3. Ultrasound Imaging(2)

Lecture 13, 14

Medical Imaging Systems

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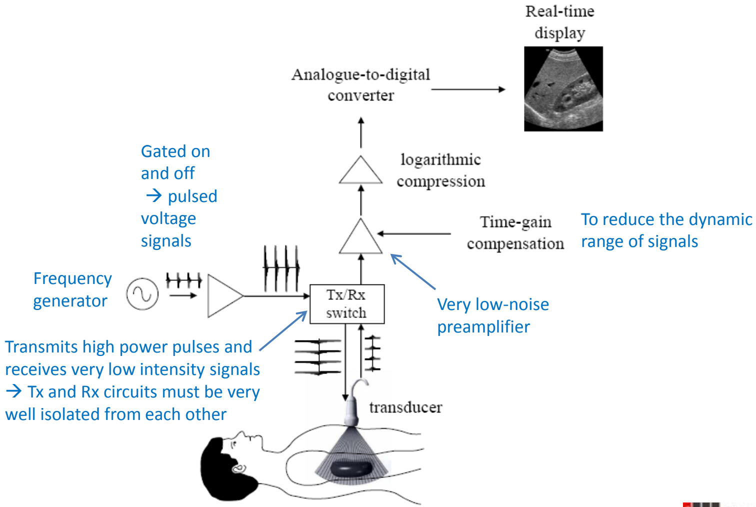
1. Instrumentation
2. Single element ultrasound transducer
3. Transducer arrays
4. Clinical diagnostic scanning modes
5. Image characteristics

Instrumentation

- The major elements of a basic ultrasound imaging system



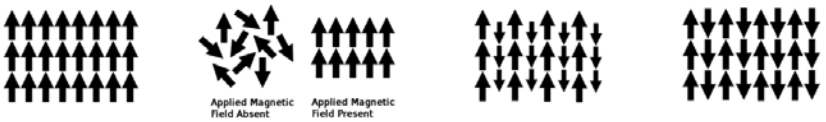
Single element US transducers

- Piezoelectric materials: produce electricity resulting from pressure and vice versa
- First application of piezoelectric material was a sonar (World War I)
- Lead zirconate titanate ($Pb[Zr_xTi_{1-x}]O_3$, $0 \le x \le 1$), PZT, is piezoelectric, pyroelectric, and also ferroelectric material which was developed by Yutaka, Gen, Etsuro at Tokyo institute of Technology ~1952

Magnetic Moments

Below T_C		Above T_C
Ferromagnetic	↔	Paramagnetic
Ferrimagnetic	↔	Paramagnetic
Antiferromagnetic	↔	Paramagnetic

Orientations of magnetic moments in materials



Ferromagnetism: The magnetic moments in a ferromagnetic material. The moments are ordered and of the same magnitude in the absence of an applied magnetic field.

Paramagnetism: The magnetic moments in a paramagnetic material. The moments are disordered in the absence of an applied magnetic field and ordered in the presence of an applied magnetic field.

Ferrimagnetism: The magnetic moments in a ferrimagnetic material. The moments are aligned oppositely and have different magnitudes due to being made up of two different ions. This is in the absence of an applied magnetic field.

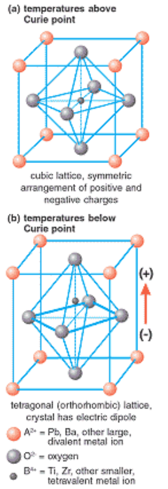
Antiferromagnetism: The magnetic moments in an antiferromagnetic material. The moments are aligned oppositely and have the same magnitudes. This is in the absence of an applied magnetic field.

https://en.wikipedia.org/wiki/Curie_temperature

How to make PZT?

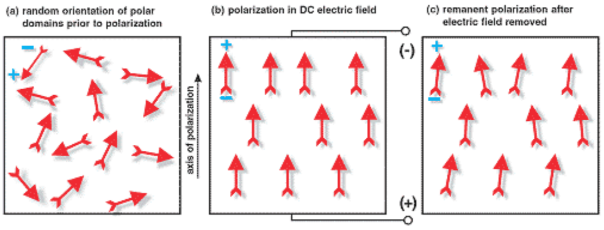
- Mix 3 metal oxides fine powders
- Heat to form a uniform powder
- Mix the powder with an organic binder and forms into a shape (rods, plates, discs, etc.) and then it is fired
- Apply a strong electric field (tens of kV per cm) just below the curie point temperature(T_c)
  PZT:320°C

Figure 1.1 Crystal structure of a traditional piezoelectric ceramic



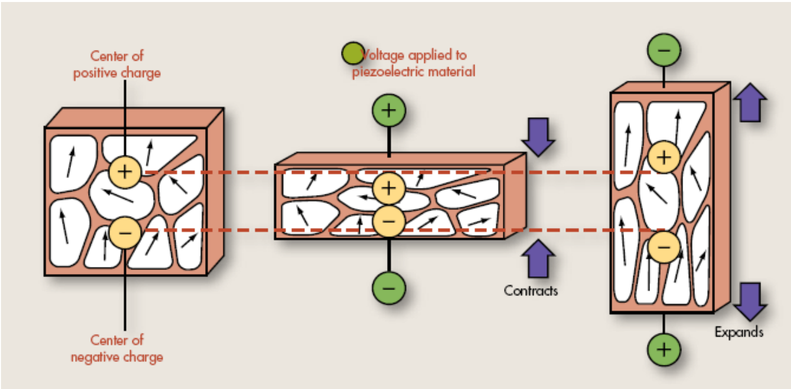
http://www.pc-control.co.uk/piezoelectric_effect.htm

Figure 1.2 Polarizing (poling) a piezoelectric ceramic*



→ ferroelectric material

How to produce voltages?

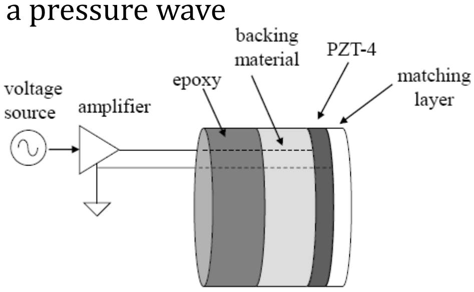


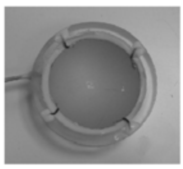
1. The piezoelectric effect causes crystal materials like quartz to generate an electric charge when the crystal material is compressed, twisted, or pulled. The reverse also is true, as the crystal material compresses or expands when an electric voltage is applied.

http://juvenilerejoice.blogspot.com/2011_09_01_archive.html

Single element US transducers

- PZT-4 is coated with a thin layer of silver and connected via bonding wires to a coaxial cable leading to transmit-receive switch
- Apply an oscillating voltage → oscillating PZT thickness at the same frequency as the applied voltage
- Place transducer to the patient's skin → transfers mechanical motion of PZT into a pressure wave





PZT Thickness

- The element has a natural resonant frequency (f_0) corresponding to its thickness (t) being ½ of the ultrasound wavelength (λ) in the crystal

$$t = \frac{\lambda_{crystal}}{2} = \frac{c_{crystal}}{2f_0}, \quad f_0 = \frac{c_{crystal}}{2t}$$

- Resonant frequency at the odd harmonics of f_0, that is, $3f_0$, $5f_0$, $7f_0$, etc.
- $c_{crystal}$ of PZT is ~4000m/sec, so the thickness of a crystal for 1.5MHz operation is ~1.3mm

Matching Layer

- Z value of PZT is ~ 30 X 10^5 g/cm^{2}s and Z value of skin/tissue is ~1.7 X 10^5 g/cm^{2}s → large amount of reflection will occur
- What will be solution?

  → have a 'matching layer'

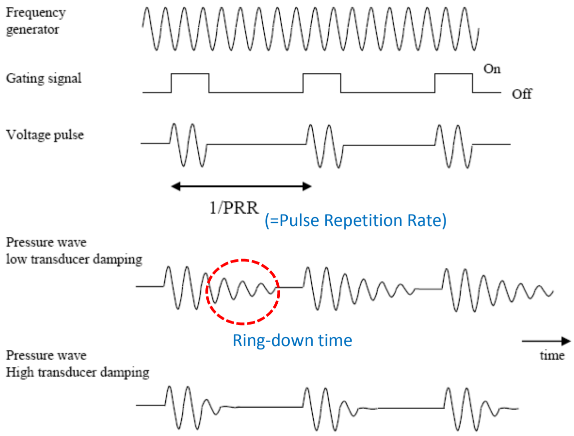
$$Z_{matching\ layer} = \sqrt{Z_{PZT} Z_{skin}}$$

- The thickness of the matching layer should be ¼ of the ultrasound wavelength to maximize energy transmission through the layer in both directions.
- PVDF(polyvinyldifluoride) can be used as a matching layer and multiple matching layers are used to increase the efficiency further.

Damping Layer

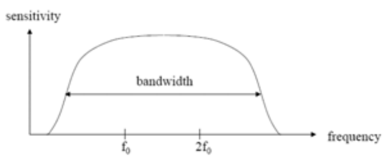
- Damping layer (backing material: small Al_2O_3 particles + epoxy) reduces the pulse duration → increase axial resolution

Damping layer is analogue to a mechanical damping to stop the long sound produced from a single strike of a bell



Transducer bandwidth

- A modern transducer has a wide bandwidth
  – Ex) central frequency (f_0) of 3MHz and 1~5MHz bandwidth
- This means
  – We can use a single transducer for many applications
  – We can receive the second harmonic signals ($2f_0$) without having a second transducer
- A quality factor (Q) = $2\pi f_0$ / bandwidth
- Q has been improved by growing small oriented PZT crystals (as opposed to small particulates being embedded in a polymer matrix)

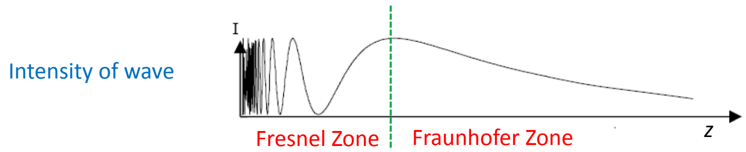


Beam geometry

- If wave propagation is in the z direction, then the on-axis, or axial, intensity $I(z)$ of the wave is given by

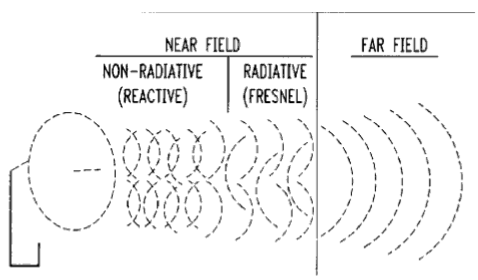
$I(z) \approx 2\rho c u_z^2 sin^2\left[\frac{\pi}{2}\left(\frac{r^2/\lambda}{z}\right)\right]$, ($I$: W/cm^2=J/sec·cm^2=kg·m^2/sec^2)

where r is the radius of the crystal



Beam geometry

- Near-field (Fresnel) zone: the wave pattern very close to the transducer face is extremely complicated → not useful for diagnostic scanning
- Far-field (Fraunhofer) zone: ultrasound beam do not oscillate in intensity but rather decays exponentially with distance



Fresnel diffraction occurs when:

$$F = \frac{a^2}{L\lambda} \geq 1$$

Fraunhofer diffraction occurs when:

$$F = \frac{a^2}{L\lambda} \ll 1$$

a - aperture or slit size,
λ - wavelength, L - distance from the aperture

http://en.wikipedia.org/wiki/Far_field

Beam geometry

- Near-field boundary (NFB) occurs at a distance (Z_{NFB}) from the transducer face which is

$$Z_{NFB} \approx \frac{r^2}{\lambda}$$

- r is radius of the transducer and λ is the wavelength of ultrasound in tissue
- At the NFB, the field has a lateral beamwidth (≅ transducer diameter)
- Beyond NFB, the beam diverges in the lateral direction with an angle of deviation (θ)
  $\theta = sin^{-1}\left(\frac{0.61\lambda}{r}\right)$, ↓$\lambda$ (= ↑ frequency), ↑r → ↓ θ → ↑ lateral resolution

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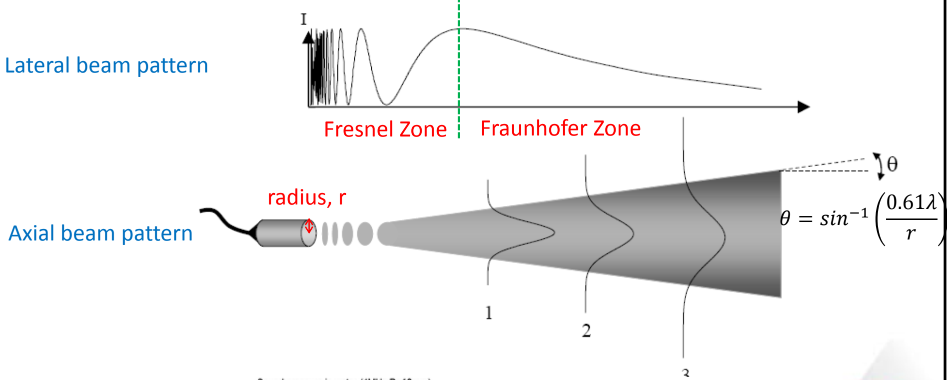
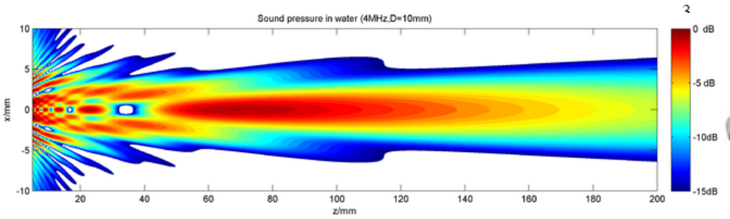
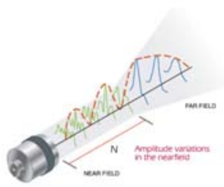
Beam geometry





http://en.wikipedia.org/wiki/File:Soundfield_Water_4MHz_TransducerRadius5mm.png



http://www.olympus-ims.com/ko/ndt-tutorials/transducers/characteristics/

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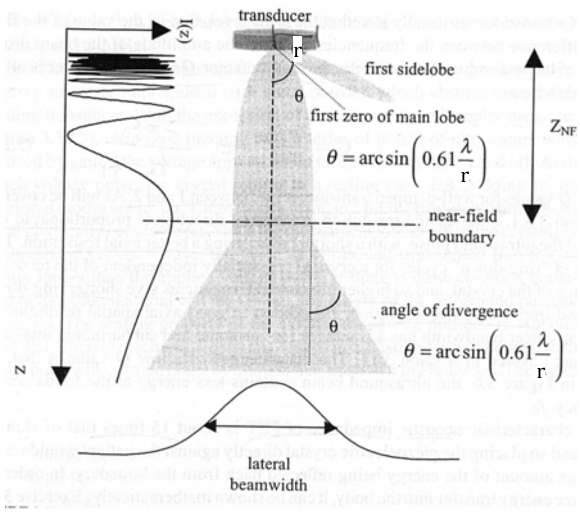
Side-lobes

- Side-lobes are also produced by a single element transducer due to the transducer acting as a diffraction grating
- The first zero of the side-lobe present at an angle φ is given by $\varphi = sin^{-1}\left(\frac{0.61\lambda}{r}\right)$
- These side-lobes are undesirable because
  1) they remove energy from the main beam
  2) they can introduce artifacts if the lobes are backscattered from tissue which is outside the region being studied
- The magnitude and the number of side lobes depend on <u>ultrasound wavelength/transducer diameter (λ/2r)</u>
- $\left(\frac{\lambda}{2r}\right)$ ↑ → the number of side lobes ↓, the closer the NFB lies to the face of the transducer

Side-lobes



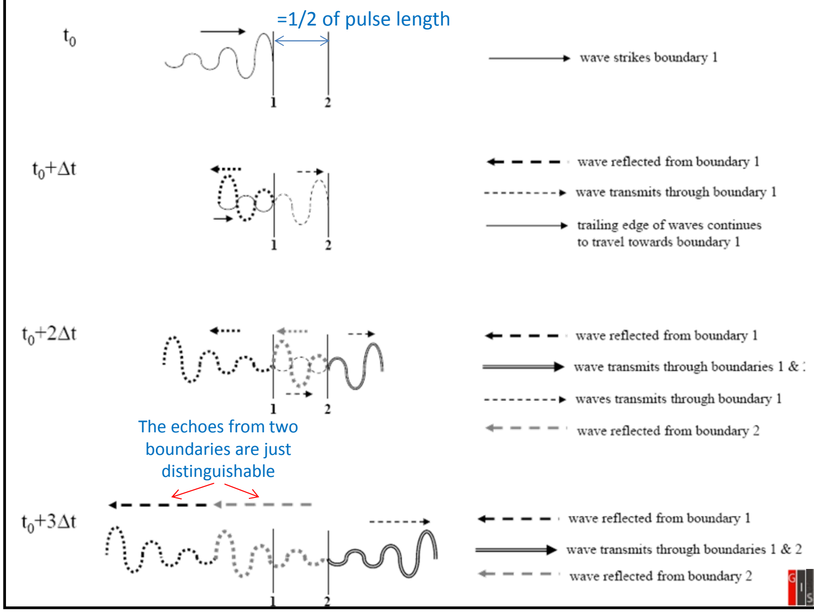
Axial resolution

- Axial resolution can be defined as the closest distance between two boundaries that can be resolved as two

$$Axial\ resolution = \frac{1}{2} P_d c$$

- where P_d is the pulse duration (sec) , therefore, axial resolution is ½ of pulse length (=$P_d c$)
- Typical value of axial resolution is 1.5mm at 1 MHz and 0.3 mm at 5 MHz
- However, higher frequency attenuates after as it penetrates tissue
- To improve axial resolution
  1) Increase frequency
  2) Increase the damping efficiency

Axial resolution



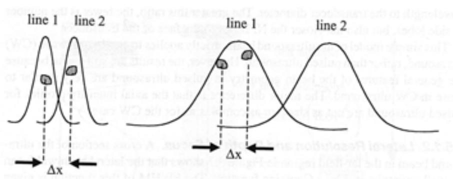
Lateral Resolution

- In the far-field, the lateral beam is approximately Gaussian and the lateral resolution is defined as the FWHM

$$\text{FWHM} = 2\sqrt{2ln2}\sigma \cong 2.36\sigma$$

where σ is the standard deviation of the Gaussian function

- Because a single crystal transducer typically has a diameter of between 1 and 5 cm, the intrinsic lateral resolution is very poor

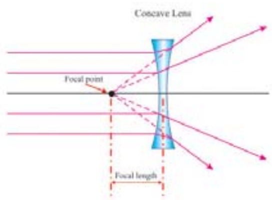


Transducer focusing

- A single flat transducer has a poor lateral resolution
- How can we improve the resolution?

  → Make it focus to produce a 'tighter' ultrasound beam
- Two ways to focus ultrasound beam in a single flat transducer
  1) Place a concave lens (usually polystyrene or an epoxy resin) in front of the piezoelectric materials

     :acoustic lens allows sound propagates at a higher speed than water or body tissues, and therefore, concave lens will converge US wave
  2) Curved surface of piezoelectric materials



Transducer focusing

- The shape of the curvature is defined in terms of $f\#$
  $f\# = \frac{R\ (the\ radius\ of\ curvature)}{2a\ (2a{:}diameter\ of\ the\ lens)}$,
  - In optics, $f\# = \frac{f\ (focal\ length)}{D\ (aperture\ diameter)} = \frac{1}{2NA}$
- $F \approx \frac{R}{1-(1/f\#)}$ where F is the focal distance
- Focal distance is slightly larger than the radius of curvature of the lens
- A transducer is normally referred to as being
  - strongly focusing (R < NFB/4)
  - medium (NFB/4 < R < NFB/2)
  - weakly focusing (R > NFB/2)

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Transducer focusing

- For a spherical focusing lens, the FWHM at the focal point is FWHM$\approx \frac{1.1\lambda R}{2a(=D)} = 1.1\lambda f\#$
- Therefore, decreasing the radius of curvature and increasing the diameter of crystal will improve the lateral resolution
- The lateral resolution is given by $\lambda F/D$ (D: transducer diameter)
- Depth of focus (DOF): the distance over which the beam intensity is at least 50% of its max value (focal point)

$$DOF = 15\left(1 - 0.01 sin^{-1}\frac{a}{R}\right) \cdot FWHM$$

- The above equation shows that a compromise has to be made between lateral resolution and depth of focus

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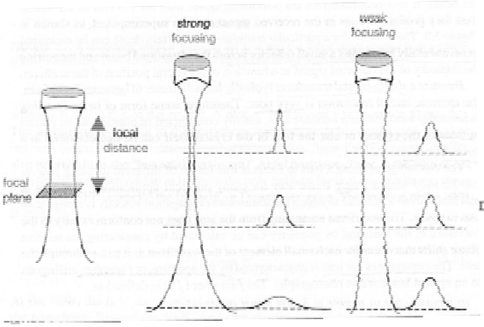
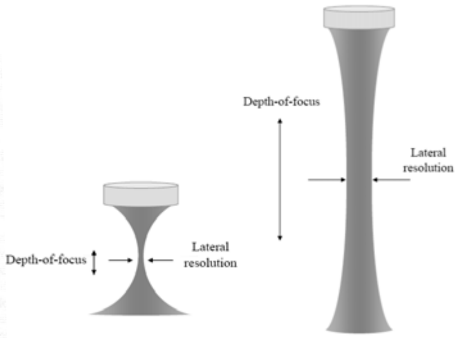
Transducer focusing

- The focal distance is approximately same as the radius of curvature of PZT element except for very strongly focusing transducer
- The lateral resolution of a focused transducer is also improved by increasing the frequency
  FWHM$\approx \frac{1.1\lambda R}{2a}$





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Transducer arrays

- Transducer arrays over a single element transducer enable to acquire two dimensional images in a fixed position
- Two basic types of array transducer, sequential and phased
- Linear/curved sequential array transducer: ~512 elements, only X elements (typically 8 to 16) of total number of elements are selectively pulsed to form a scan line and simply moves the same X elements pulse sequence along the entire array to form the parallel focused scan lines
- Linear/curved phased array transducer: 16 and 256 elements, all the array elements must be selectively pulsed to form the wavefront for a single scan line. It requires a unique total element pulse sequence for each scan line since each line has its own unique angle with respect to the transducer face in the sector format

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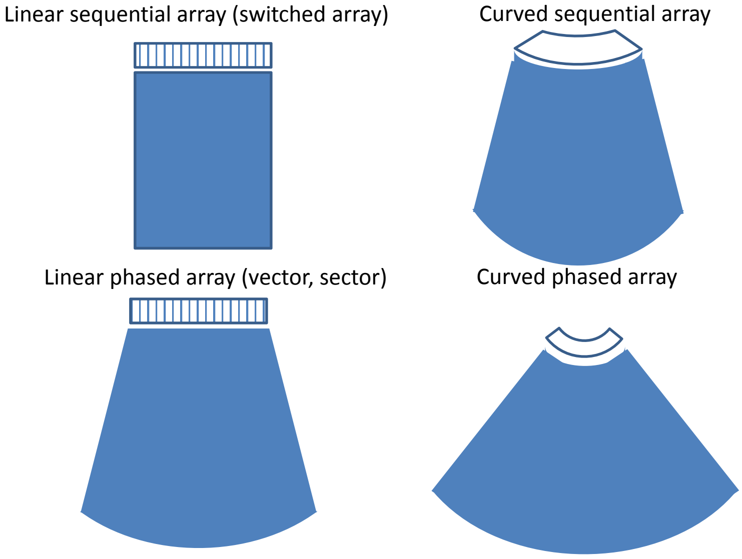
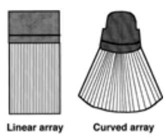
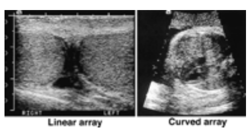
Transducer arrays

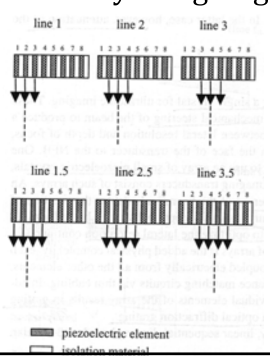


Linear sequential arrays

- Consists of a large number, 128~512, of rectangular shaped piezoelectric elements (1cm in width and 10-15cm in length)
- The width of the ultrasound beam is determined by the number of elements excited
- Linear sequential array is essentially unfocused device and, if required, focusing can be introduced by designing a curved array or adding a cylindrical lens







Linear sequential arrays

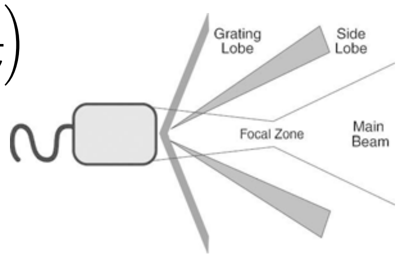
- Grating lobes are present with an angle ϕ_g,

$$\phi_g = sin^{-1}\left(\frac{n\lambda}{g}\right)$$

  where g is the gap between the elements and n=±1, ±2, ±3, etc.
- It also produces the normal side lobes and the angle at which the main beam intensity first reaches zero is given by

$$\theta = sin^{-1}\left(\frac{\lambda}{w}\right)$$

  where w is the width of the array



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Linear sequential arrays

- The magnitude of the grating lobes can be reduced by introducing small random variations into the spacing between adjacent elements of the array
- Alternatively, the spacing between elements can be made so small that the value of ϕ_g is close to 90^o, and the first grating lobe falls close to the edge of the FOV of the imaging
- Linear arrays are used when a large field of view is required close to the surface of the array

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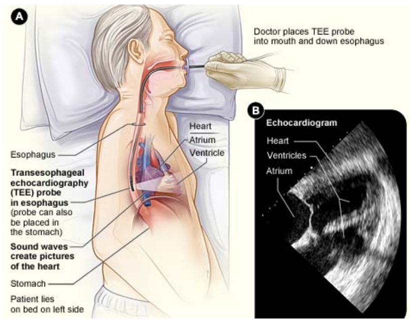
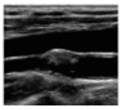
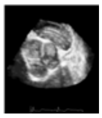
Phased arrays

- Phased arrays are much smaller than linear arrays (1-3cm length and 1cm width)
- Each element is less than 1 mm in width
- Phased arrays are widely used in applications that there is a small 'acoustic window' such as the space between ribs

Transesophageal phased array

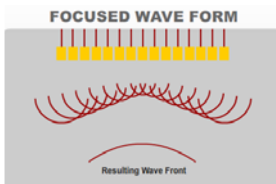






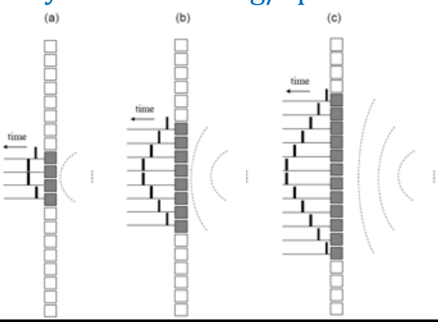
Phased arrays

- Beam forming: all elements in the array are excited by voltage pulses which is applied at slightly different times and the sum of all of the individual waves makes an effective wavefront shown in the right

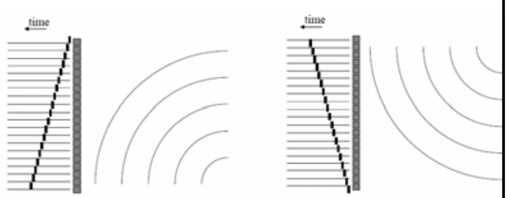


http://www.olympus-ims.com/ko/ndt-tutorials/transducers/generating/

Dynamic focusing/aperture



Beam steering

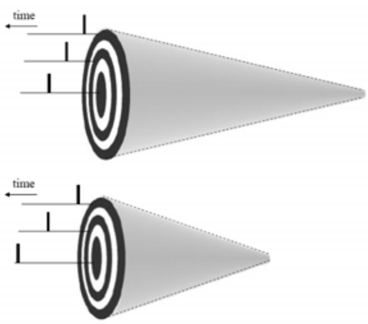


Phased arrays

- Slice thickness: is decided by the length of each element of phased arrays and is typically 2~5mm.
- To improve the resolution in this dimension, a curved concave lens can be incorporated into the transducer

Annular arrays

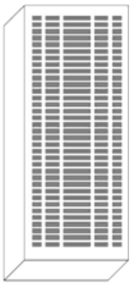
- Linear sequential and linear phased arrays are hard to construct for very high frequencies (> 20 MHz)
- Annular array is capable of two dimensional dynamic focusing, but has a fewer elements (5~10) than linear or phased array
- Disadvantage: mechanical motion is required to sweep the beam to form an image
- However, highly accurate mecha-nical unit can be integrated into the design



Multi-dimensional arrays

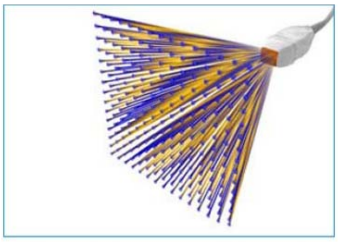
- Multi-dimensional arrays enable to focus in the elevation dimension, but add complexity to transducers

1.5D array



2D array

xMATRIX array technology utilizes 2400 fully-sampled elements for 360-degree focusing and steering.



Live Volume imaging allows the acquisition and rendering of full volume data at true real-time frame rates with unparalleled isovoxel resolution.

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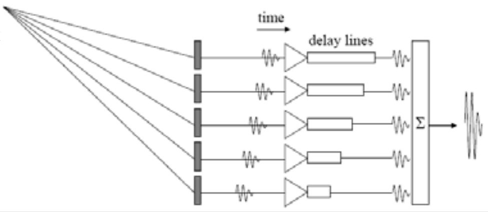
Receiver Beam-Forming

- In a phased array transducer, the effective focal length and aperture of the transducer can also be changed dynamically while the signal is being received → 'receiver beam-forming'
- This is a reverse of dynamic focusing
- Summation of the signals from focal point without time delays will have partial destructive interference and signal loss → therefore, after amplification, each signal is delayed by a time specified by the pathlength from the focal point to the transducer. After passing the various delay lines, the signals are now in-phase and are co-added to produce the max signal

Focal point



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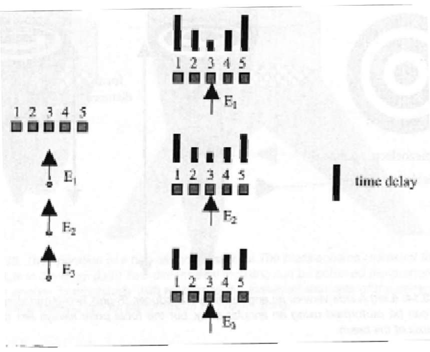
Receiver Beam-Forming

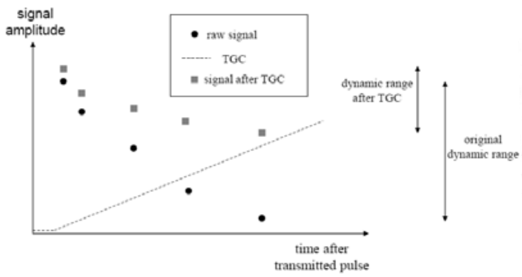
- Three backscattered ultrasound wave E1, E2, and E3 arrive at the transducer at surface different times
- As the first echo(E_1) reaches the transducer, time delays for the voltages from elements 1-5 are introduced to produce the best lateral resolution at the depth at which E_1 was formed.
- Values of the time delays are dynamically varied to optimize the lateral resolution for echoes E_2 and E_3

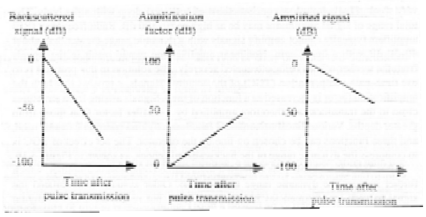


Time gain compensation

- The received signals have large range of amplitudes: very strong signals from reflections at fat/tissue boundaries close to the transducer, and very weak signals from soft tissue/soft tissue boundaries in a deep position
- The total range can be ~100dB
- The amplifier has linear dynamic range for 40~50dB → the weak signals can be lost
- What can we do?
- Apply time gain compensation (TGC) to reduce the dynamic range of the signals
- TGC is the slope of the graph of amplifier gain vs. time (dB/sec)

Time gain compensation

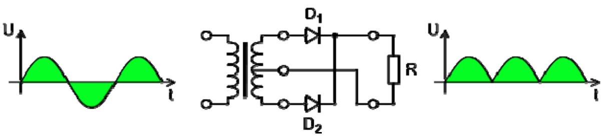




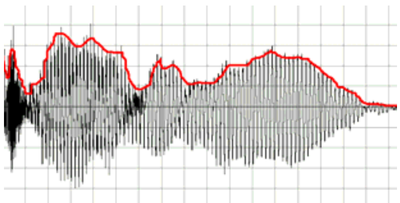
The effects of TGC in reducing the dynamic range of the signals received from close to the transducer surface and deep in tissue

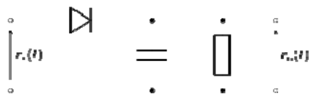
Rectification

- The final step before digitization of the signals is rectification: i.e. to take the complex waveform and transform it into a magnitude-mode signal
- Rectification is performed via a quadrature demodulator followed by envelope detection



Full-wave rectifier using a center tap transformer and 2 diodes.

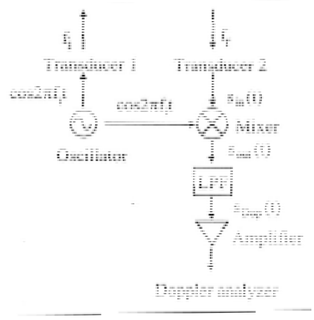




Envelope demodulator circuit

Continuous wave Doppler

- Homodyne demodulation scheme:
  $S_{in}(t) = Acos[2\pi(f_i + \Delta f)t]$
  $S_{out}(t) = Acos[2\pi(f_i + \Delta f)t]\cos(2\pi f_i t)$
  $S_{out}(t) = \frac{1}{2}A\{cos[2\pi(2f_i + \Delta f)t] + \cos(2\pi\Delta f t)\}$
- The signal is then pass through a low-pass filter with a cutoff frequency f_{co} given by $\Delta f << f_{co} << f_i$
  $$S_{Dop}(t) = \frac{1}{2}A\cos(2\pi\Delta f t)$$



- Then the signal passes through a high-pass filter (50~1000Hz) to remove high-intensity reflected signals from the relatively slow movement of vessel walls during the cardiac cycle.
- The final signal is amplified, digitized and stored.
- Fourier transform the time domain signal gives the frequency spectrum, corresponding to the range of blood velocities

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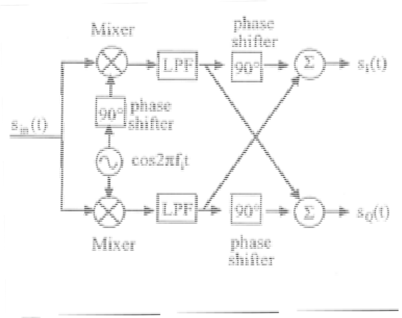
Continuous wave Doppler

- Quadrature or Heterodyne demodulation: to overcome the problem of homodyne detector (there is a directional ambiguity in the output signal)
- Ex) when there is blood flow away from the transducer, Δf will be replaced with - Δf, which will give
  $$S_{Dop}(t) = \frac{1}{2}A\cos(-2\pi\Delta f t)$$



- However, cos(t)=cos(-t) and thus we don't know the direction of the blood flow.
- Therefore, a quadrature or heterodyne detection is used.
- Positive Doppler shift: $S_I(t) = A\cos(\Delta f t)$ and $S_Q = 0$.
- Negative Doppler shift: $S_I(t) = 0$ and $S_Q = A\cos(\Delta f t)$

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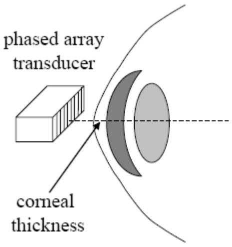
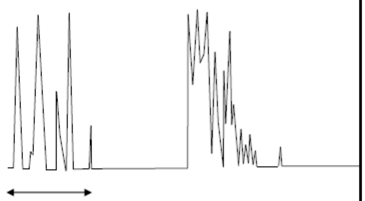
Clinical diagnostic scanning modes

- A mode scanning: acquires a one-dimensional 'line-image' which plots the amplitude of backscattered echo vs time using a small, high frequency (10-20MHz) probe
- A major application is opthalmic corneal pachymetry
- Corneal pachymetry is used in glaucoma, corneal transplants and refractive surgery

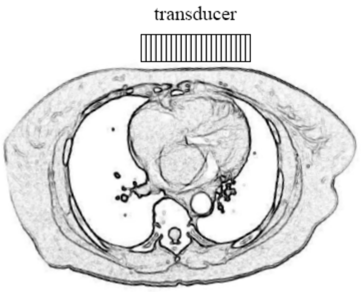
A typical ultrasound pachymeter
(source: wikipedia)

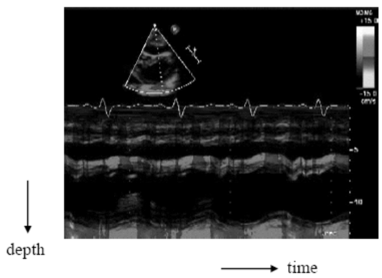




Clinical diagnostic scanning modes

- Motion (M)-mode echocardiography acquires a continuous series of A-mode lines and displays them as a function of time
- The brightness represents the amplitude of the backscattered echo
- Several thousands of lines can be acquired per second
- Mostly used in cardiac and fetal cardiac imaging



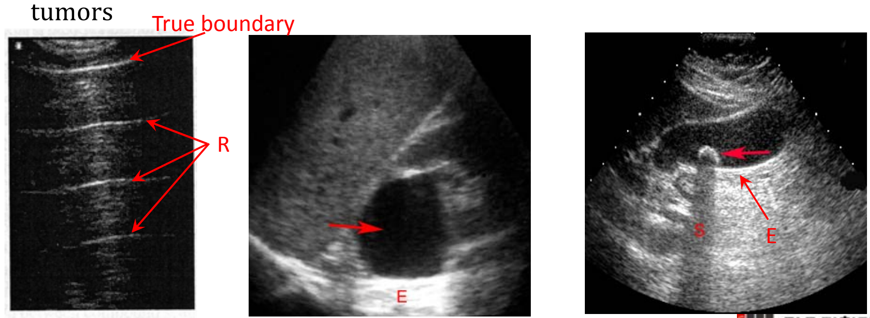


Clinical diagnostic scanning modes

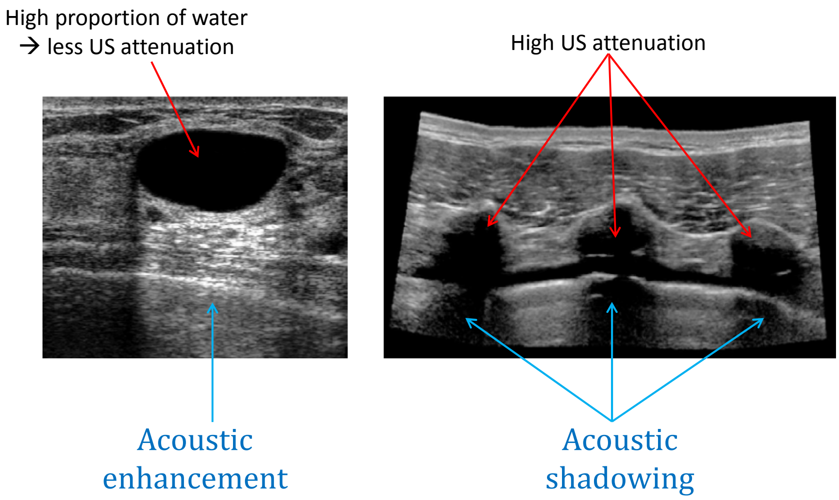
- 2D brightness (B)-mode scanning: most commonly used scanning in clinical diagnosis
- Each line in the image is an A-mode line

Image Artifacts

- Reverberation: occurs if there is a very strong reflector close to the transducer. Multiple reflections occur between the surface of the transducer and the reflector. It occurs when the ultrasound beam encounters bone or air.
- Acoustic enhancement: occur from low attenuation (high proportion of water such as cysts) to surrounding tissue
- Acoustic shadowing: occur from highly attenuating medium such as solid tumors

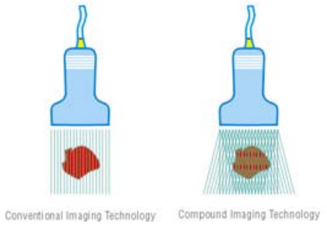


Artifacts in Ultra Sound Imaging



Compound scanning

- Compound scanning acquires images from multiple angles and combine them together
- Advantages: 1) it reduces the speckle in the image, 2) it can show the boundary or structure which are parallel to the beam, 3) it can reduce artifacts such as acoustic enhancement or shadowing ← possible by using multi-angle scanning
- Disadvantage: takes much longer time to image since it scans from multiple angles



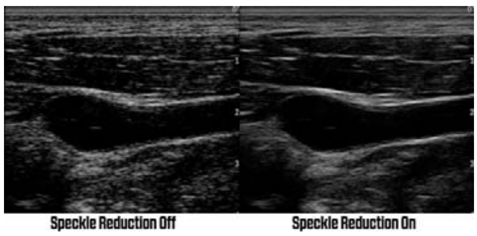


Image characteristics

- Signal to noise
  - The signal intensity is affected by
  1) The transmitted ultrasound intensity
     higher intensity → higher signal
  2) The operating frequency
     higher frequency → higher attenuation at deep tissue
     → lower signal from the large depth
  3) The type of focusing used
     better focusing at one point → higher signal from that point → higher SNR → lower signal at out of focus

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Image characteristics

- Spatial resolution
  1) Lateral resolution
  - single element transducer: higher focus → better lateral resolution → but causes a reduced depth-of-focus
  - Higher frequency → better lateral resolution for both single and phased array transducers
  2) Axial resolution
  - Axial resolution is the half of the length of ultrasound pulse
  - Higher damping, higher frequency → better axial resolution

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Image characteristics

- Contrast-to-noise

  Factors affecting SNR also affects CNR.
  Noise sources such as clutter and speckle reduce the image CNR