM - *represents the audible signal*
- *provides information about direction of flow, how fast the flow is traveling (velocity), quality of flow (normal vs. abnormal*

DIRECTION OF FLOW - *flow towards the transducer is reflected above the baseline*
- *flow away from the transducer is reflected below the baseline*





(A) higher-frequency Doppler signal obtained if the beam is aligned more to the direction of flow
(B) lesser-frequency Doppler signals due to malalignment
(C) almost 90° very poor Doppler signal
(D) Doppler is away from the beam and there is a negative signal



DOPPLER IMAGING PRINCIPLES

DIFFERENT FORMS OF DOPPLER ECHOCARDIOGRAPHY

1. Continuous wave Doppler (CW)
2. Pulsed wave Doppler (PW)
3. Multigate pulsed wave Doppler - High PRF mode

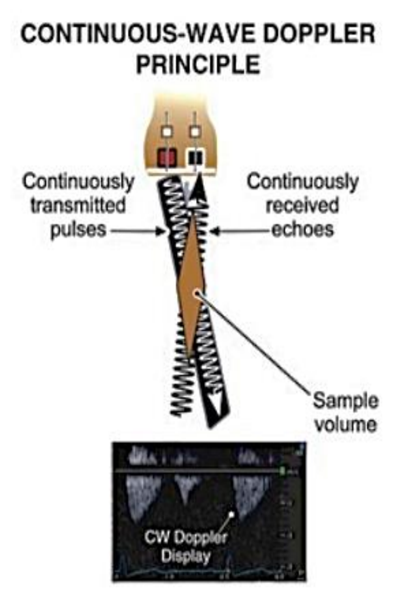
SPECTRAL DOPLER (items 1–3)

4. Colour Doppler flow mapping
5. Colour Doppler M-Mode
6. Three Dimensional (3D) colour Doppler flow mapping

DOPPLER IMAGING PRINCIPLES

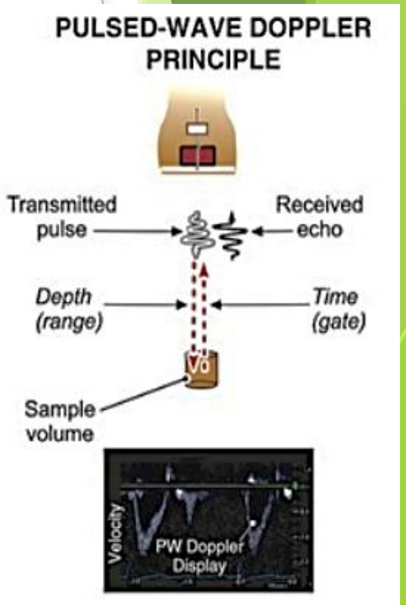
CONTINUOUS WAVE DOPPLER

- requires a transducer containing two separate ultrasound crystals (one continuously transmitting and the other continuously receiving the signal
- performed using image or non-image guided (pencil probe) transducers
- ADVANTAGE: ability to accurately measure maximum velocity without limitation of aliasing phenomenon
- DISADVANTAGE: cannot recognize where the velocity (along the beam) has been recorded
- NO RANGE RESOLUTION



PULSED WAVE DOPPLER

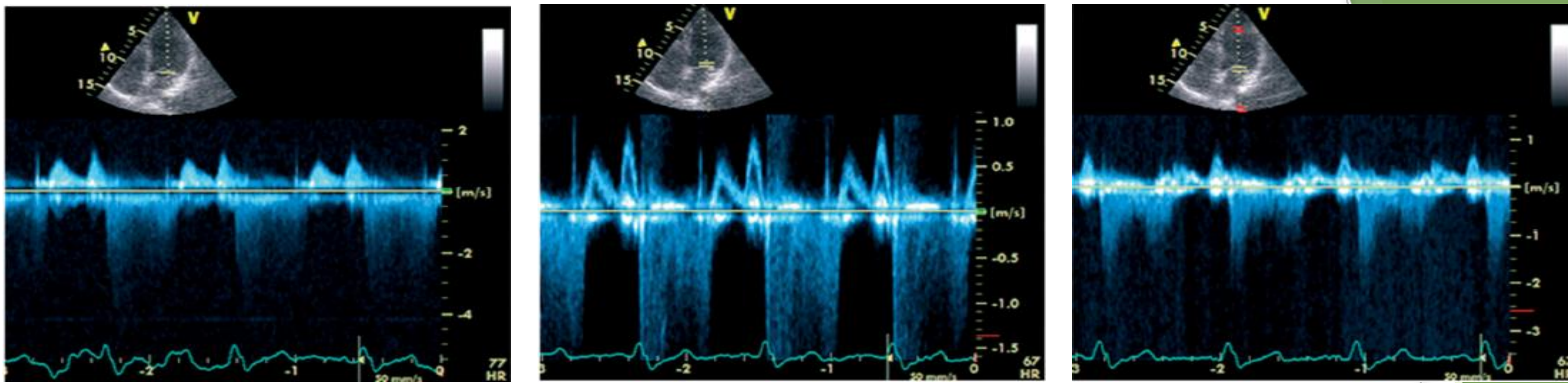
- requires a single crystal that sends short, intermittent bursts of ultrasound then waits to receive the returning signal
- ADVANTAGE: ability to accurately measure velocities from specific location in the heart using a sample volume controllable on a reference 2D image panel
- DISADVANTAGE: aliasing occurs in high velocity signals
- GOOD RANGE RESOLUTION



MULTIGATE PULSED WAVE DOPPLER

- high pulse repetition PW Doppler
- ADVANTAGE: offers higher Nyquist limit for correct recording of flow
- DISADVANTAGE: additional sample volume zones appear resulting in a more blurred spectral information
- LOSS OF SPATIAL SPECIFICITY

DOPPLER IMAGING PRINCIPLES



(A) CWD with full velocity range along the dotted sampling line, information on laminar versus turbulent flow is lost

(B) PWD with clear distinction of laminar inflow (above the baseline, empty spectrum envelope) and turbulent mitral regurgitant flow (below the baseline, filled envelope)

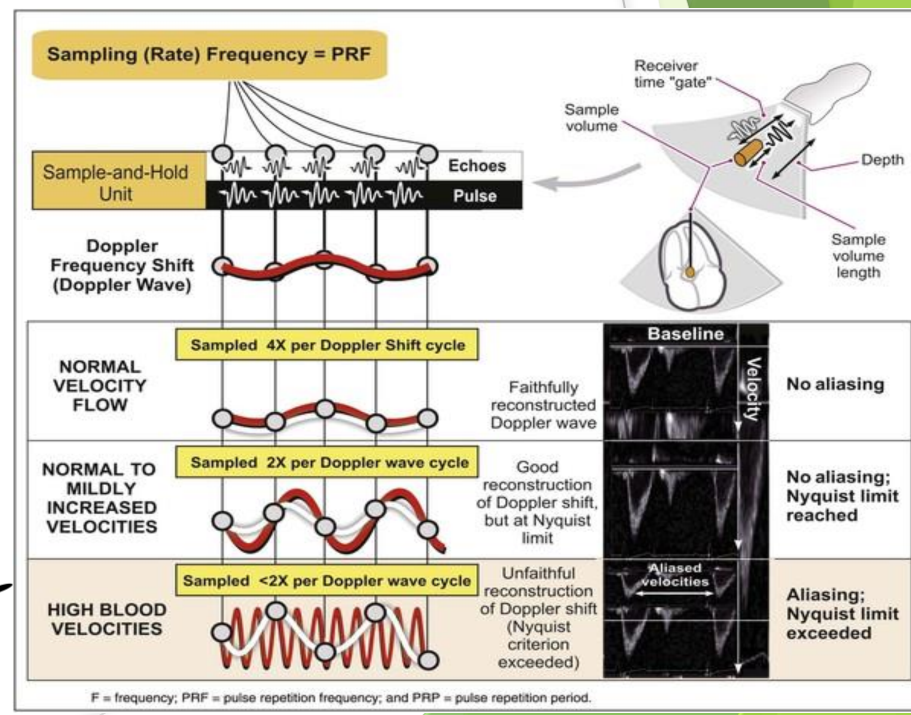
(C) HPRF PWD offers higher Nyquist limit for correct recording of flow but losing spatial specificity as two additional red sampling zones appear resulting in more blurred spectral information

NYQUIST LIMIT

- defines when aliasing will occur using PW Doppler
- specifies that measurements of frequency shifts (and, thus, velocity) will be appropriately displayed only if the pulse repetition frequency (PRF) is at least twice the maximum velocity (or Doppler shift frequency) encountered in the sample volume.

FORMULA:

$$\text{Nyquist limit} = \frac{\text{No. of pulses per second}}{2}$$



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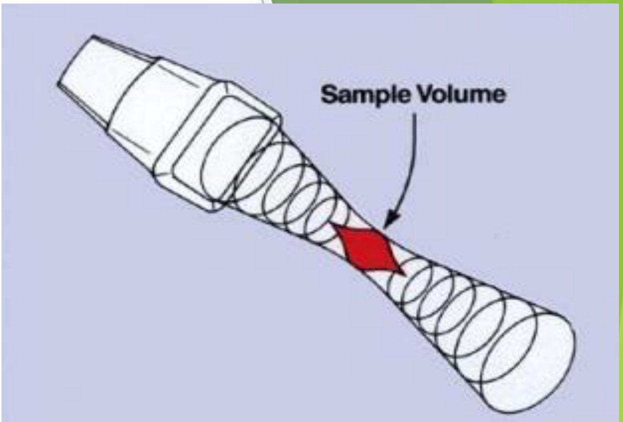
DOPPLER IMAGING PRINCIPLES

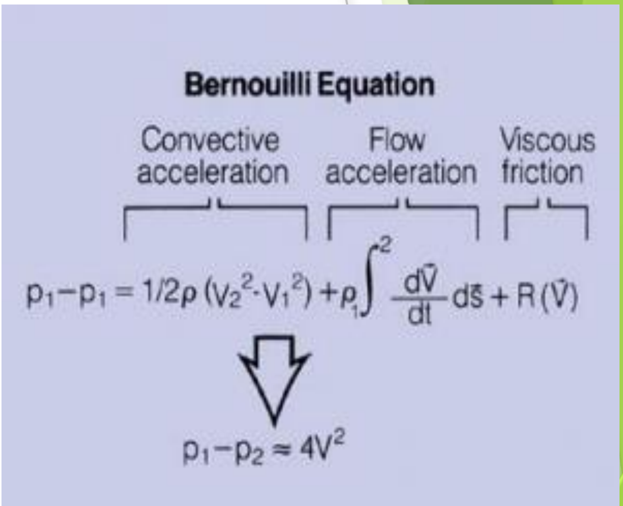
SAMPLE VOLUME

- a real three dimensional, tear drop shaped portion of the ultrasound beam
- volume varies with different Doppler machines, different size and frequency transducers and different depths into the tissue
- width is determined by the width of the ultrasound beam at selected depth and length is determined by the length of each transmitted ultrasound pulse



BERNOULLI EQUATION

- a complex formula that relates the pressure drop (or gradient across an obstruction to many factors
- full Bernoulli equation requires velocity data from Below (V1) and above (V2) any given obstruction, V1 can usually be ignored in calculation of a pressure gradient
- reduced to simplest equation
- in cases that beam is orient as parallel to flow as possible so that the full velocity recording is obtained (this assumes cosine Ø = 1)



example of a CW spectral recording of aortic stenosis, given velocity (V1) on the ventricular side of the valve that is accelerated, (V2) as blood is ejected through the stenotic orifice

FORMULA:
$P1 - P2 = 4V^2$

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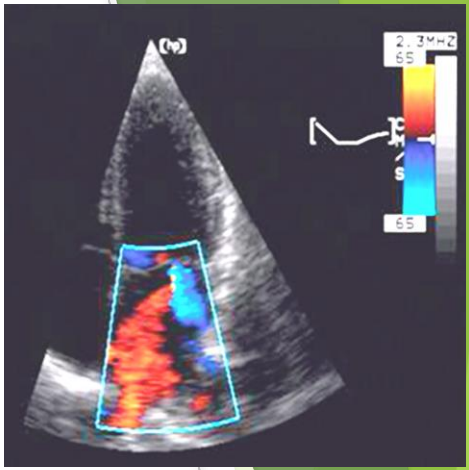
DOPPLER IMAGING PRINCIPLES

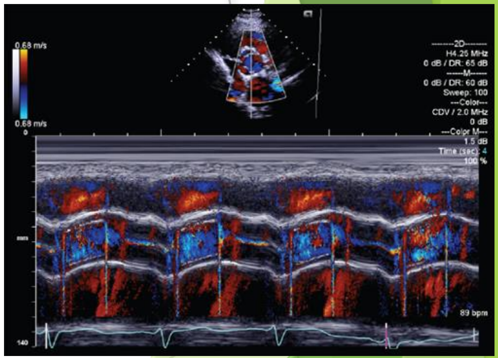


COLOUR DOPPLER DISPLAY

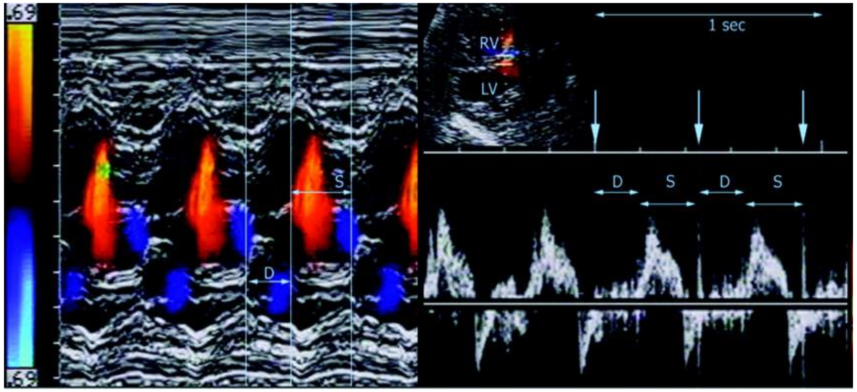
- presents information on the presence, direction, speed and character of blood flow by Colour coded patterns
- Colour spectrum display is superimposed on the 2D image
- red and yellow represent increasingly positive Doppler shifts above the baseline (towards the transducer), blue and cyan represent increasingly negative Doppler shifts below the baseline (away from transducer)

COLOUR M-MODE

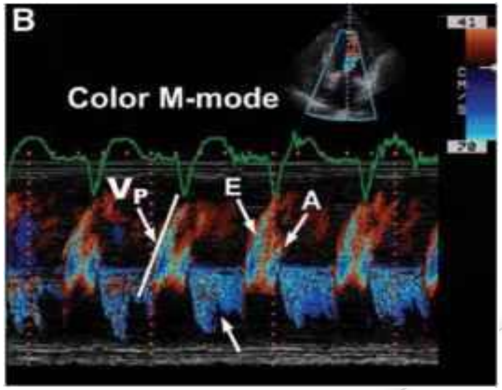
- a combination of M-Mode and Color Doppler
- good in assessing shunts, velocity progression on functional analysis, valve regurgitation



Assessment of Aortic regurgitation



Assessment of VSD

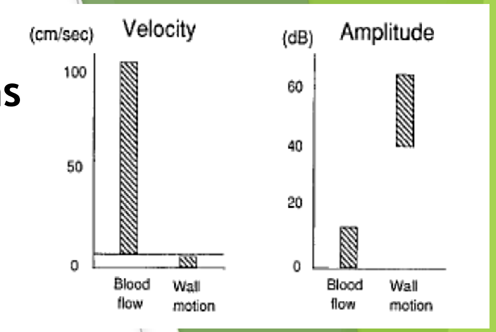


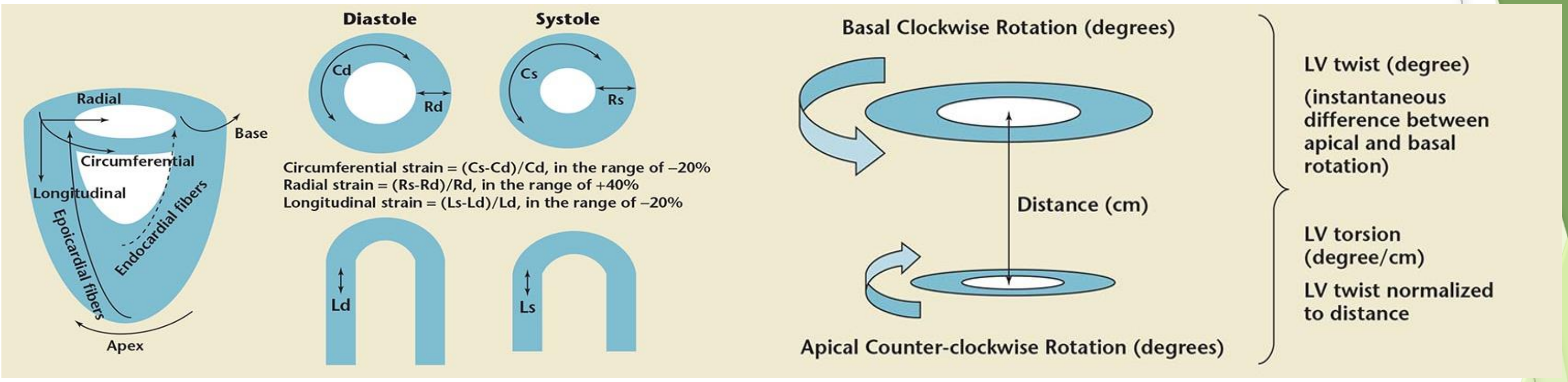
Assessment of Velocity of propagation of MV flow in functional analysis

ADVANCED IMAGING PRINCIPLES

TISSUE DOPPLER AND DEFORMATION ECHOCARDIOGRAPHY

- allows measurement of myocardial tissue velocity with several options
- assessment of tissue deformation in aid of early detection of wall motion abnormalities
- assessment is based on concept that tissue has high amplitude, low velocities (blood = high velocity, low amplitude)





3D ECHOCARDIOGRPHY

- used in evaluation of chamber volumes and mass, avoiding geometric assumptions, regional wall motion and quantification of systolic dyssynchrony
- very good in presentation of realistic views of heart valves and volumetric evaluation of regurgitant lesions and shunts
- 3D x-matric transducers are composed of nearly 3,000 piezoelectric elements
- 2-4MHz for transthoracic echo transducers
- Used in conjunction with 2D transthoracic echo

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ARTIFACTS

IMAGING ARTIFACTS

- extraneous ultrasound signals resulting in appearance of structures that are not usually present (at least in the specific imaging location)
- Failure to visualize structures that are present or a structure that differs in size or shape or both from the actual appearance
- width is determined by the width of the ultrasound beam at selected depth and length is determined by the length of each transmitted ultrasound pulse
- Most common artefact is from sub-optimal image quality

Artifact	Mechanism	Example(s)
Suboptimal image quality	Poor ultrasound tissue penetration	Body habitus (obesity, lung disease) Postcardiac surgery
Acoustic shadowing	Reflection of entire ultrasound signal by a strong specular reflector	Prosthetic valve Calcification
Reverberations	Reverberation between two strong parallel reflectors	Prosthetic valve
Beam width	Superimposition of structures within the beam profile (including side lobes) into a single tomographic image	Aortic valve "in" LA Atheroma "in" aortic lumen
Lateral resolution	Displayed width of a point target varies with depth	Excessive width of calcified mass or prosthetic valve
Refraction	Deviation of ultrasound signal from a straight path along the scan line	Double aortic valve or LV image in short-axis view
Range ambiguity	Echo from previous pulse reaches transducer on next cycle	Second, deeper heart image
Electronic processing	Instrument specific	Variable

ULTRASOUND IMAGING ARTIFACTS

ARTEFACTS

ASSUMPTIONS MADE DURING 2D ECHO IMAGING

1 Ultrasound travels in a straight line.
2 A structure or object (reflector) generates reflection (echo) only once.
3 Echoes are generated only from reflectors located within the main ultrasound beam.
4 The intensity of the echoes is related to the acoustic characteristics of the reflector.
5 The position of the reflector on the display monitor is proportional to the round trip travel time of the ultrasound beam.
6 The speed of sound in human tissue is constant.

COMMON 2D ECHOCARDIOGRAPHY ARTEFACTS

1. Shadowing and Enhancement artefact
2. Reverberation, Comet Tail, Ringdown artefact
3. Mirror-image (Reflection) and Ghost (Refraction) artefact
4. Beam width artefact
5. Side and Grating lobe artefact

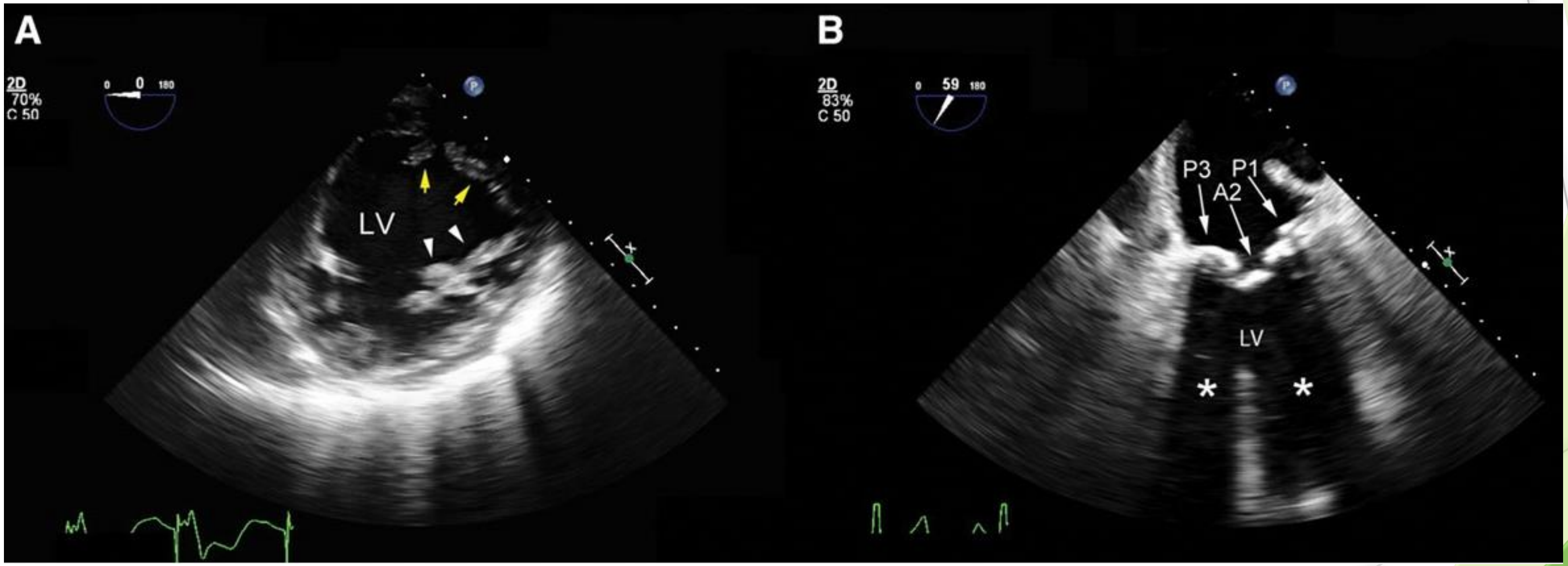
CATEGORY OF ARTEFACTS

1. Ultrasound beam characteristics: Side lobe, Grating lobe, Beam width, Near field clutter
2. Multiple echoes: Reverberation, Comet Tail, Ring down, Mirror Image (Reflection)
3. Velocity errors: Ghost (Refraction)
4. Attenuation errors: Shadowing, Enhancement

ARTEFACTS

SHADOWING AND ENHANCEMENT ARTEFACT

- anechoic or hypoechoic regions may be a result of shadowing
- hyperechoic areas on an image maybe a result of enhancement (sometimes resulting to extra-anatomic features
- Shadowing occurs when transmitting beam encounters a structure with high attenuating properties
- Enhancement occurs when tissue attenuates less that its surroundings
- can be resolved in by using TGC, or changing transducer position (use other imaging planes)



The left ventricle (LV) is shown in the short-axis view (A). The posteromedial papillary muscle (yellow arrows) is displayed with the correct grayscale. Propagation of the ultrasound inside the fluid-filled LV results in a relatively brighter anterolateral papillary muscle (white arrowheads) because of enhancement. B, Shadowing (white asterisks) from heavily calcified posterior (P1 and P3) and anterior (A2) mitral valve leaflets. The anechoic area distal to these structures prevents a thorough assessment of LV wall motion.

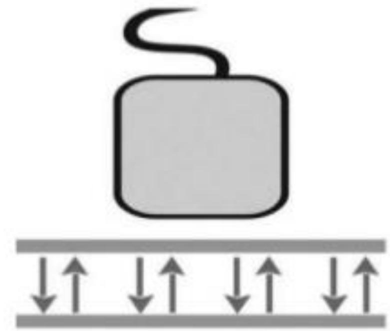
ARTEFACTS

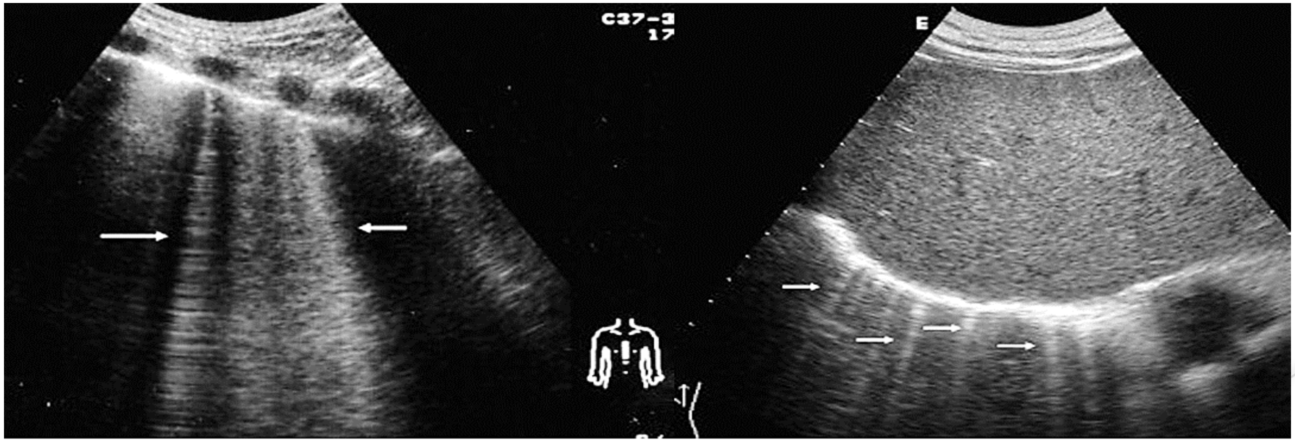
REVERBERATION ARTEFACT

- assumes that an echo returns to the transducer after a single reflection and that depth of an object is related to the time for this round trip
- results in a pattern of regularly spaced artefacts, spacing represents the distance between proximal and distal reflector
- intensity of reverberation is directly related to the difference in acoustic impedance between the reflector and its surroundings
- can be resolved by changing transducer or transducer position, reduce the gain

RING DOWN
- occurs when bubbles within a fluid background reflect or resonate sound waves
- presence should alert echocardiographer to presence of gas (i.e. air embolism or post-cardiopulmonary bypass air

COMET TAIL
- linear artefacts in that extends longitudinally
- usually occurs in the presence of closely spaced reflectors





Ring-down artefact continues to the back of the image and gets wider with depth

Comet-tail artifacts are shorter in length and taper with depth

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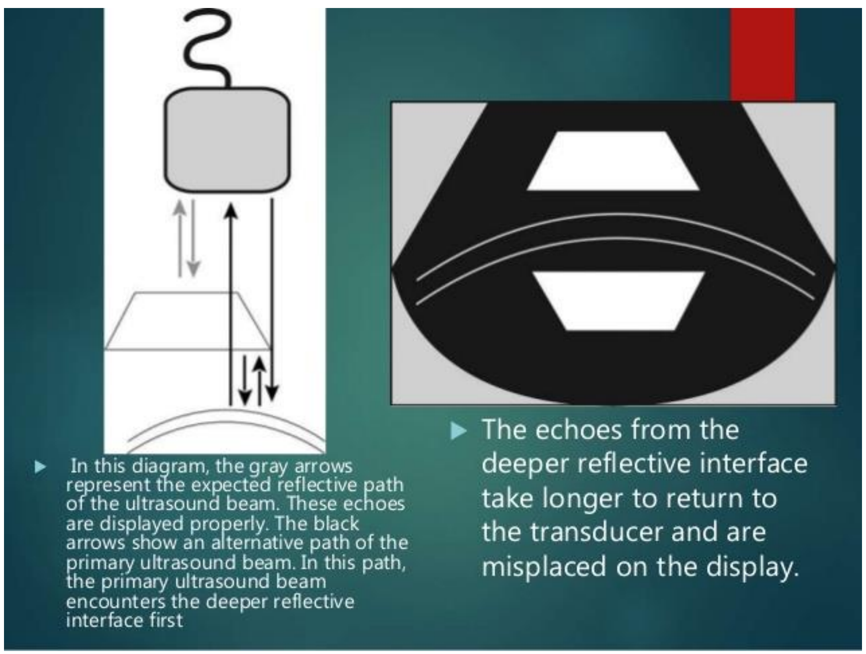
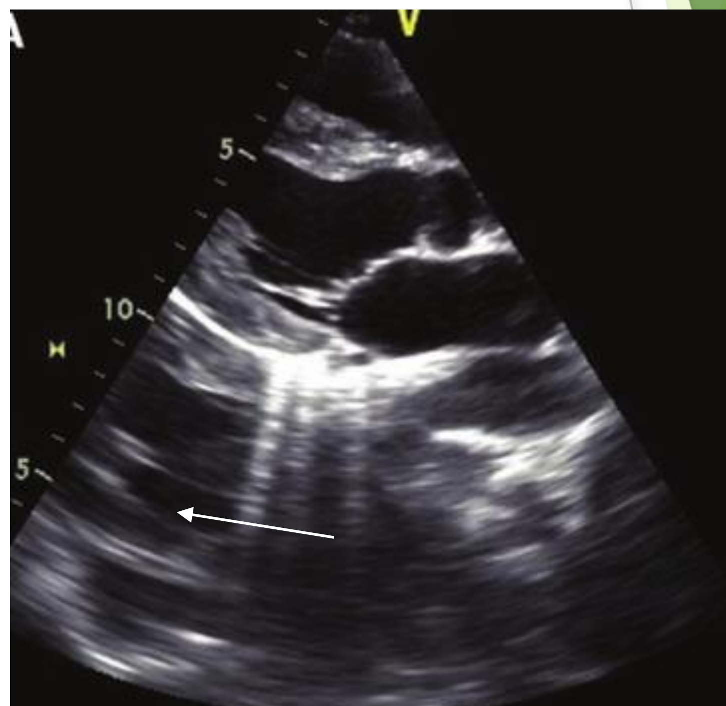
ARTEFACTS

MIRROR-IMAGE (REFLECTION) ARTEFACT

- create an appearance of additional structures on the display
- duplicated structure is deeper, equidistant and occasionally lateral to the reflector
- occur when the assumption that the ultrasound echo returns to the transducer after only a single reflection is violated
- the ultrasound beam first hits a large, smooth (mirror-like) reflector during the transmission phase, which directs it to a second reflector, then bounces from the target back to the mirror-like surface on its return to the probe
- can be resolved by changing frequency (ie harmonics) and angulation of transducer

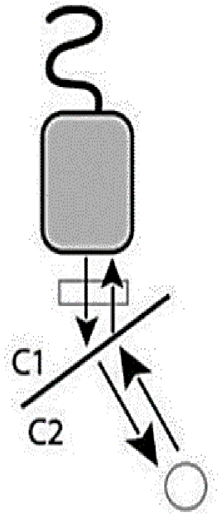


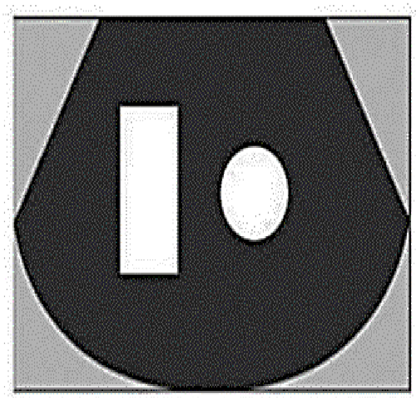


ARTEFACTS

GHOST/DOUBLE IMAGE (REFRACTION) ARTEFACT

- create an appearance of additional structures on the display
- duplicated structure is lateral to the original image
- ultrasound display assumes that the beam travels in a straight line and thus misplaces the returning echoes to the side of their true location
- maybe produced due to change in velocity of the ultrasound beam as it travels through two adjacent tissues with different density and elastic properties
- non-perpendicular incident ultrasound energy encounters an interface between two materials with different speeds of sound resulting to beam changing direction
- can be resolved by changing frequency (ie harmonics) and angulation of transducer

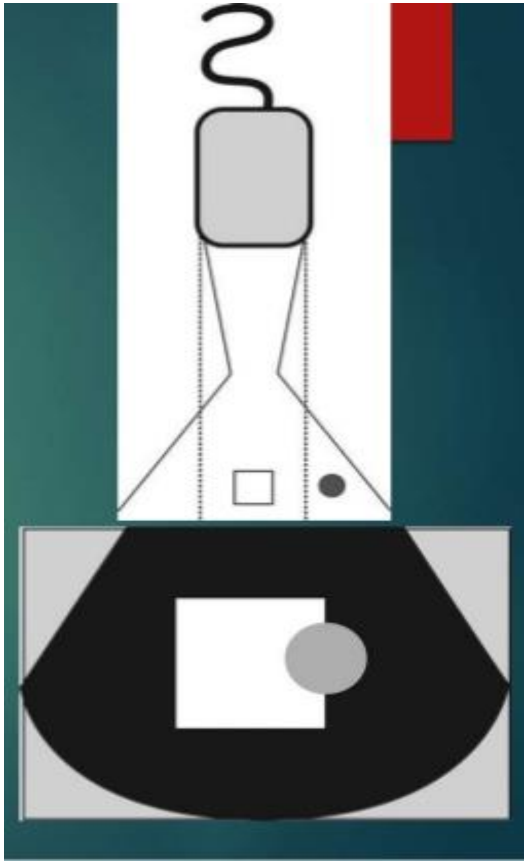


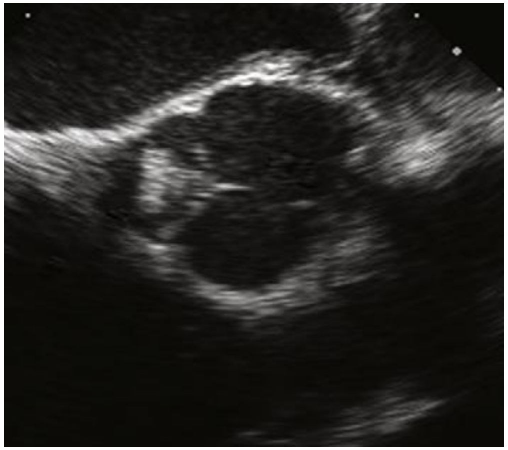


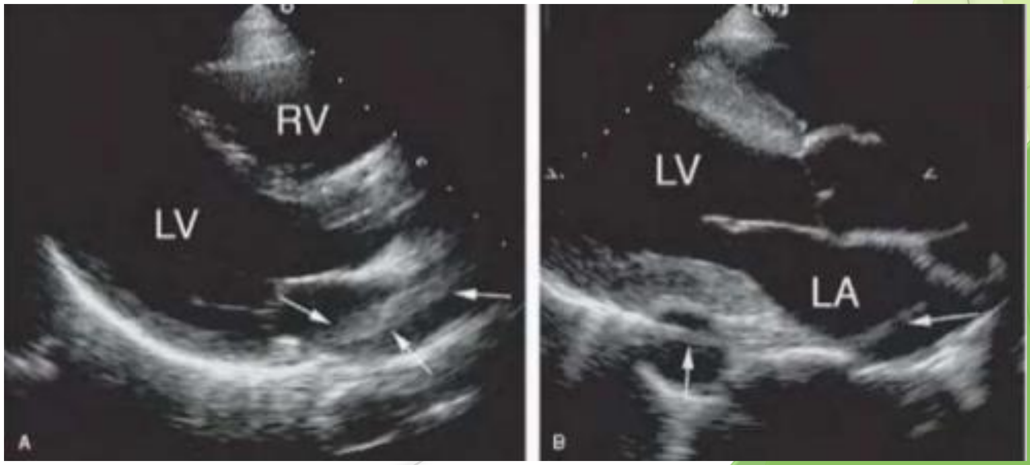
ARTEFACTS

BEAM WIDTH ARTEFACT

- distal beam may widen beyond the actual width of transducer
- highly reflected object located within the widened beam beyond the margin of transducer may generate detectable echoes
- artefact can be seen in an echo-free area as spurious echoes
- can be resolved in positioning the focus (by narrowing the beam)







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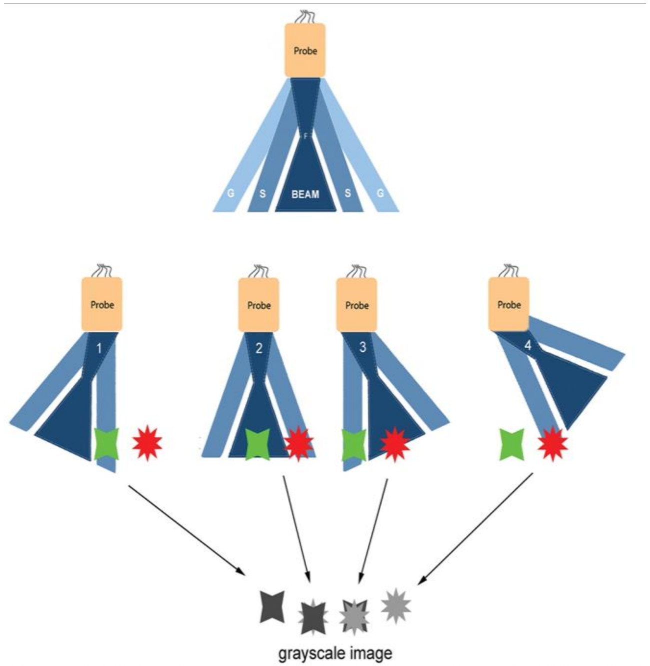
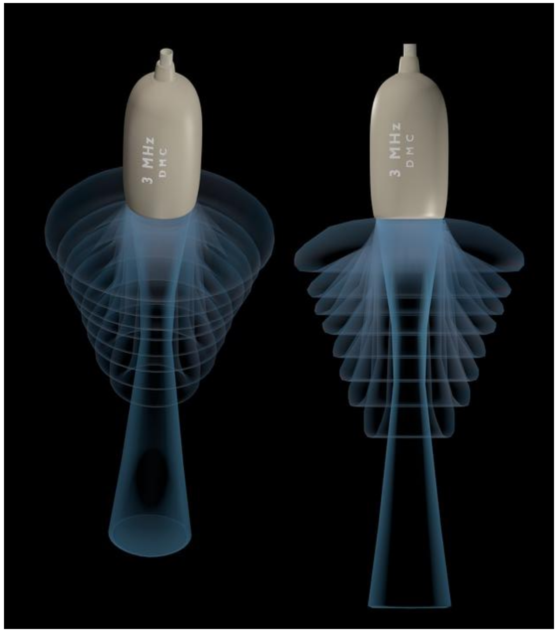
ARTEFACTS

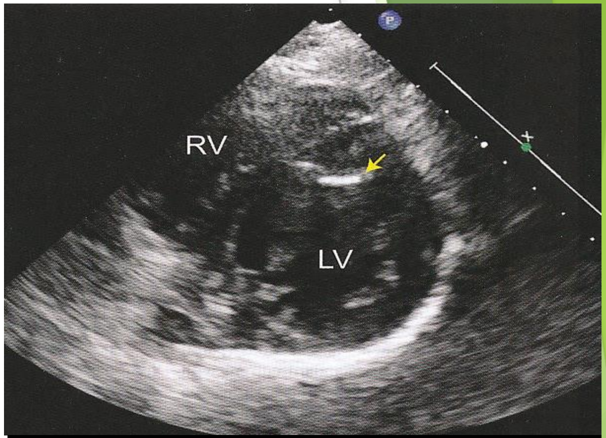
SIDE AND GRATING LOBE ARTEFACT

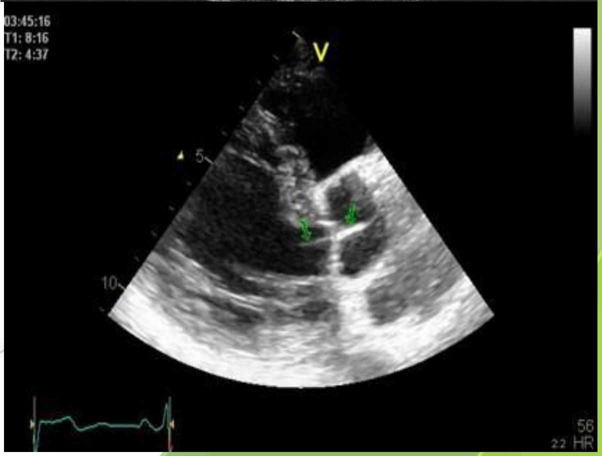
- result in blurring of the edges of a displayed object (reduce lateral resolution)
- assumption that ultrasound waves are infinitely thin is violated
- grating and side lobes appear similarly around the main beam but mechanisms of origin differ, had to differentiate in clinical practice
- are secondary beam around the central ultrasound beam and are produced by non-axial vibrations of the piezoelectric elements
- can be resolved by changing transducer frequency from high to low or using tissue harmonics



In the far field, distal to the focal zone (F), the beam widens (beam). Side lobes (S) and grating lobes (G) are lateral to the main beam. The bottom panel shows how each target may potentially have 2 additional echoes in the display



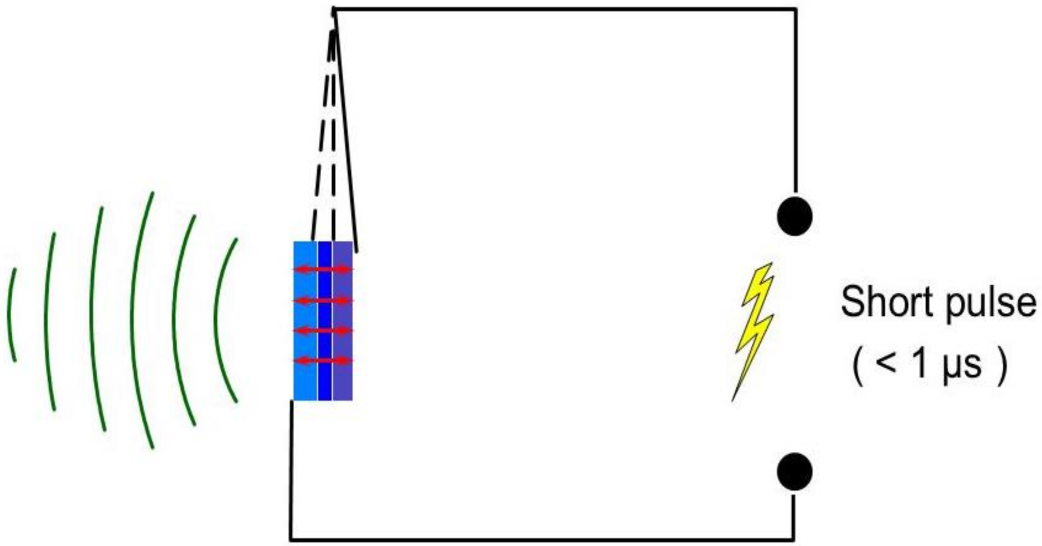


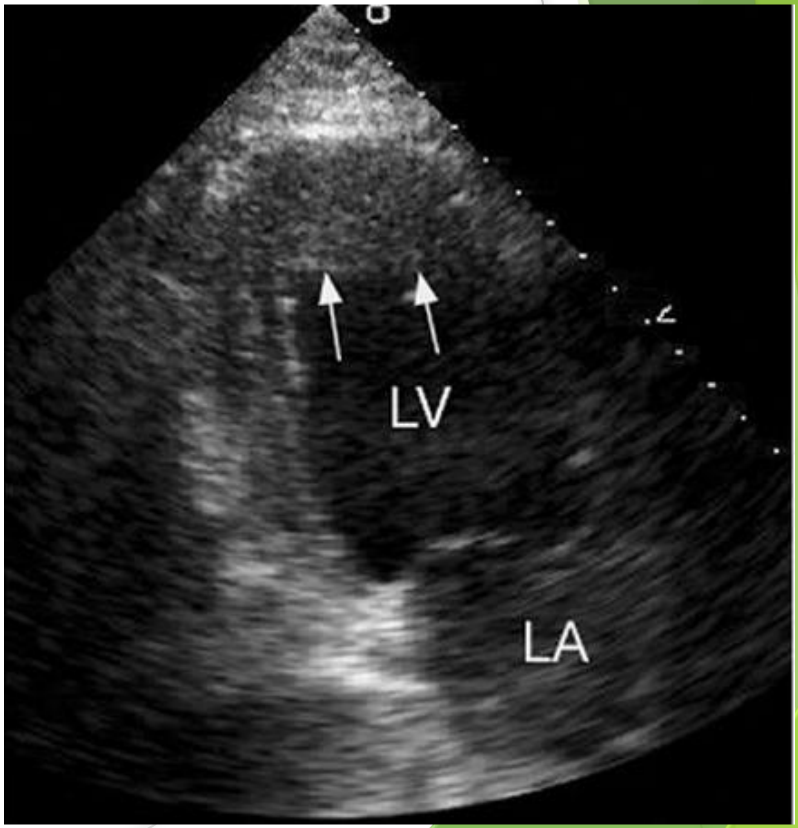


ARTEFACTS

NEAR FIELD CLUTTER ARTEFACT

- arises from the high amplitude oscillations of the piezo-electric crystals, called ringing effect
- only involved in near field
- quite troublesome when trying to identify close structures to the transducer
- Significantly resolved by modern day transducers, can also be avoided by changing to high frequency transducer, decreasing depth and adjusting focal zone





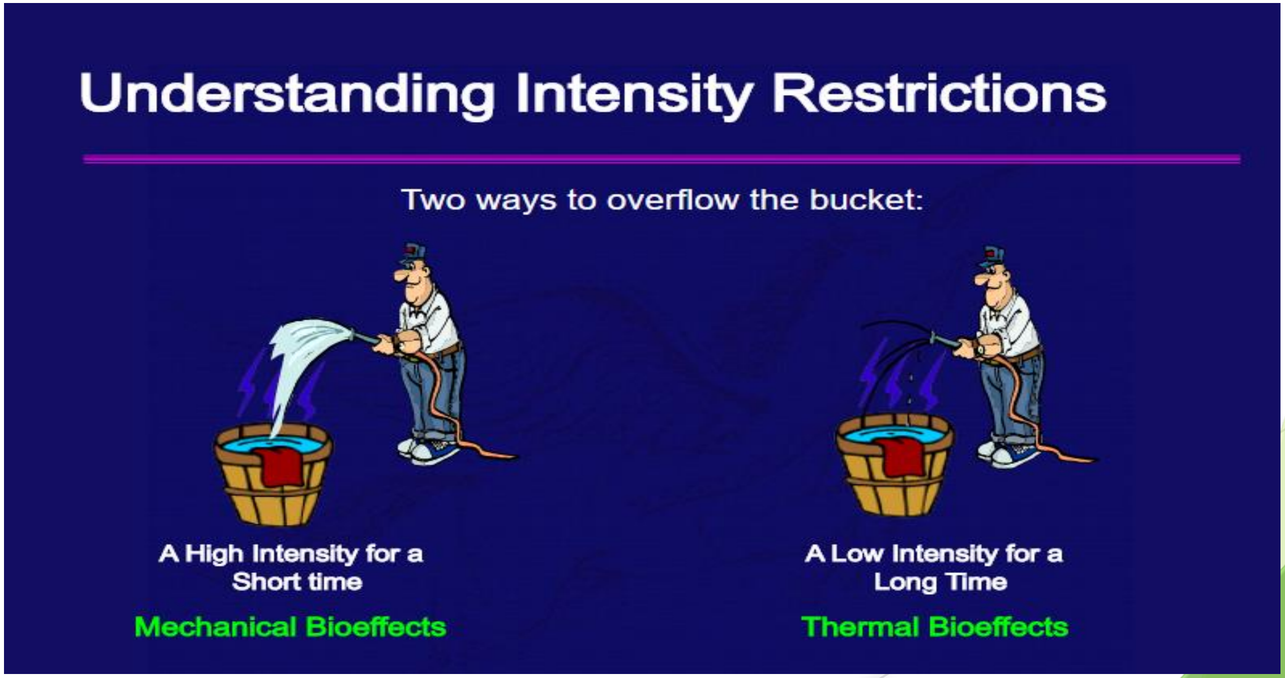
BIOEFFECTS

BIOEFFECTS

- the amount of acoustic energy applied depends upon two factors, the intensity of the beam being applied and the amount of time the pulse is on
- too much acoustic energy can result in tissue damage
- Acoustic energy gives better signal and better image quality (better signal-to-noise ratio)

THERMAL INDEX (TI) - monitors heating
- should remain below 2

MECHANICAL INDEX (MI) - monitors cavitation (i.e. formation of small gas bubbles with subsequent bubble collapse associated with high pressures/temperatures locally
- should remain below 1.9




That's all Folks!