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Ultrasound Imaging System

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Today's Topics

- History
- What is Ultrasound?
- Physics of ultrasound
- Ultrasonic echo imaging
    - Focusing technique
    - A-mode signal and B-mode image
    - Features of echo image
- Transducers Design and Modeling
- Applications and Transducers

History

- 1877
  - Discovery of piezoelectricity (Pierre and Jacques Curie)
- 1913
  - First sonar patent filed after Titanic disaster
- 1917
  - Sonar used for detecting range of u-boats during WWI
  - Developed by the French government (Langevin – student of Curies)
  - Hydrophone hung over side of ship
- 1929-1935
  - Use of ultrasound waves in detecting flaws in metals (Sokolov -USSR)
  - First patent for using ultrasound waves to detect flaws in solids (Mulhauser, 1931)
- 1930s
  - Ultrasound used for physical therapy for Europe’s football teams
  - Ultrasound used for sterilization of vaccines and for cancer therapy
- 1940s
  - Ultrasound was seen as a “cure-all” therapy tool
  - Used for arthritis, gastric ulcers, and eczema

History

- Post WWII
  - Surplus Naval sonar equipment used for medical applications
  - Based on radar and sonar techniques
  - Japanese led the development of medical sonography in 1940s
    - Pulse-echo measurements on oscilloscopes
    - Detection of gallstones, breast masses, and tumors
  - Austrian group generated images of brain tumors and cerebral ventricles through skull
  - Researchers from US Naval Medical Research Institute imaged gallstones
- 1950s
  - Simple 2D imaging devices developed by researcher in US and Japan
    - Some were as large as a room
  - The profession “sonographer” was created by the AMA
  - Echocardiography (Sweden)
  - Doppler measurements of tissue motion and blood flow (Japan, US)
  - Focused ultrasound ablation (Fry, U. Illinois Urbana)

History

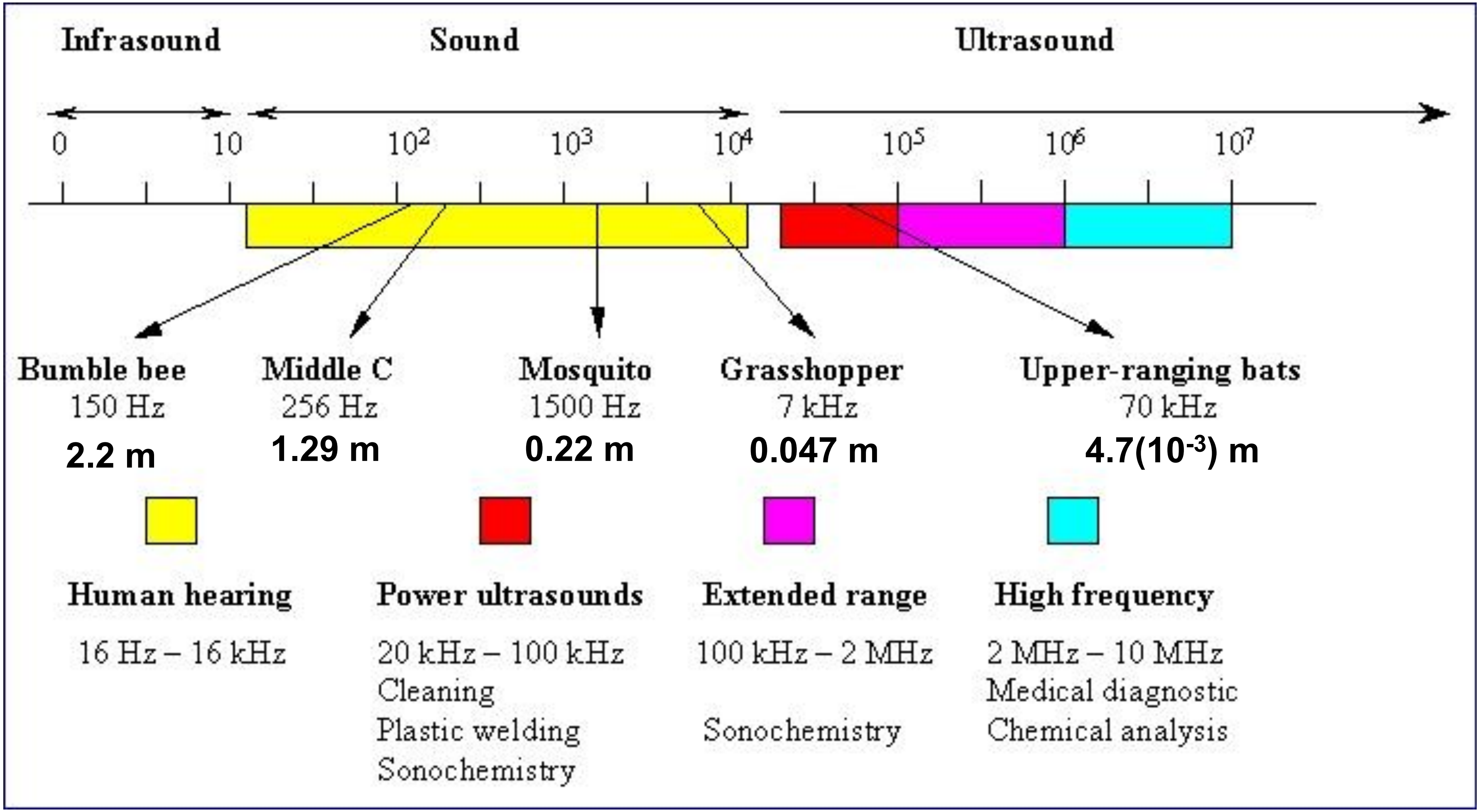
- 1970s – “Sonic Boom”
    - Medical sonography became accepted for several clinical applications
        - Sector scanners, arrays
        - Static 2D grayscale images, then real-time images
        - CW and PW Doppler
        - Fetal heart monitoring
    - Dept. Education defined curriculum for sonographers
    - Growth of NDT
- 1980s
    - Maturation of NDT
    - Real-time ultrasound
        - Greatly facilitated practical use of ultrasound
        - Operator could better recognize what they were looking at
    - Color Doppler for viewing tissues and fluids in motion
    - Higher resolution
    - Smaller probes and systems

History

- 1990s
  - Intravascular ultrasound (IVUS)
  - Early 3D and 4D ultrasound imaging systems
  - First commercial HIFU system
- 2000s
  - Portable ultrasound systems
  - MR-guided focused ultrasound
- Today
  - Diagnostic ultrasound is second most popular medical imaging modality after X-ray

What is Ultrasound?

Sound Spectrum

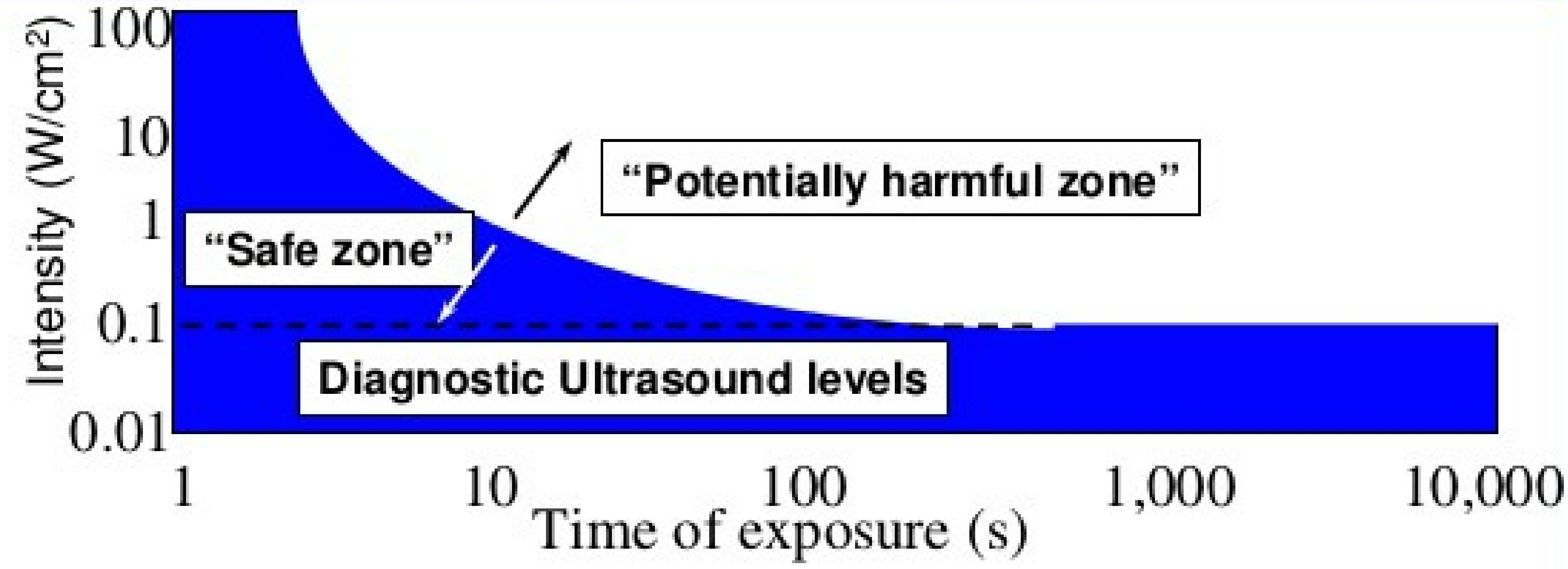


Bumble bee	150 Hz	**2.2 m**
Middle C	256 Hz	**1.29 m**
Mosquito	1500 Hz	**0.22 m**
Grasshopper	7 kHz	**0.047 m**
Upper-ranging bats	70 kHz	**$4.7(10^{-3})$ m**

🟨	**Human hearing**	16 Hz – 16 kHz
🟥	**Power ultrasounds**	20 kHz – 100 kHz Cleaning Plastic welding Sonochemistry
🟪	**Extended range**	100 kHz – 2 MHz Sonochemistry
🟦	**High frequency**	2 MHz – 10 MHz Medical diagnostic Chemical analysis

Ultrasound Safety

- High intensity ultrasound causes heating
- Could damage body tissues
- Low intensity ultrasound is always used for diagnostics ultrasound



Ultrasound

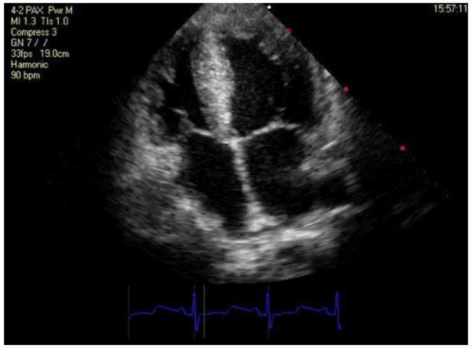
- Acoustic waves are mechanical pressure waves
- Ultrasound waves are pressure waves that travel through a medium at a frequency greater than 20 kHz
- Humans
  - Can typically hear frequencies between 20 Hz to 20 kHz
  - Children can detect higher frequencies than adults
- Animals
  - Many animals can detect higher frequencies
    - Dogs – up to 22 kHz
    - Fish – up to 180 kHz
  - Other animals detect lower frequencies
    - Infrasound – below 20 Hz
- Attenuation vs Resolution
  - Higher frequency has smaller wavelength $c = f\lambda$
    - Better spatial resolution
  - Higher frequency waves degrade faster with distance
  - Trade-off between penetration depth and spatial resolution

Basics of Ultrasound

- Propagation of ultrasound waves are defined by the theory of *acoustics*
    - Ultrasound moves in a wavelike fashion by expansion and compression of the medium through which it travels
    - Ultrasound waves travel at different speeds depending on material
    - Ultrasound waves can be absorbed, refracted, focused, reflected, and scattered.

Basics of Ultrasound

- Process Overview
  - Transducer (electrical signal ⇄ acoustic signal) generates pulses of ultrasound and sends them into patient
  - Organ boundaries and complex tissues produces echoes (reflection or scattering) which are detected by the transducer
  - Echoes displayed on a grayscale anatomical image
    - Each point in the image corresponds to an anatomical location of an echo-generating structure
    - Brightness corresponds to echo strength

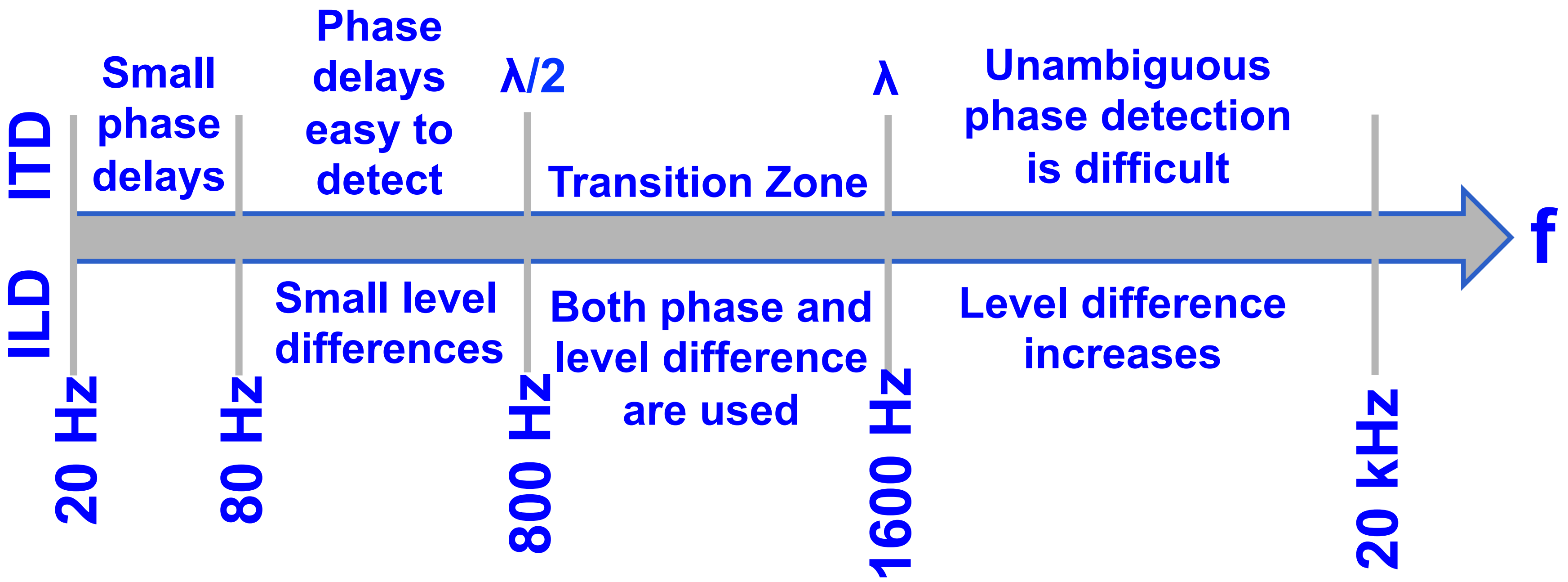


Echolocation

- “Biosonar” or “Active navigation”
- Animals emit sounds and listen for echoes
    - Used to navigate or to hunt
    - Bats, toothed whales and dolphins, shrews, and cave-dwelling birds use biosonar
    - Ultrasound, audible, and infrasound frequencies
    - Many other animals use “passive” biosonar
- Humans
    - Listening is equivalent to passive biosonar

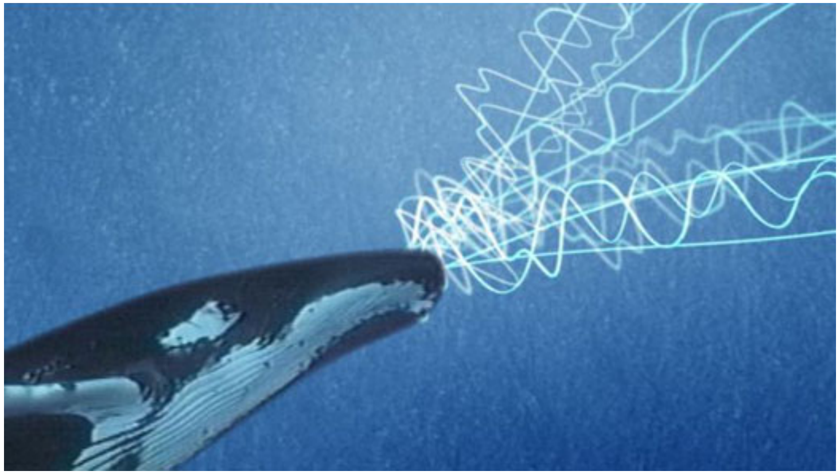
Sound Localization

- Localization determined by interaural space
  - Interaural time difference (ITD)
    - Depends on head size
  - Interaural sound level difference (ILD)
    - Head shadows
    - Frequency dependent



Sound Localization

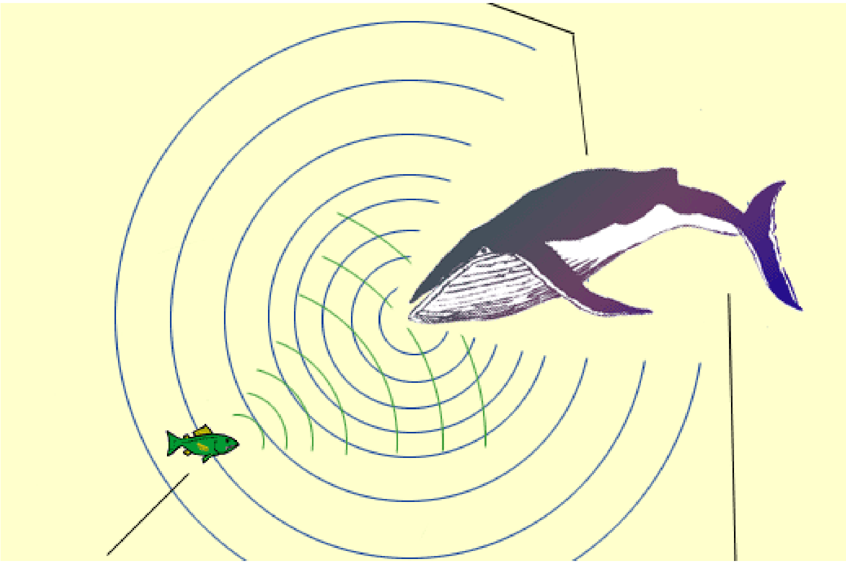
- Why do animals communicate within different audible frequency ranges?
  - Head size impacts audible frequency range of animals
    - Also affected by ear position and movement
    - Larger animals utilize lower frequencies
  - Larger animals communicate over longer distances
    - Lower frequencies have less acoustic loss with distance
  - Smaller animals need to resolve smaller objects
    - Higher frequencies have better spatial resolution



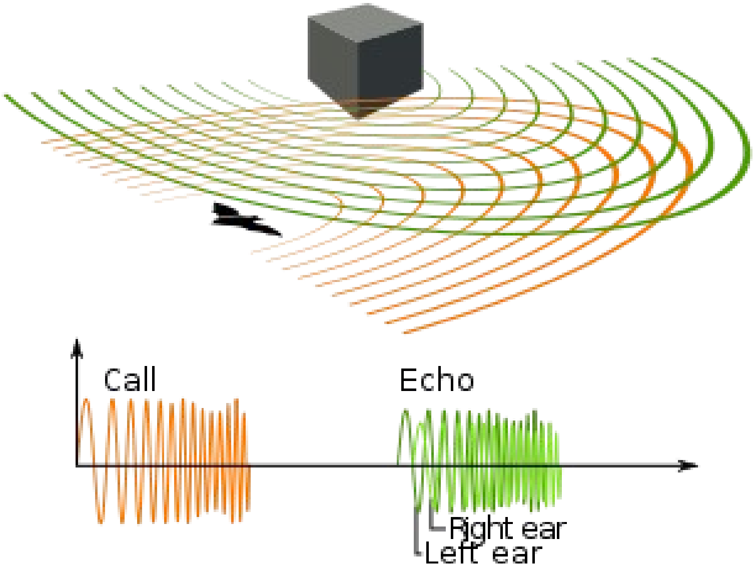
Sound Localization

- Range Detection
  - Pulse-echo ranging

$$R = \frac{c\Delta t}{2} \qquad \Delta x = \frac{c\Delta t}{2}$$

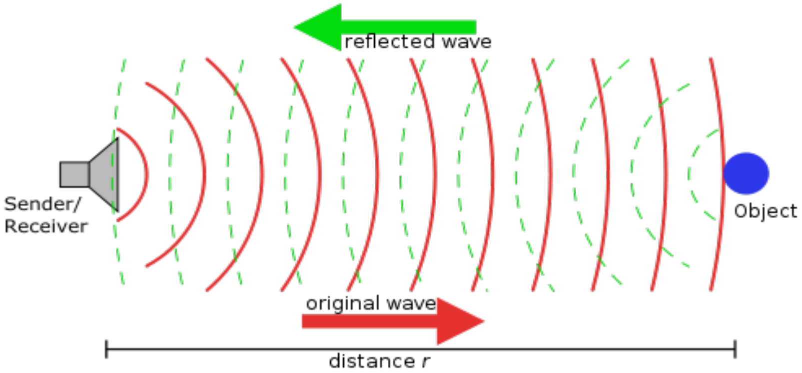


- Localization
  - Difference in arrival time to ears
  - Difference in sound level between ears
  - Interaural space

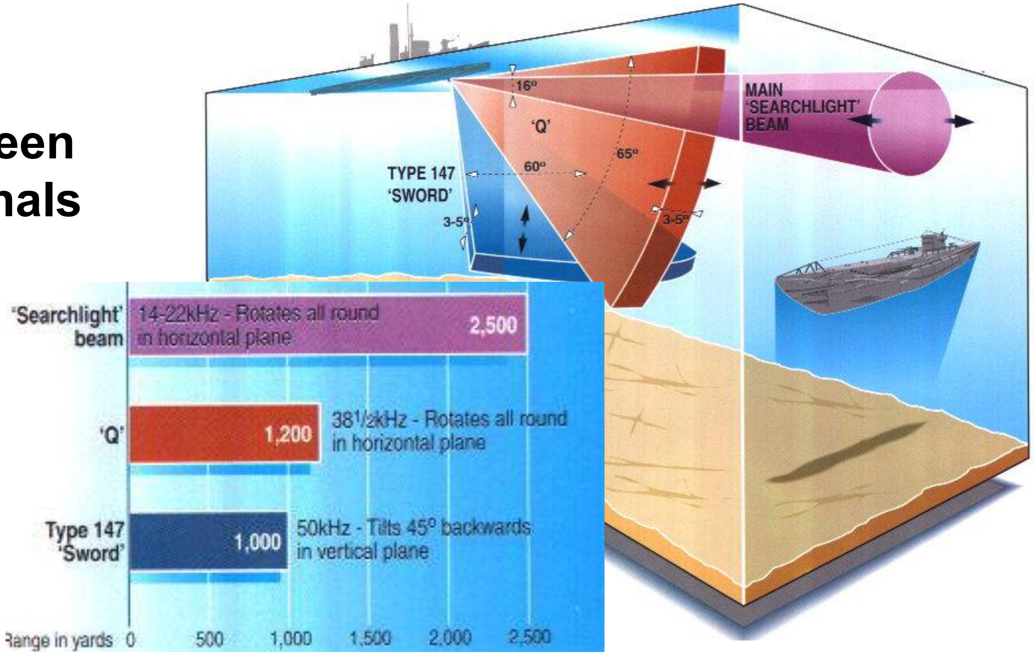


Sonar

- SOund Navigation And Ranging
- Distance
  - Pulse-echo ranging
- Bearing
  - Similar to Localization
  - Relative arrival times measured
    - multiple hydrophones or array
- Speed
  - Doppler effect
  - Difference in frequency between transmitted and received signals
    - Converted to velocity
  - Speed of transmitter must be accounted for
- Several sonar beams used



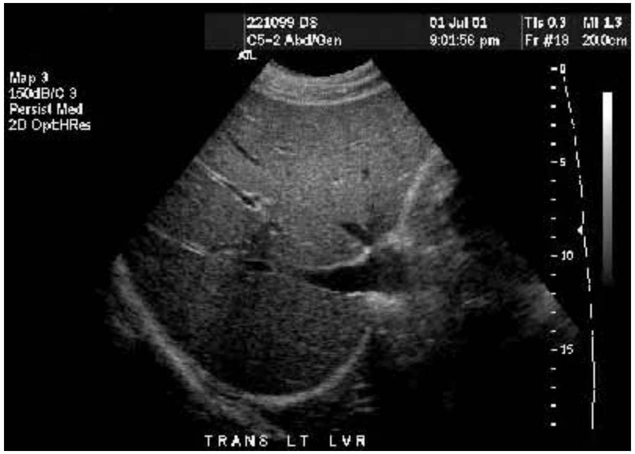
Pulse-echo ranging



Sonar

Uses of Ultrasound

- Ultrasound Imaging / Detection
  - Medical Sonography
    - 3-20 MHz
  - Sonar
    - Hz–kHz range
  - Non-destructive testing (NDT)
    - kHz–low MHz range
    - Detection of cracks in materials



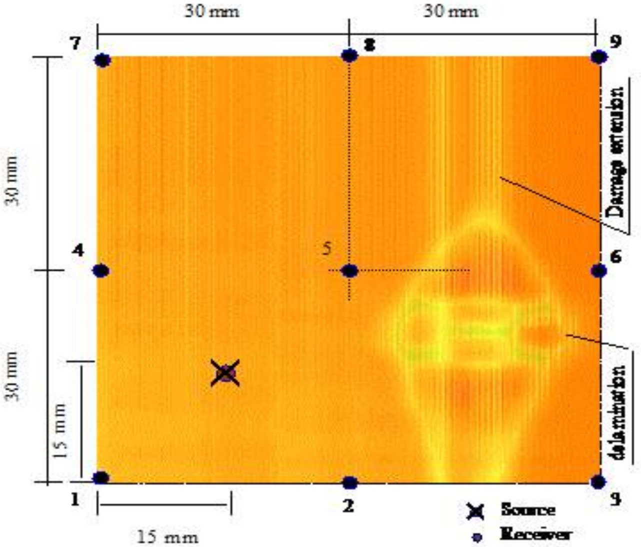
Medical sonography

Non-destructive testing

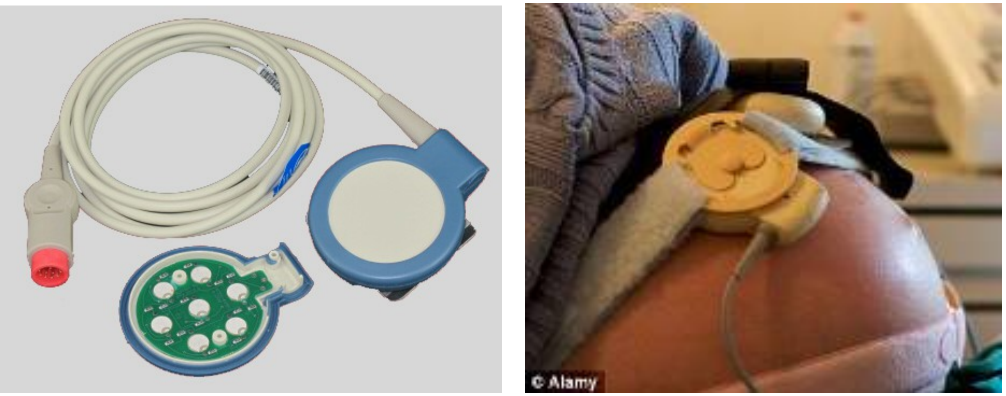
Uses of Ultrasound

- Monitoring
  - Structural Health Monitoring
    - Long term damage detection
    - Infrastructure, aircraft
    - Embedded sensor networks
    - kHz-low MHz range
  - Fetal Heart Monitoring
    - Continuous detection and monitoring of fetal heart beat
    - Low MHz range

Sensor network for SHM



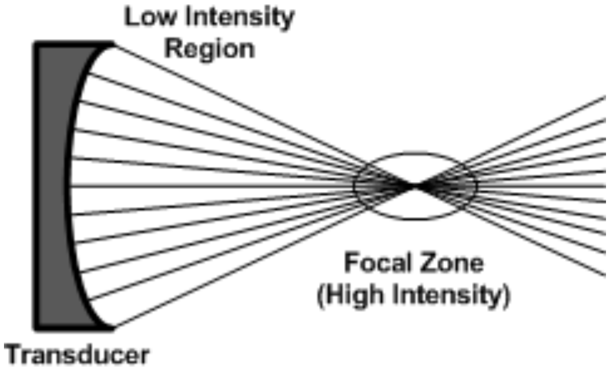
Damage detection with sensor network



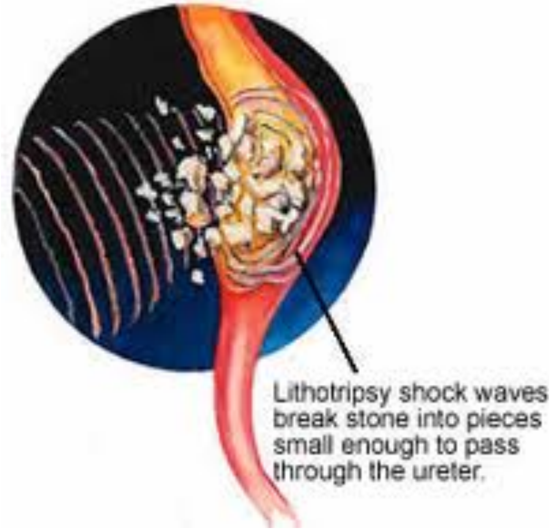
External fetal heart monitoring

Uses of Ultrasound

- Ablation/Destruction of Tissues
    - Lithotripsy – ablation of kidney stones
    - Uterine fibroids (FDA approved)
    - Tumor ablation
    - MRI or Ultrasound guided
- Ultrasound Hyperthermia Treatment
    - Low level heating (<45 °C)
    - Combined with radiation/chemotherapy
- Other Uses
    - Drug activation (Focused heating of drugs)
    - Vibration (Wire bonding)
    - Tissue cutting and hemostasis (Harmonic scalpel)
    - Water treatment



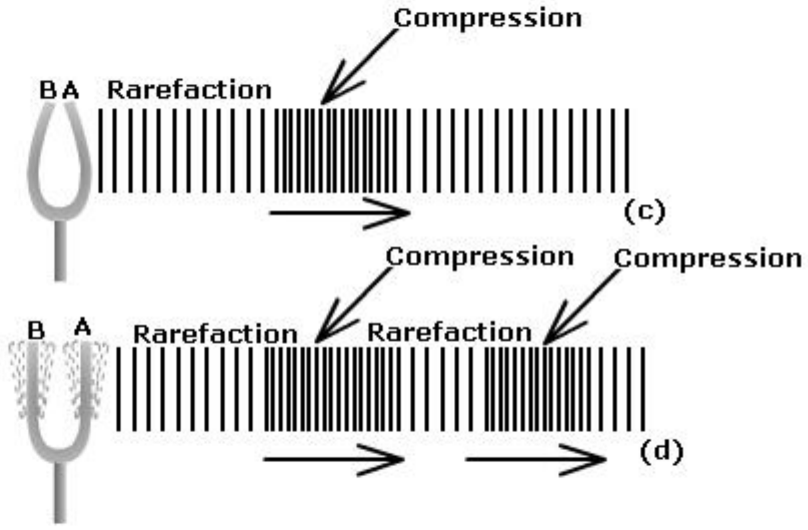
Focused ultrasound

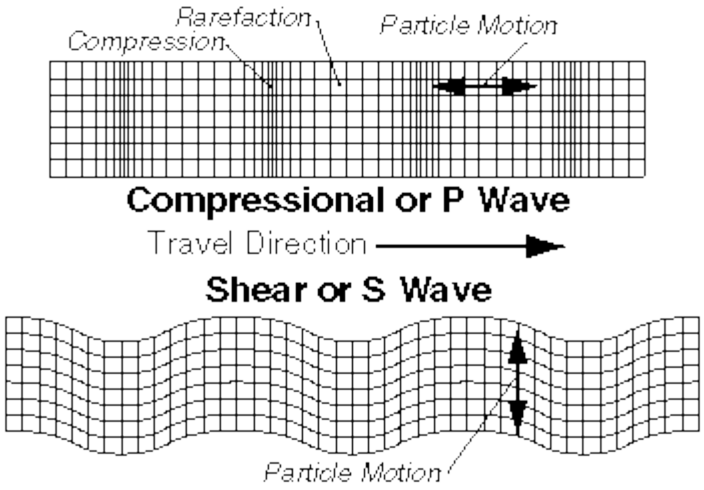


Lithotripsy

Physics of Ultrasound

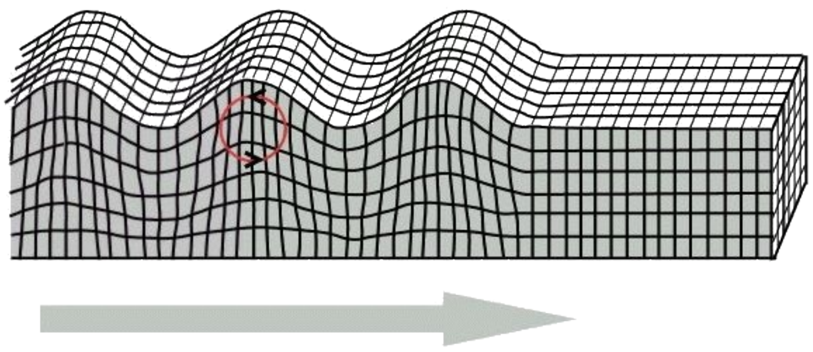
Wave Propagation

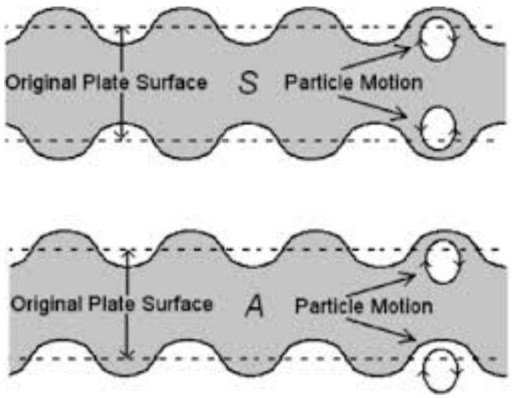




- Compressional (Longitudinal) Waves
  - Oscillation occurs in direction of wave propagation
    - Zones of compression & rarefaction
    - Speaker analogy
  - Used in medical sonography
- Shear (Transverse) Waves
  - Oscillation occurs normal to direction of wave propagation
    - Cork bobbing in water analogy
  - Only supported by hard tissues
  - Important in transducer design and NDT

Wave Propagation

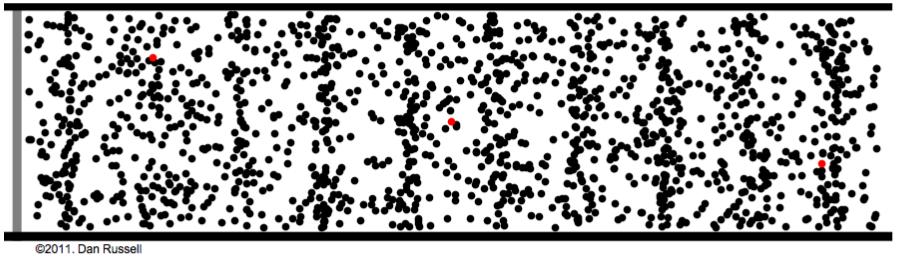




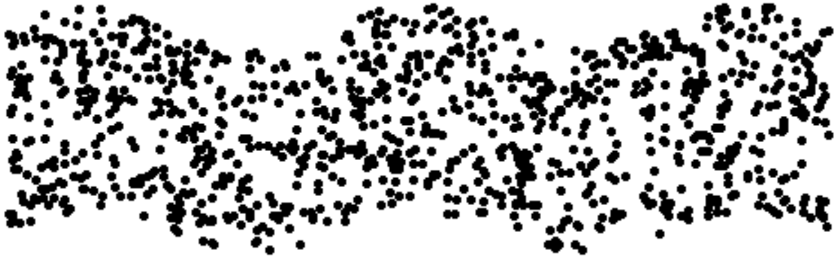
- Surface (Rayleigh) Waves
  - Travel along surfaces of hard materials, up to 1 λ depth
  - Elliptical motion, combines compressional & shear motion
- Lamb (Guided) Waves
  - Travel within thin plates or layers
  - Important in NDT

Wave Propagation Animation

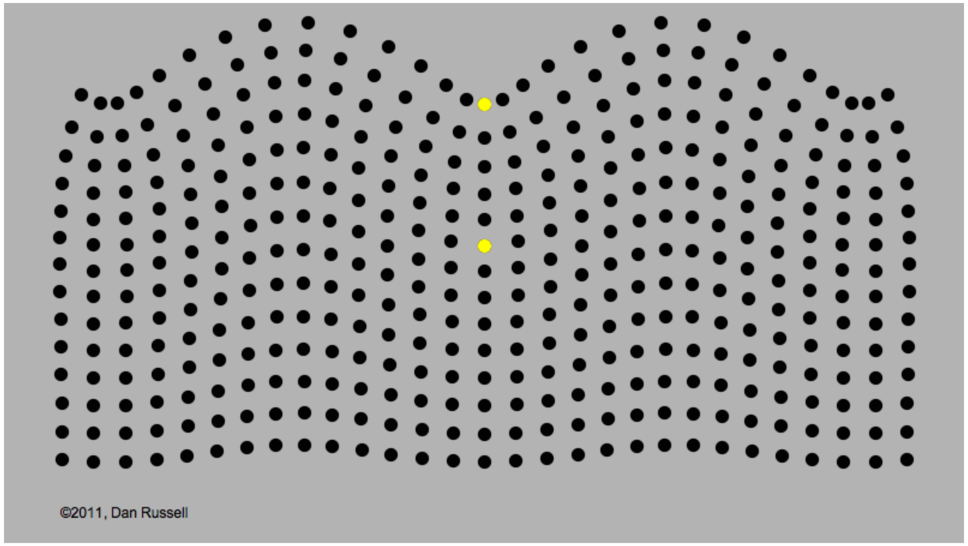
Compressional or Longitudal Wave



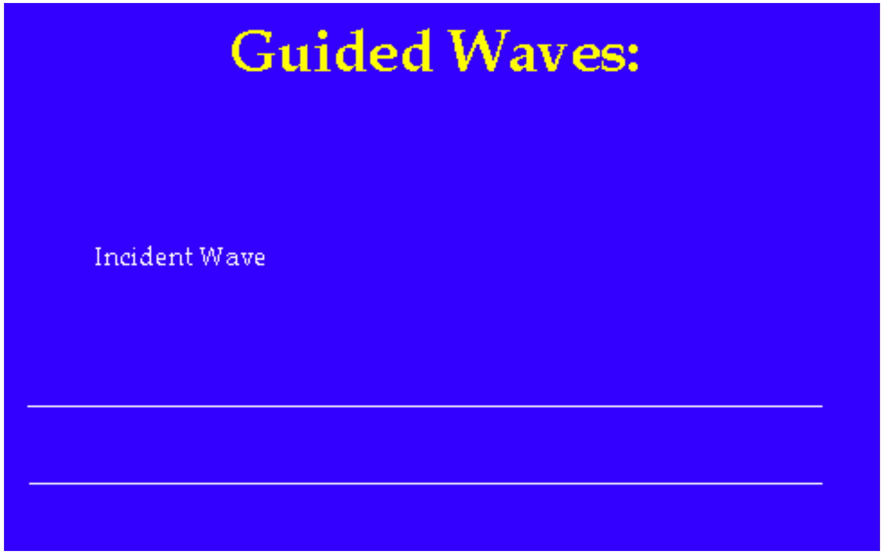
Shear or Transverse Wave



Rayleigh or Surface Wave

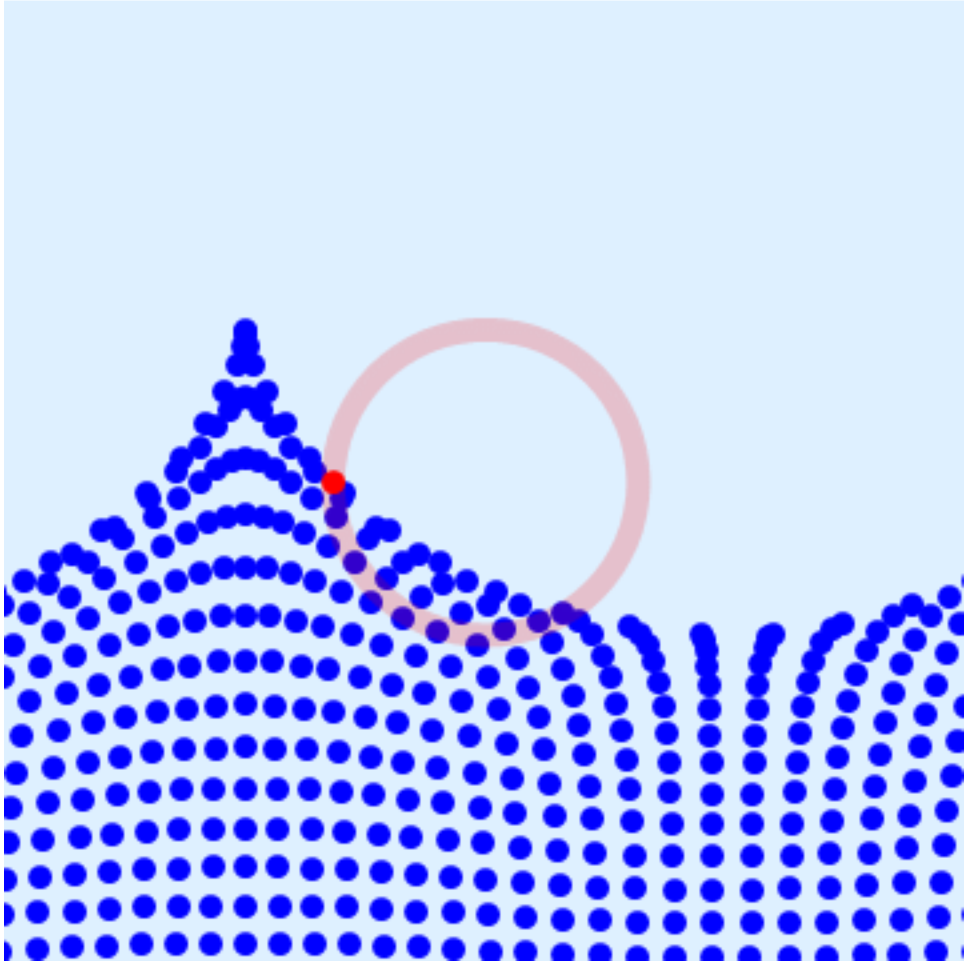


Lamb or Guided Wave



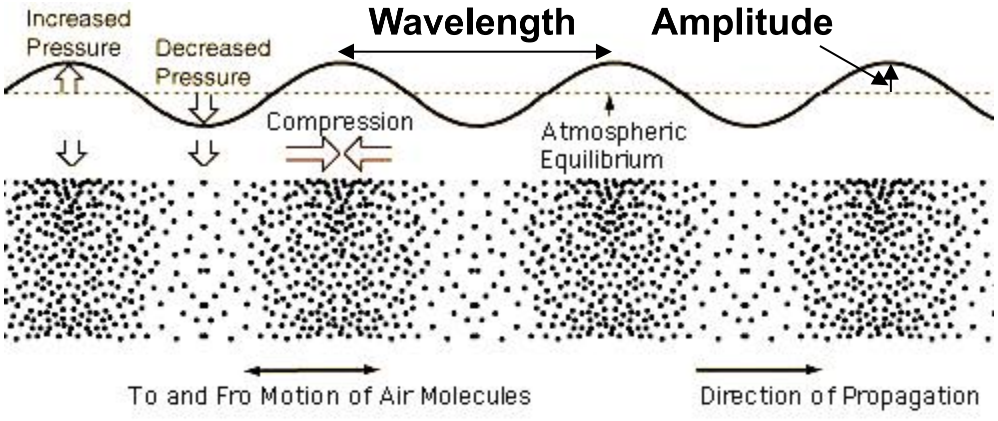
Animation courtesy of Prof. Daniel Russell, Penn State University, and NDT.net

Particle Velocity and Phase Velocity



Anatomy of a Wave

- Amplitude
  - Change in magnitude
  - Units of pressure (Pa or N/m^2)
- Wavelength (λ)
  - One complete wave cycle
  - Unit of distance (m)
  - ½ λ, ¼ λ thicknesses are important in acoustics
    - Constructive/Destructive interference
- Frequency (f)
  - Number of vibrations that a molecule makes/second
  - Unit of cycles/s (Hz)
- Period ($T = 1/f$)
  - Elapsed time between compression zones
  - Units of time (s)



Speed of Sound

- Often called velocity, but is a scalar value
- Depends on medium
  - Air = 330 m/s
  - Water = 1480 m/s
  - Average Soft Tissue = 1540 m/s
  - Bone = 4080 m/s
  - Steel = 5960 m/s
- Speed of sounds varies slightly with Temperature
  - Water = 1480 m/s at 20°C
  - Water = 1570 m/s at 37°C
- Depends on elasticity of the material through which it travels
- Particle velocity
  - Velocity of individual oscillating particles
- Phase velocity
  - Rate at which the phase of the wave propagates in space
    - Speed of any one frequency component

Speed of Sound

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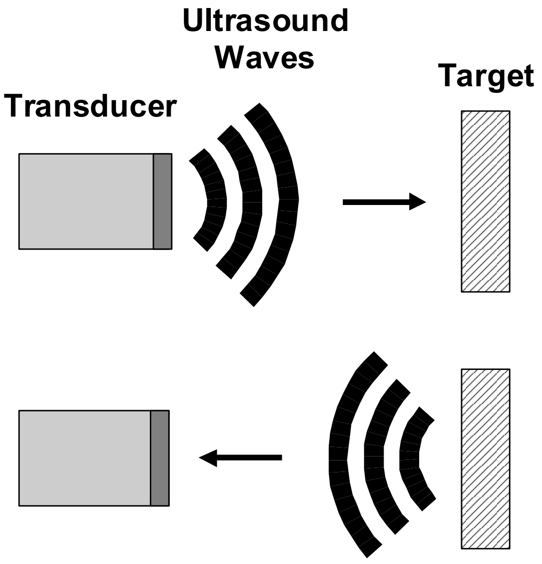


Propagation of compressional waves

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

$$\Delta x = \frac{c\Delta t}{2}$$

- Wave propagation
  - Pressure waves travel thru a medium at a frequency range of 3-20 MHz in medical ultrasonography
  - Transducer transmits and receives ultrasound energy
  - Speed of sound (c) depends on the medium
  - Velocity and round-trip time must be known to measure range (R) or thickness (D_x)

Speed of Sound

- Density (ρ)
  - Mass of the medium/volume
- Compressibility (κ)
  - Decrease in volume when pressure is applied to a material
- Bulk Modulus (β)
  - Stress-strain ratio, under isotropic conditions
  - Similar to the stiffness, or Young’s Modulus (E)
  - $\beta = 1/\kappa$

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

- Change in ρ often associated with larger change in κ
  - Therefore as ρ increases, c generally also increases

Speed of Sound

- Speed is constant, so a change in *f* results in a change in *λ*
- When sound travels from one medium to another, *f* remains constant
  - *λ* must change with changing c
- <u>λ highly influences spatial resolution</u>

$$c = f\lambda \quad \Rightarrow \quad \lambda = \frac{c}{f} = \frac{1}{f}\sqrt{\frac{\beta}{\rho}}$$

Ultrasound Waves in Soft Tissue (c = 1480 m/s)

Frequency (MHz)	Wavelength (mm)	Period (μs)
1	1.54	1
5	0.31	0.2
10	0.15	0.10
20	0.08	0.05

Acoustic Interactions

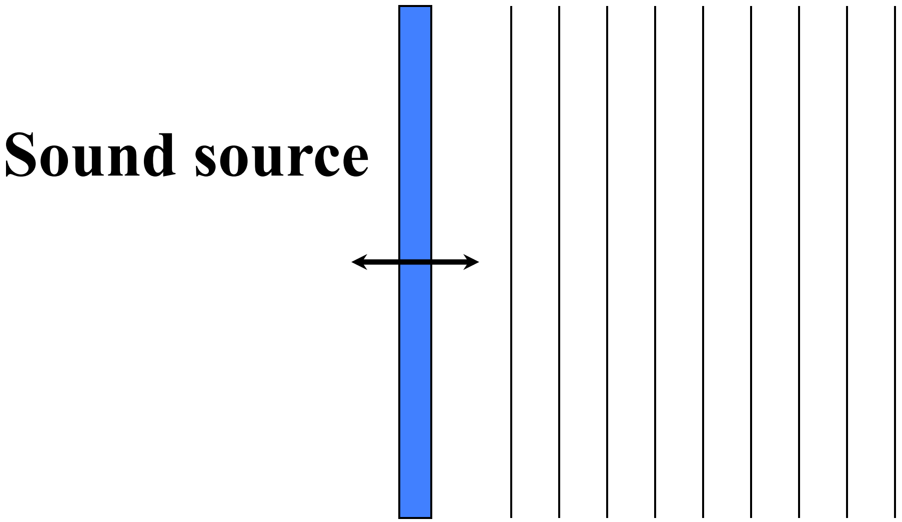
- Mechanisms of acoustic interaction with tissue
  - Reflection
  - Refraction
  - Diffraction
  - Divergence
  - Interference
  - Scattering
  - Absorption
    - All can reduce beam intensity

Physics of ultrasound
- propagation -

- Velocity of propagation
- About 1540[m/s] in human body
- Each tissue has its own velocity.
  - Ultrasonic diagnostic equipment assumes that sound velocity is constant in the body.
    - This assumption causes artifacts in echo image
- Wavelength
  - About 0.437[mm] in the body (3.5MHz)

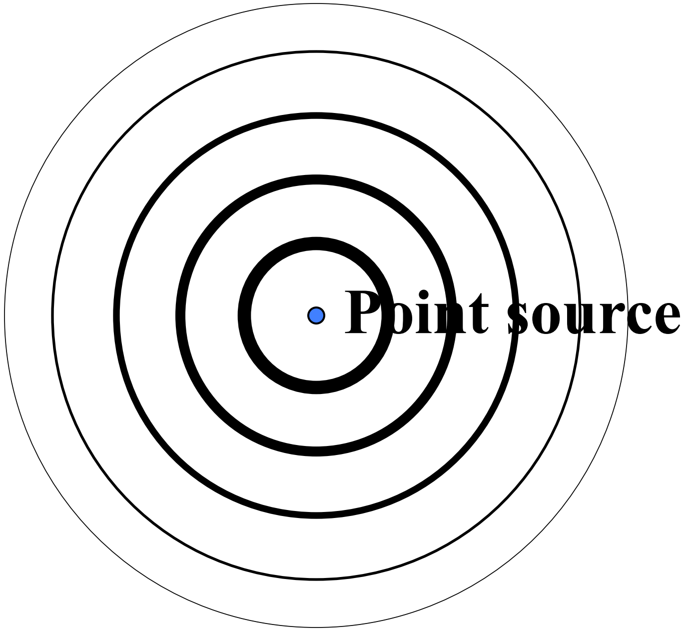
Physics of ultrasound
- propagation -

- Plane wave
  - Line sound source, infinite length
  - No diffusion attenuation



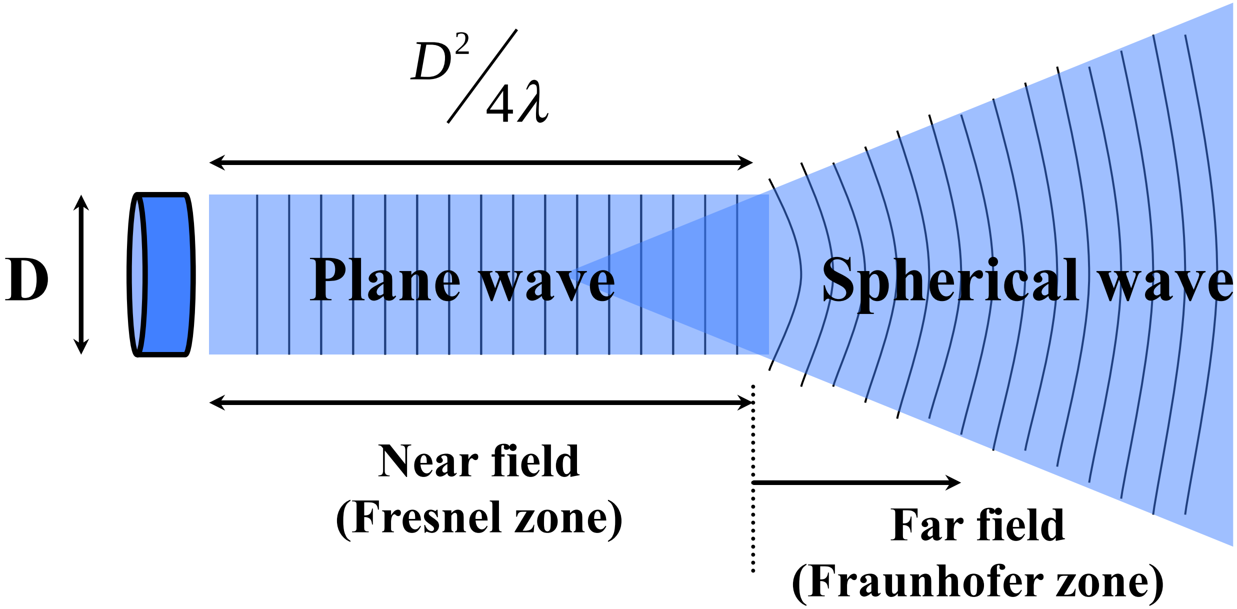
Physics of ultrasound
- propagation -

- Spherical wave
  - Point sound source
  - Diffuse sound field



Physics of ultrasound
- propagation -

- Practical condition –ultrasonic element-
  - Finite element size (about 0.3mm)
  - Not plane wave, not spherical wave

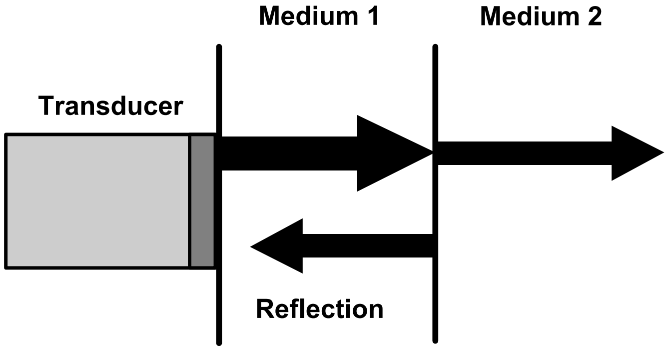


λ: wavelength
= 0.437mm
D:diameter
= 0.3mm

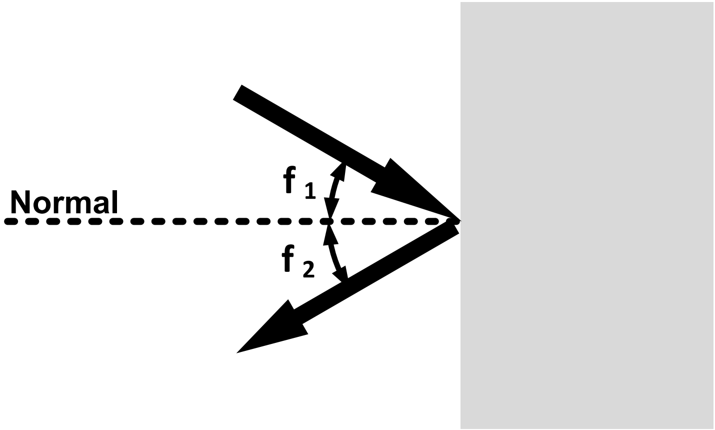
Fresnel zone
= 0.052mm

Reflection

- Normal incidence on a specular reflector
  - A portion of the beam is reflected, a portion is transmitted through the interface
  - Normal reflection



- Angled incidence on a specular reflector
  - Incident angle $\phi_1 = \phi_2$
  - During imaging, it is important to minimize incident angle
    - Maximize probability of detection
    - Best odds when perpendicular
    - This is why scanning is so important



Refraction

- Refraction
    - Beam not normal to interface
    - Transmitted beam bends (refracts) away from normal
    - Reflected beam does not reflect directly back to transducer
    - Refracted beam results in misregistration of object
        - Example: Swimming pool
        - Results in unwanted image artifacts in medical sonography
- Snell's Law
    - Relationship between angle of incidence and refraction

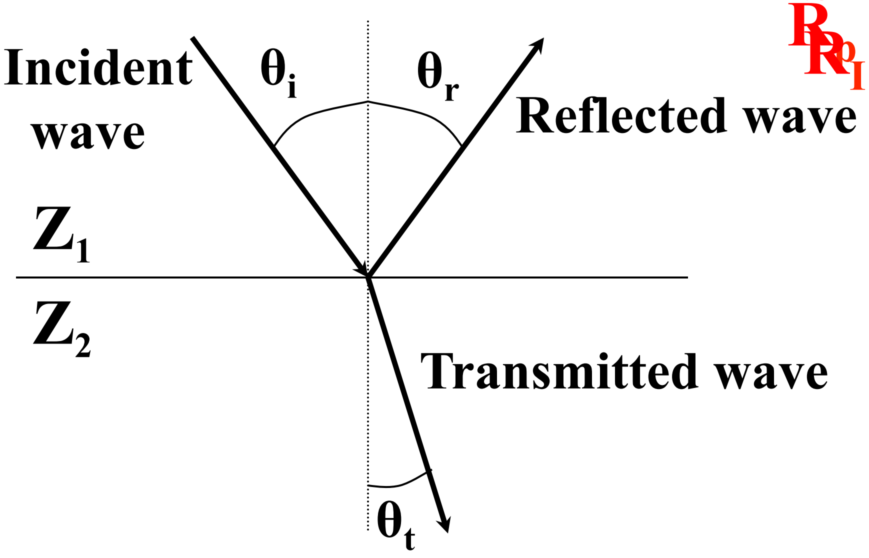
$$\frac{\sin\phi_1}{\sin\phi_2} = \frac{c_1}{c_2}$$

Refraction

- Snell's Law
  - Example 1 ($c_1>c_2$)
    - Bone/Tissue
    - Bends towards the normal
  - Example 2 ($c_1<c_2$)
    - Tissue/Bone
    - Bends away from the normal
  - Example 3 ($c_1<c_2$, $\phi_2\geq 90°$)
    - Critical angle ϕ_c is determined by setting $\phi_2=90°$
    - Total internal reflection
    - Reflected wave travels along surface at $\phi_2=90°$

Physics of ultrasound
- characteristics-

- Reflection and transmission
  - Acoustic impedance : Z = *ρc*
  - *ρ*: density, *c* : sound velocity



For sound pressure

For sound intensity

Snell's law

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

R_p : reflectance, T_p : transmittance

R_I : reflectance, T_I : transmittance

$$R_p = \frac{Z_2\cos\theta_i - Z_1\cos\theta_t}{Z_2\cos\theta_i + Z_1\cos\theta_t}$$

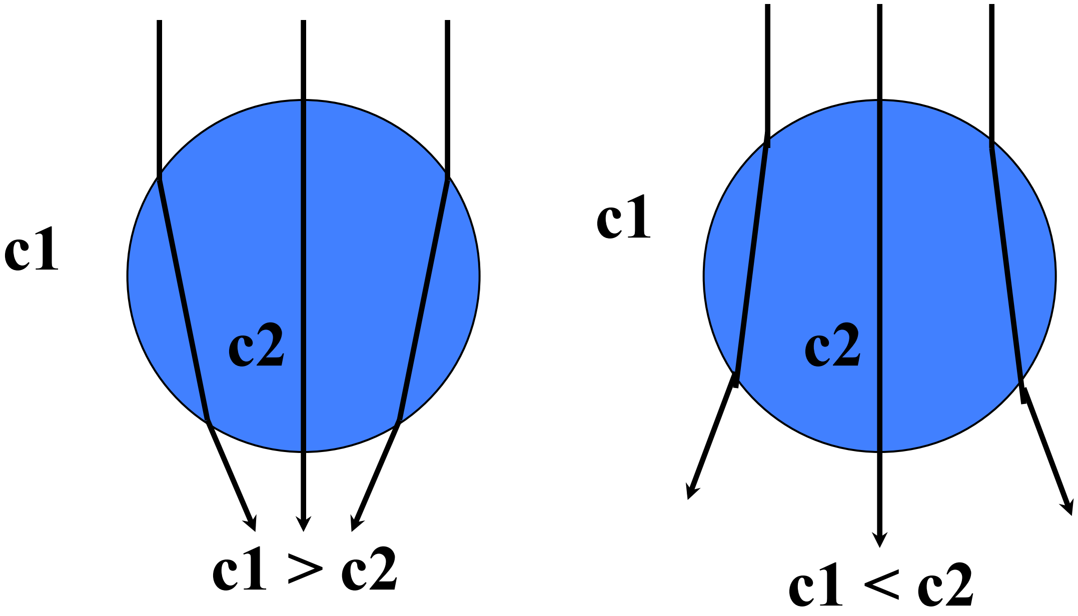
$$R_I = \left|R_p\right|^2$$

$$T_p = \frac{2Z_2\cos\theta_i}{Z_2\cos\theta_i + Z_1\cos\theta_t}$$

$$T_I = 1 - R_I$$

Physics of ultrasound
- characteristics-

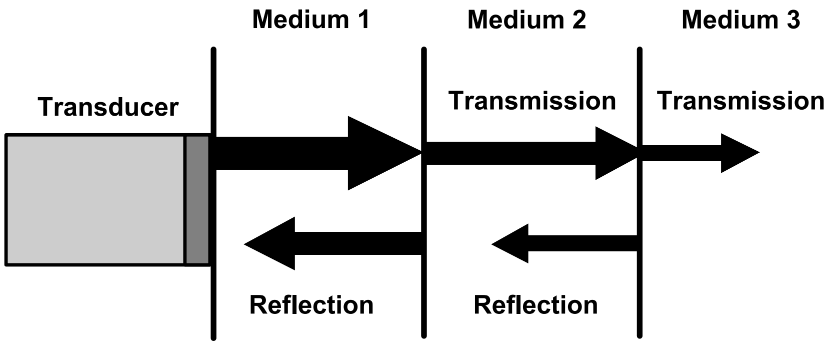
- Refraction (snell’s law)
  - c : sound velocity



This phenomenon causes artifacts in medical echo image.

Acoustic Impedance

- Reflected Intensity
  - Ultrasound energy reflects off of each interface between different media in the path



  - Reflected echo intensity depends on the acoustic impedances of the two adjoining media

Acoustic Impedance

- Acoustic Impedance (*Z*)
  - Resistance to sound passing thru a medium
  - Analogous to electrical resistance
    - Degree of difficulty experienced by electrons passing thru a material
  - Analogous to index of refraction (n) in optics
  - Depends on the density (ρ) and speed (c) of a material

$$Z = \rho c$$

  - Units are kg/m^2/s (rayl)
- Examples of acoustic impedance matching
    - Ear directly against train track
    - Acoustic scanning gel between body and probe

Acoustic Impedance

- Acoustic Coupling
  - Air gaps or bubbles between the transducer and body result in large reflections and prohibitive acoustic losses
  - Liquid or gel coupling must be used to minimize air gaps
  - The acoustic impedance of the couplant must be between that of the transducer and body
  - To minimize reflections, the ideal impedance of the couplant (Z_2) is the square root of the product of the transducer's impedance (Z_1) and the body (Z_3)

$$Z_2 = \sqrt{Z_1 Z_3}$$

Reflection Coefficient (Boundary Interaction)

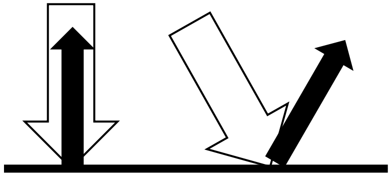
- Reflection Coefficient (Γ)
  - A measure of the fraction of acoustic pressure reflected at an interface
- Transmission Coefficient
  - Transmitted pressure thru an interface = 1- Γ
- Intensity Reflection Coefficient
  - Reflected intensity = Γ^2
- Intensity Transmission Coefficient
  - Transmitted intensity = 1- Γ^2
- All independent of *f* and layer thickness
- Examples
  - Detection of submarine vs. whale
  - Bone vs. soft tissue
  - Lungs and air bubbles
  - Multiple interfaces
    - Transmission into brain

$$\Gamma = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

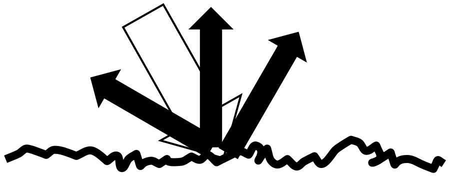
Medium 1	Medium 2	G
Air	Water	0.99
Gel	Skin	0.04
Muscle	Bone	0.65
Salt Water	Steel	0.93

Diffuse Reflection (Boundary Interaction)

- Specular Reflection
  - Reflection off of smooth/flat objects
  - These echoes are relatively intense and angle dependent. (i.e.valves) - Reflection from large surfaces
- Diffuse Reflection
  - Most surfaces are rough
  - Parts of beam are redirected due to in multiple directions
  - Loss of coherence of beam
  - Decreased beam intensity
  - Increased acoustic clutter
  - echoes originating from relatively small, weakly reflective,
- Therefore, Influences on Reflectivity are:
  - Impedance mismatch
  - Angle of incidence
  - Size, shape, texture of structure relative to λ



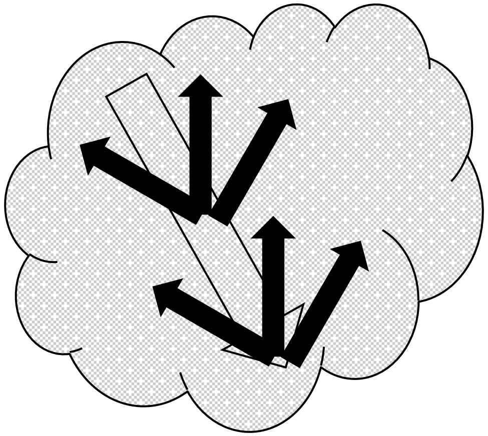
Smooth or Specular Reflection



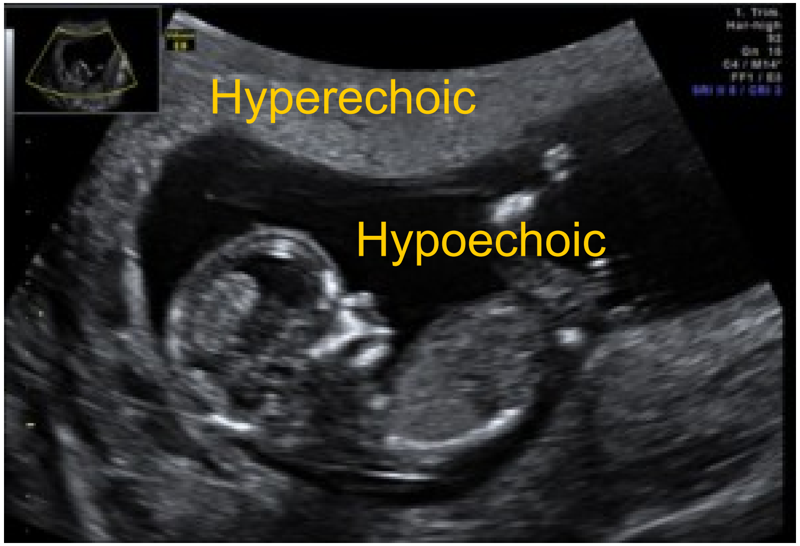
Diffuse or non-Specular Reflection

Scattering (Tissue Interaction)

- **Similar to diffuse reflection, but smaller scale**
- **Rough surfaces or materials with particle dimensions with size ≤ λ**
- **Responsible for internal texture of organs**
- **Dependent on**
  - **Number of scatterers/volume**
  - **Size of scatterers (relative to λ)**
  - **Acoustic impedance Z**
  - **Frequency *f***
- **Hypoechoic**
  - **Dark regions in ultrasound images lack scatterers**
  - **Fluids**
- **Hyperechoic**
  - **Bright regions have many scatterers**

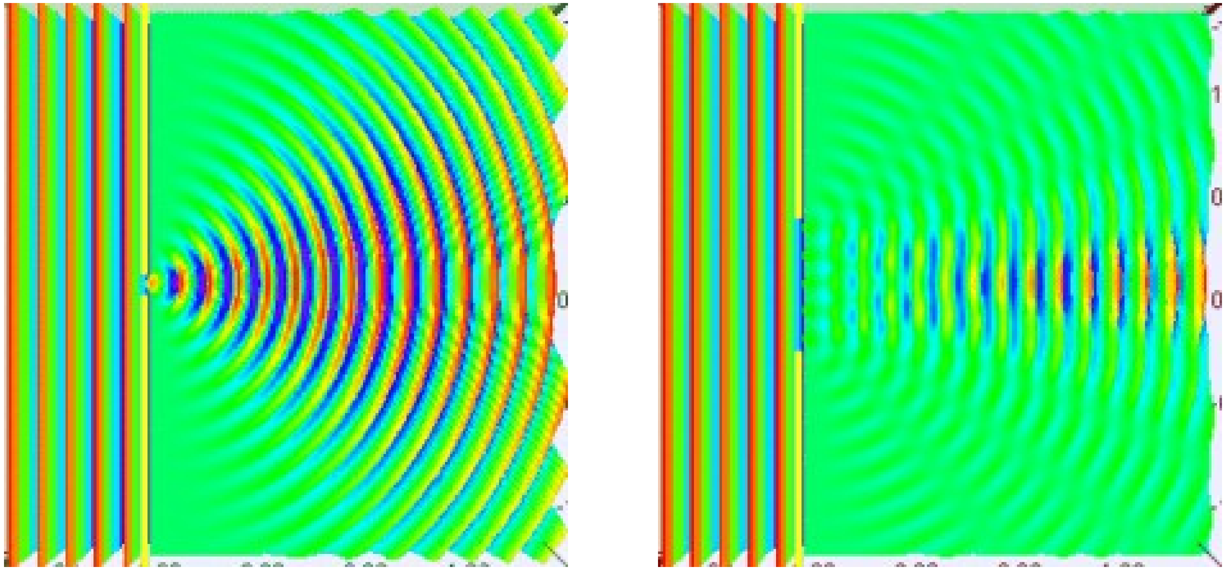


Small object reflections with size ≤ λ



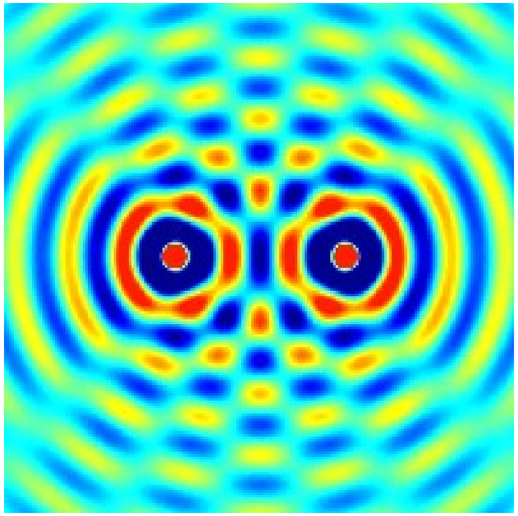
Diffraction & Divergence

- Diffraction
  - Causes divergence of beam
  - Rate of divergence increases as aperture decreases
  - Divergence also occurs as waves pass thru small aperture, close to the size of a λ



Interference

- Superposition of waves
  - When two or more waves are incident on the same point, total displacement is equal to vector sum of the displacements of individual waves
- Constructive Interference
  - All waves in phase
- Destructive Interference
  - All waves out of phase



Reverberation

- Reverberation
  - Persistence of sound in a particular space after the original sound is removed
  - Similar to multiple echoes in a cave
  - Common in cavities with large impedance mismatches
  - Energy is released over time as successive passes are absorbed by adjacent layers
  - Can clutter ultrasound signals

Absorption

- Conversion of ultrasound energy to heat energy
  - Basis of therapeutic ultrasound
  - Only loss mechanism where acoustic energy is dissipated into the medium
  - Other mechanisms cause loss by redirection of the beam
- Depends on *f*, viscosity, and relaxation time of the medium
  - Viscosity
    - Ability of molecules to move past one another
    - More heat is produced with greater resistance to flow (high viscosity)
  - Relaxation Time
    - Rate at which molecules return to their original positions after being displaced by a force
    - More energy is required to counteract molecular movement of a material with a high relaxation time
      - Leads to more heat loss
  - Frequency affects both viscosity and relaxation time
    - Molecules must move more often with higher f

Intensity & Power

- Intensity
  - Amount of energy flowing through a cross-sectional area/ second
  - Rate at which energy is transmitted by the wave over a small area
  - c, f, λ are not affected by I
  - Proportional to the square of the pressure amplitude
    - Instantaneous Intensity

$$I = \frac{p_i^2}{\rho c} = \frac{p_i^2}{Z}$$

Average Intensity

$$I = \frac{p_0^2}{2\rho c} = \frac{p_0^2}{2Z}$$

  - Units of I are W/m^2
- Power
  - Total energy transmitted/unit time summed over the entire cross-sectional area of the beam
  - Intensity x Area

Absorption

- Absorption causes an exponential decrease in pressure
- Absorption coefficient (α) depends on medium and frequency

$$p_0 = p_{\max} e^{\alpha z}$$

Physics of ultrasound
- characteristics-

- Attenuation
  - Diffusion attenuation [dB/m]
    - Inverse proportion to distance from source
  - Absorption attenuation [dB/m/MHz]
    - Frequency dependent attenuation
    - Reflected wave from deep region has lower center frequency and longer wavelength than incident wave.
- Attenuation causes low resolution of echo image.

Attenuation

- Includes both absorption and scattering
- Attenuation coefficient (a) is the sum of the absorption (α) and the scattering coefficient (a_s)
- Expressed in Np/cm or dB/cm or dB/cm/MHz
- Also dependent on frequency and medium

$$p_0 = p_{\max} e^{-(a_s + \alpha) z} = p_{\max} e^{-az}$$

Attenuation

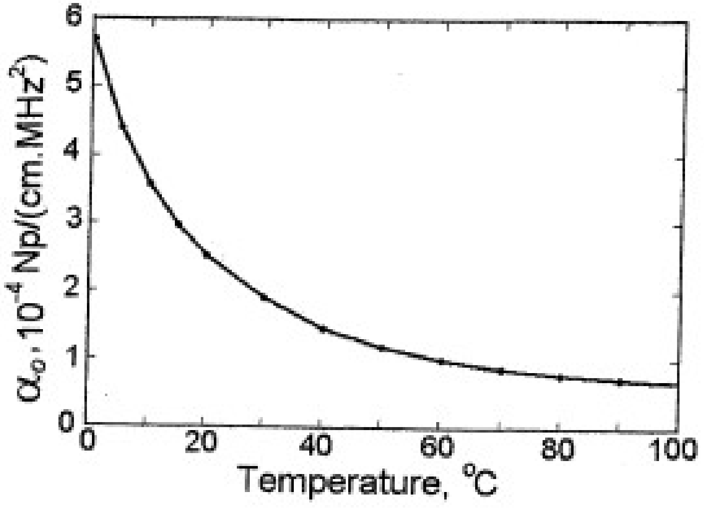
- Can also be expressed in Intensity
  - Also has exponential decrease with distance

$$I_0 = I_{\max} e^{-2(a_s + \alpha)z} = I_{\max} e^{-2az}$$

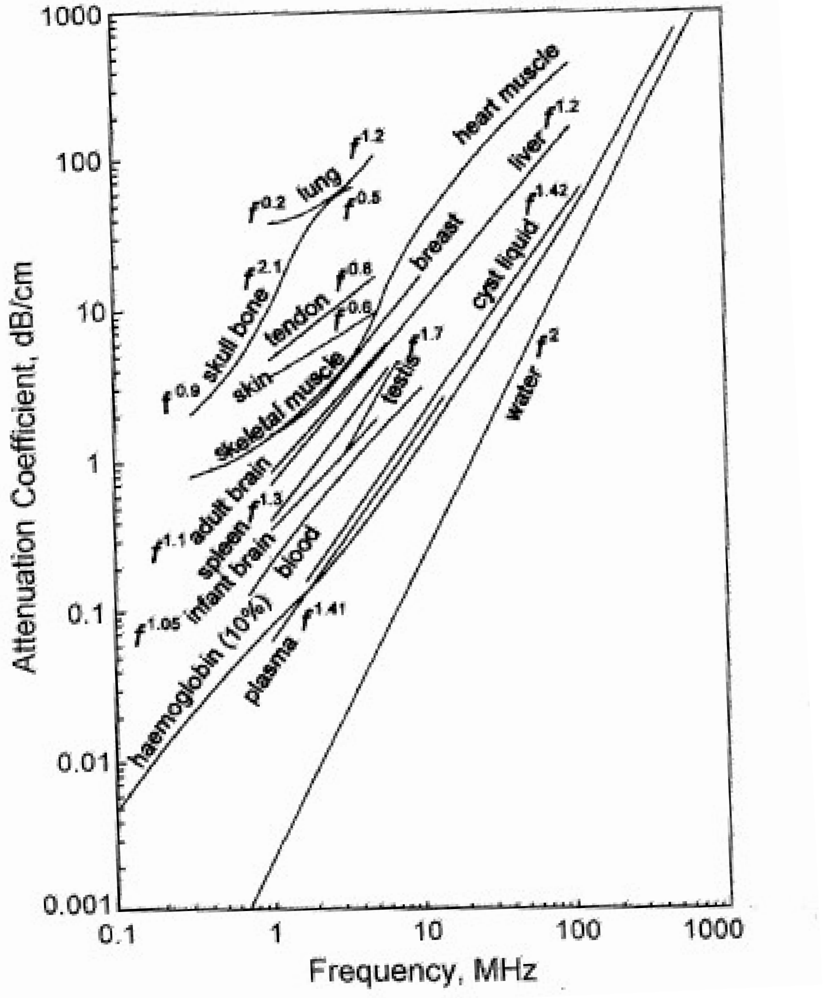
  - Function of frequency
    - Soft tissues – increases linearly with *f*
    - Liquids – increases with f^2
    - Hard tissues – less dependent on *f*

Attenuation

- **Attenuation is dependent on temperature and frequency (courtesy Cobbold RSC)**



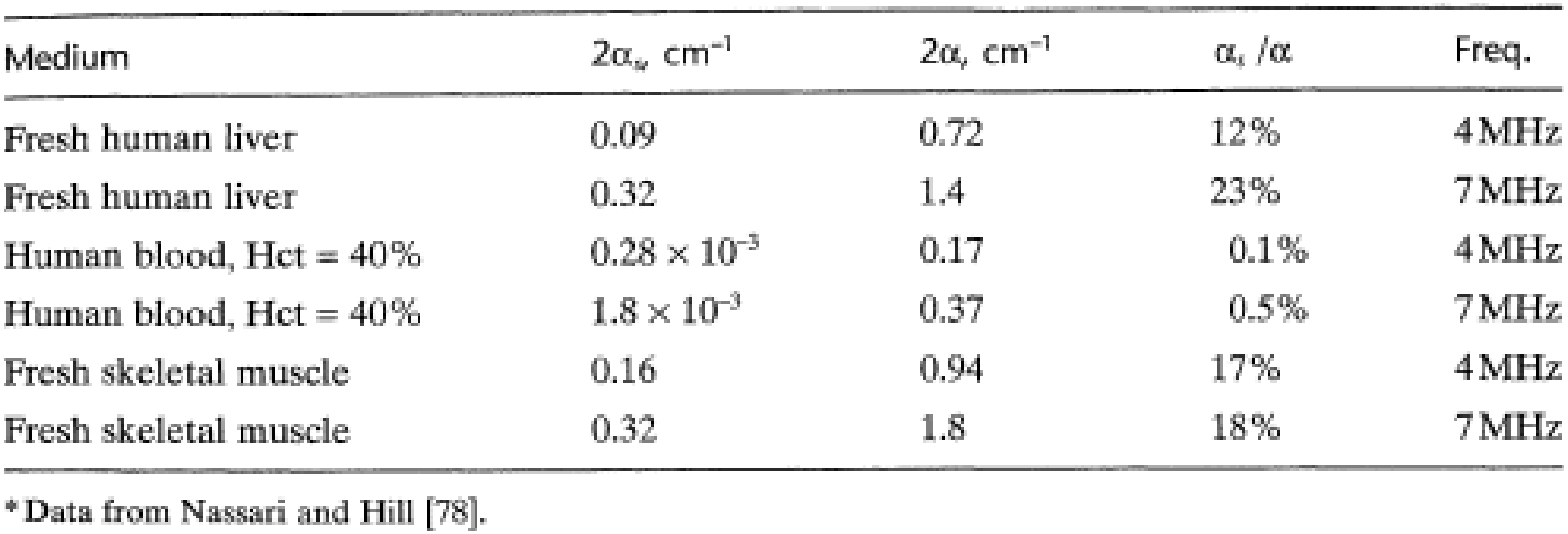
Temperature dependence of attenuation, for water



Frequency dependence of attenuation, for various tissues

Attenuation

- **Relative contributions of scattering to the attenuation coefficient (courtesy Cobbold RSC)**



Medium	$2\alpha_s$, cm^{-1}	2α, cm^{-1}	α_s/α	Freq.
Fresh human liver	0.09	0.72	12%	4 MHz
Fresh human liver	0.32	1.4	23%	7 MHz
Human blood, Hct = 40%	0.28×10^{-3}	0.17	0.1%	4 MHz
Human blood, Hct = 40%	1.8×10^{-3}	0.37	0.5%	7 MHz
Fresh skeletal muscle	0.16	0.94	17%	4 MHz
Fresh skeletal muscle	0.32	1.8	18%	7 MHz

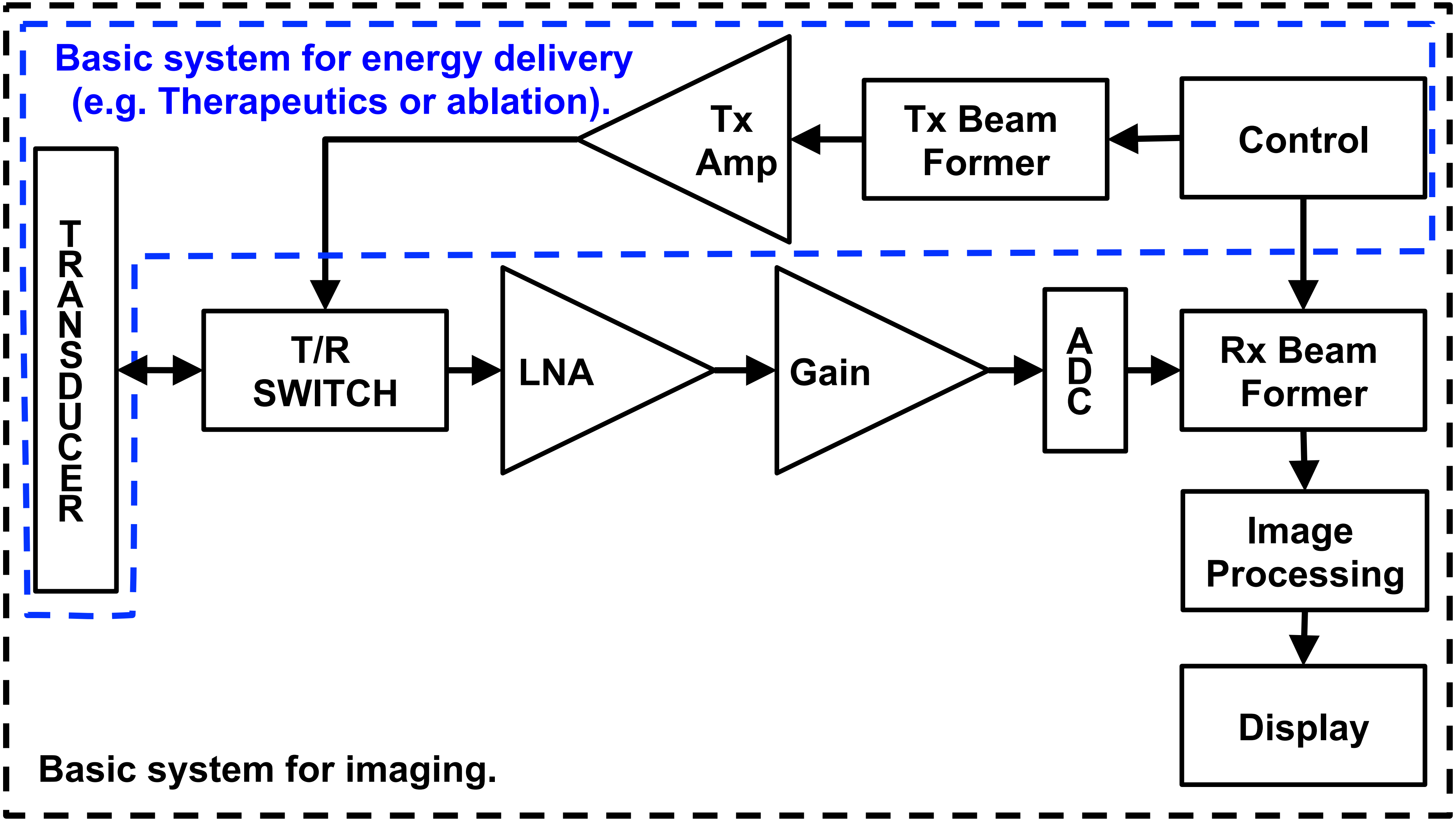
* Data from Nassari and Hill [78].

Tissue Phantoms

- Materials used to mimic biological tissues
  - Training of sonographers
  - Characterization of ultrasound systems
  - Comparison to computer models
  - Development of new probes and systems
- Tissue mimicking materials
  - For individual tissues
- Tissue phantoms
  - Composed of one or more tissue mimicking materials
  - Homogenous / heterogeneous
- Acoustic properties
  - Speed of sound
  - Acoustic impedance
  - Attenuation
  - Backscatter coefficient
  - Nonlinearity parameter
  - Must be stable across multiple f, T

Ultrasound Imaging

Basic Ultrasound System



Signaling Basics

- Pulse wave output
  - Pulses are transmitted, transmission is paused to allow transducer to listen
- Pulse Repetition Frequency (PRF)
  - Number of times an element is pulsed or electrically stimulated per second
  - Limited by listening time
  - Maximum PRF is limited by depth and velocity of medium

$$PRF_{\max} = \frac{v_{ac}}{2R}$$

  - Because R = V_{ac} t

$$PRF_{\max} = \frac{v_{ac}}{2R} = \frac{v_{ac}}{2v_{ac}t} = \frac{1}{2t}$$

  - In tissue 13us are required to detect an interface for each cm of depth

Signaling Basics

- Pulse Repetition Period (PRP)
  - Time required to transmit a pulse and listen for the received echo

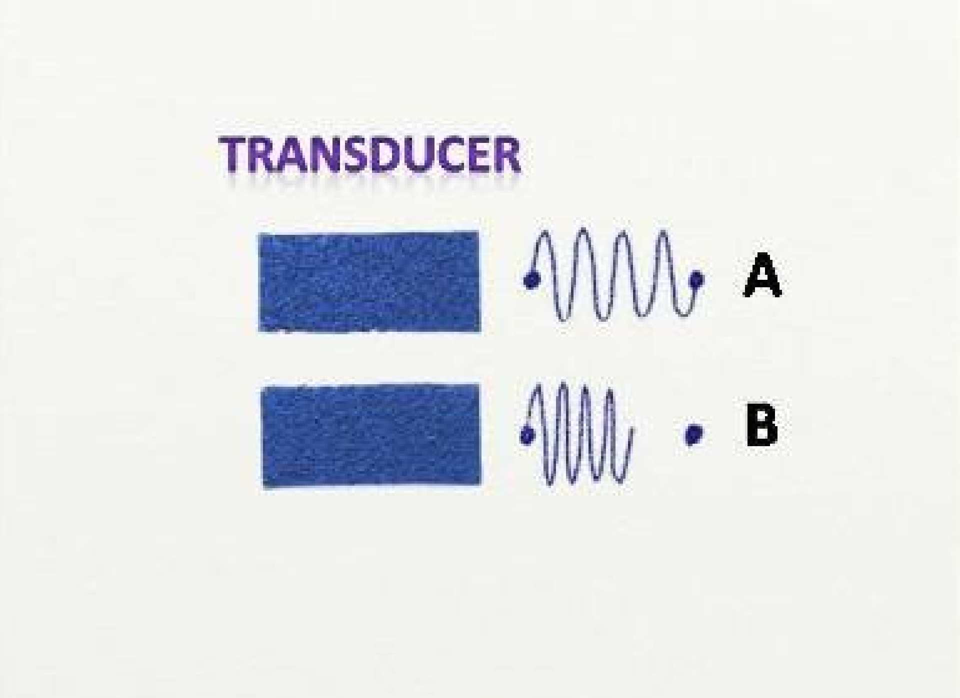
$$PRP = \frac{1}{PRF}$$

- Spatial Pulse Length (SPL)
  - Based on wavelength and number of cycles at the center frequency

$$SPL = n\lambda$$

  - SPL must be reduced to improve axial resolution
    - Decrease number of cycles or increase *f*
  - Often between 2-5 cycles are used in medical ultrasound signaling

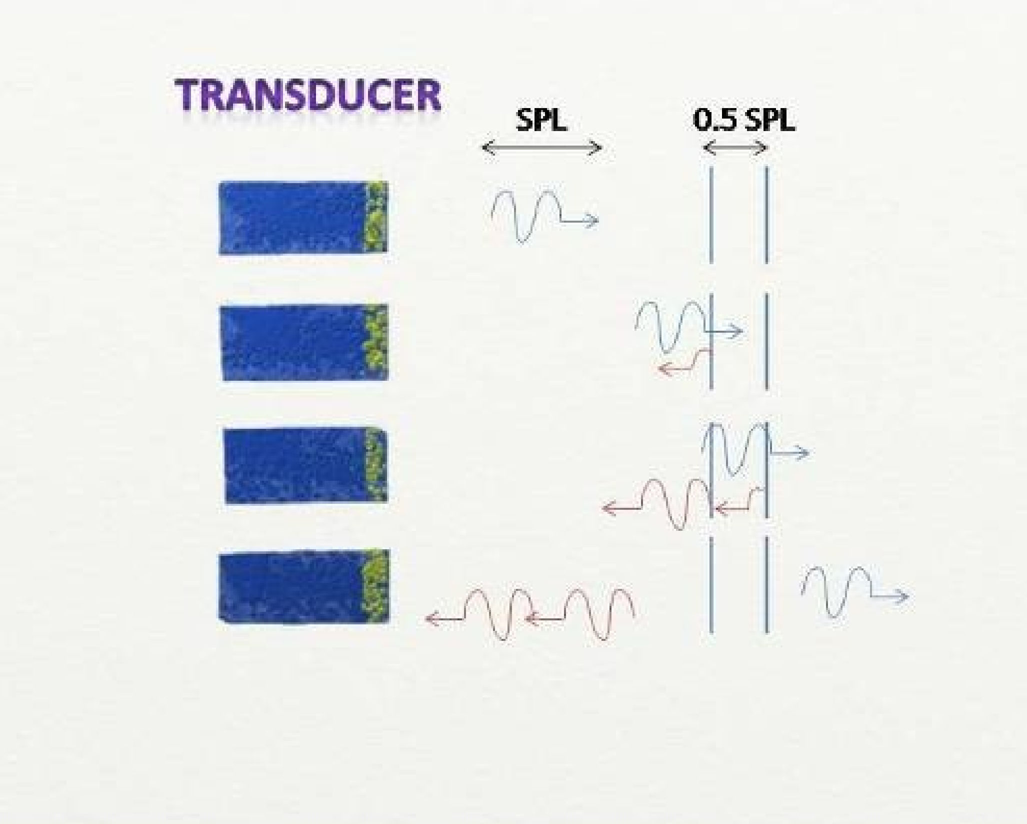
Signaling Basics



Pulse length shortened by increasing the frequency.

- **A. The four-cycle pulse from a low-frequency transducer includes both objects within the SPL.**
- **B. The four-cycle pulse from a high-frequency transducer has a shorter spatial pulse length and can resolve objects located more closely together.**

Signaling Basics



- SPL with maximal resolution.
- Two objects (vertical lines) are separated by 0.5 SPL.
- The echo from each interface is shown by dashed lines. The objects are just resolvable.

Signaling Basics

- Pulse Duration (PD)
  - Temporal pulse length
  - Influenced by matching and backing layers
  - Defined from initiation to 20 dB decrease in V_{pp}

$$PD = nt$$

- Duty Factor (DF)
  - Fraction of time that the ultrasound system is actively transmitting

$$DF = \frac{PD}{PRP} = PD\left(PRF\right)$$

  - Important indicator for how much intensity is delivered to tissue, particularly during therapeutic ultrasound

Signaling Basics

- **Q-Factor**

$$Q = \frac{f_c}{\Delta f}$$

- **Bandwidth**
  - **Full width half maximum (FWHM)**
  - **3dB Bandwidth**

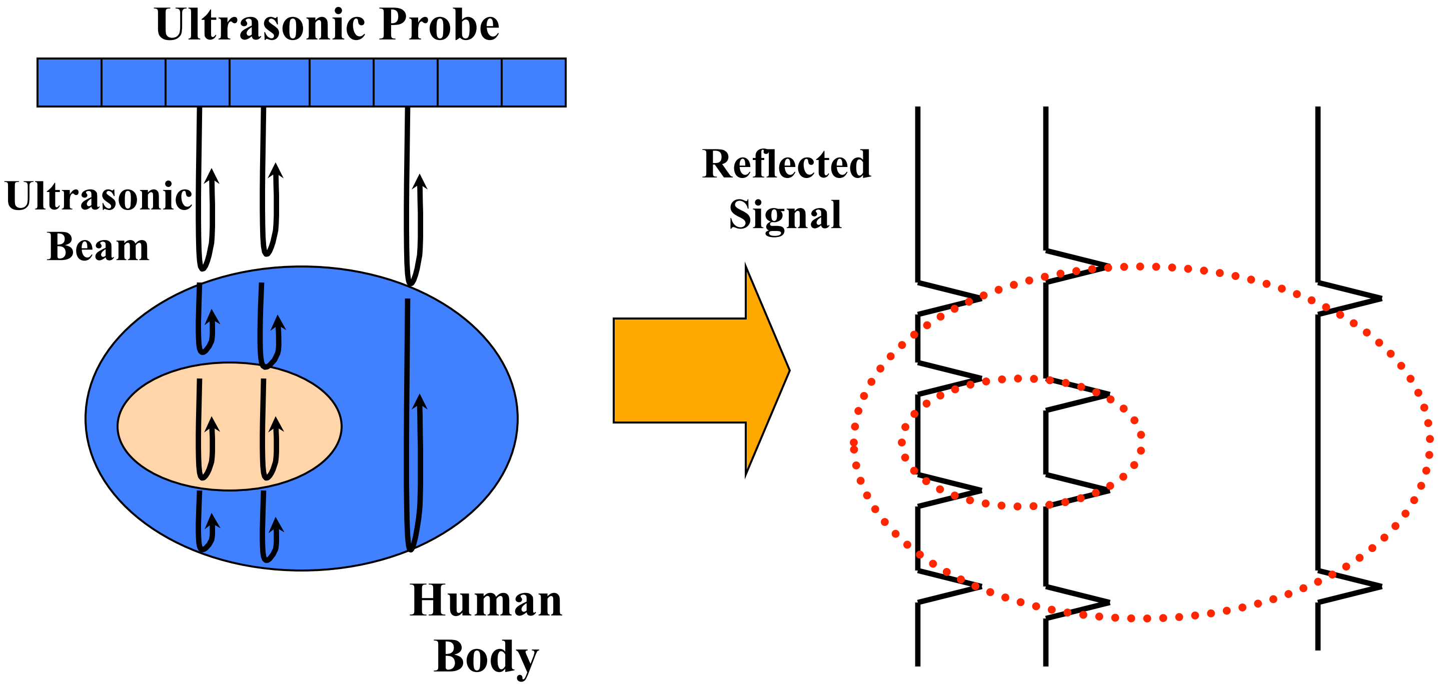
$$\Delta f = \frac{1}{PD(\mu s)}$$

- **Fractional bandwidth**
  - **Bandwidth expressed as a fraction of the center frequency**

$$FractionalBandwidth = \frac{1}{Q} = \frac{\Delta f}{f_c}$$

Ultrasonic echo imaging
- basic principle-

- Same principle as echo among the hills.
- Estimate the distance from the sound reflection and the sound velocity.

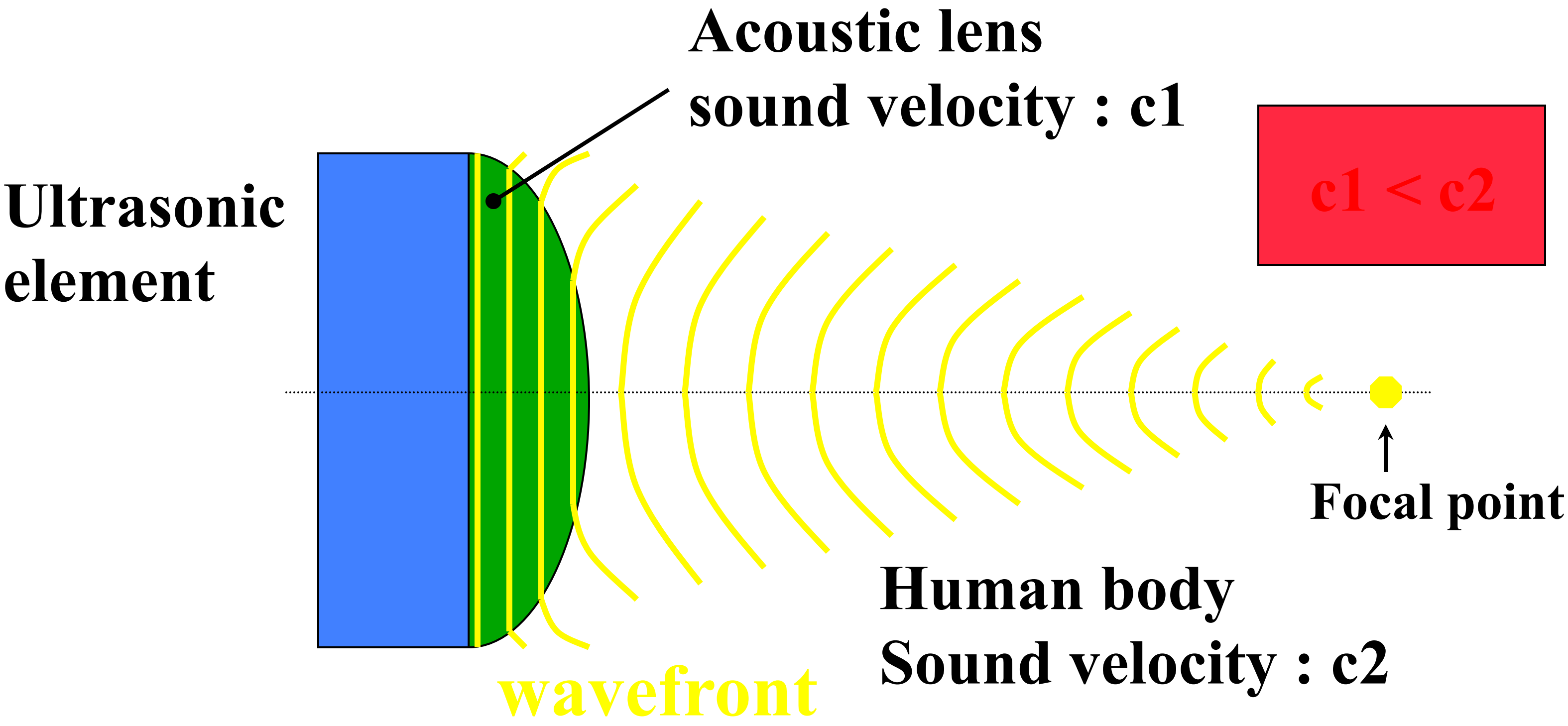


Ultrasonic echo imaging
- focusing technique-

- Ultrasonic beam is needed for imaging.
- Ultrasonic wave is widely spread in human body!
- It propagates as spherical wave, not beam!
- How to form ultrasonic beam ?
  - Acoustic lens
  - Electronic focus

Ultrasonic echo imaging
- focusing technique-

- Acoustic Lens



Acoustic lens
sound velocity : c1

Ultrasonic
element

c1 < c2

Focal point

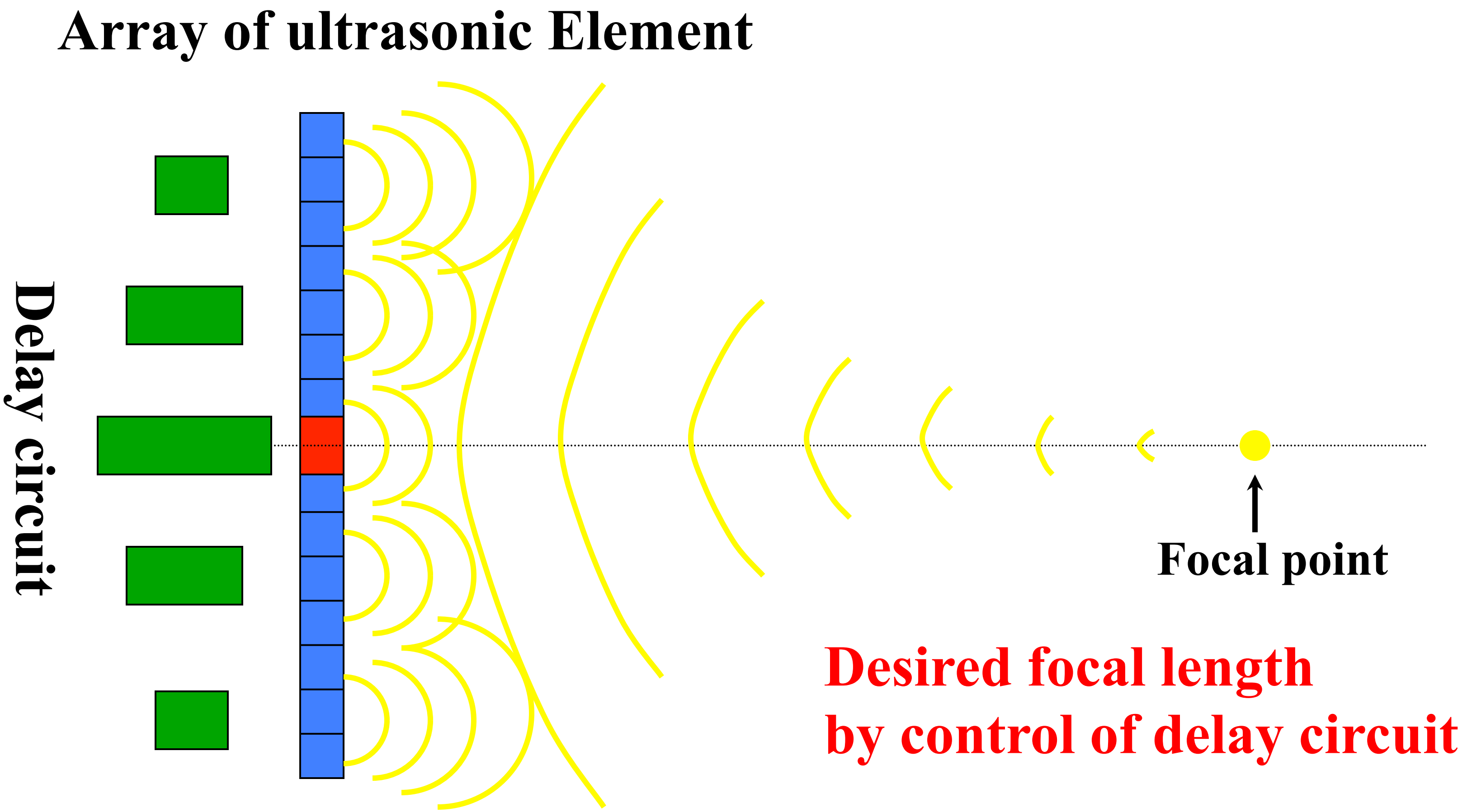
Human body
Sound velocity : c2

wavefront

Weak point : a fixed focus

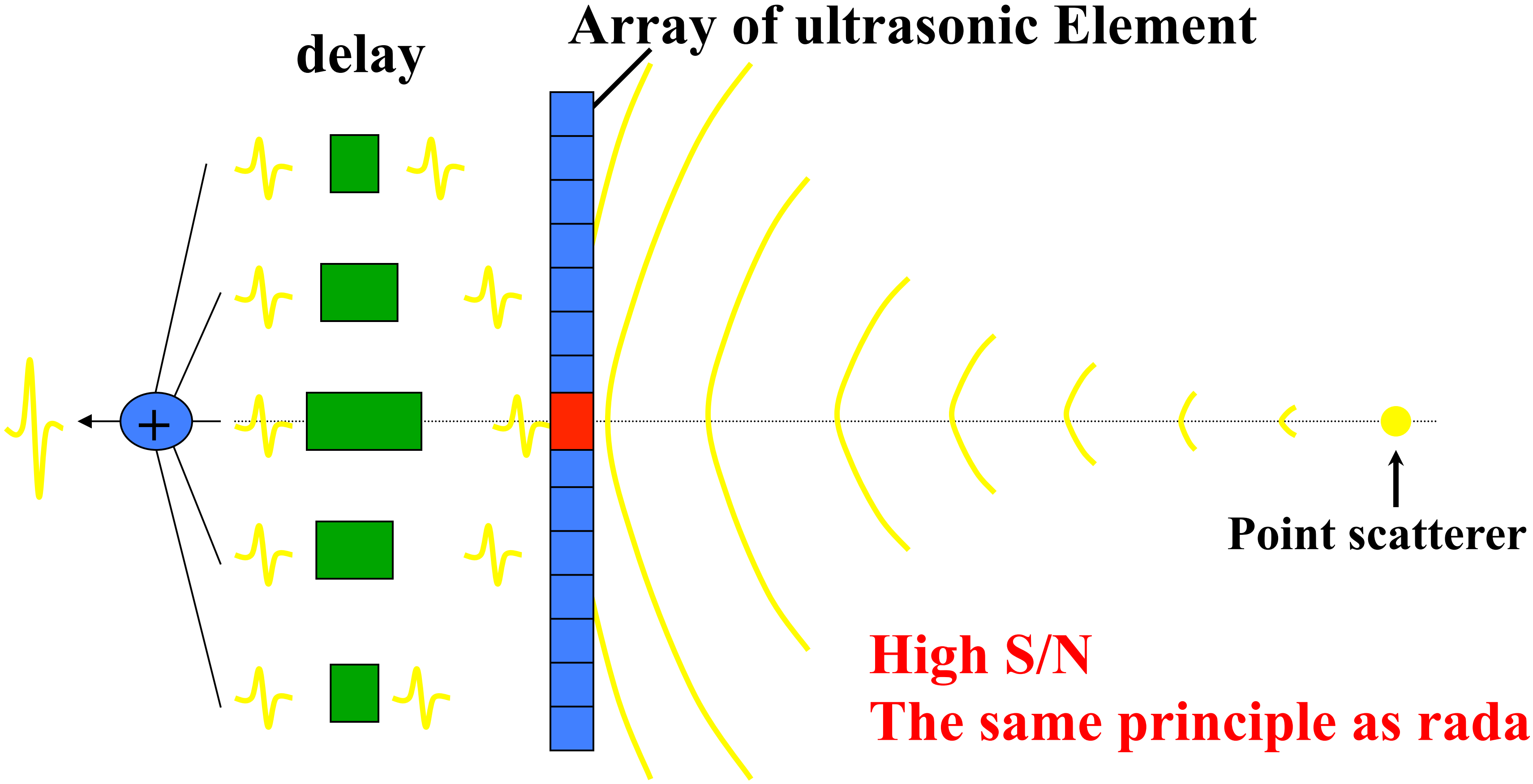
Ultrasonic echo imaging
- focusing technique-

- Electronic focus (transmission)



Ultrasonic echo imaging
- focusing technique-

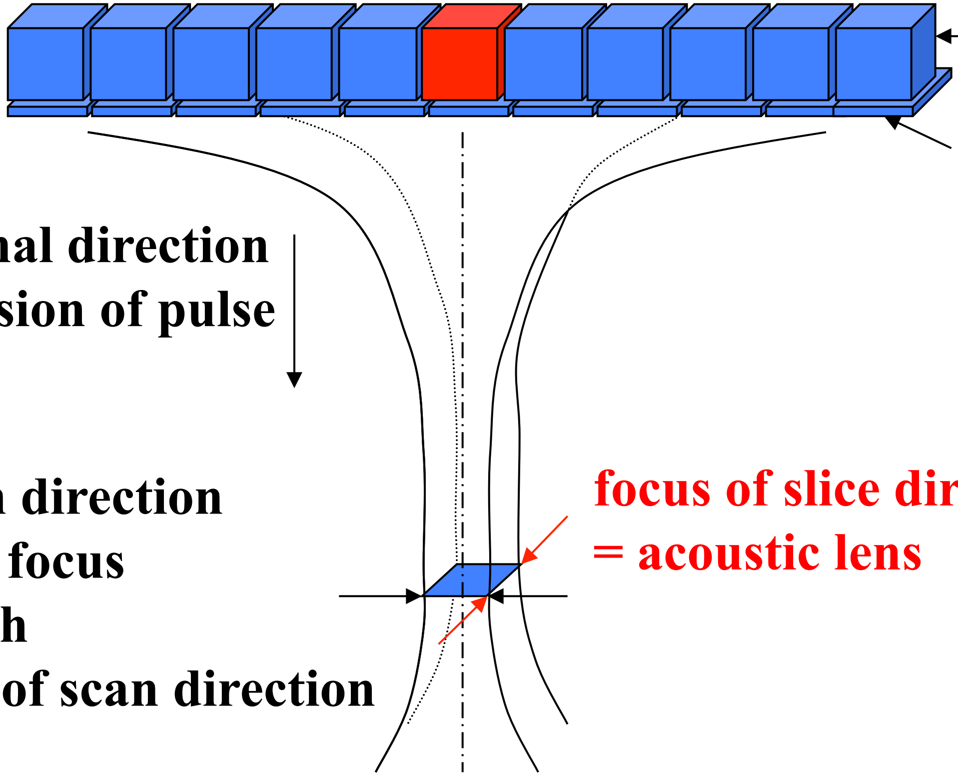
- Electronic focus (receiving)



High S/N
The same principle as radar

Ultrasonic echo imaging
- focusing technique-

- Electronic focus (beam profile)
  - Use of several elements



element

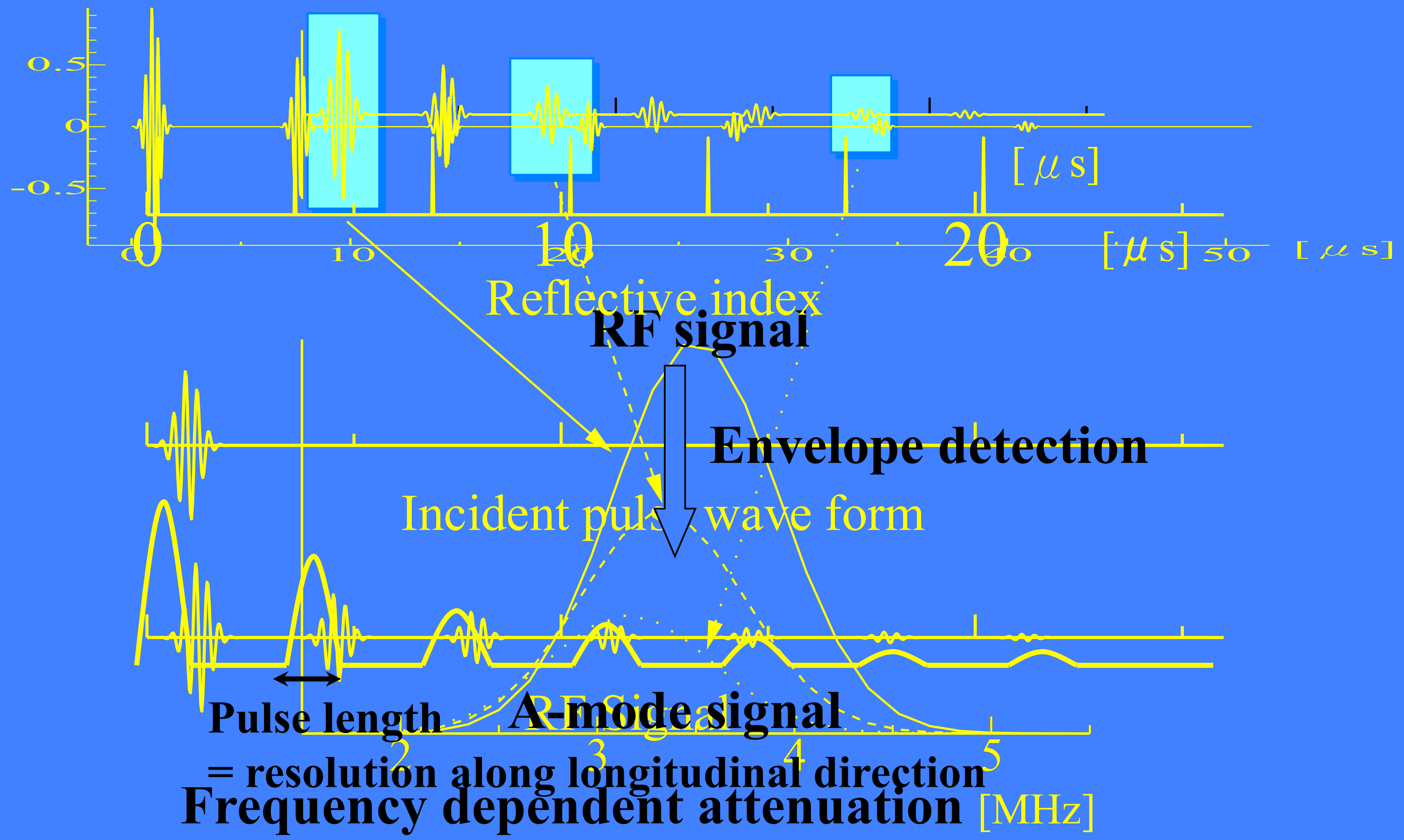
acoustic lens

longitudinal direction
= progression of pulse

focus of scan direction
= Electronic focus
= beam width
= resolution of scan direction

focus of slice direction
= acoustic lens

Ultrasonic echo imaging
- A-mode signal-



Reflective index

RF signal

Envelope detection

Incident pulse wave form

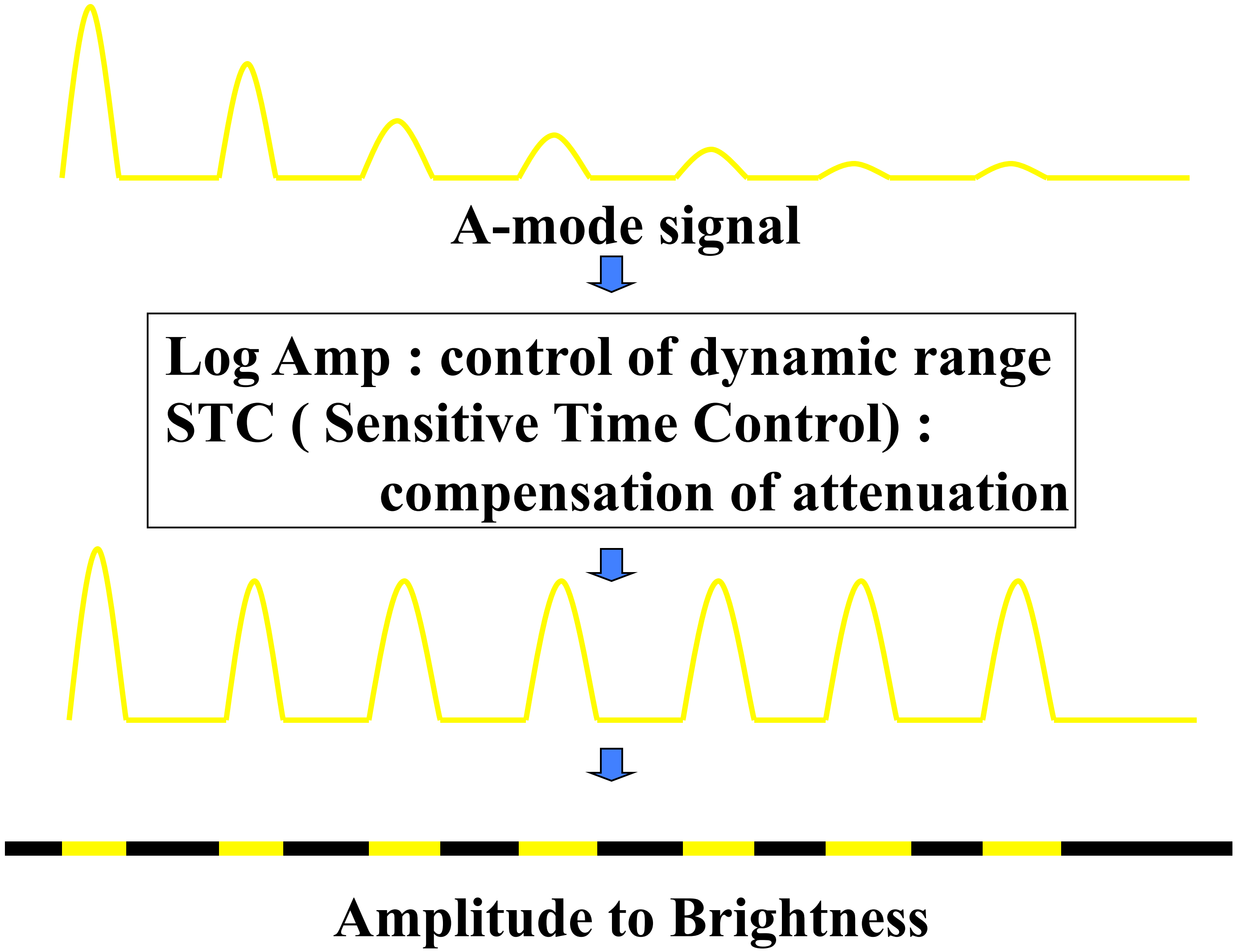
A-mode signal

RF Signal

Pulse length
= resolution along longitudinal direction

Frequency dependent attenuation

Ultrasonic echo imaging
- A-mode signal to B-mode image -



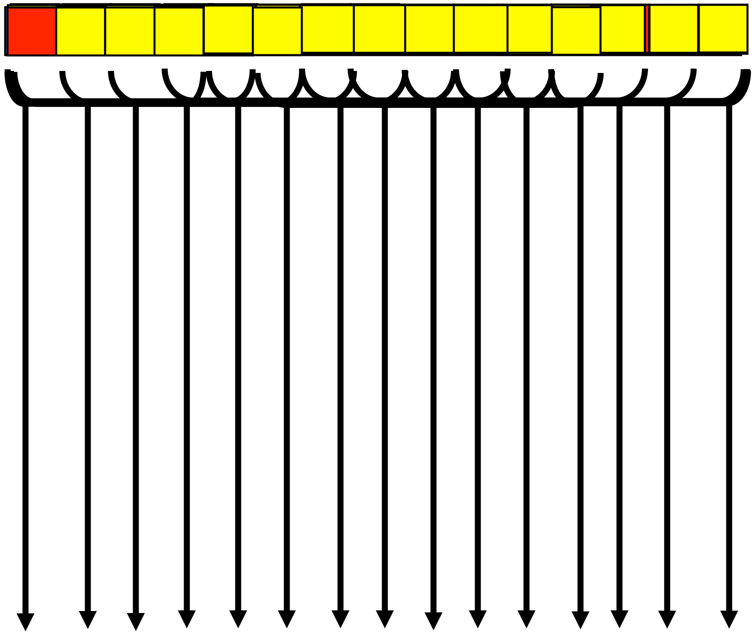
A-mode signal

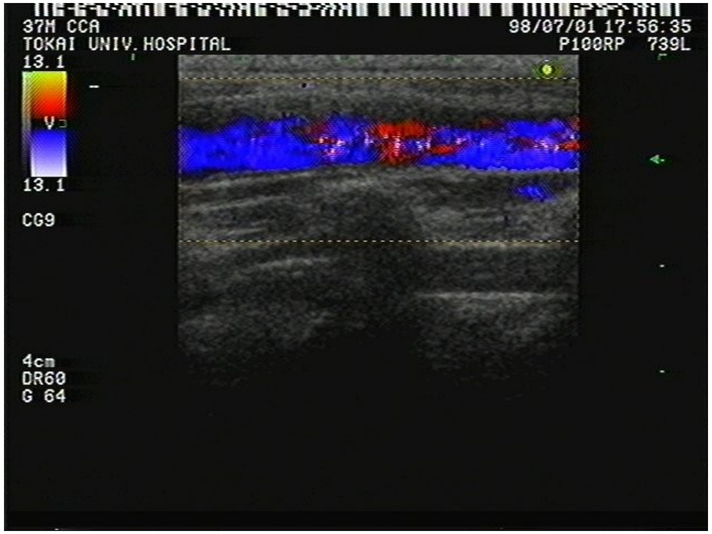
Log Amp : control of dynamic range
STC (Sensitive Time Control) :
compensation of attenuation

Amplitude to Brightness

Ultrasonic echo imaging
- scanning techniques -

- Control of beam direction : switched array
- Scanning : linear

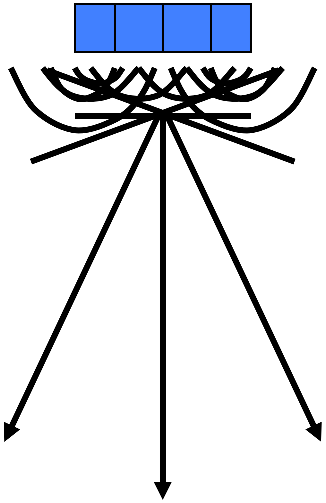


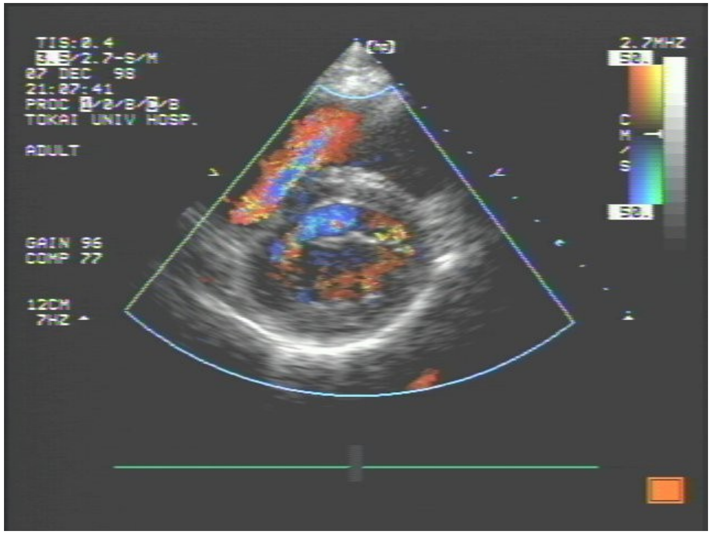


Thyroid image

Ultrasonic echo imaging
- scanning techniques -

- Control of beam direction : phased array
- Scanning : sector

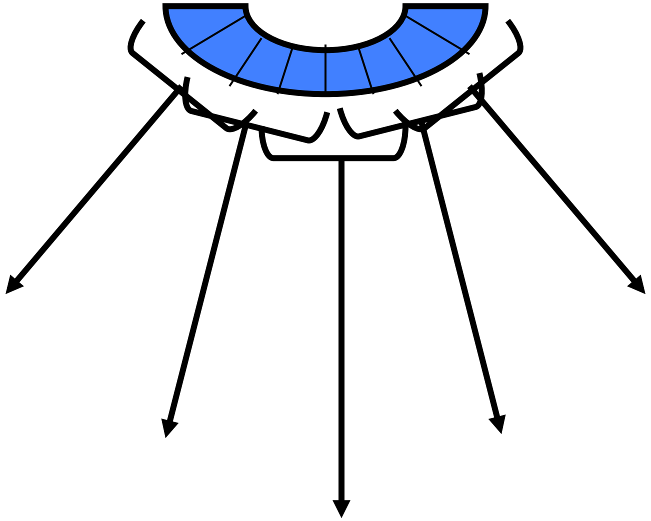




Heart image

Ultrasonic echo imaging
- scanning techniques -

- Control of beam direction : switched array
- Scanning : offset sector



Liver image

Ultrasonic echo imaging - grouping -

Element array	linear	convex	linear	annular
Control of beam direction	Switched array method		Phased array method	mechanical
scan	linear	Offset sector	sector	
Probe form	linear	convex	sector	
Region of image	thyroid, breast	Abdominal region	heart	

Ultrasonic echo imaging
- features -

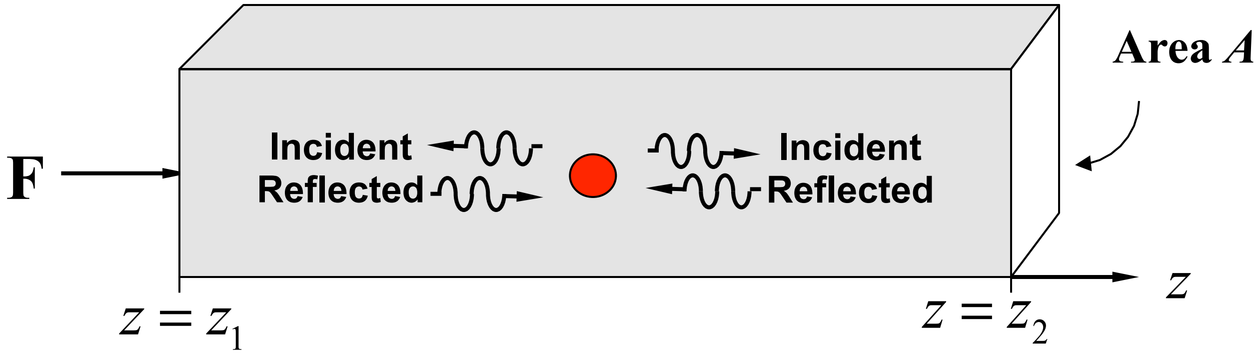
- Resolution
    - Direction of pulse propagation : pulse width : 1-2mm
    - Direction of scanning : beam width : 2-3mm
    - Low resolution and low S/N in deep region
- Ability of imaging of soft tissue
- Imaging in real time
- Doppler image
- Not quantitative image
- Artifacts due to wave properties

Transducers Design and Modeling

Transducers

- **<u>Mechanical Probe</u>: seldom used now**
- **<u>Electronic Probe:</u>**
  - **Linear array transducers**
    - **piezoelectric elements linearly arranged**
    - **sequentially activated to produce an image**
  - **Phased array transducers**
    - **smaller scanning surface (foot print)**
    - **good for echocardiography**
    - **more expensive**
    - **elements are activated with phase differences to allow steering of the ultrasound signal**

Modeling: Stress & Strain in 1D



Red dot denote specific particle in the bar

Hooke's Law: $T = c \cdot S$

Stress → T ; c ← **stiffness matrix**

Strain: $S = \lim_{\Delta z \to 0} \frac{\Delta U}{\Delta Z} = \lim_{\Delta z \to 0} \frac{U(Z+\Delta Z) - U(Z)}{\Delta Z} = \frac{\partial U}{\partial Z}$

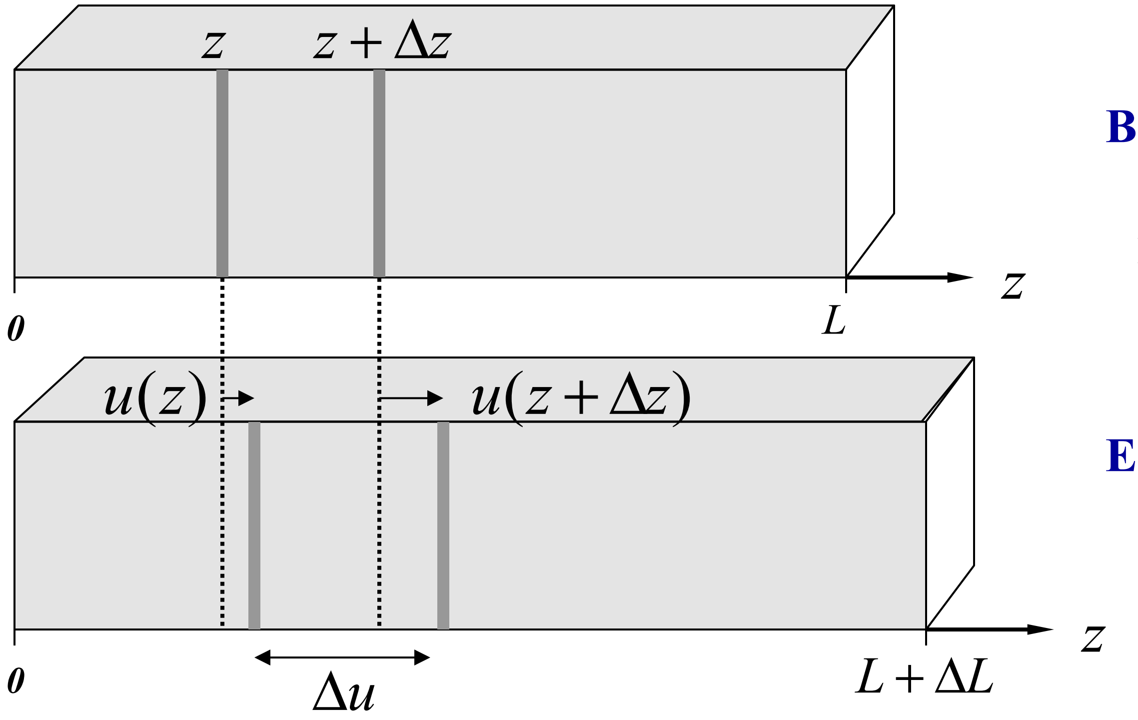
$$\boxed{S = \frac{\partial U}{\partial Z}}$$

Newton's Law: $\Delta F \equiv ma = (\rho_m A \Delta Z) \frac{\partial^2 U}{\partial t^2}$ ⇒ $\boxed{\frac{\partial T}{\partial z} = \rho_m \frac{\partial^2 U}{\partial t^2}}$

$$F = T \cdot A$$

Modeling: Strain in 1D

Strain *S*. ***Dimensionless, a relative change in length***



Bar in equilibrium

Gray lines denote specific particles in the bar

Elongated by external force

Applied force induces internal stress and deforms the bar, shifting particles

$$S \equiv \lim_{\Delta z \to 0} \frac{\Delta u}{\Delta z} = \lim_{\Delta z \to 0} \frac{u(z+\Delta z)-u(z)}{\Delta z} = \frac{\partial u}{\partial z}$$

$$\boxed{S = \frac{\partial u}{\partial z}}$$

[†]Note: the definition of strain here is valid only for small deformations of the type we will encounter in piezoelectric transducers

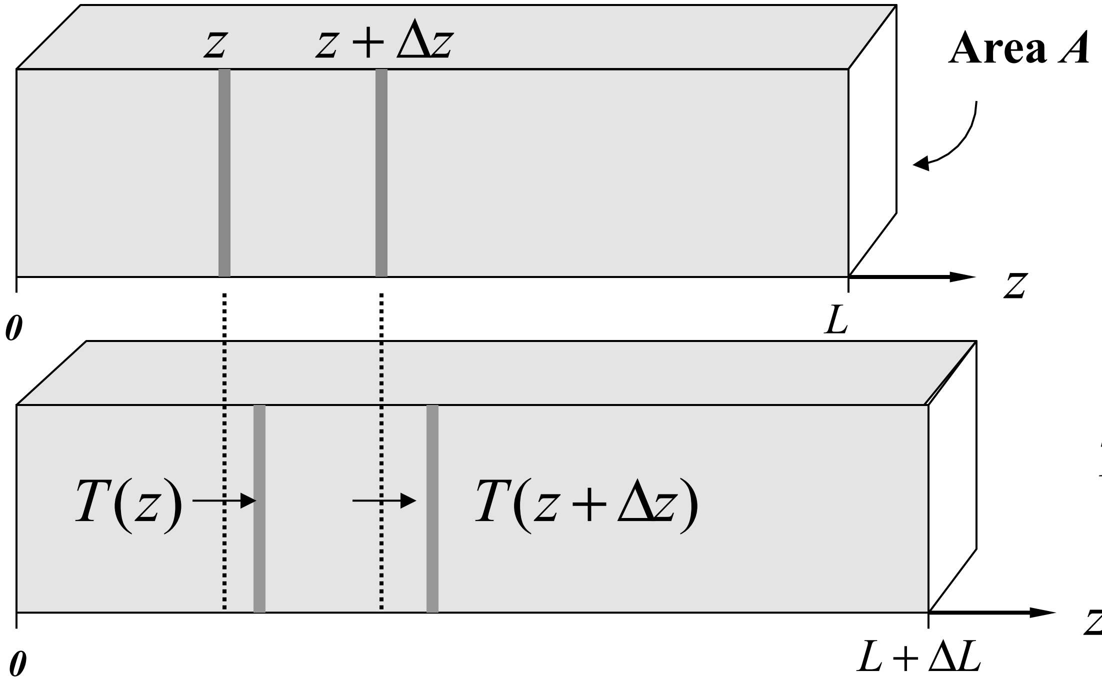
Modeling: Stress in 1D

Stress *T, has units of force per unit area*

$$\boxed{\begin{array}{l} T = c \cdot S \\ \\ F = T \cdot A \end{array}}$$

Hooke's law

stiffness matrix



Bar in equilibrium

Gray lines denote specific particles in the bar

Elongated by external force

$T(z) \rightarrow$ **Describes the internal stress (force per unit area) in the material**

$$F(z) = T(z)A$$

Net force on slice: $$\Delta F \equiv F(z+\Delta z) - F(z) = A\frac{\partial T}{\partial z}\Delta z$$

Newton's Law: $$\Delta F \equiv ma = (\rho_m A\Delta z)\frac{\partial^2 u}{\partial t^2}$$

Mass density

$$\Longrightarrow \boxed{\frac{\partial T}{\partial z} = \rho_m \frac{\partial^2 u}{\partial t^2}}$$

Acoustic Waves in 1D

The equations of motion are:

$$\frac{\partial T}{\partial z} = \rho_m \frac{\partial^2 u}{\partial t^2}$$ **Newton's law**

$$T = c_m S$$ **Hooke's law**

$$S = \frac{\partial u}{\partial z}$$ **Stress definition**

$$v = \frac{\partial u}{\partial t}$$ **Particle velocity**

where: $T = \text{Stress [N/m}^2\text{]}$

$S = \text{Strain [unitless]}$

$u = \text{Displacement [m]}$

$v = \text{Particle velocity [m/s]}$

$c_m = \text{Stiffness constant [N/m}^2\text{]}$

$\rho_m = \text{Mass density [kg/m}^3\text{]}$

Eliminating variables we find a 1D wave equation for each of these functions

$$\frac{\partial^2 u}{\partial z^2} = \left(\frac{\rho_m}{c_m}\right)\frac{\partial^2 u}{\partial t^2}$$

So the acoustic phase velocity is: $v_p = \sqrt{\frac{c_m}{\rho_m}}$

Note potential confusion between the particle velocity and wave phase velocity

Transmission-Line Analogy

The 1D equations of motion can be rearranged as follows, with *F=TA* and *v* as the independent variables:

$$\frac{\partial F}{\partial z} = \rho_m A \frac{\partial v}{\partial t}$$

$$\frac{\partial v}{\partial z} = \frac{1}{c_m A} \frac{\partial F}{\partial t}$$

$v \leftrightarrow$ ***current***

$F \leftrightarrow$ ***voltage***

Compare with the telegraphers equations:

$$\frac{\partial V}{\partial z} = L' \frac{\partial I}{\partial t}$$

$$\frac{\partial I}{\partial z} = C' \frac{\partial V}{\partial t}$$

One-dimensional acoustic waves can be modeled using transmission-line theory, with force (stress) playing the role of voltage, and particle velocity playing the role of current

Define an acoustic impedance as the ratio of force to particle velocity

$$Z_0 = \frac{F}{v} = A\sqrt{\rho_m c_m} = \rho_m A v_p$$

Propagation Constant:

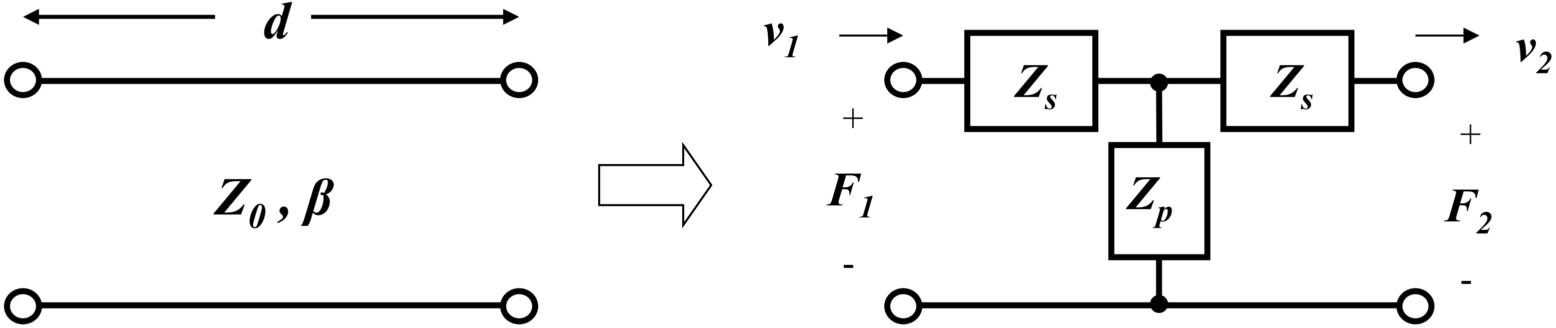
$$\beta = \frac{\omega}{v_p} \Longrightarrow Z = A\frac{\rho_m \omega}{\beta} = A\frac{c_m \beta}{\omega}$$

T-line Equivalent

A section of transmission-line can be modeled by an equivalent lumped-element circuit

The following "tee-circuit" equivalent can be established using z-parameters from network theory

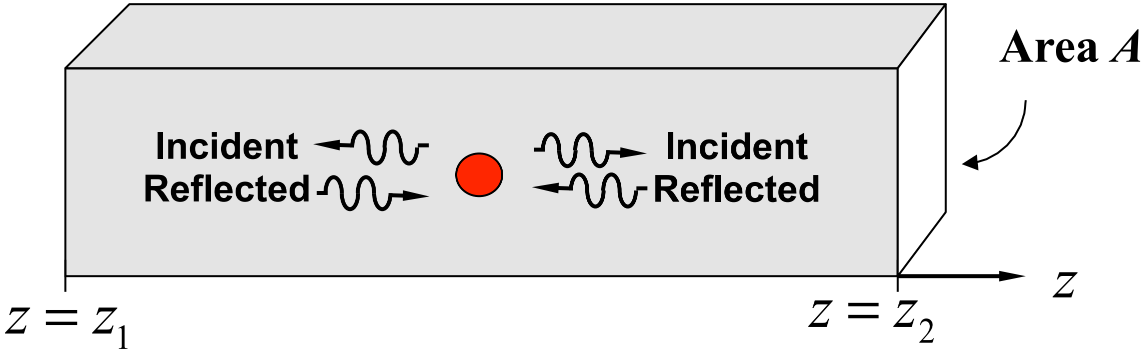
$$Z_s = jZ_0 \tan\left(\frac{\beta d}{2}\right) \qquad Z_p = \frac{-jZ_0}{\sin(\beta d)}$$



Using the equivalent circuit we can relate the terminal parameters as follows

$$F_1 = Z_s v_1 + \underbrace{Z_p(v_1 - v_2)}_{\text{"Voltage" across shunt branch}} \qquad F_2 = -Z_s v_2 + Z_p(v_1 - v_2)$$

Acoustic Wave in Non-piezoelectric Material



$$a = \frac{V_1 e^{\gamma Z_2} - V_2 e^{\gamma Z_1}}{2\omega \sinh \gamma d}$$

$$b = \frac{V_2 e^{-\gamma Z_1} - V_1 e^{-\gamma Z_2}}{2\omega \sinh \gamma d}$$

Red dot denote specific particle in the bar

Particle Displacement: $u = ae^{-j\beta z} + be^{j\beta z}$

Traveling to the right

Traveling to the left

Particle Velocity: $v = \frac{du}{dt}$

Phase Velocity: $v_p = \lambda f = \frac{\lambda}{T} = \frac{\omega}{\beta}$

Modeling : Model non-piezoelectric material

$$T = c \cdot \frac{\partial U}{\partial Z} \qquad F = T \cdot A \qquad \Rightarrow \qquad F = cA\frac{\partial U}{\partial Z}$$

Force equations:

$$\boxed{\begin{aligned} F_1 &= (v_1 - v_2)\frac{Z_o}{\sinh \gamma d} + v_1 Z_o \tanh\frac{\gamma d}{2} \\ F_2 &= (v_1 - v_2)\frac{Z_o}{\sinh \gamma d} - v_2 Z_o \tanh\frac{\gamma d}{2} \end{aligned}}$$

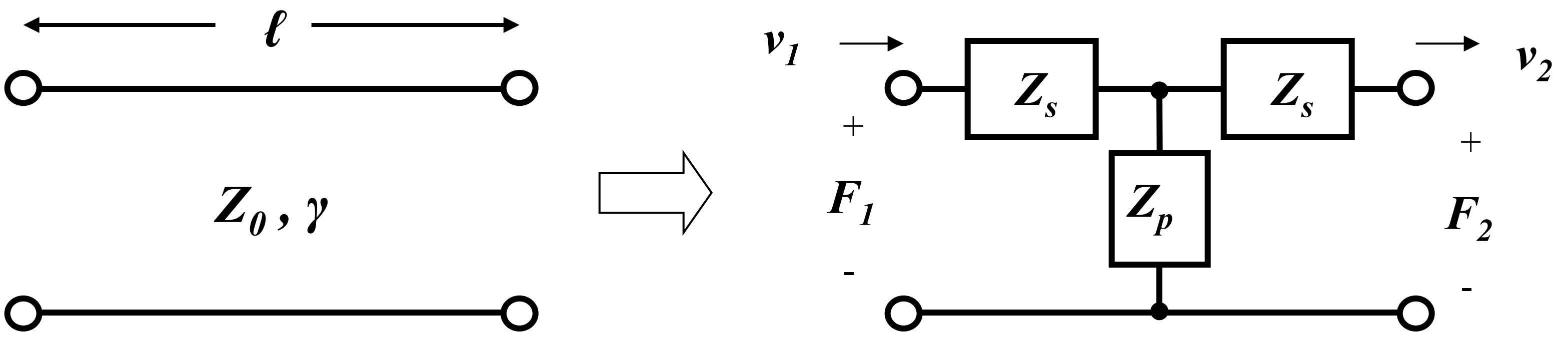
v to current

F to voltage

Equivalent circuit:

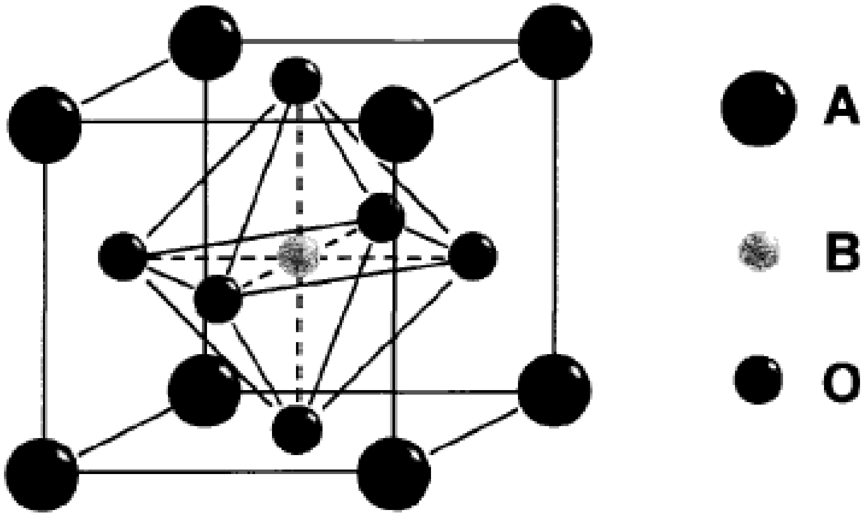
$$\boxed{Z_s = Z_o \tanh\left[\frac{\gamma d}{2}\right]} \qquad \boxed{Z_p = \frac{Z_o}{\sinh[\gamma d]}}$$

Using Kirchoff's Current Law

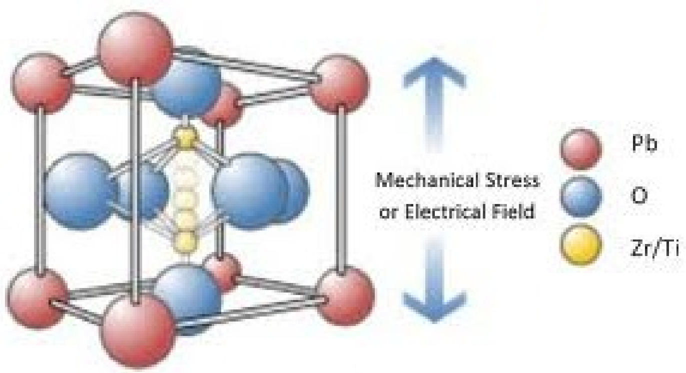


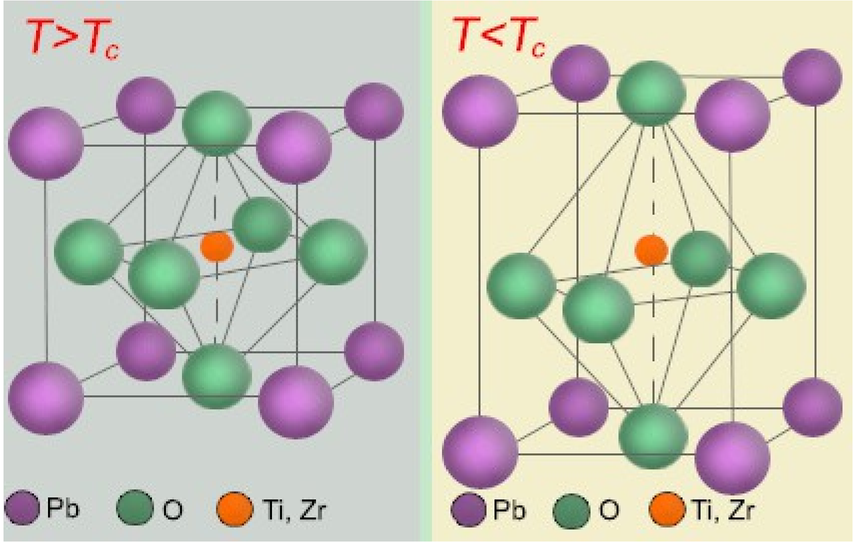
Ferroelectrics

- Ferroelectric materials (mid-1600’s)
  - Exhibit a spontaneous polarization
  - Rochelle salt (mild purgative medicinal properties).
- Brewster in the early 1800’s (Pyroelectric effect).
- Curie brothers 1880 identified piezoelectric effect.
- Piezoelectricity: “Pressing” electricity.
- 1940’s several simple oxide crystals with a perovskite structure were discovered.
- All Ferroelectric materials are piezoelectric but not all piezoelectric materials are ferroelectric.
  - $BaTiO_3$, $Ba,SrTiO_3$, PZT, $PbTiO_3$, $SrTiO_3$
  - SiO_2, AlN, ZnO, GaN

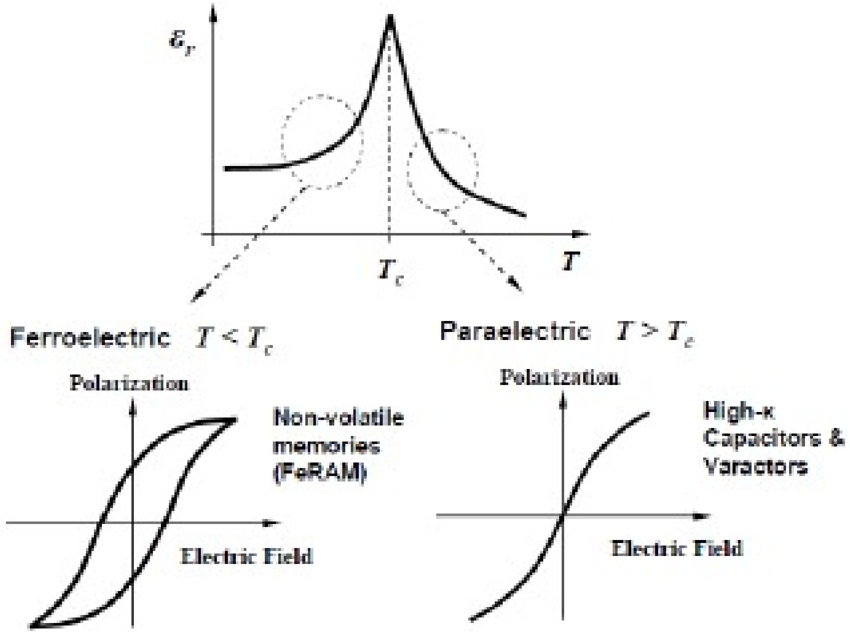


Ferroelectrics

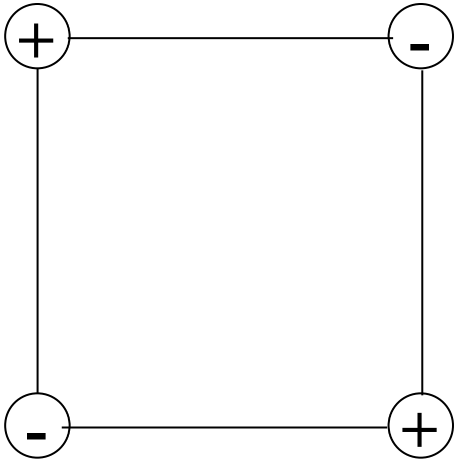


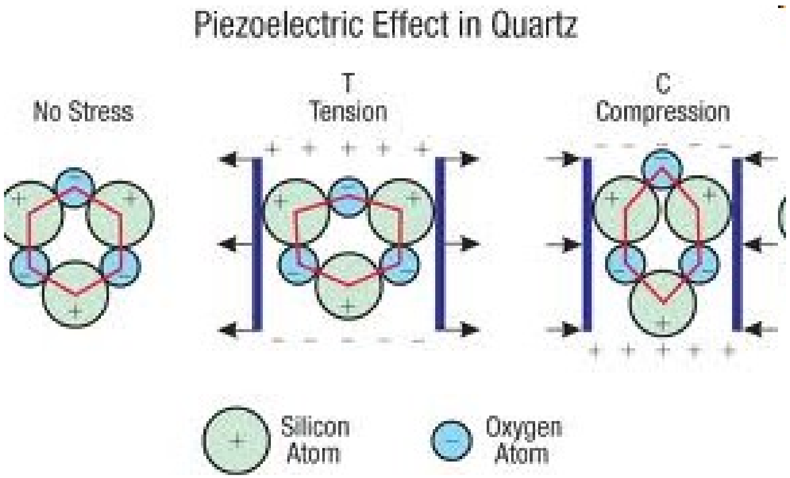


- **Lead Zirconium Titanate**
  - **PZT**
  - **$Pb(Zr_xTi_{1-x})O_3$**
- **1952 Shirane, Suzuki : $Pb(Zr,Ti)O_3$ solid solutions**
- **1955 Jaffe, Cook, Berlincourt, Gerson: Complete Study of PZT formulations**
- **Curie temperature 170-360**

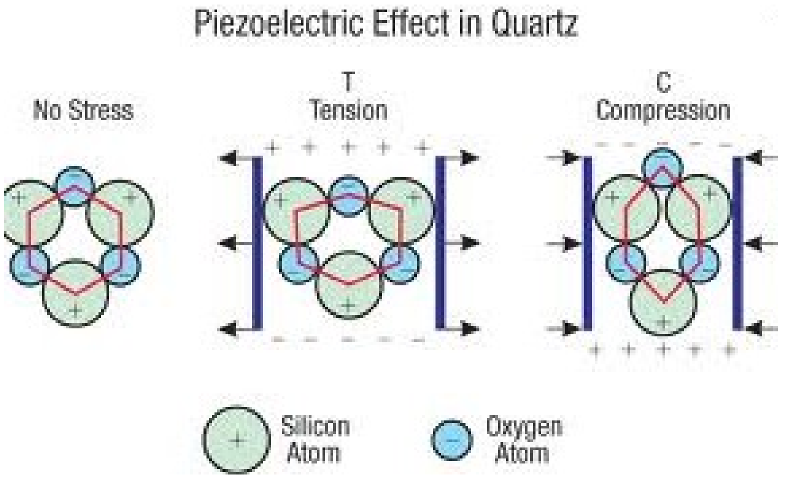


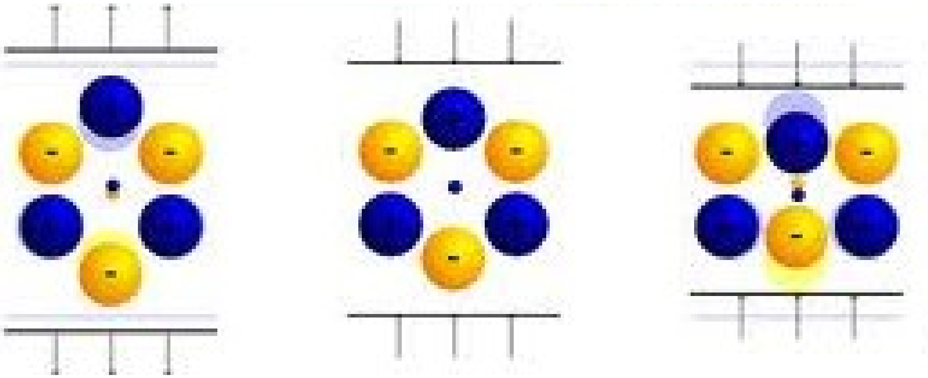
Centrosymmetric and Non-Centrosymmetric

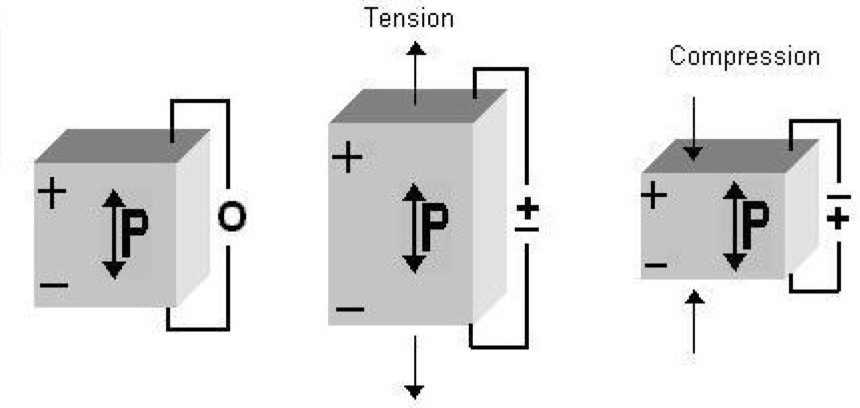


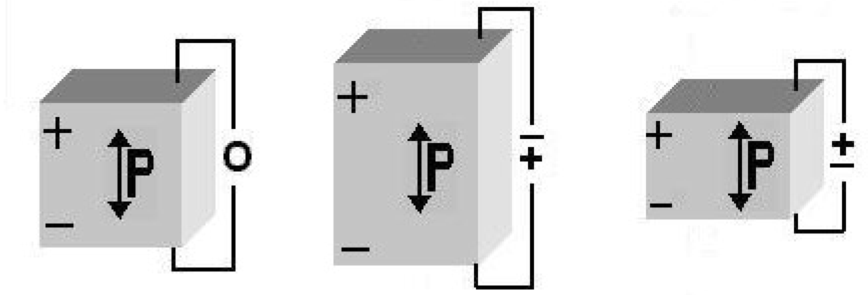


Piezoelectricity









Electrical and Mechanical Variables

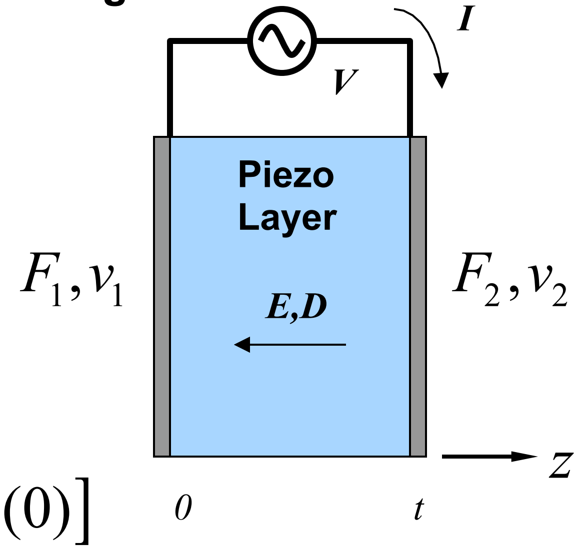
- Electric field strength E (V/m)
  - Field induced by an electric charge when in free space
  - Electric dipole: The center of symmetry of the electron cloud is altered with respect to the nucleus with one pole of the atom becoming more positively charged and the other pole becoming more negatively charged
  - Polarization : distortion process.
- Electric Displacement or Flux Density D (C/m^2)
  - $D = \varepsilon E$
- Stress T (N/m^2)
- Strain S, normalized deformation (Dimensionless)
- Electric permittivity ε (Dimensionless)
- compliance s (m^2/N)
- Stiffness c (N/m^2)
- Piezoelectric strain constant d (C/N also m/V)

Electrical and Mechanical Variables Cont’d

- Piezoelectric strain constant d (C/N also m/V)
  - Transmission constant representing the resulting change in strain per unit change in electric field with unit of coulombs per newton
- Piezoelectric stress constant e (N/V-m or C/m^2)
  - Stress change per unit change in electric field with units of newton per volt-meter or coulombs per square meter
- Receiving constant g (V-m/N)
  - Represents the change in electric filed per unit change in applied stress.
- Dielectric constant (Clamped/Free)
  - Material is clamped so that it cannot move in response to an applied field or the strain is zero ε^s.
  - Material is free to move without restriction ε

Piezoelectric Resonators

An acoustic resonator driven by an external sinusoidal electric signal



Since there is an externally applied field, there will be a net displacement vector inside the material related to the charge on the electrodes. The terminal current is

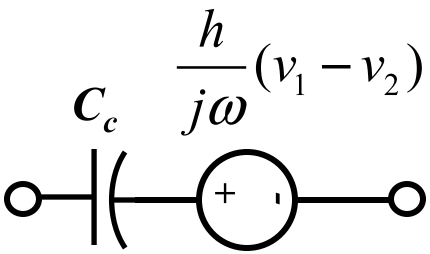
$$I = j\omega Q = j\omega D A$$

To find the I-V relation we use the *D(E,S)* relation

$$E = \frac{D}{\varepsilon^S} - \frac{e_m}{\varepsilon^S}\frac{\partial u}{\partial z} \qquad V = \int_0^t E\,dz = \frac{Dt}{\varepsilon^S} - \frac{e_m}{\varepsilon^S}\left[u(t) - u(0)\right]$$

Writing in terms of current and particle velocity gives

$$V = \underbrace{\frac{I}{j\omega C_c}}_{\text{Normal voltage across capacitor}} + \underbrace{\frac{h}{j\omega}\left[v_1 - v_2\right]}_{\text{Additional voltage due to piezoelectric energy conversion}} \qquad h = \frac{e_m}{\varepsilon^S}$$

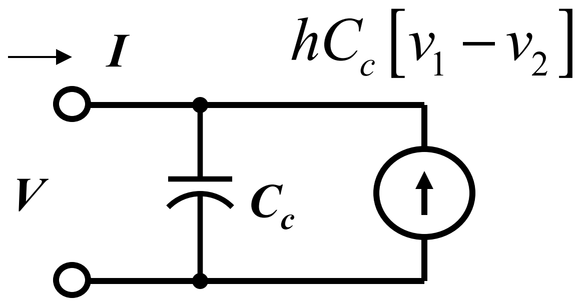


Series equivalent circuit

Equivalent Circuits

We could also rearrange the I-V relationship to give a parallel equivalent

$$I = j\omega C_c V - hC_c\left[v_1 - v_2\right]$$



Parallel equivalent circuit

We need to know the particle velocities at the boundaries, which we get from the acoustic transmission-line model. This model also needs to be modified to account for piezoelectric conversion. For example, the terminal force becomes

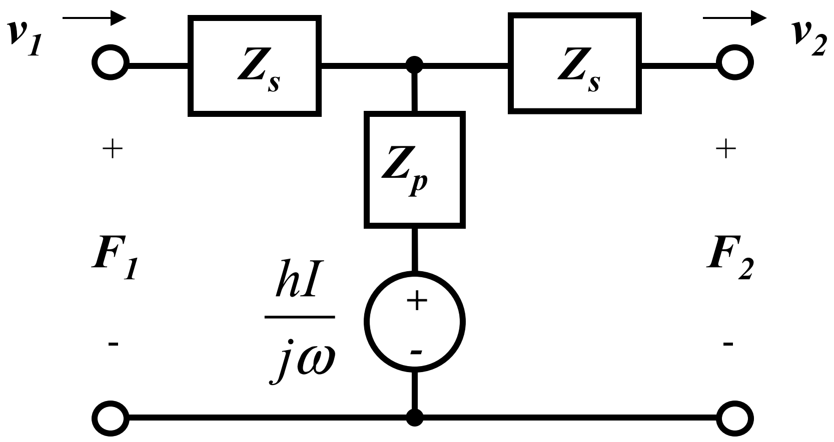
$$F_1 = TA = c_m' SA - \frac{e_m D}{\varepsilon^S} A$$

With a similar result for F_2. The term on the right is the only new term, so

$$F_1 = Z_s v_1 + Z_p\left(v_1 - v_2\right) + \frac{h}{j\omega} I$$

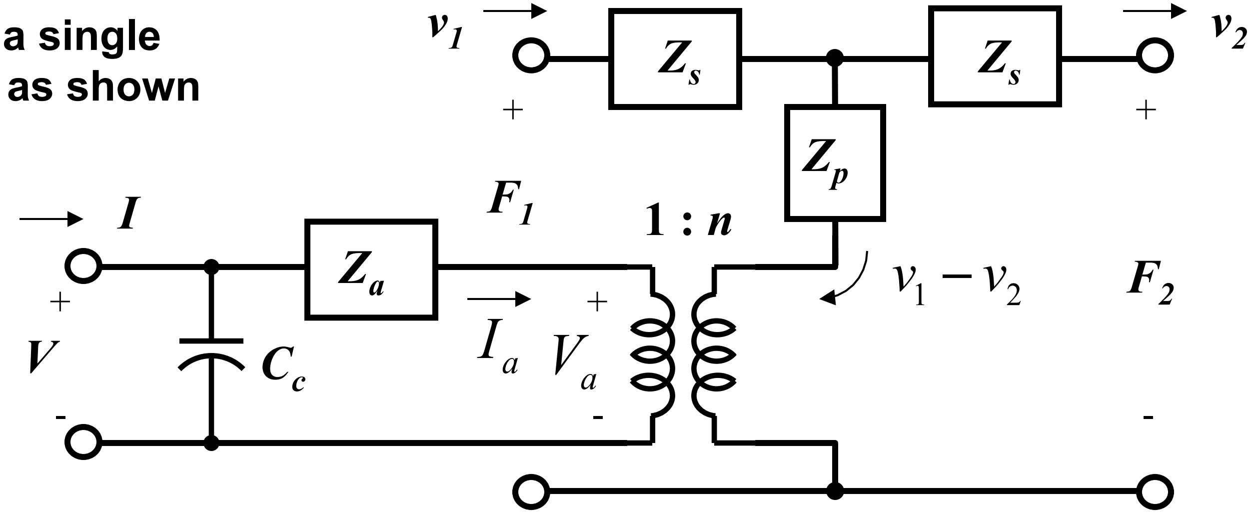
$$F_2 = -Z_s v_2 + Z_p\left(v_1 - v_2\right) + \frac{h}{j\omega} I$$

Equivalent circuit for acoustic quantities



Mason’s Model

Mason showed that the dependent sources can be replaced by a single three-port equivalent circuit as shown



The ideal transformer requires that $I_a = -n(v_1 - v_2)$ which means $\boxed{n = hC_c}$

From the requirement that $nV_a = \frac{hI}{j\omega}$ we get $\boxed{Z_a = -\frac{1}{j\omega C_c}}$

The other parameters are unchanged:

$$C_c = \frac{\varepsilon^S A}{t} \qquad h = \frac{e_m}{\varepsilon^S} \qquad Z_p = \frac{Z_0}{\sinh(\gamma l)} \qquad Z_s = Z_0 \tanh\left(\frac{\gamma l}{2}\right)$$

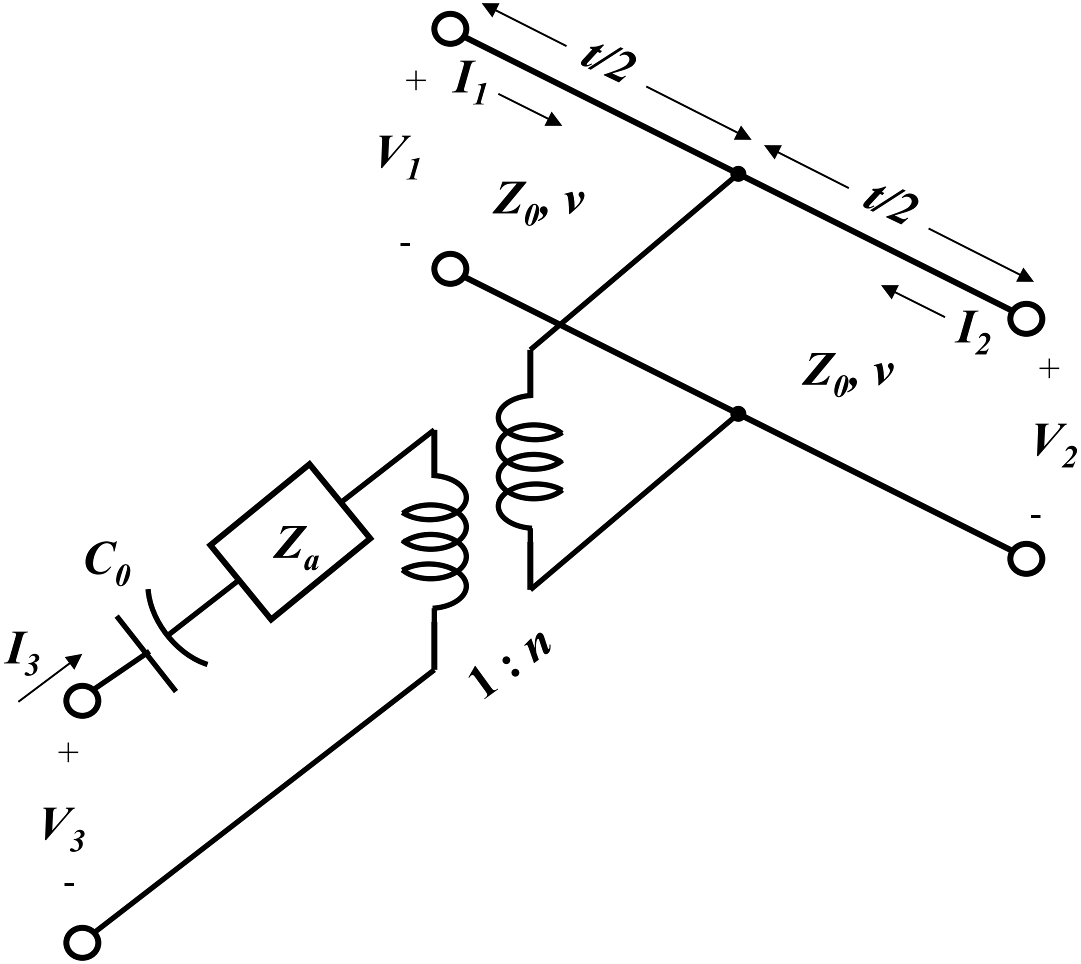
Mason's Model

$$Z=\frac{1}{j\omega C_0}\left[1-K^2\frac{\tan\phi}{\phi}Z_m\right]\qquad Z_m=\frac{\left(z_r+z_l\right)\cos^2\phi+j\sin 2\phi}{\left(z_r+z_l\right)\cos 2\phi+j\left(z_rz_l+1\right)\sin 2\phi}$$

$$Z_{in}=Z_0\left[\frac{Z_t\cos\theta+jZ_0\sin\theta}{Z_0\cos\theta+jZ_t\sin\theta}\right]$$

$$Z=\cfrac{1}{j\omega C_0+\cfrac{1}{\cfrac{-1}{j\omega C_0}+n^2\left(-jZ_p\csc 2\phi+\cfrac{1}{\cfrac{1}{jZ_p\tan\phi+Z_l}+\cfrac{1}{jZ_p\tan\phi+Z_r}}\right)}}$$

Alternative model: KLM Model



$$C_0 = \varepsilon \frac{A}{t}$$

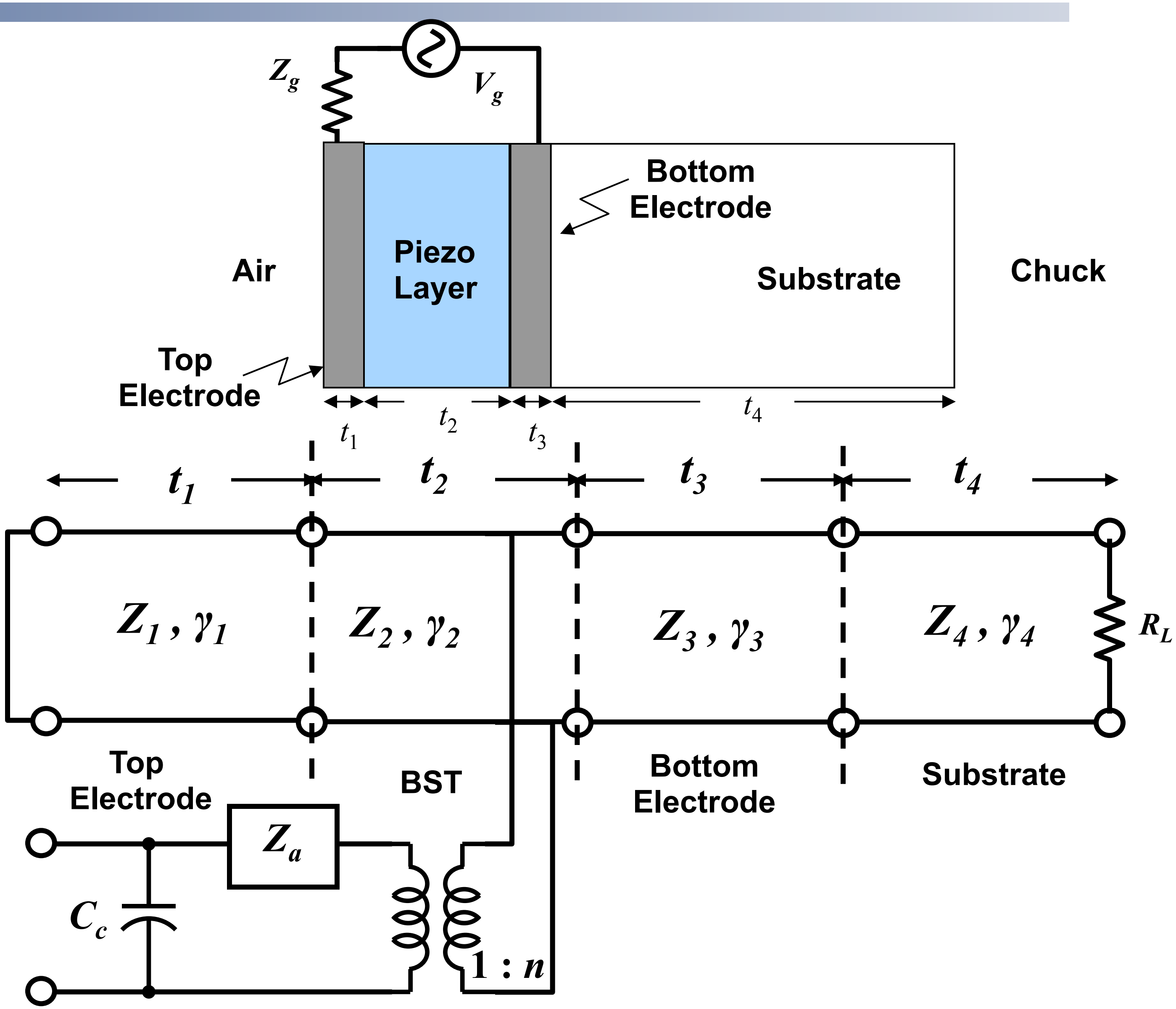
$$Z_a = \frac{h_{33}^2}{\omega^2} \frac{\sinh(\gamma t)}{Z_0}$$

$$n = \frac{j\omega}{2h_{33}} \frac{Z_0}{\sinh\left(\frac{\gamma t}{2}\right)}$$

$$Z_0 = A\rho v_a$$

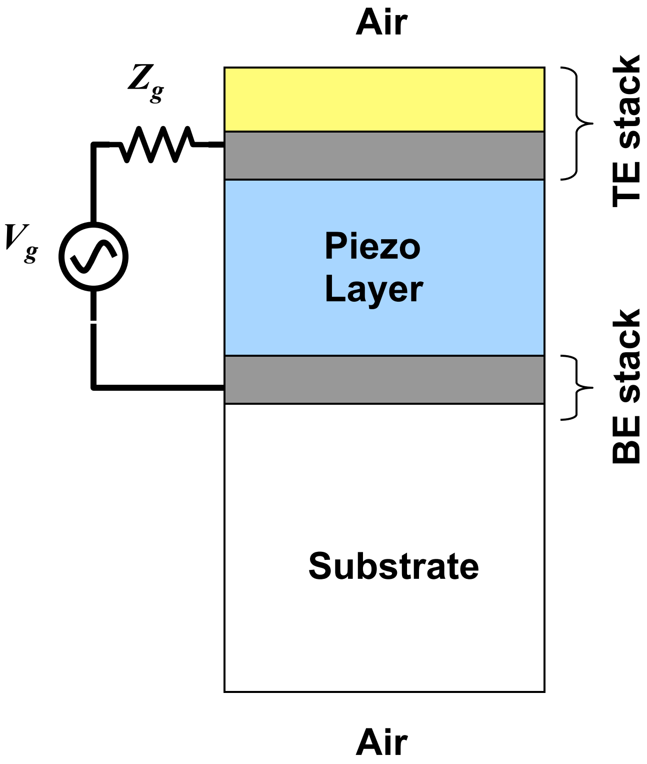
†R. Krimholtz, D. A. Leedom, G. L. Matthaei, "New equivalent circuits for elementary piezoelectric transducers," *Electron Letters*, pp. 398-399, May 1970

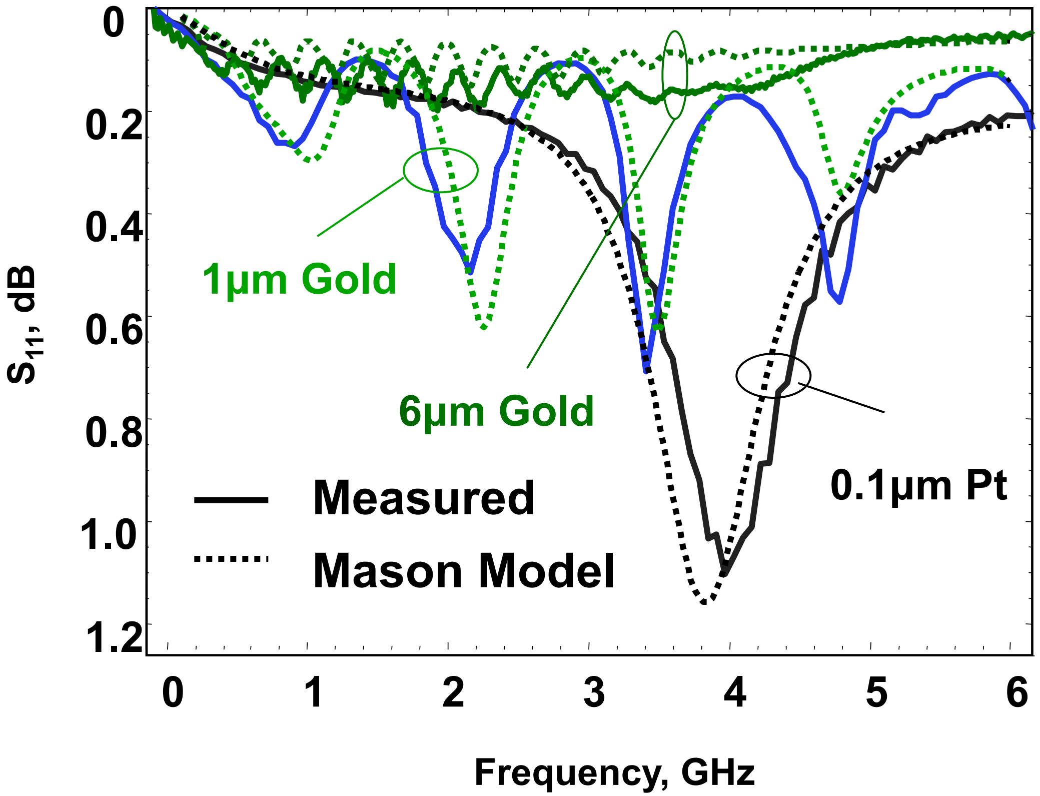
Modeling: Mason's Model



Modeling: Mason's Model

Modeling acoustic resonances

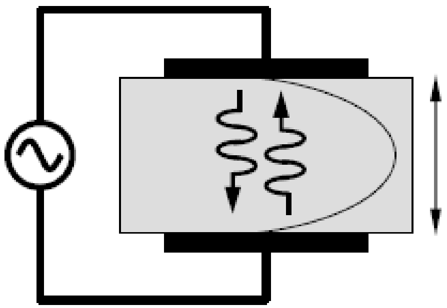




†R.A. York, Tunable Dielectrics for RF Circuits, Chap. 6 of Multifunctional Adaptive Microwave Circuits and Systems M. Steer, D., Wiley 2008

Modeling BAW Resonators

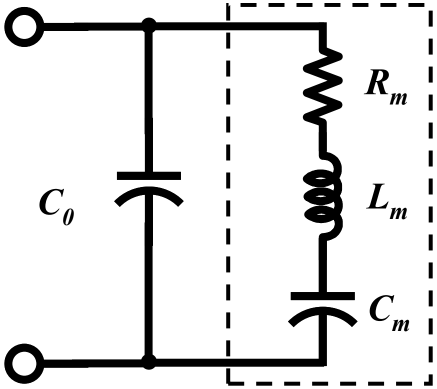
BAW Resonator



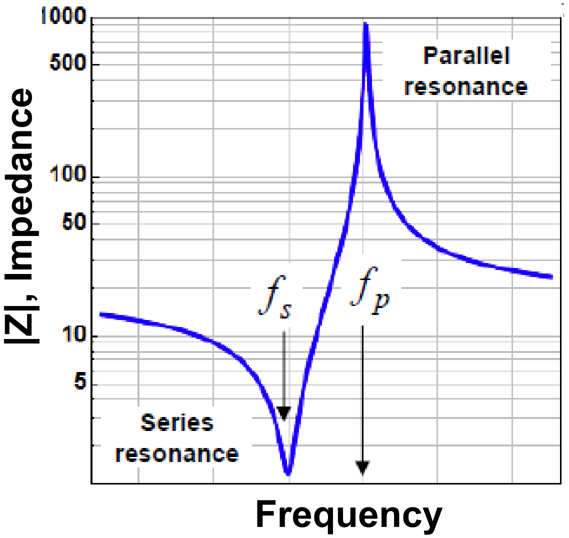
$$t \approx \frac{\lambda_a}{2}$$

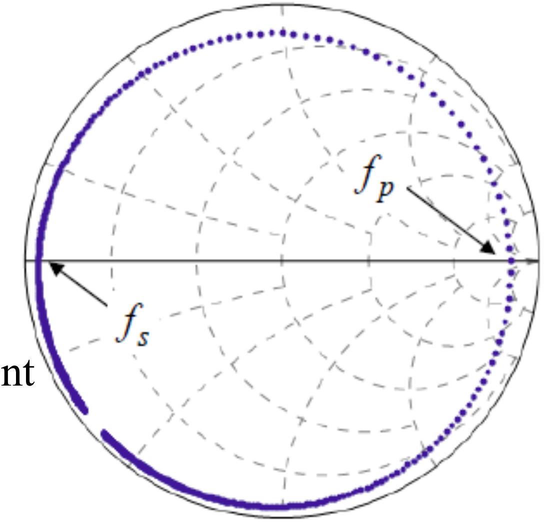
Applied electric field couples to an acoustic standing wave in the piezoelectric material

Butterworth-Van Dyke (BVD) model



Models acoustic response





$$R_m = \frac{(\pi/2)^2 \alpha_{eff}}{k_{t,eff}^2 \omega C_0} \qquad k_{t,eff}^2 = \frac{\pi}{2}\frac{f_s}{f_p}\cot\left(\frac{\pi}{2}\frac{f_s}{f_p}\right)$$

$$C_m = \frac{8k_{t,eff}^2}{\pi^2} C_0 \qquad Q_{s,p} = \frac{f_0}{2}\frac{d\phi}{df}$$

$$L_m = \frac{1}{C_m (2\pi f_s)^2}$$

$k_{t,eff}^2$: Effective electromechanical coupling coefficient

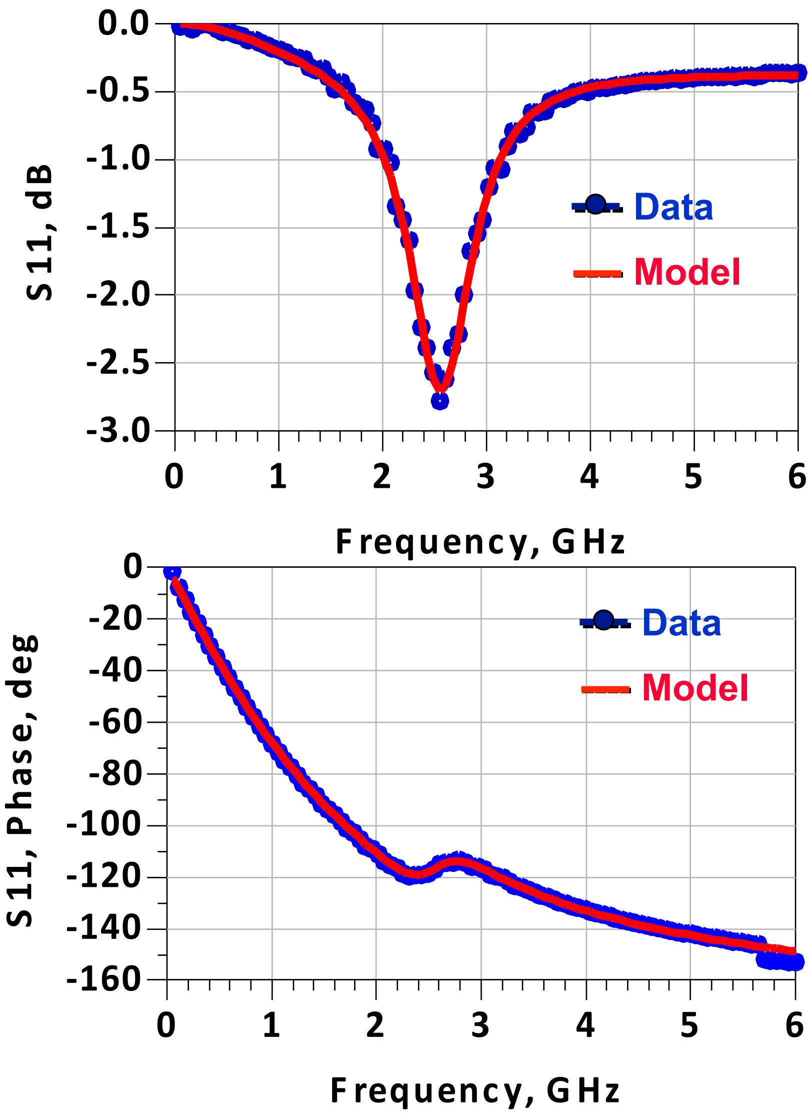
α_{eff} : Effective acoustic attenuation factor

†J. Rosenbaum ,Bulk Acoustic Wave Theory and Devices, Artech House, Boston, Ma, 1988.

Piezoelectric Resonator

UCLA

R_m	103.3 Ω
L_m	23.5 nH
C_m	0.171 pF
R_s	1.1 Ω
C_o	1.94 pF
K_t	0.33
F_r	2.5 GHz

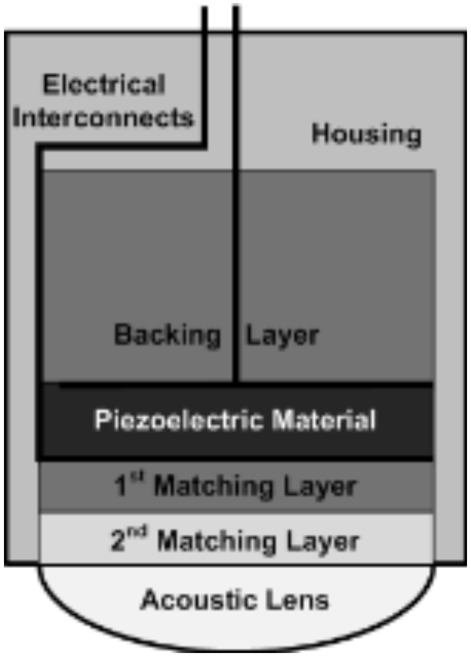


Material Parameters

Acoustic Properties of Materials

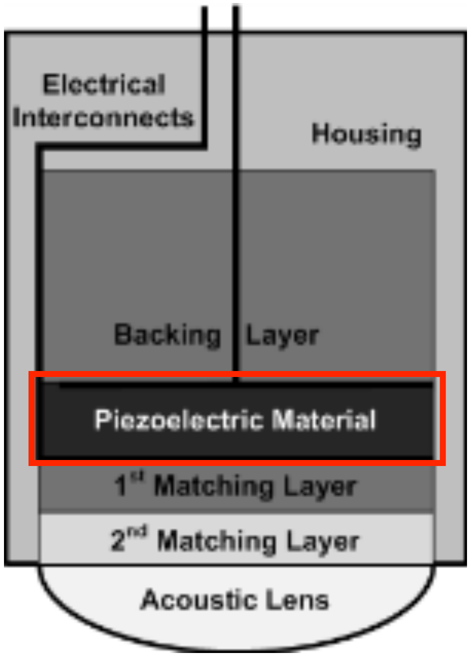
Material	Speed of Sound (m/s)	Density (kg/m^3)	Attenuation (dB/cm MHz)	Acoustic Impedance (MRayl)
Air	330	1.2	-	0.0004
Water	1480	1000	0.0022	1.48
Blood	1584	1060	0.2	1.68
Bone, Cortical	3476	1975	6.9	7.38
Fat	1478	950	0.48	1.40
Muscle	1547	1050	1.09	1.62
Tendon	1670	1100	4.7	1.84
Soft tissue, Average	1561	1043	0.54	1.63
PZT	4350	7500	-	33.0
PVDF	2300	1790	-	4.2
Epoxy	2640	1080	4-8 @ 2 MHz	2.85
Silicone rubber	1050	1180	2.5 @ 0.8 MHz	1.24
Tungsten	5200	1940	-	101.0

Ultrasound Transducer



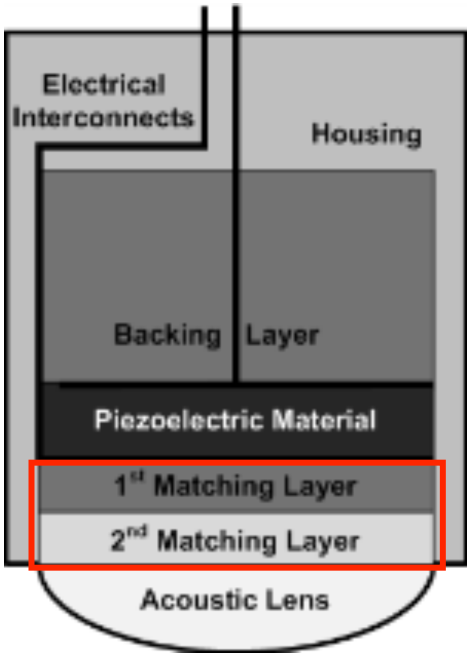
- Comprised of a number of layers and materials.
    - Piezoelectric material (most important)
    - Matching Layer (maximize transmission)
    - Backing Layer (Mechanical damper)
    - Acoustic lens (focus transducer, optional)

Piezoelectric Layer



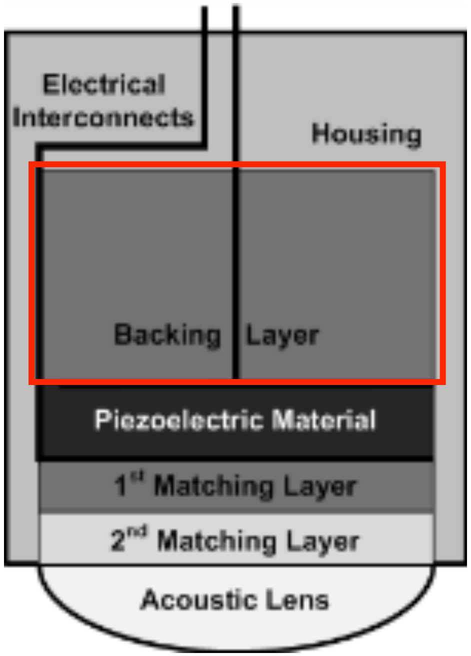
- Generate and detect acoustic wave.
    - Material properties
    - Thickness and speed of sound
    - Resonant frequency (influences resolution, penetration depth, and beam characteristics)

Matching Layer



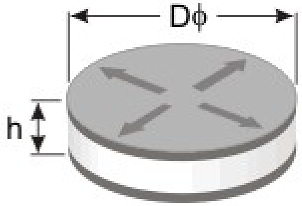
- Transfer energy out of the transducer.
  - Low attenuation
  - Single Layer or multiple Layer
  - Multiple layer to increase transmission
  - Thickness is very important

Backing Layer

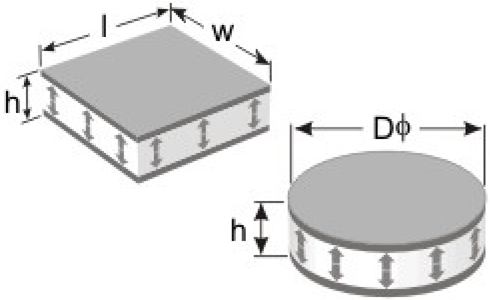


- Reduce internal reverberations
  - Reverberations cause ringing
  - Ringing elongates the pulse and reduces axial resolution.
  - Damping layer
  - Broader bandwidth, and narrowed acoustic pulse width.
  - Backing layer reduces efficiency

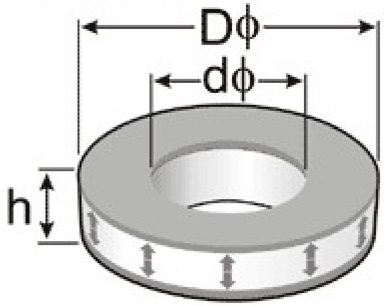
Piezoelectric Modes



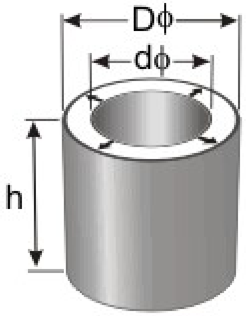
Radial Mode
(Thin Disc, $D\phi > 3.16\ h$)



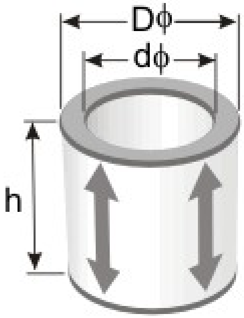
Thickness Mode
(Thin Disc or Plate, l, w, $D\phi > 3.16\ h$)



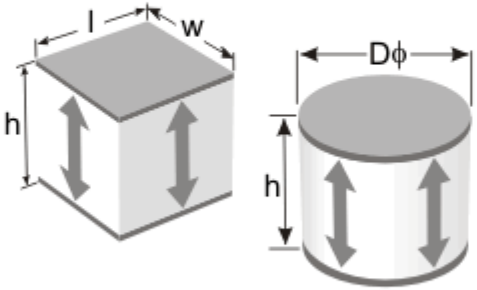
Thickness Mode



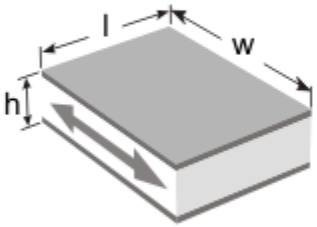
Circumferential Mode (Thin Wall Tube)



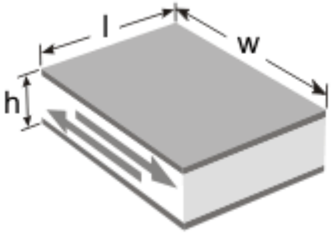
Length Mode (Thin Wall Tube)



Longitudinal Mode
(Cylinder or Block, $h > 3.16\ l$, w, $D\phi$)



Length Mode
(Thin Bar, $l > 4w$, $w > h$)



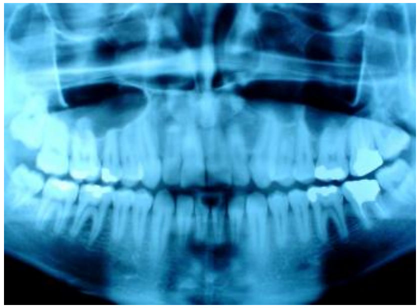
Shear Mode Plate

Applications and Transducers

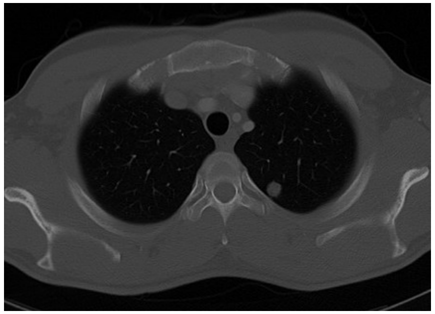
Medical Imaging

- Ultrasound
  - Pressure waves
  - Soft tissue imaging
  - Non-ionizing
  - Real-time
  - Provides depth information
  - $
- X-ray
  - Beam directed through body
  - A portion of the photons are attenuated by body tissues
  - Radiolucency depends on atomic #, thickness, ρ
  - Hard tissues
  - Projection
  - $
- Computed Tomography (CT)
  - 2D X-ray slices
  - Hard tissues
  - Contrast agents (I, Ba) – vessels, organs
  - $$$

Sonogram of fetus



Dental X-ray radiograph

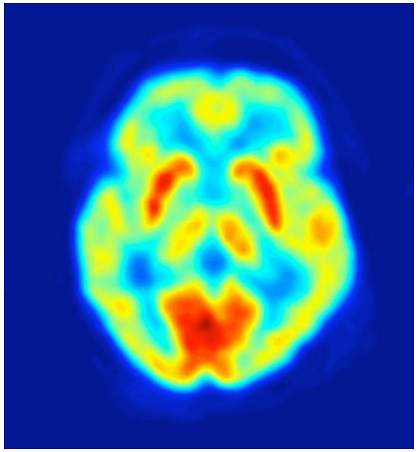


CT image of chest

Medical Imaging

- MRI
  - Magnetic field interacts with radio waves
  - Relaxation of H nuclei in water or lipids
  - Soft tissues
  - $$$$
- Nuclear Medicine
  - Functional imaging
  - Radionuclides are introduced and emitted radiation measured
  - Positron Emission Tomography (PET)
  - Single Photon Emission Computed Tomography (SPECT)
  - $$$$

MRI image of head



PET scan of brain

Ultrasound Imaging Systems

- Bedside ultrasound

GE Medical Systems

Ultrasound transducers

GE Logiq 700

- Portable ultrasound

Sonosite portable ultrasound systems

Ultrasound System Manufacturers

- GE (USA)
  - 26% market share (23% five years ago)
  - Manufacturers largest variety of probes
- Philips/ATL (Netherlands)
  - 18-20% market share (20% five years ago)
  - Focus on high end probes
- Siemens/Acuson (Germany)
  - 12% market share (22% five years ago)
  - Focus on high end probes
- Hitachi/Aloka (Japan)
  - 7% market share
- Toshiba (Japan)
  - 7% market share
- Esaote (Italy)
  - Large in Europe
- Samsung Medison (S. Korea)
- Sonosite (USA)
  - Portable ultrasound
- Mindray (China)
  - Rapid growth
- Zonare (USA)
  - Does not manufacture probes
- Ultrasonix (Canada)
  - Does not manufacture probes
- Many other Chinese companies
  - SIUI, Chinson, Sonoscape, Landwind (~30 companies)
  - Low end markets

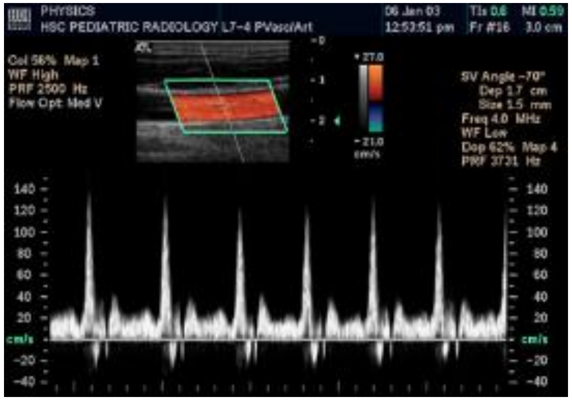
Independent Probe Manufacturers

- USA
  - STI (#1 in world)
  - Blatek
  - Tetrad (recently acquired by STI)
- Japan
  - Panasonic (#2 in world)
  - UJRC
  - NDT
  - Okusonic
- China
  - 10+ companies
- EU
  - Vermon (France, #3 in world)
  - Imasonic (France)
  - Odelft (Netherlands, specialize in TEE probes)
- Korea
  - Human Scan (Single crystal probes)
  - Prosonic (Samsung)

Medical Ultrasound

- Tissue interfaces
- Fluid and air-filled organs
- Some common applications
  - Abdominal ultrasound
  - Thoracic ultrasound
  - Pelvic ultrasound
  - Neurosonography (infants)
  - Mammography
  - Ocular
  - Thyroid
- Doppler
- Color Flow

Medical ultrasound system



Doppler ultrasound of carotid artery

Fetus at 12 weeks

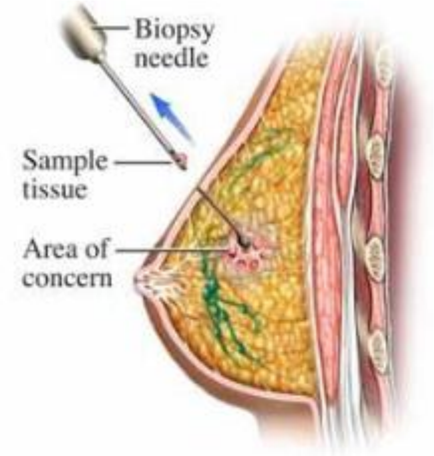
4D Ultrasound

- 3D imaging in real time = “4D”
- Designs
    - 2D array
        - Expensive
    - Rotating linear array
        - Lower cost

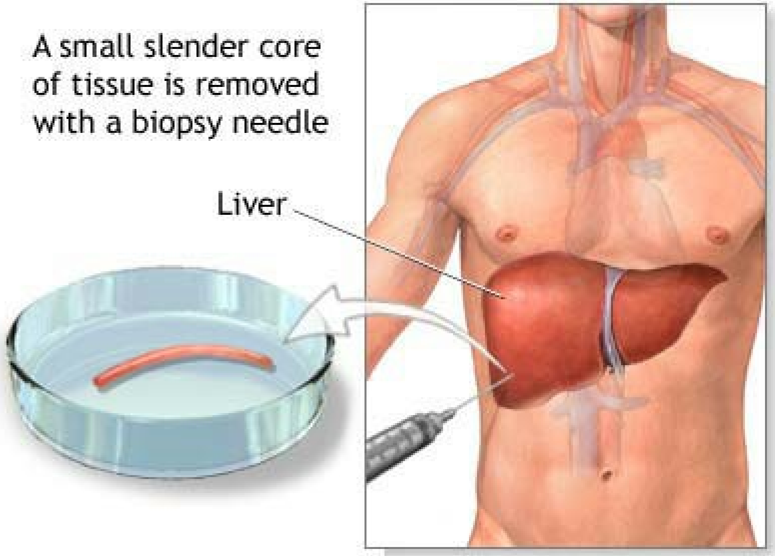
4D ultrasound image of fetus at 28 weeks

Ultrasound Guidance

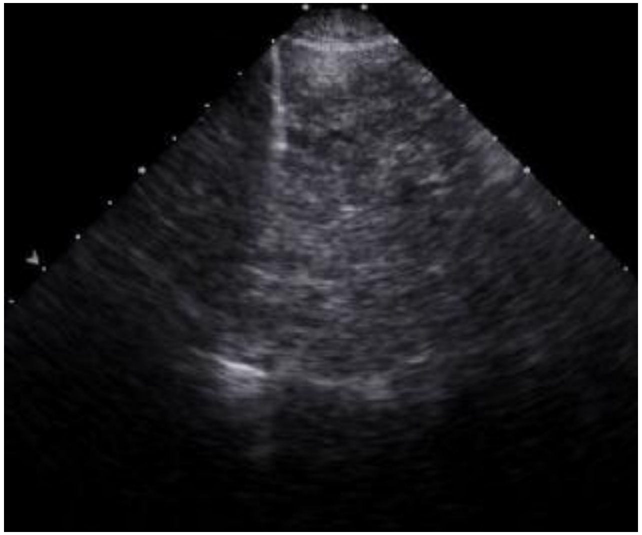
- Ultrasound used in real-time to guide minimally invasive procedures
  - Liver biopsy – hepatitis, tumors, scarring
  - Thyroid biopsy - tumors
  - Breast biopsy - tumors



Breast biopsy



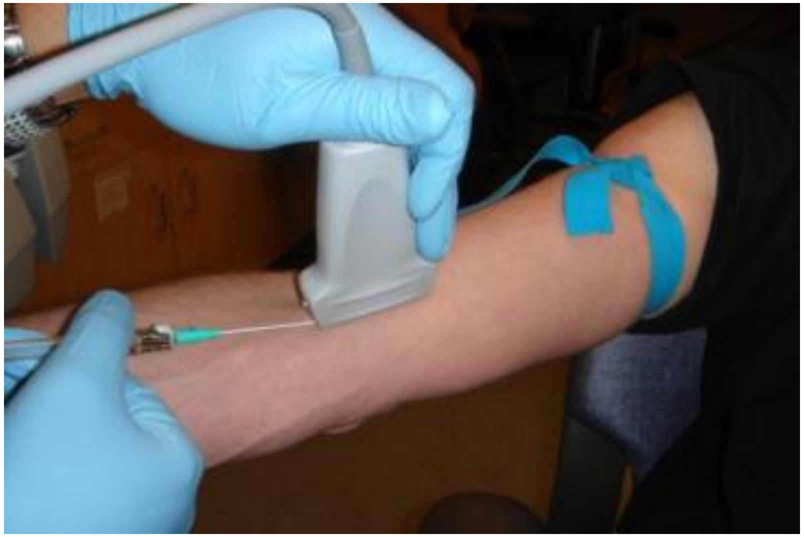
Liver biopsy



Ultrasound needle guidance of liver

Ultrasound Guidance

- Central line placement
  - Insertion thru needle and sheath
  - Access thru neck (jugular vein), chest (subclavian vein), groin (femoral vein), or arm
  - Uses
    - Monitoring of central venous pressure
    - Delivery of antibiotics, medications, chemotherapy agents
    - Dialysis

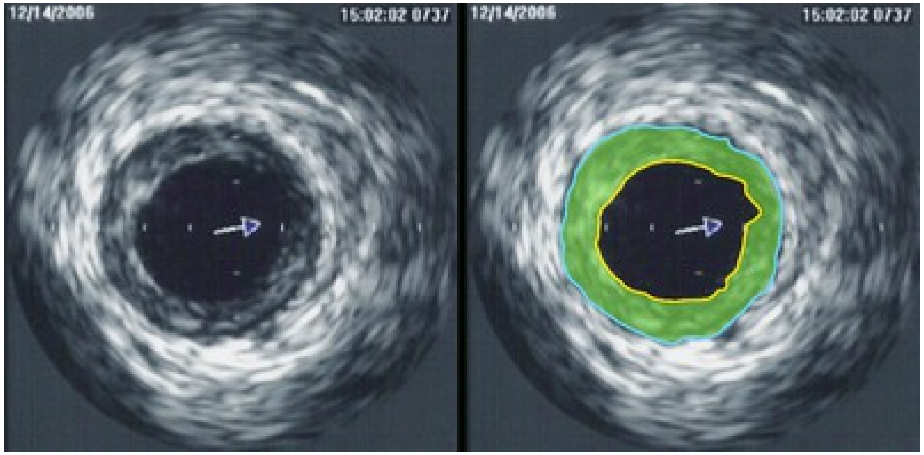


Ultrasound guidance of picc line placement

Central line kit

Intravascular Ultrasound (IVUS)

- Catheter-based transducers
- Common applications
  - Vascular endothelium
  - Plaques in coronary arteries
  - Angioplasty screening
  - Stent monitoring
- Types
  - Single-element rotational
  - Linear arrays
  - Circumferential arrays



IVUS image with color-coded plaque (green)

IVUS transducers

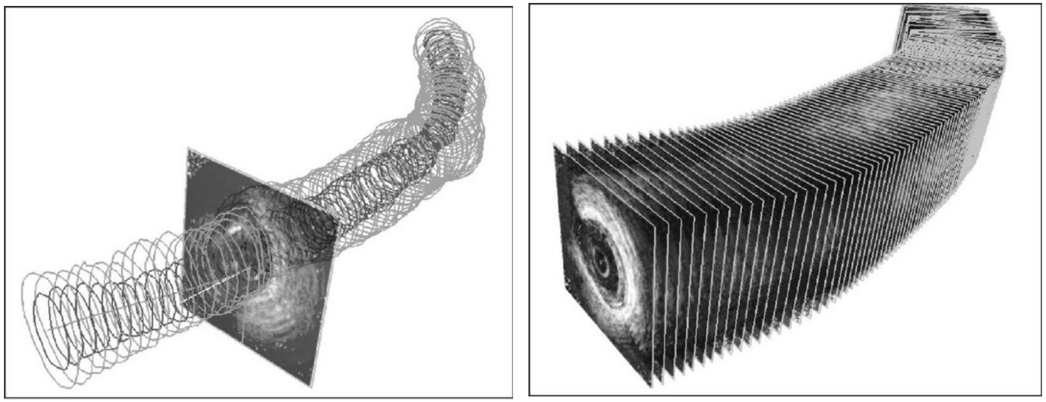


Image reconstruction from rotational IVUS transducer

Laparoscopic Ultrasound

- Laparoscopic tool
  - Inserted through cannula as in minimally invasive surgery
  - Forward looking transducer
  - Articulation
  - Liver tumors
  - Kidney tumors
  - Guidance during cyroblation
  - Guidance during biopsy

Laparoscopic probe insertion

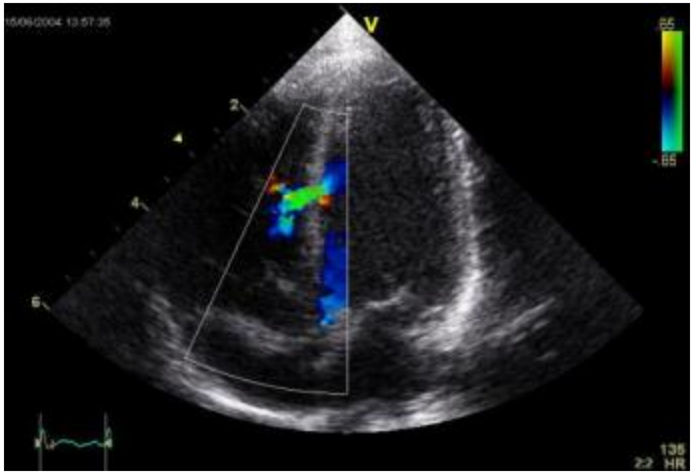
Laparoscopic ultrasound tool with biopsy needle

Laparoscopic ultrasound during MIS

Transesophageal Ultrasound

- Transesophageal Echocardiograph (TEE) Probes
  - Provides imagery of the heart structure and blood flow in the heart
  - For diagnosis of blood clots, aneurysms, valve dysfunction, septal wall defects, backflow of the blood through the valves, infections of the heart valve and cardiac masses
  - Inserted through esophagous
  - Rotating array – complicated
- Endobrachial ultrasound (EBUS)
  - Imaging of lungs through larnxy

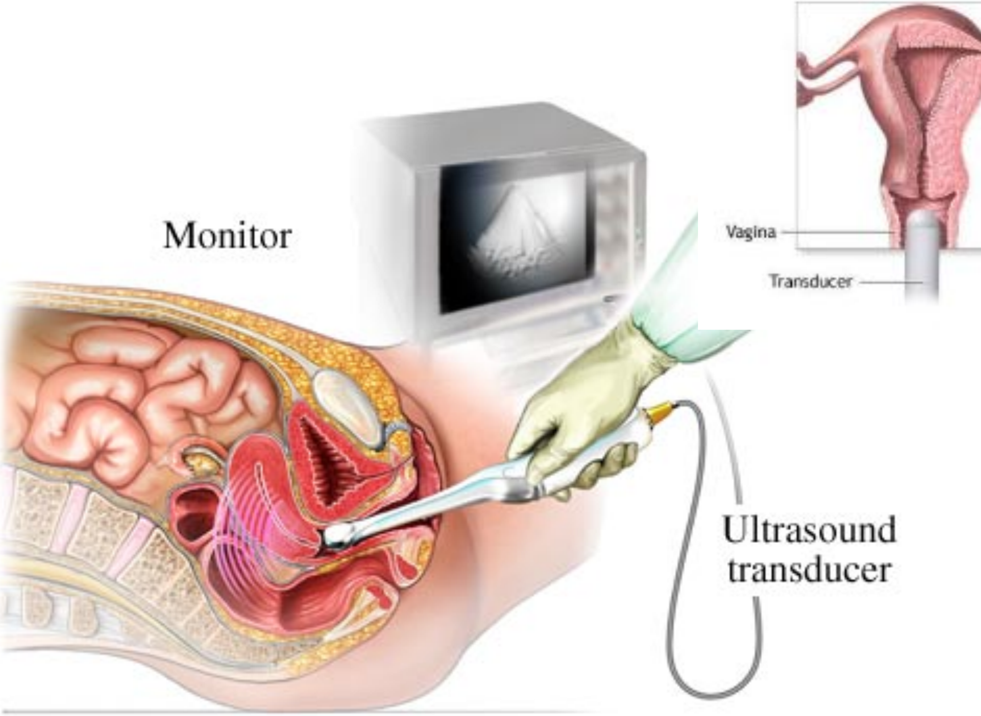
TEE Probe

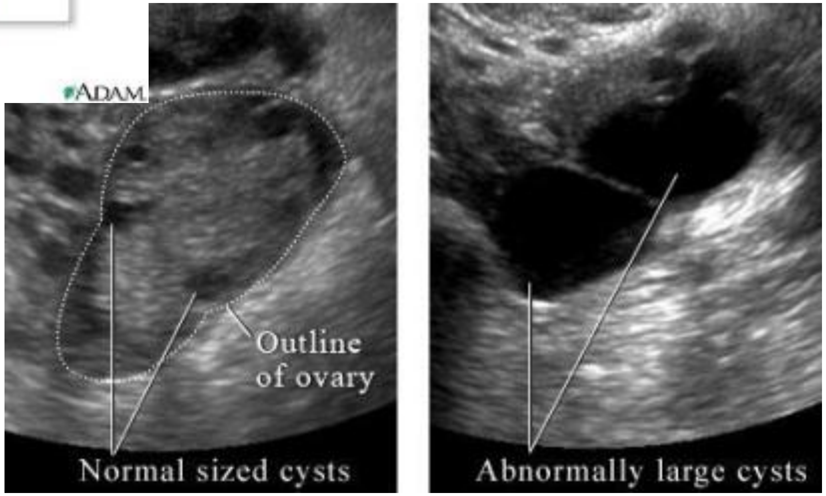


TEE Image

Transvaginal ultrasound

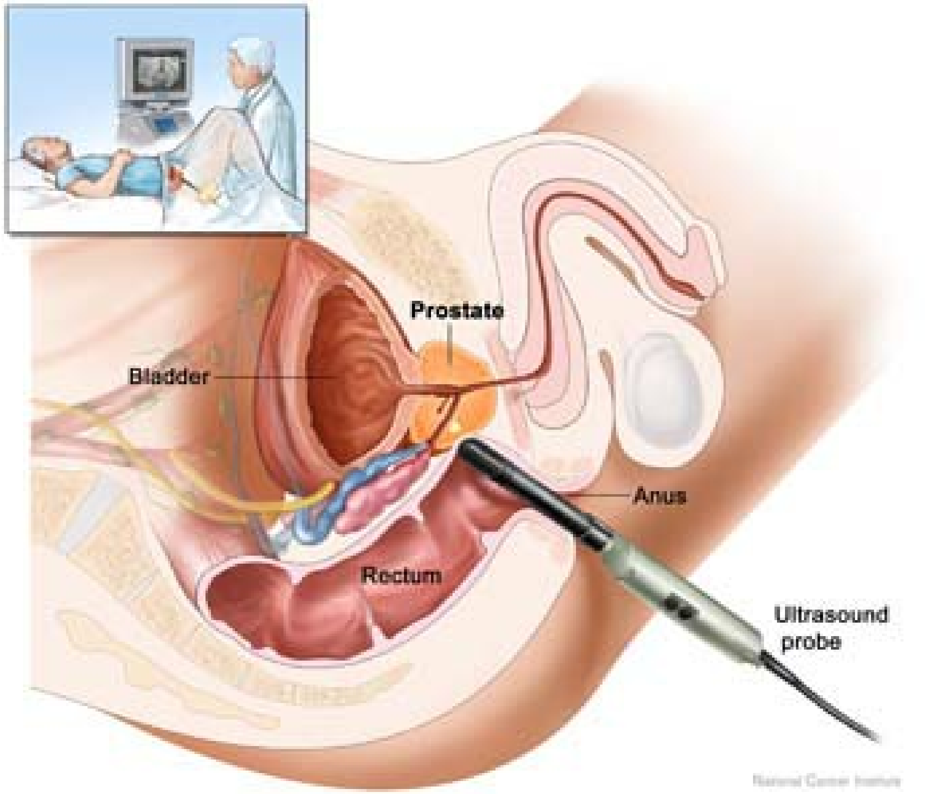
- Organs
    - Ovaries, uterus
- Uses
    - Typically for fertility problems
    - Endometrial biopsy
    - Remove eggs from ovaries



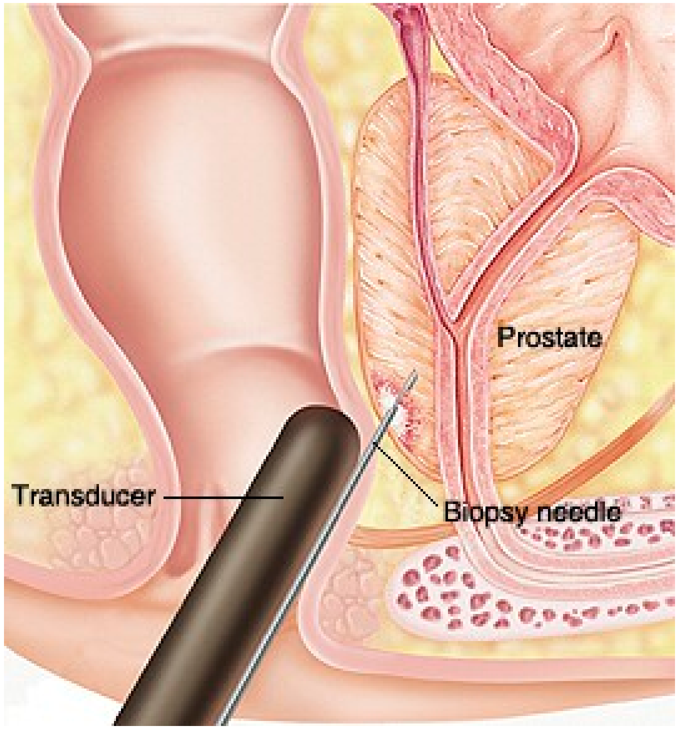


Transrectal Ultrasound

- Organs
  - Prostate, bladder, seminal vesicles
- Uses
  - Prostate biopsy
  - Prostate size, infertility



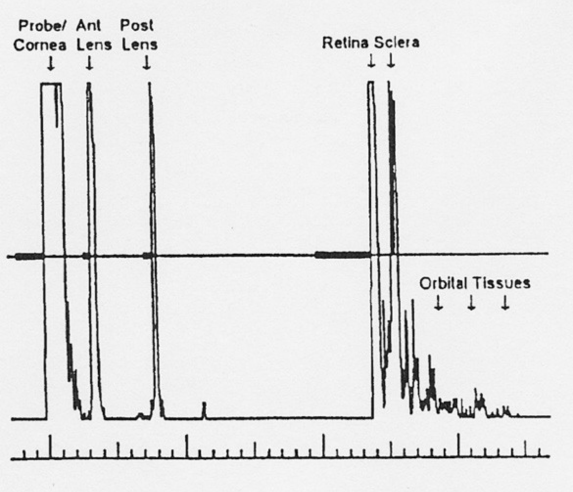
National Cancer Institute



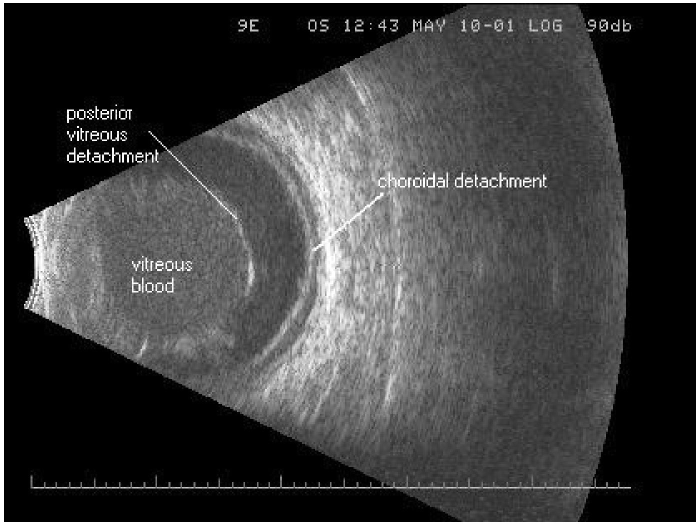
Ophthalmic Ultrasound

- First application of ultrasound in 1940s
- A-scan ocular ultrasound
  - View tissue interfaces
  - Measures tissue depths and densities
- B-scan ocular ultrasound
  - 2D map of the eye

Ophthalmic ultrasound



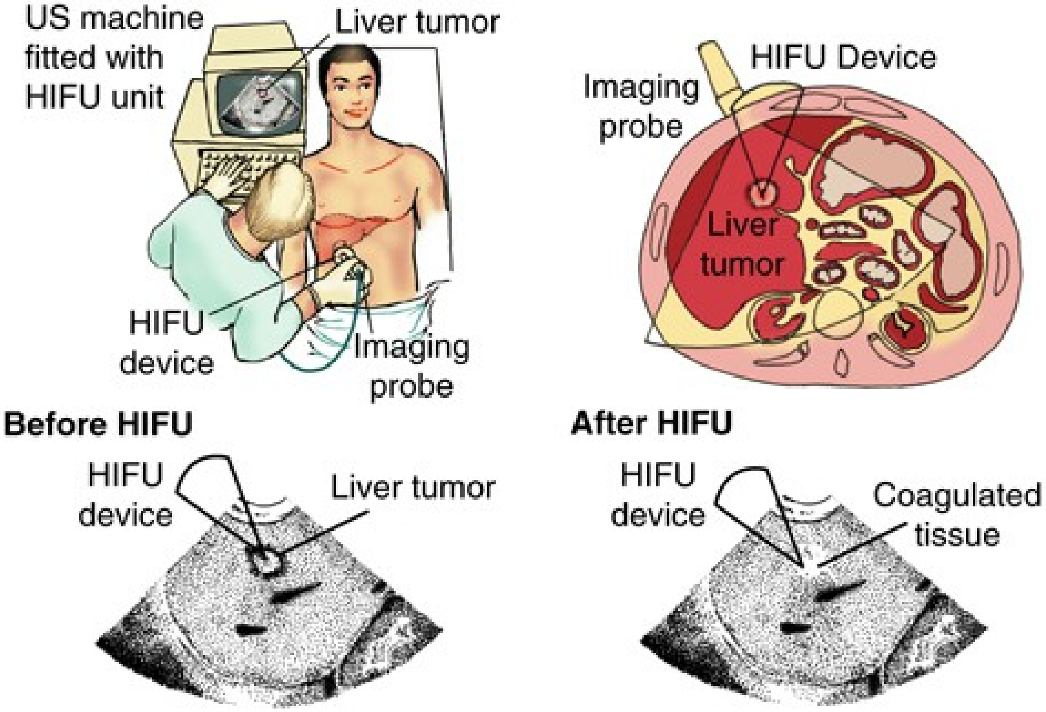
A-scan ocular ultrasound



B-scan ocular ultrasound

High Intensity Focused Ultrasound

- HIFU currently approved only for few applications
- MRI guidance most common for HIFU
- Most applications involve moving targets
  - Respiration, heart beat, pulse, muscle contractions
  - Need real-time imaging modality for guidance
  - Need adaptive control system to move beam focus



HIFU ablation of liver tumors

HIFU transducer

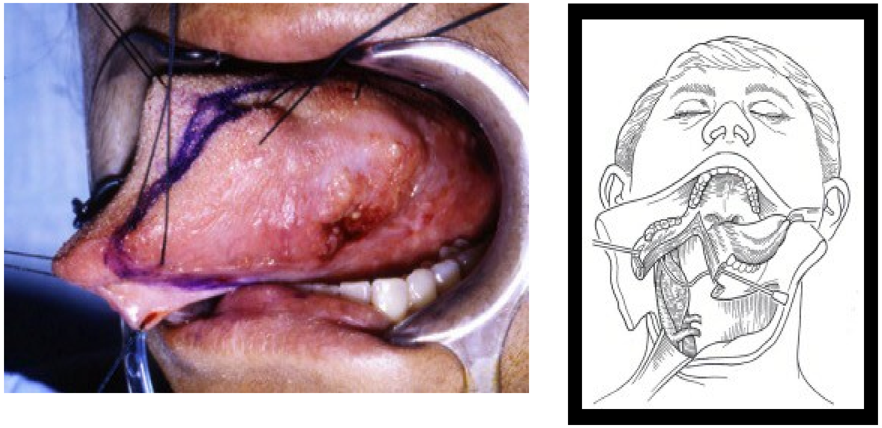
Vibro-Acoustography

Elastography

- Imaging technique that investigates the elastic properties of materials.
  - Medical applications: Tissue soft or hard
- Methods: Palpation, Ultrasound, MRI
- Ultrasound Elastography Methods
  - Strain imaging
  - Acoustic Radiation Force Imaging
    - Vibro-acoustography
  - Shear Wave Elasticity Imaging
  - Supersonic Shear Imaging
  - Transient Elastography
- Magnetic Resonance Elastography

Background: What is the problem?

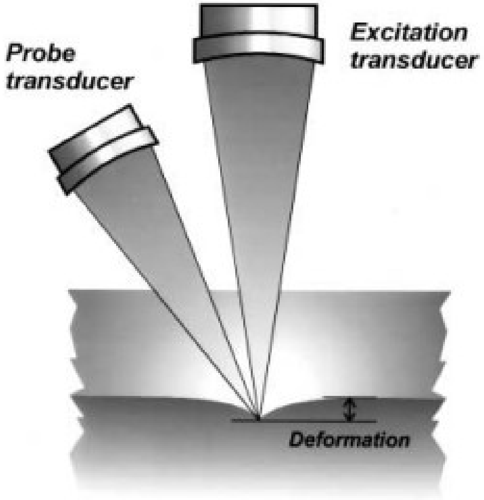
- **Need for a precise and accurate diagnostic of oral cancers**
  - To provide enhanced and clear boundary region
    - Ability to distinguish healthy from diseased region
  - **To assist surgeons with tumorectomy intra-operatively**
    - Contrast mechanism (Viscoelastic Properties)

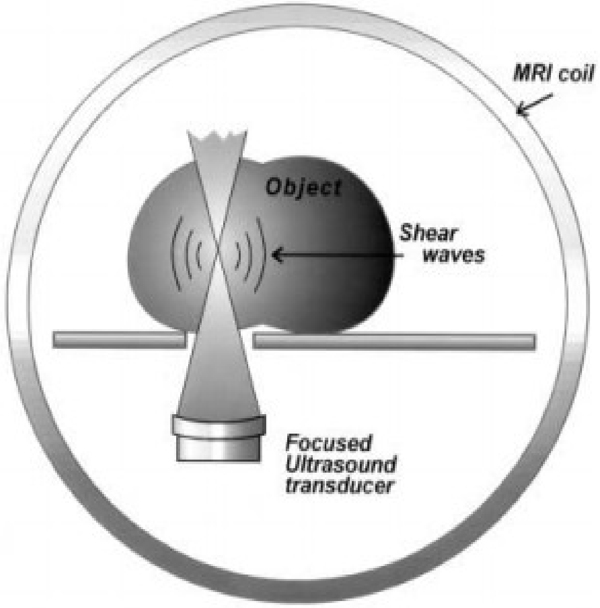


Oral squamous cell cancer (left) and surgical procedure (right)

What's out there?

- **Palpation, Gold Standard**
  - Probing for relative changes in deformation from an applied pressure
- **Transient Method**
  - Pulse-Echo Ultrasound
    - Doppler Ultrasound as a detector
    - Bulk Modulus
- **Shear-Wave Method**
  - Shear-wave Elastography
    - Use or radiation force of focused ultrasound
    - Phase-sensitive MRI as detector
      - Shear Modulus





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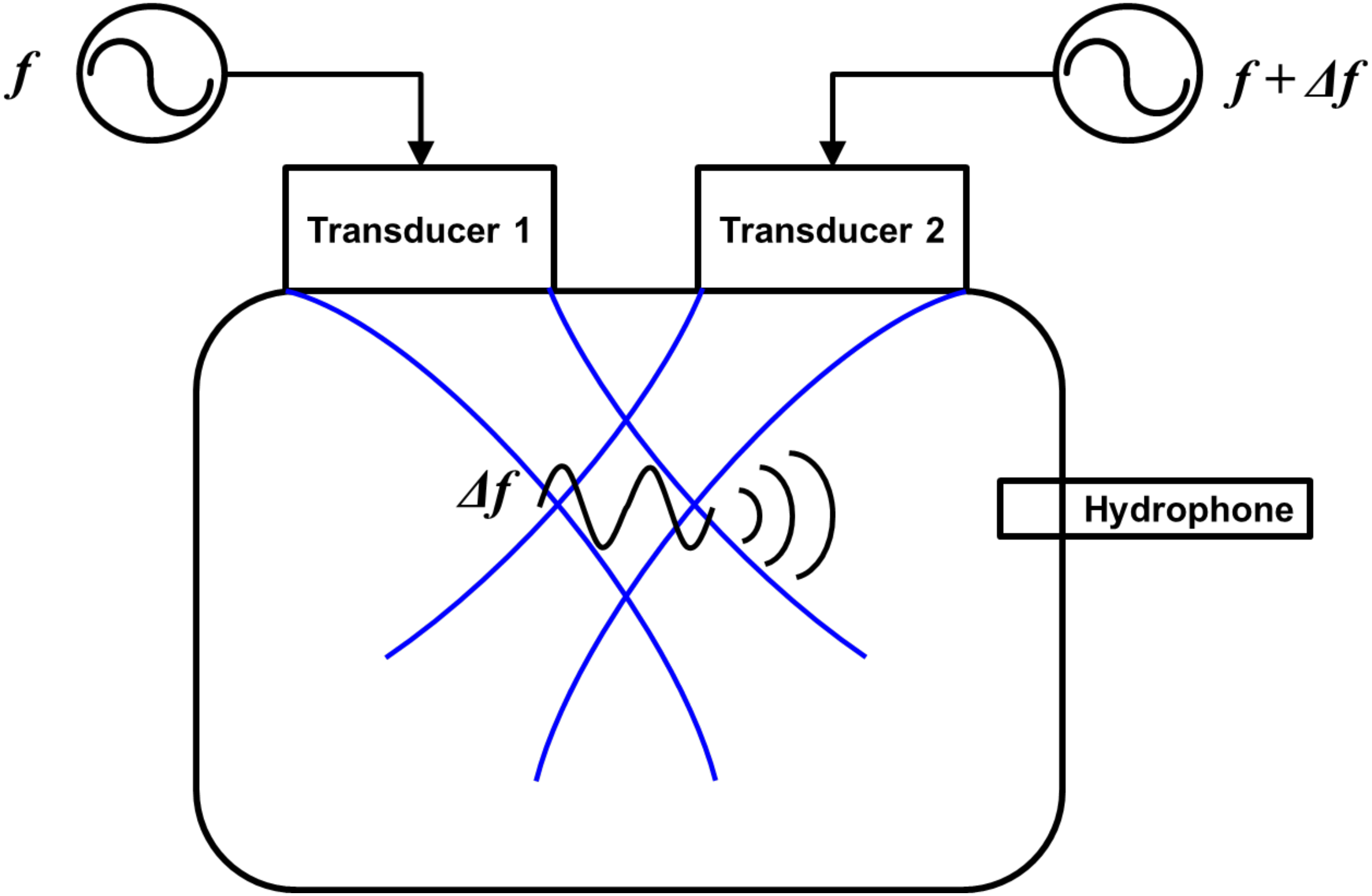
1. Fatemi, Mostafa, and James F. Greenleaf. "Imaging the viscoelastic properties of tissue." *Imaging of Complex Media with Acoustic and Seismic Waves*. Springer Berlin Heidelberg, 2002. 257-276.

Limitations of Current Methods

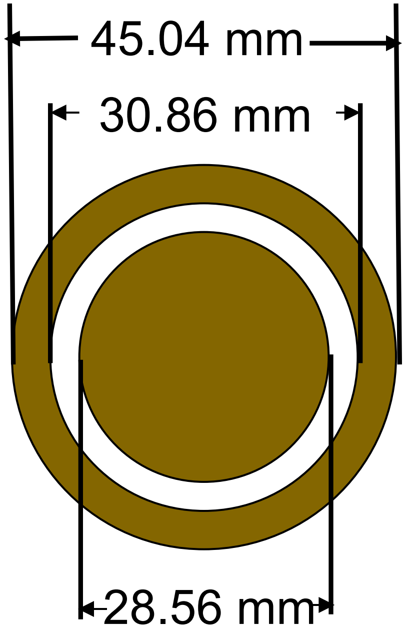
- **Palpation, Visual Assessment**
  - Very subjective, qualitative results
  - Lack of depth of penetration
  - No defined boundaries
- **Transient Method**
  - Poor sensitivity coupled with low contrast, speckle
  - Bulk modulus → compressional wave speed → less variation
- **Shear-Wave Method**
  - Long duration of scan time, Not OR Friendly, expensive, patient discomfort
  - Lack of penetration depth due to signal attenuation

1. http://www.nidcr.nih.gov/imagegallery/oralhealth/OralCancerExam.htm

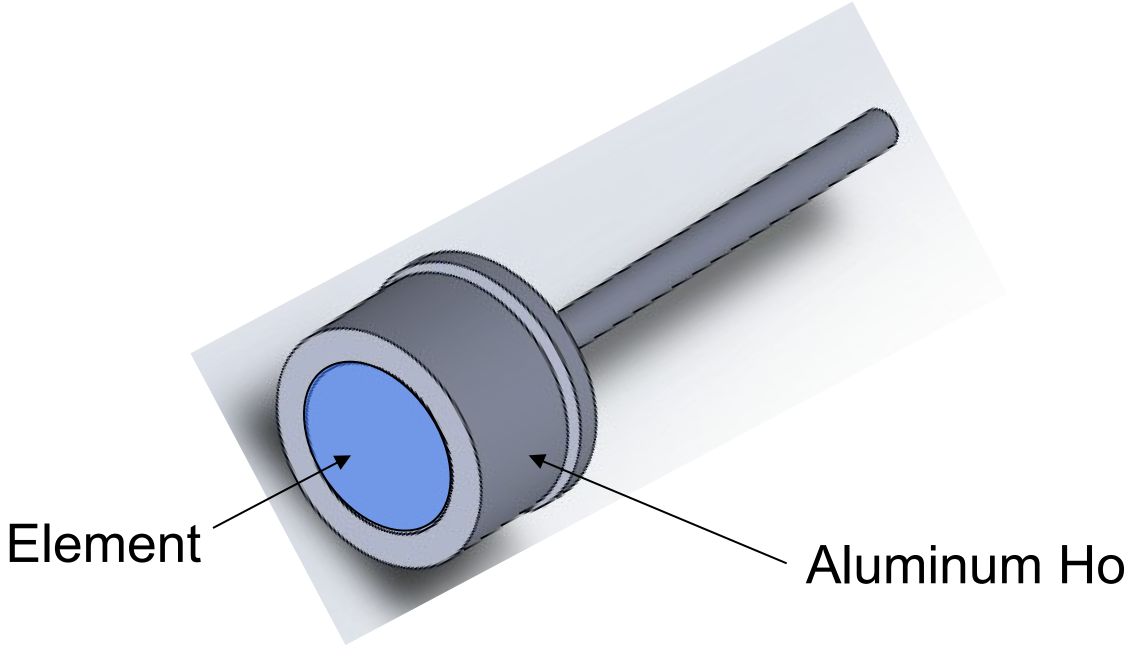
Basis of Vibro-Acoustography



Transducer Design



Confocal Curved Element

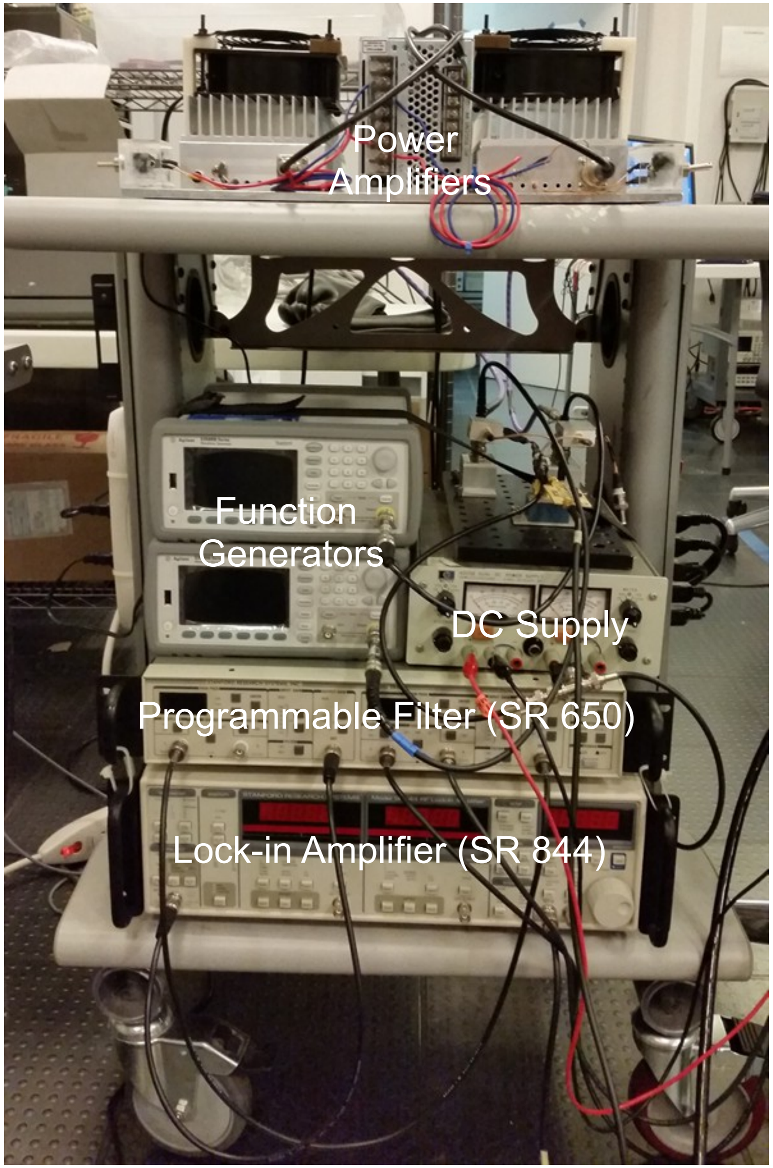


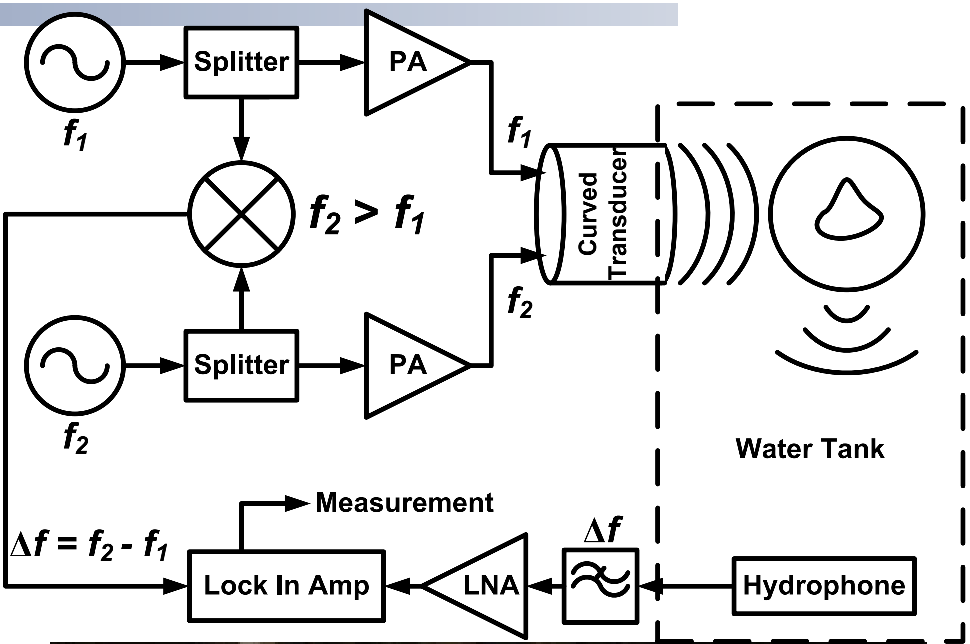
Transducer Housing

- Confocal piezoelectric Lead Zirconium Titanate (PZT) element
  - Two relatively close ultrasonic tone in low MHz and overlap at focus point to low kHz signal

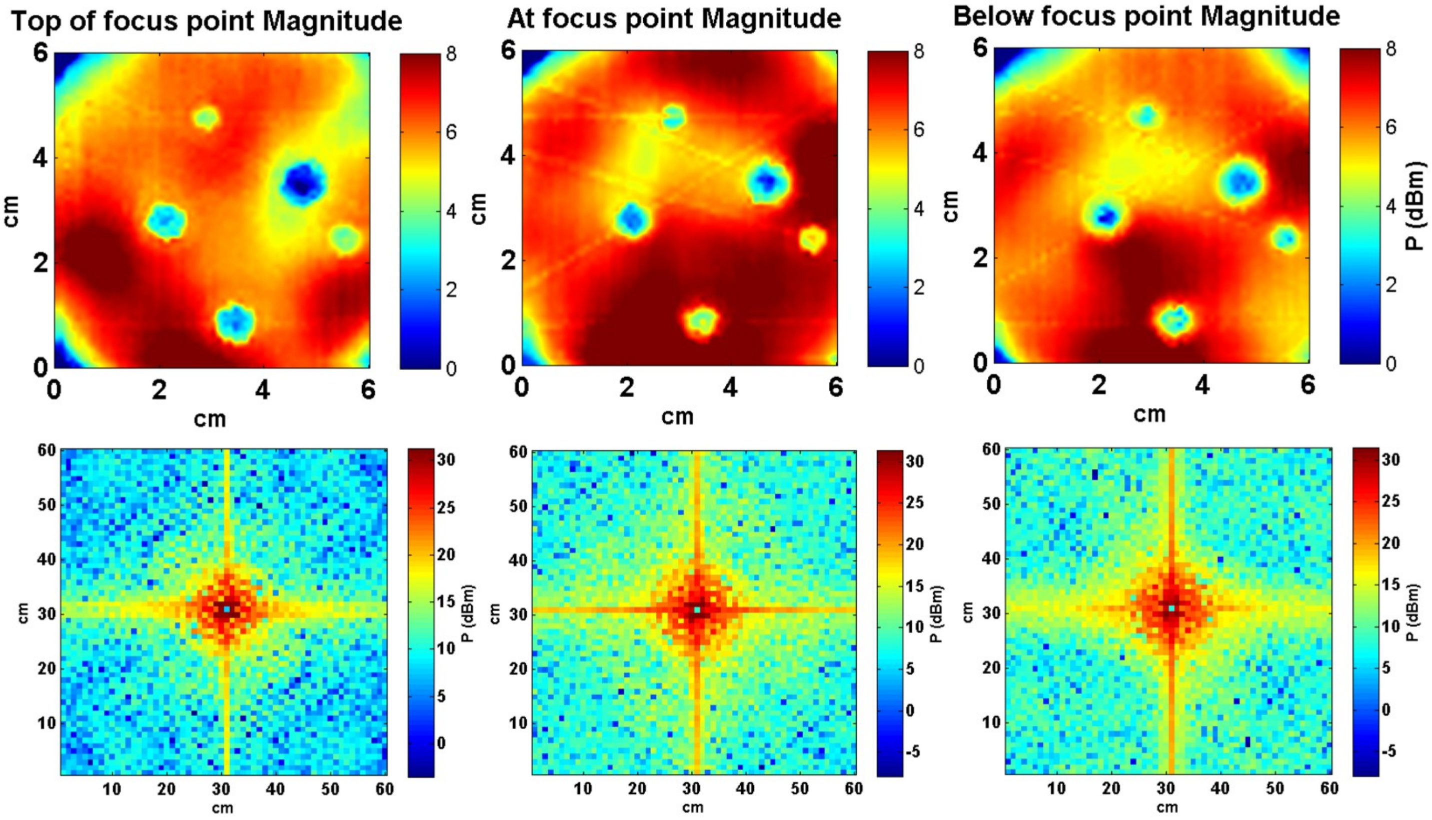
* 6 cm radius of curvature

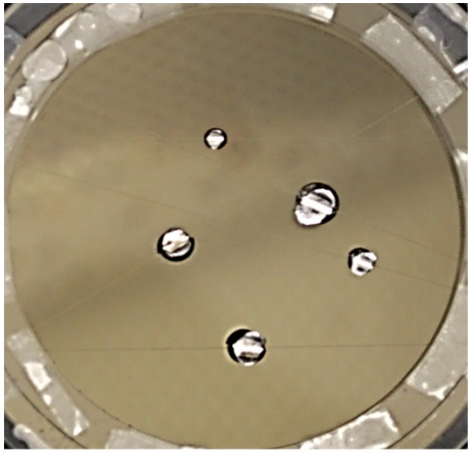
Visible images of USVA system and Setup





Fishing Weights Experiment





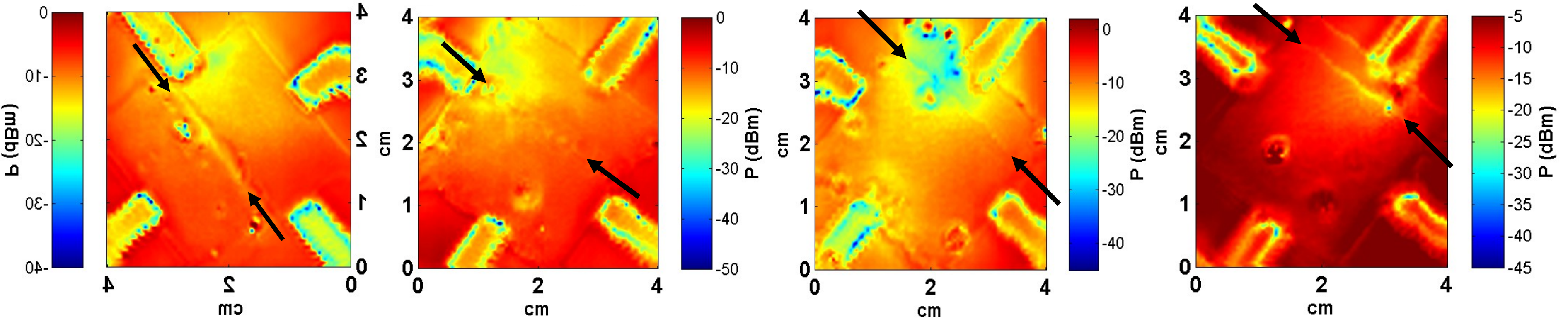
Visible Image

**1 mm fishing line

- 4 mm up and 4mm down the focus point
- 5.38,6.0,7.97,8.71,9.79 mm BB dimensions

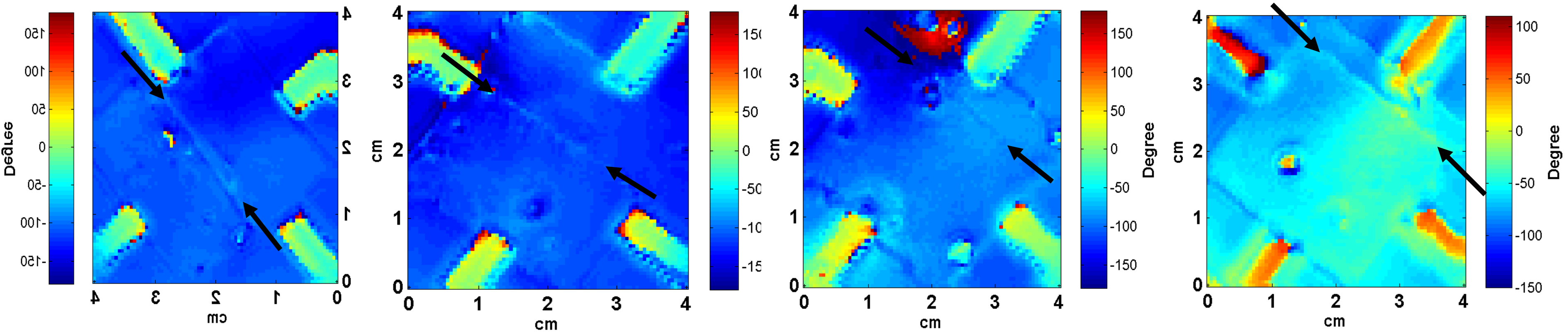
Agar Line pair Phantoms

Magnitude



15mm 2% & 15mm 4% 12mm 2% & 18mm 4% 8mm 2% & 23mm 4% 5mm 2% & 25mm 4%

Phase

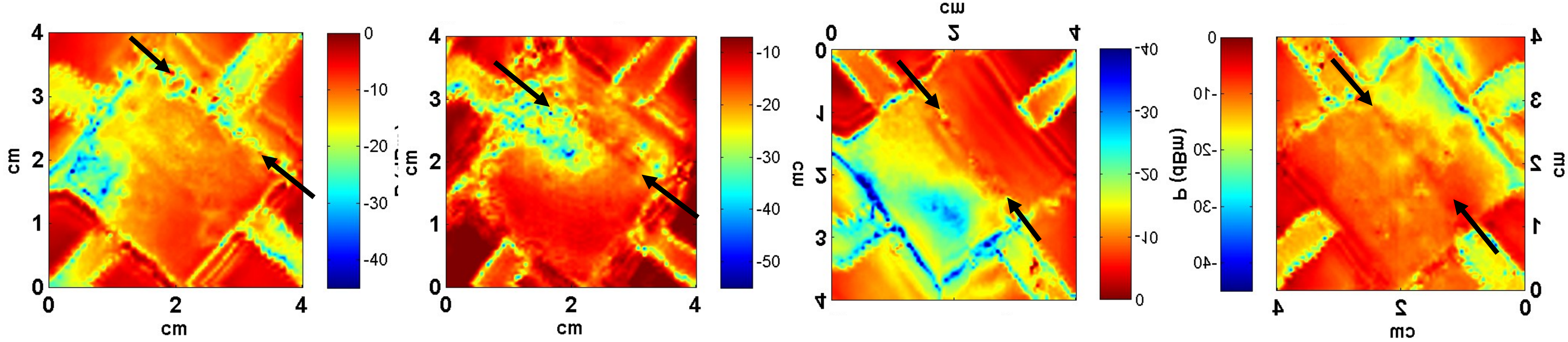


15mm 2% & 15mm 4% 12mm 2% & 18mm 4% 8mm 2% & 23mm 4% 5mm 2% & 25mm 4%

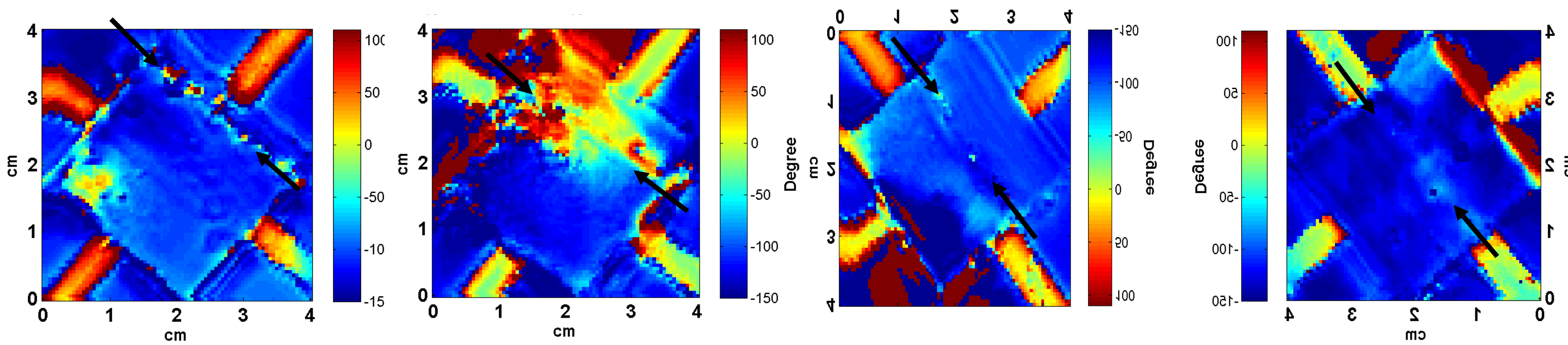
Sensitivity testing for homogeneous medium

Gelatin Line pair Phantoms

Magnitude

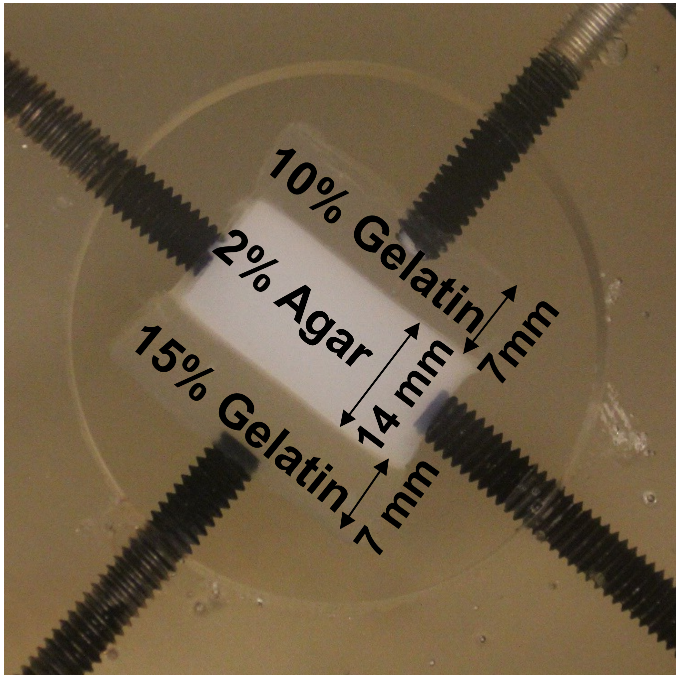

5 mm 10% & 25mm 20% — 7mm 10% & 24mm 20% — 11mm 10% & 21mm 20% — 16.5mm 10% & 15mm 20%

Phase


5mm 10% & 25mm 20% — 7mm 10% & 24mm 20% — 11mm 10% & 21mm 20% — 16.5mm 10% & 15mm 20%

Sensitivity testing for homogeneous medium

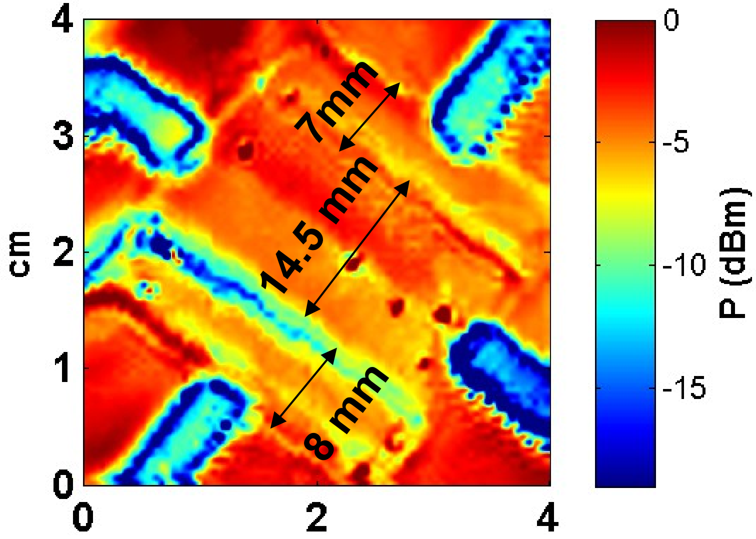
Multilayered Phantom



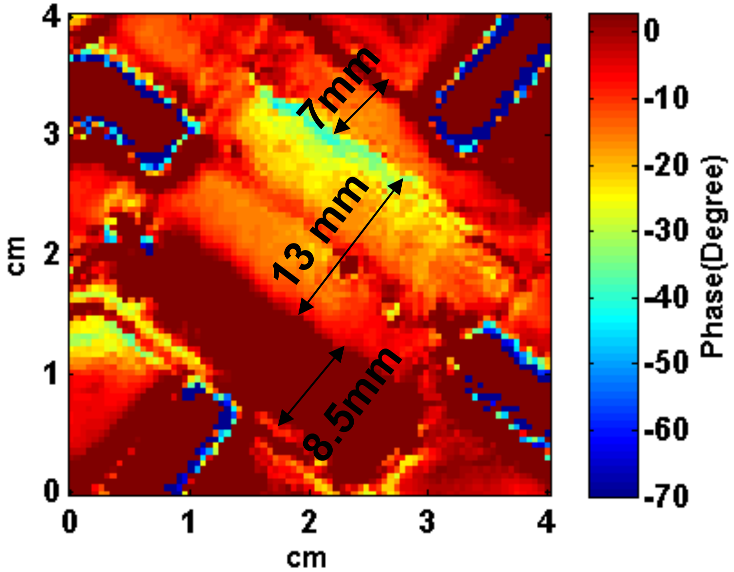
Visible Image

- Enhanced boundaries and edge detection
- Quantitative material distinction
- 500 µm motor step-size in X and Y axis

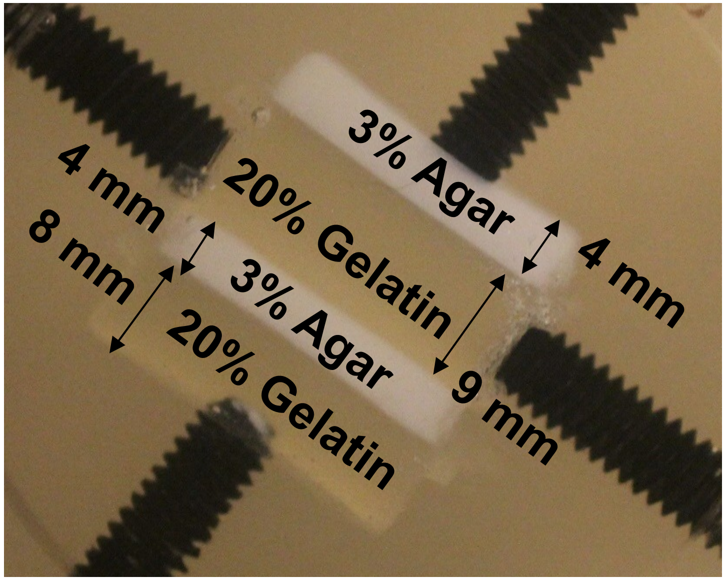
10% Gelatin, 2% Agar, 15% Gelatin Cube Magnitude



10% Gelatin, 2% Agar, 15% Gelatin Cube Phase

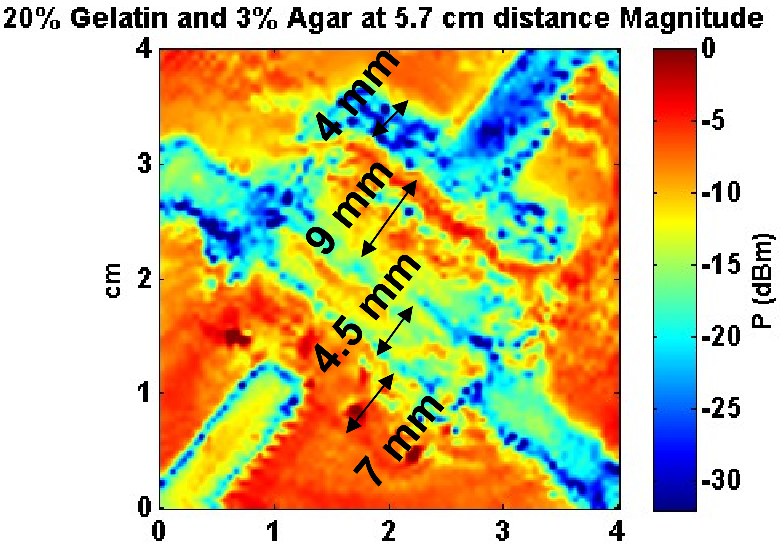


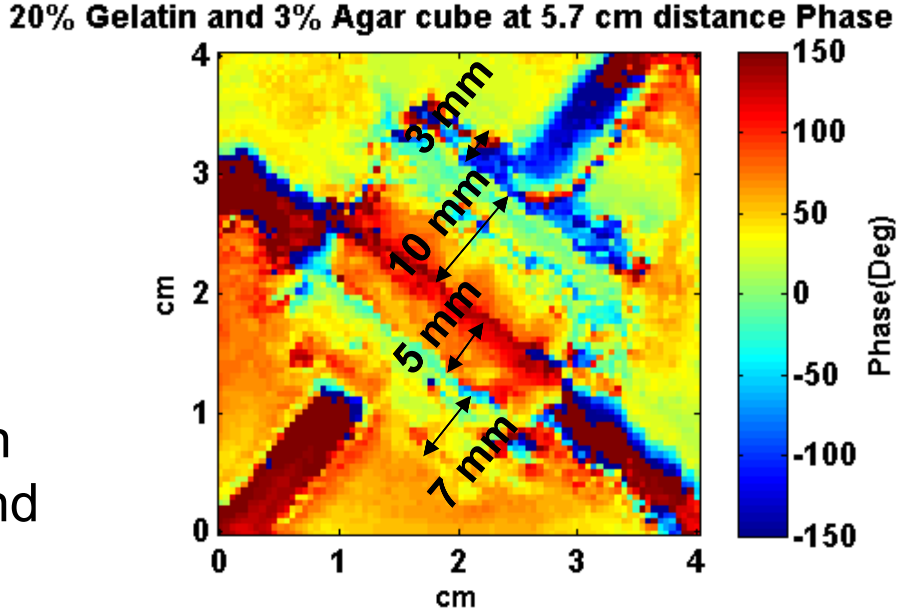
Multilayered Phantom Cont.



Visible Image

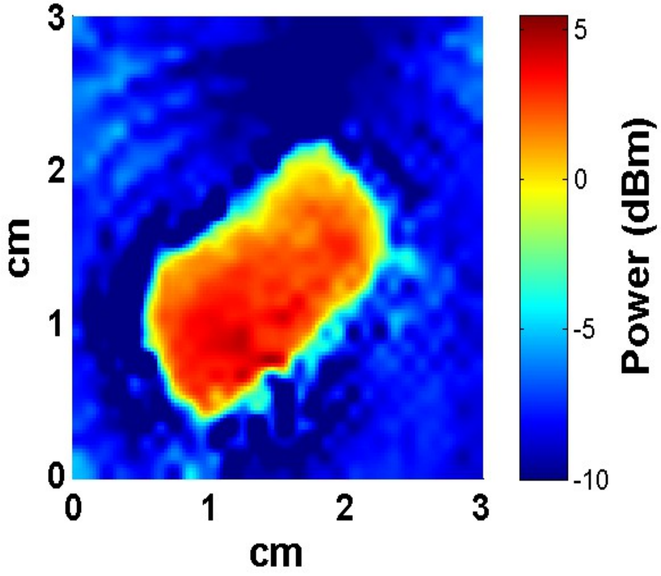
- Enhanced boundaries using Magnitude and Phase for edge detection
- Quantitative material distinction
- 500 µm motor step-size in X and Y axis


20% Gelatin and 3% Agar at 5.7 cm distance Magnitude


20% Gelatin and 3% Agar cube at 5.7 cm distance Phase

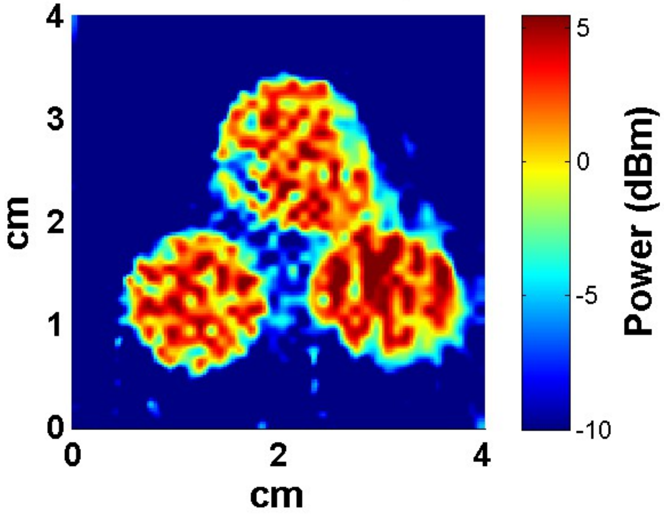
Phantom Registration Success

5% PVA Phantom, Vibro

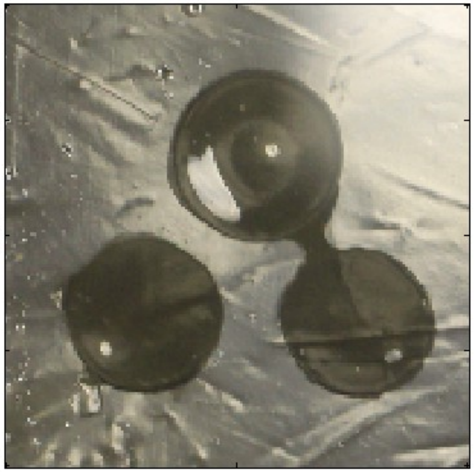


5% PVA Phantom, Visible

PVA Phantom, Vibro

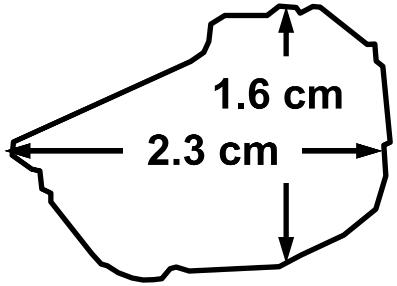


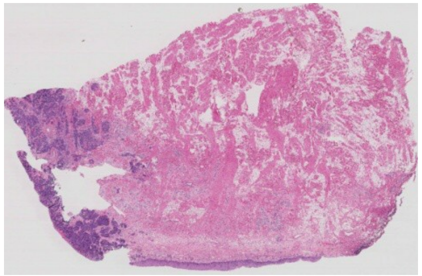
5% PVA Phantom, Visible

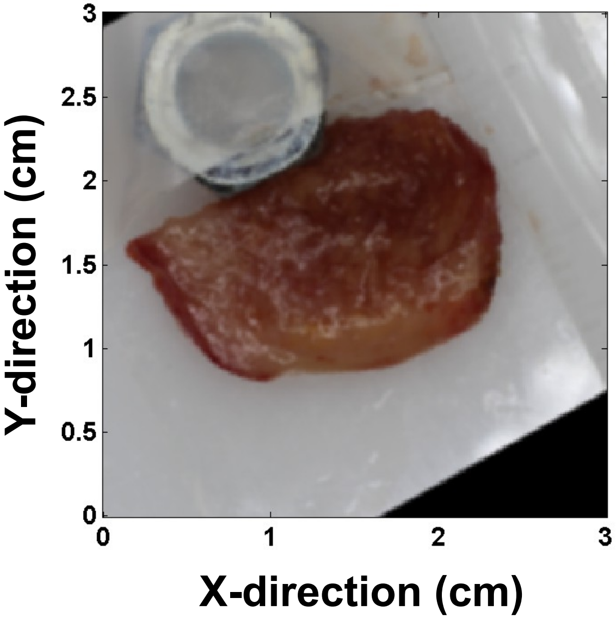


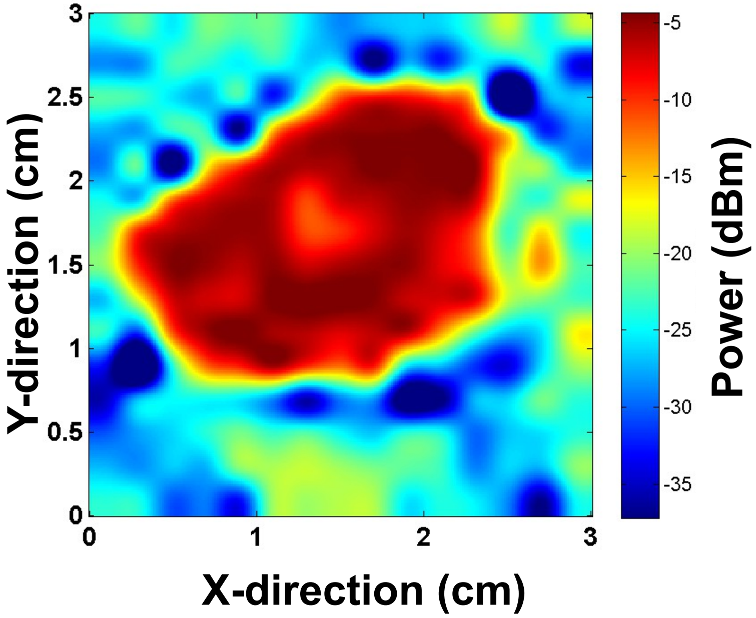
Clinical Tongue Squamous Cell Carcinoma (SCC)

Tongue Sample A2





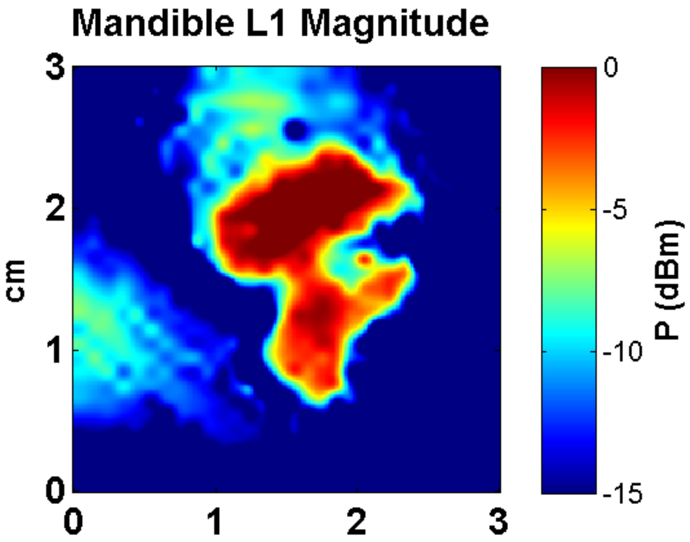




Ex-vivo Clinical SCC Evaluation

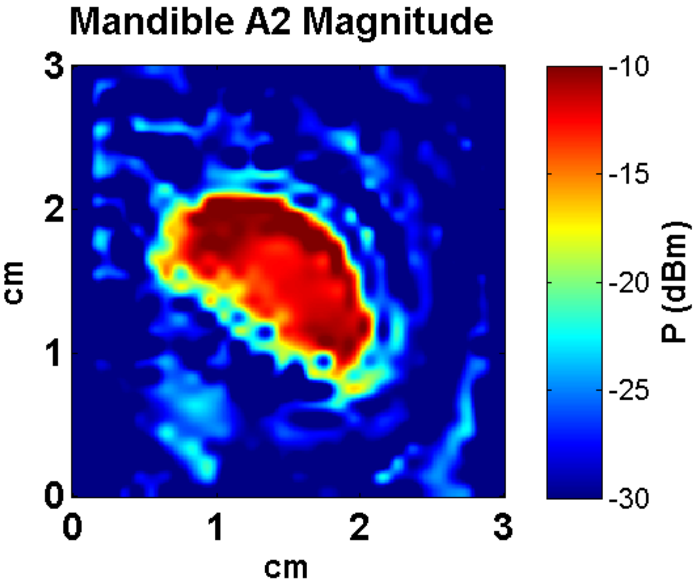
11/04/2013

Visible


Mandible L1 Magnitude

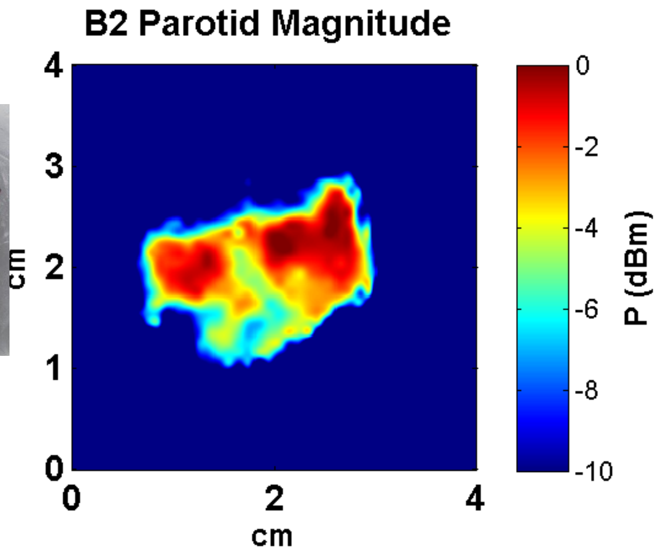
11/06/2013

Visible

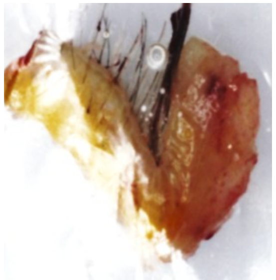

Mandible A2 Magnitude

10/07/2013

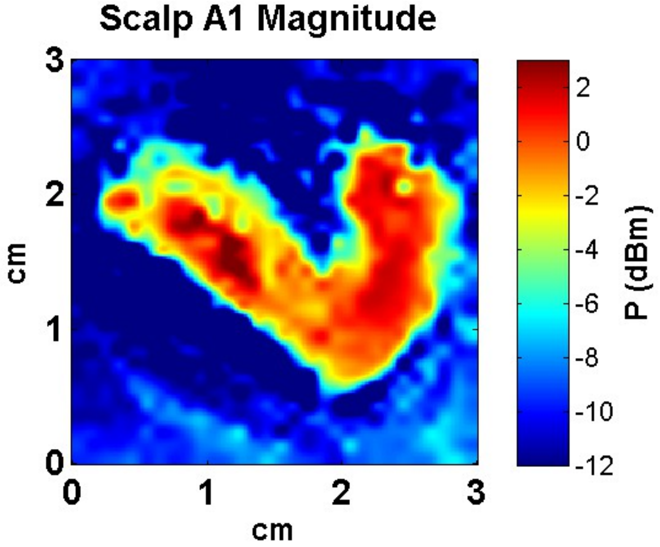
Visible


B2 Parotid Magnitude

11/18/2013



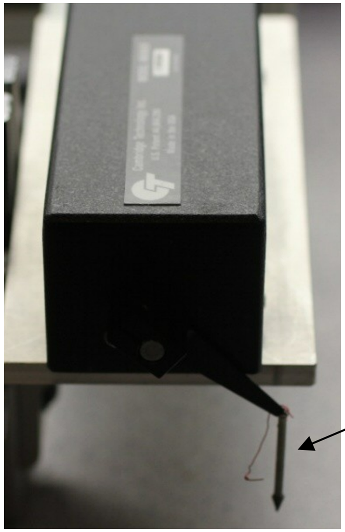
Visible

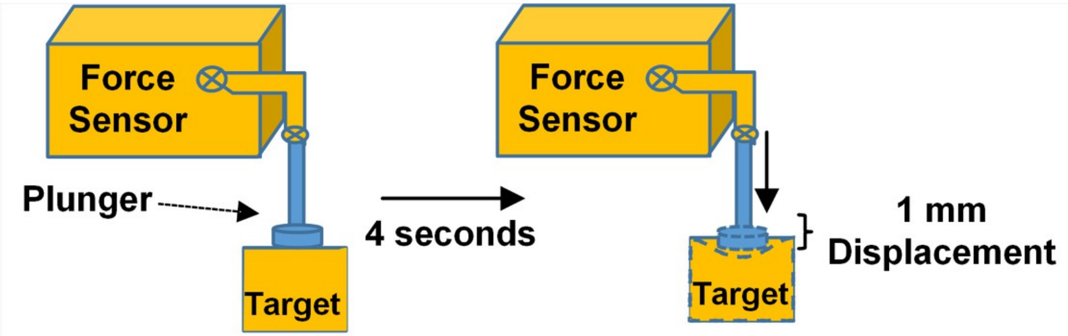

Scalp A1 Magnitude

Evaluation of the contrast mechanism

- **Validation of mechanical properties as contrast**
  - To deliver a true meaning of USVA images
  - Need for modality to provide direct elasticity measurements

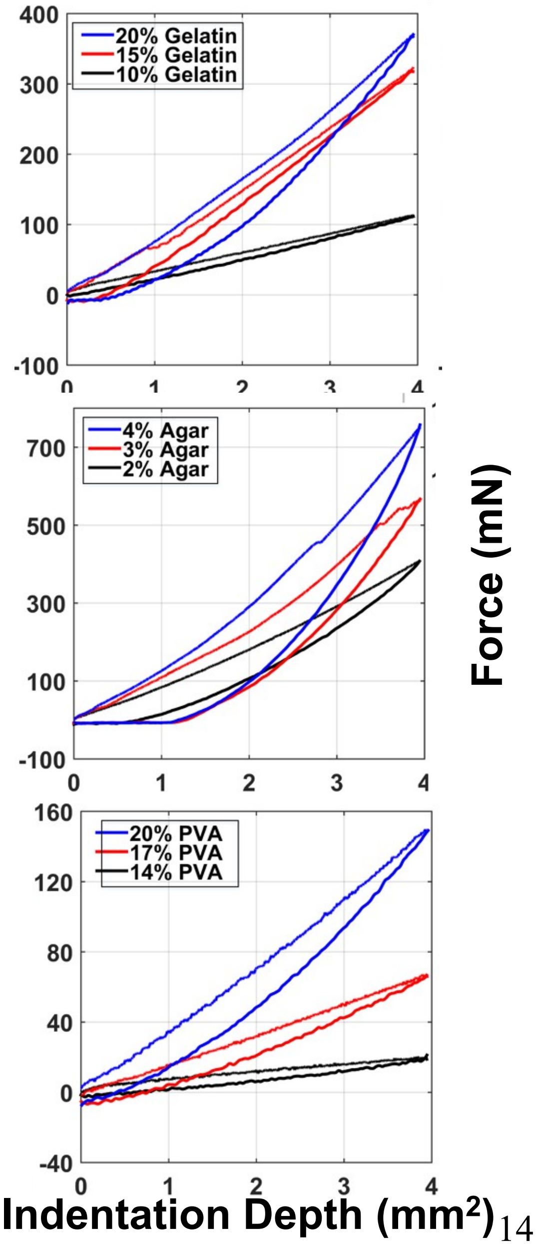
- **Muscle Lever System:**
  - Absolute elasticity measurements of a target
  - Impose a length change (strain) and measure force (stress)
  - Delivers true meaning of USVA images
    - Is it viscosity or elasticity or both?
  - Provides a model to explain USVA images



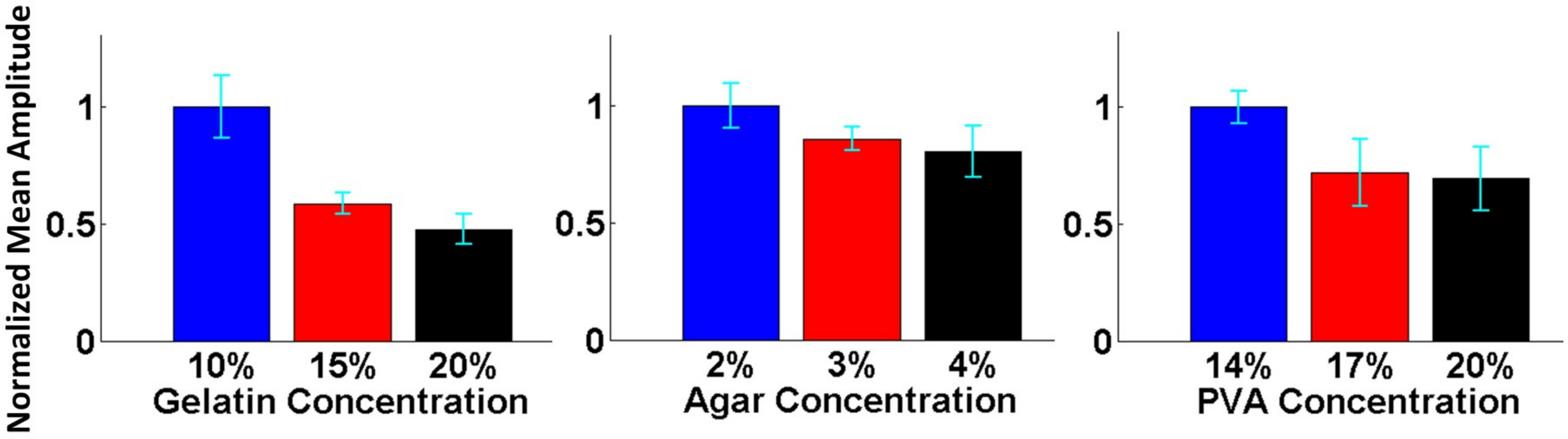


Experimental Setup

Phantom Characterization



Modulus Calculations



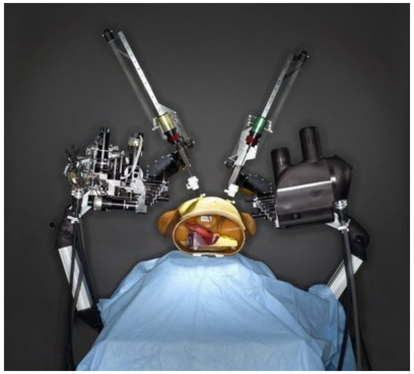
Agar		PVA		Gelatin	
	Modulus (kPa)		Modulus (kPa)		Modulus (kPa)
2%	102.90	14%	4.44	10%	27.00
3%	145.40	17%	17.23	15%	82.01
4%	187.90	20%	37.46	20%	92.39

- Inverse-correlation between modulus and intensity measurements

Recent/Future Advances

- Smaller, more portable systems
  - Cell-phone sized systems (GE V-scan)
- High resolution systems
- Multi-modality imaging or ablation systems
  - Fusion with pre-operative MR or CT data
  - Real-time guidance
  - Neurosurgery, prostate surgery, others
- Hybrid imaging/therapy systems
- 4D imaging gaining acceptance
- Robotically/remote-controlled ultrasound
  - Automated surgical tasks
  - Remote diagnostics and therapies
  - Battlefield trauma
  - Lower earth orbit
  - Long distance space travel

GE V-scan System



RAVEN robotic system

DARPA Trauma Pod

Thank You!

Additional Slides

Static Piezoelectricity I

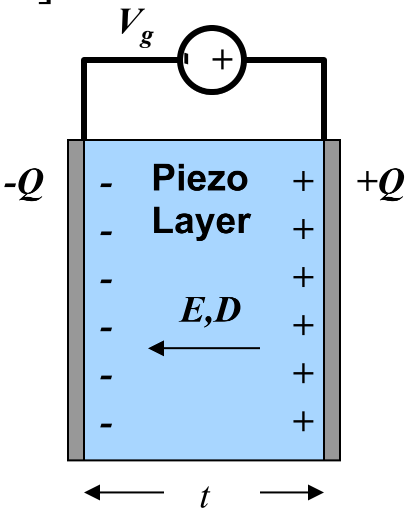
Piezoelectricity: “Pressing” electricity

Mechanical stress is coupled to electrical polarization

- *An applied mechanical stress induces a polarization* → $P \propto T$
- *An applied field induces a mechanical deformation* → $S \propto E$

It can be shown by thermodynamic arguments that the proportionality factor is the same in both cases:

$$\left.\begin{aligned} P &= d_m T \\ S &= d_m E \end{aligned}\right\} \quad d_m = \text{piezoelectric strain constant [C/N]}$$



Case 1: Applied field, no stress

$$P = \varepsilon_0 \chi E \qquad S = d_m E$$

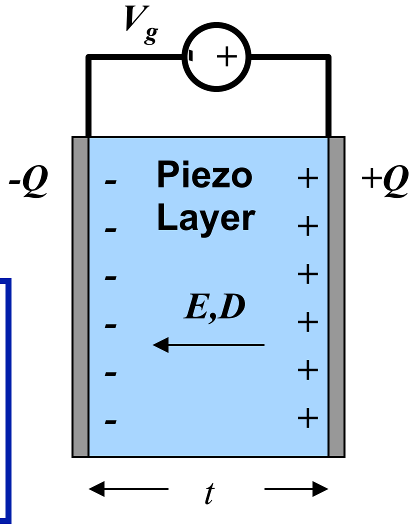
Case 2: Applied stress, no field

$$P = d_m T \qquad S = \frac{1}{c_m} T$$

Static Piezoelectricity II

Assuming a linear system (superposition)

$$D = \varepsilon E + d_m T \qquad S = d_m E + \frac{1}{c_m} T$$



Rewrite using E and S as independent variables

$$D = \left(\varepsilon - c_m d_m^2\right) E + c_m d_m S$$
$$T = -c_m d_m E + c_m S$$

or

$$\boxed{\begin{aligned} D &= \varepsilon^S E + e_m S \\ T &= -e_m E + c_m S \end{aligned}}$$

where $\varepsilon^S = \left(\varepsilon - c_m d_m^2\right) \qquad e_m = c_m d_m$

Piezoelectric stress constant

Note that ε is the permittivity under conditions of *no stress*.

If the crystal is clamped to prevent deformation, then $S = 0$ and the measured permittivity would be ε^S. Thus clamping the bar changes the capacitance.

"Clamped" capacitance

$$\boxed{C_c = \frac{\varepsilon^S A}{t}}$$

Static Piezoelectricity II

$$\begin{tikzcd}
\left(\frac{\mathrm{C}}{\mathrm{m}^2}\right) & D & \sigma \arrow[l, Rightarrow, "d"'] \arrow[d, Rightarrow, "s_E"] & \left(\frac{\mathrm{N}}{\mathrm{m}^2}\right) \\
\left(\frac{\mathrm{V}}{\mathrm{m}}\right) & E \arrow[u, Rightarrow, "\varepsilon_\sigma"] \arrow[r, Rightarrow, "d"'] & \delta & \left(\frac{\mathrm{m}}{\mathrm{m}}\right)
\end{tikzcd}$$

$$\begin{tikzcd}
(\mathrm{C}) & Q & f \arrow[l, Rightarrow, "d"'] \arrow[d, Rightarrow, "k_v^{-1}"] & (\mathrm{N}) \\
(\mathrm{V}) & v \arrow[u, Rightarrow, "C_f"] \arrow[r, Rightarrow, "d"'] & u & (\mathrm{m})
\end{tikzcd}$$

Piezoelectric Actuator

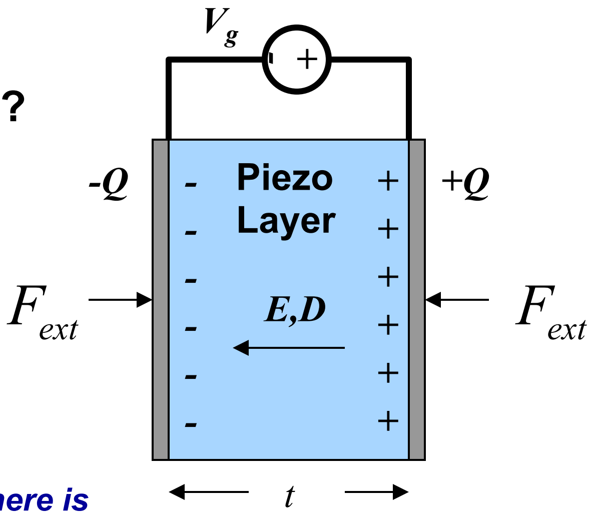
Apply a voltage to generate mechanical motion

Simple problem: how much deformation for a given voltage?

No external force, free boundaries, steady state:

$$\frac{\Delta t}{t} = S = d_m E = d_m \frac{V}{t}$$

$$\boxed{\Delta t = d_m V}$$



In steady-state, stress is a constant throughout the material, and since there is no externally applied force, that means T=0 throughout

In a real mechanical actuator, the material would meet some resistance as it tries to elongate. What external force would have to be applied to prevent the material from deforming? (ie. what is the maximum force the actuator can exert?)

In this case *S*=0 (clamping condition)

$$T = F_{ext} / A = -c_m d_m E$$

$$\boxed{F_{ext} = \kappa V \qquad \kappa = \frac{c_m d_m A}{t}}$$

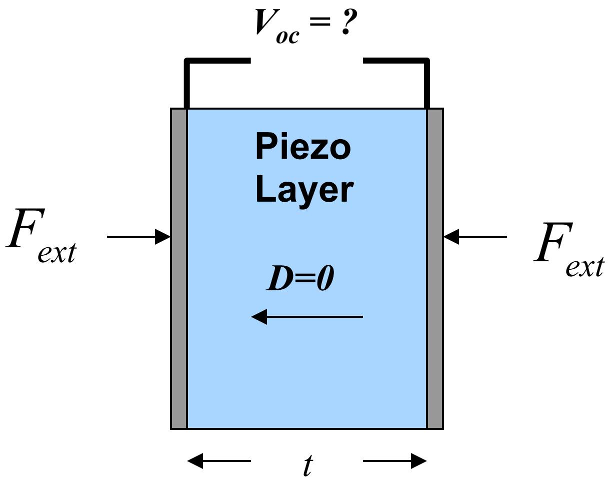
Electromechanical coupling coefficient

Piezoelectric Transducer

UCLA

Converts a mechanical force to a measurable voltage (eg. microphone)

What is the open-circuit voltage for a given force?



Assume a constant stress such that: $T = F_{ext} / A$

If there is no free-charge in the material,

$$\nabla \cdot \bar{D} = 0 \xrightarrow{D \text{ must be constant}} D = Q / A$$

However, under open-circuit conditions, there can be no accumulation of free charge on the electrodes (no current can flow on the wires to supply the charges). Therefore

$$\bar{D} = 0$$

Under open-circuit conditions

$$\Rightarrow \quad 0 = \varepsilon E + d_m T \qquad S = d_m E + \frac{1}{c_m} T$$

Solving for E and T

$$E = -\frac{d_m T}{\varepsilon} \qquad T = \left(c_m + \frac{e_m^2}{\varepsilon^S} \right) S$$

"Free" capacitance

$$V_{oc} = -Et = \frac{d_m}{C_0} F_{ext} \qquad C_0 = \frac{\varepsilon A}{t}$$

Piezoacoustic Waves, Briefly

For open-circuit conditions in the static case we found

$$T = \left(c_m + \frac{e_m^2}{\varepsilon^S} \right) S = c_m' S \qquad c_m' = c_m + \frac{e_m^2}{\varepsilon^S}$$

The stiffness constant is modified by the piezoelectric coupling (the material is "piezoelectrically stiffened")

To model waves in piezoelectric materials, we just insert the new stiffness constant into the transmission-line model for acoustic wave propagation. Thus

$$v_p' = \sqrt{\frac{c_m'}{\rho_m}} = \sqrt{\frac{c_m}{\rho_m}\left(1 + \frac{e_m^2}{c_m \varepsilon^S}\right)} = v_p \sqrt{1 + K^2}$$

Piezoelectric coupling constant

$$K^2 = \frac{e_m^2}{c_m \varepsilon^S} = \frac{c_m d_m^2}{\varepsilon^S}$$

For most materials, $K^2 < 0.3$

Similarly

$$Z_0' = A\sqrt{\rho_m c_m'} = A\rho_m v_p' = Z_0 \sqrt{1 + K^2} \qquad \beta' = \frac{\omega}{v_p'} = \frac{\beta}{\sqrt{1 + K^2}}$$

Coupling Constants

There are two coupling constants that arise in piezo devices:

Piezoelectric coupling constant

$$K^2 = \frac{e_m^2}{c_m \varepsilon^S} = \frac{c_m d_m^2}{\varepsilon^S}$$

Electromechanical coupling constant

$$k_t^2 = \frac{e_m^2}{c_m' \varepsilon^S}$$

The electromechanical coupling constant uses the modified stiffness constant discussed in the previous slide:

$$k_t^2 = \frac{e_m^2}{\left(c_m + \frac{e_m^2}{\varepsilon^S}\right)\varepsilon^S} = \frac{K^2}{K^2 + 1}$$

Electromechanical Coupling Constant

The ability of a material to convert one form of energy into another

Electromechanical coupling constant

$$k_t^2 = \frac{e_m^2}{c_m' \varepsilon^S}$$

$$k_t^2 = \frac{\text{Stored Mechanical Energy}}{\text{Total Stored Energy}}$$

The total stored energy includes mechanical and electric energy.

Note: This quantity should not be confused with the efficiency of the transducer.

Waves in Piezoelectric Materials I

Here is a closer look at the effect of the electromechanical coupling on the acoustic wave propagation. We have

$$D = \varepsilon^S E + e_m S \qquad (1)$$

$$T = -e_m E + c_m S \qquad (2)$$

$$\frac{\partial T}{\partial z} = \rho_m \frac{\partial^2 u}{\partial t^2} \qquad (3)$$

$$\frac{\partial S}{\partial t} = \frac{\partial v}{\partial z} \qquad (4)$$

Note, T is no longer a constant inside the material, as in the static case

Lets assume 1D wave propagation in the z-direction, and consider source-free fields ($D = 0$, or open-circuit conditions)

From (2) and (4) we find

$$\frac{\partial T}{\partial t} = -e_m \frac{\partial E}{\partial t} + c_m \frac{\partial v}{\partial z}$$

Differentiating Newton's law (3) with respect to z (converts u to v) and substituting the above gives the *Christoffel equation:*

$$\longrightarrow \boxed{\frac{\partial^2 v}{\partial z^2} + d_m \frac{\partial^2}{\partial z^2}\left(\frac{\partial \Phi}{\partial t}\right) = \frac{\rho_m}{c_m} \frac{\partial^2 v}{\partial t^2}}$$

This is a wave equation with a piezoelectric coupling term

….continued on next page

Waves in Piezoelectric Materials II

Some comments about the internal electric fields: note that we've made a quasi-static approximation (no magnetic field):

$$\nabla \times \bar{E} = 0 \longrightarrow \bar{E} = -\nabla \Phi$$

The acoustic wave generates an electric field which also "travels" at the acoustic velocity. Furthermore it can be longitudinally polarized, and there is a negligible magnetic field. Thus it is not a conventional EM wave, we might call it an "electrostatic wave".

Differentiating (1) with time we also find

$$\varepsilon^S \frac{\partial}{\partial t}\left(\frac{\partial \Phi}{\partial z}\right) = e_m \frac{\partial v}{\partial z}$$

Assuming plane waves of the form $e^{-j\beta' z}$ **gives**

$$\beta'^2 \left(c_m v + j\omega e_m \Phi\right) = \omega^2 \rho_m v \qquad \omega \varepsilon^S \Phi = -j e_m v$$

Eliminating variables we find the propagation constant

$$\beta'^2 \left(c_m + \frac{e_m^2}{\varepsilon^S}\right) = \omega^2 \rho_m \qquad \text{or} \qquad \beta' = \frac{\omega}{v_p'}$$

Same form as a non-piezo material, but using the "stiffened" constant c'