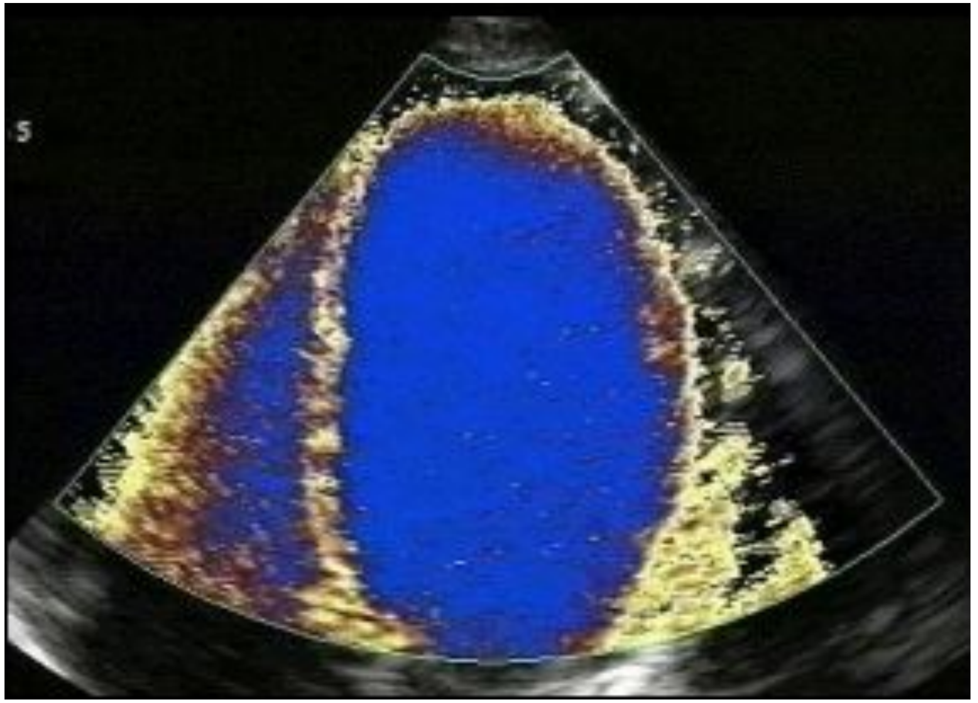
Medical Ultrasound

Contents:

- Physics of sound waves
- Ultrasound devices
- Signal processing
- Medical applications



Recommended book:
«Introduction to Medical Imaging»
Nadine Barrie Smith, Andrew Webb
Cambridge University Press, 2014

Based on lecture material from Prof. Urs Moser

Medical Ultrasound Imaging – Physicist's perspective

Ultrasound is an acoustic wave;
i.e., mechanical perturbation traveling in medium
(it is not an EM wave)

Acoustic frequency range:

- Infrasound: <20 Hz
- Audible sound: 20 Hz – 20 kHz
- Ultrasound: >20 kHz

(Medical ultrasound: 1 to 20 MHz)

Probe → Matter interaction → Detection → Image production

Probe	Matter interaction	Detection	Image production
γ-photons	**Reflection**	**Piezo-electric**	**Back projection**
X-ray photons	**Refraction**	Scintillator	Filtered back projection
RF waves	**Scattering**	Photo diode	Inverse transform
Acoustic waves	**Absorption**	RF Antennae	Constrained transforms
Light	**Attenuation**	CCD chips	…
	Spin transitions	…	
	…		

Applications

Frequency range: 1 to 50 MHz

Wavelength in tissue: 1 mm down to 0.03 mm

Intensity: temporal mean 100mW/cm^2 (eye 20mW/cm^2), pulsed: 10 W/cm^2

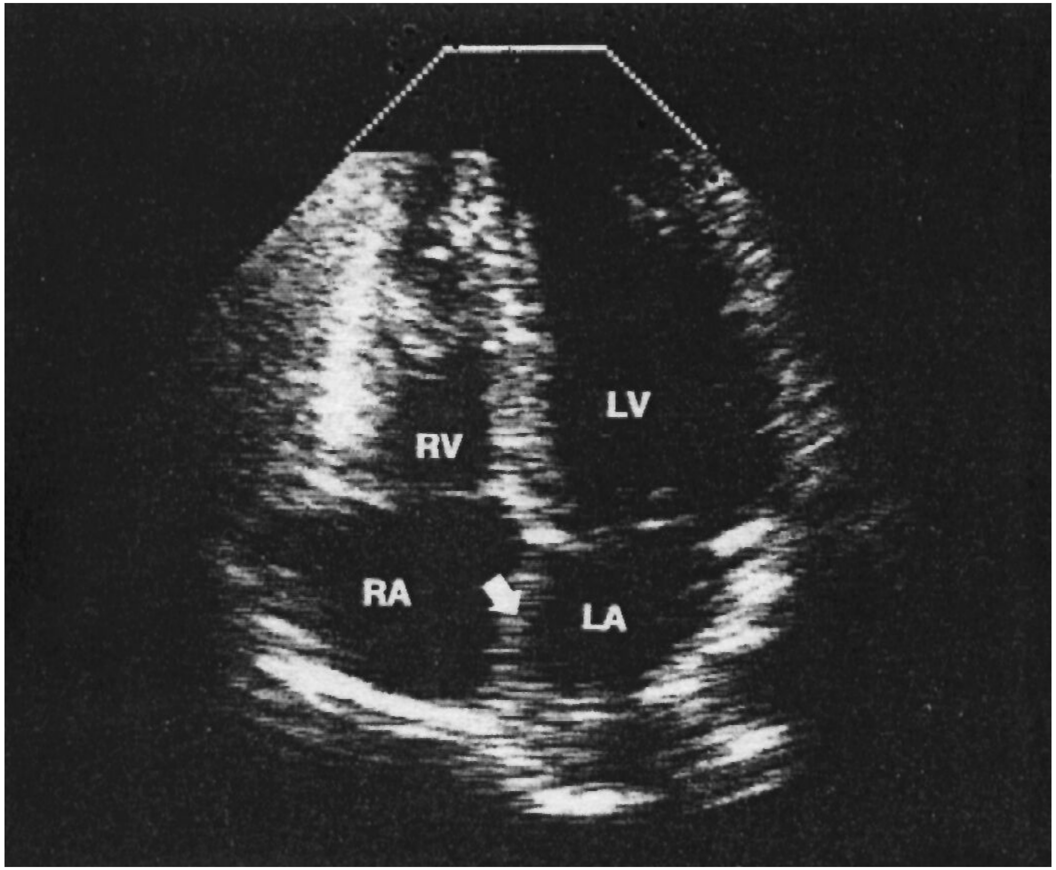
Diagnostic:

- Echography (imaging tissue cross-sectional slices, > 30 frames/second)
- Doppler (measuring blood flow velocity)
- Bone densitometry

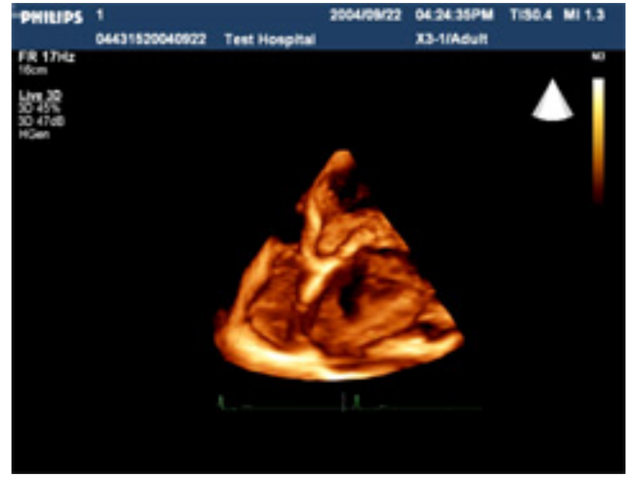
Therapeutic:

- Lithotripter (disintegration of kidney stones)
- Thermal therapy (ligaments, tendons etc.)
- Ultrasound surgery (e.g. in the deep brain)

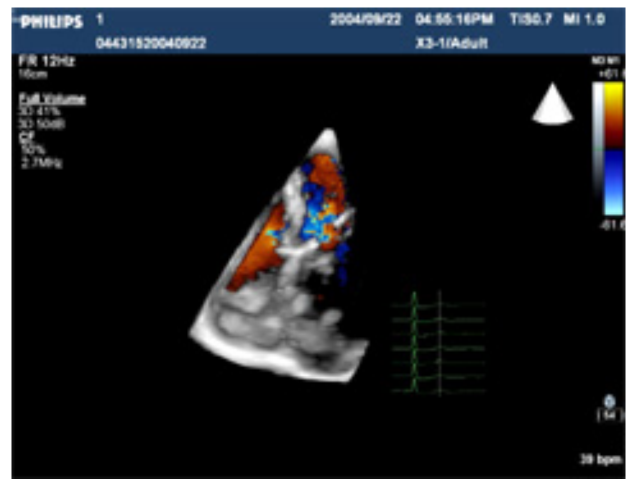
Diagnostic Ultrasound: Cardiology



2D image



3D image



3D image + Doppler

Diagnostic Ultrasound : Vessels

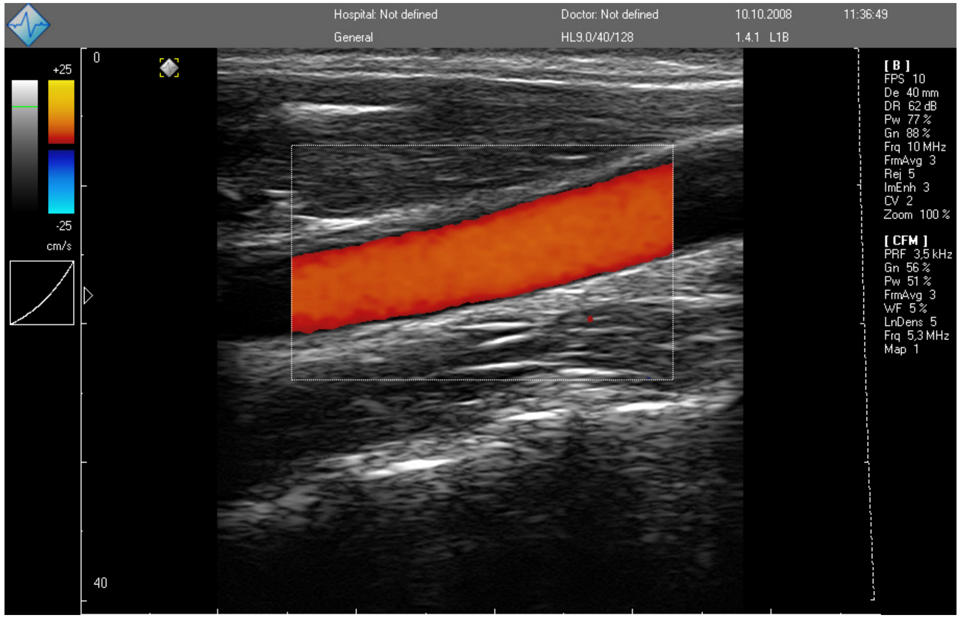
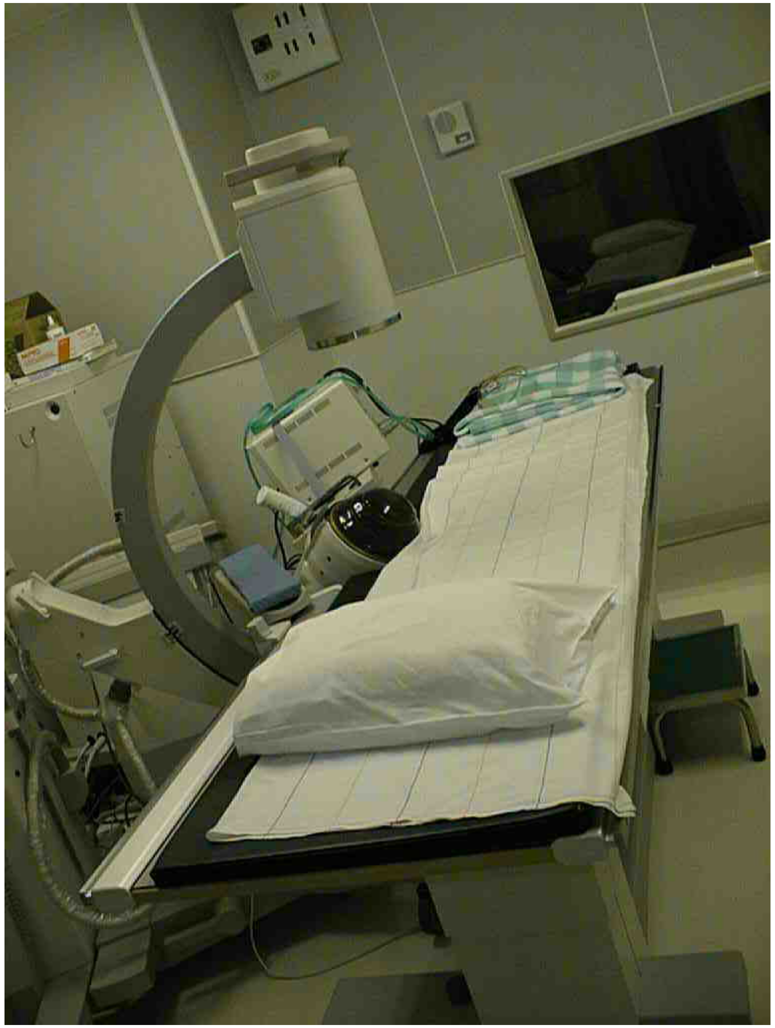


Image of vessel with blood flow

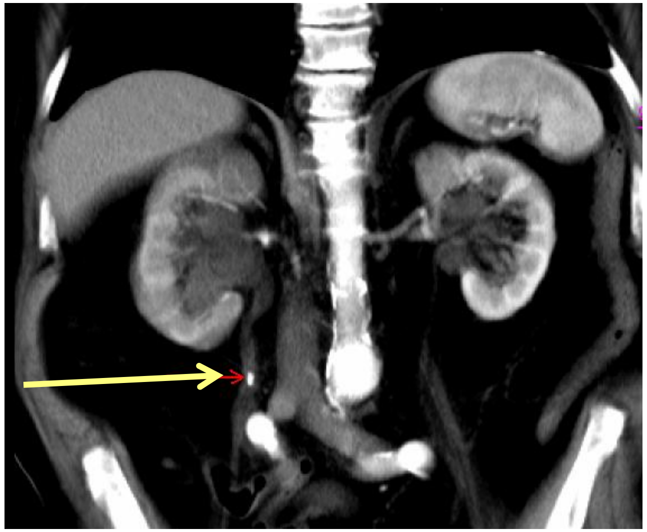
Vessel bifurcation

Therapeutic Ultrasound: Lithotripter



Lithotripsy device

Kidney stone
blocking ureter

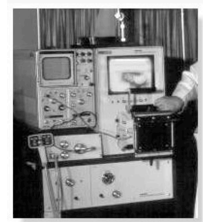


Kidney stone

History and Evolution

Year	Event
1822	Daniel Coladon: measures speed of sound in lake Geneva
1877	Lord Rayleigh: «The Theory of Sound»
1880	Curie brothers: discover piezoelectric effect
1914	Langevin: First ultrasound generator using piezoelectric effect (first practical use in WWI for detecting submarines)
1937	Dussik: first clinical use for locating brain tumors
1954	Edler, Hertz: first use in cardiology (echocardiography)
1965	First real-time grayscale images (Siemens)
1968	Electronic beam-steering with phased-array technology
>1970	Ultrasound “boom” with progress in microelectronics … still growing, with new applications continually discovered

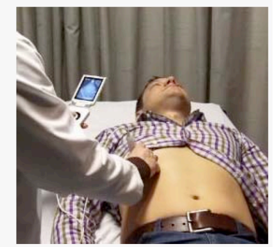

Portable / low-power systems



Scan converter incoporated in early American model. Image is being displayed on the TV monitor on the right Note the black-on-white display format

about 1970

2005



2010: pocket device

Clinical Applications

- Obstetrics/Gynaecology
- Cardiology (echocardiography, real-time imaging)
- Internal Medicine: liver, gall bladder, kidney etc.
- Musculoskeletal system: muscles, tendons, ligaments
- Breast imaging
- Urology: prostate
- Blood flow measurement (US Doppler)

Advantages and Drawbacks

Advantages:

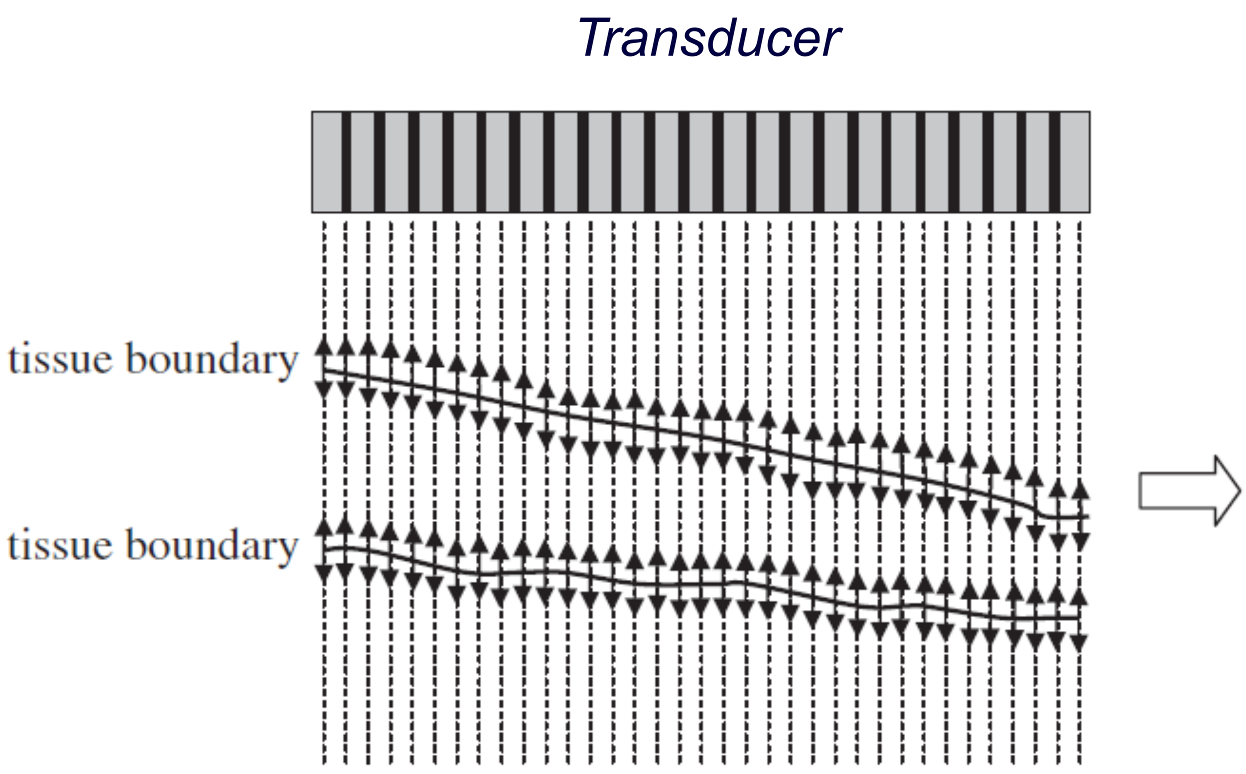
- no ionizing radiation (!!)
- minimal safety requirements, to operator & patient (!)
- real-time capability
- small, mobile, bed-side
- comparatively low costs

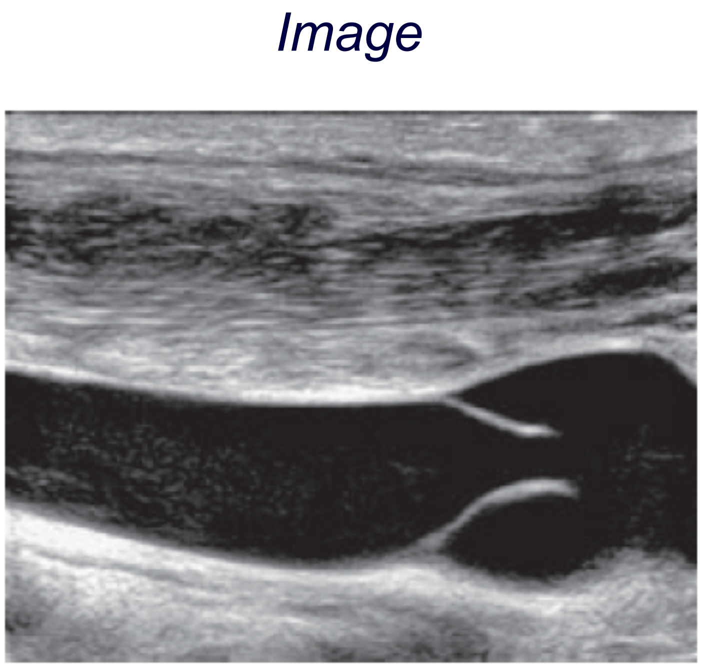
Drawbacks:

- moderate image quality, poor contrast (low SNR)
- need for “acoustic window” (gas & bone impede ultrasound)
- needs operator training and experience
  (to navigate the probe & to interpret images)

Basic Principle of Ultrasound Imaging

Chapter 5.18

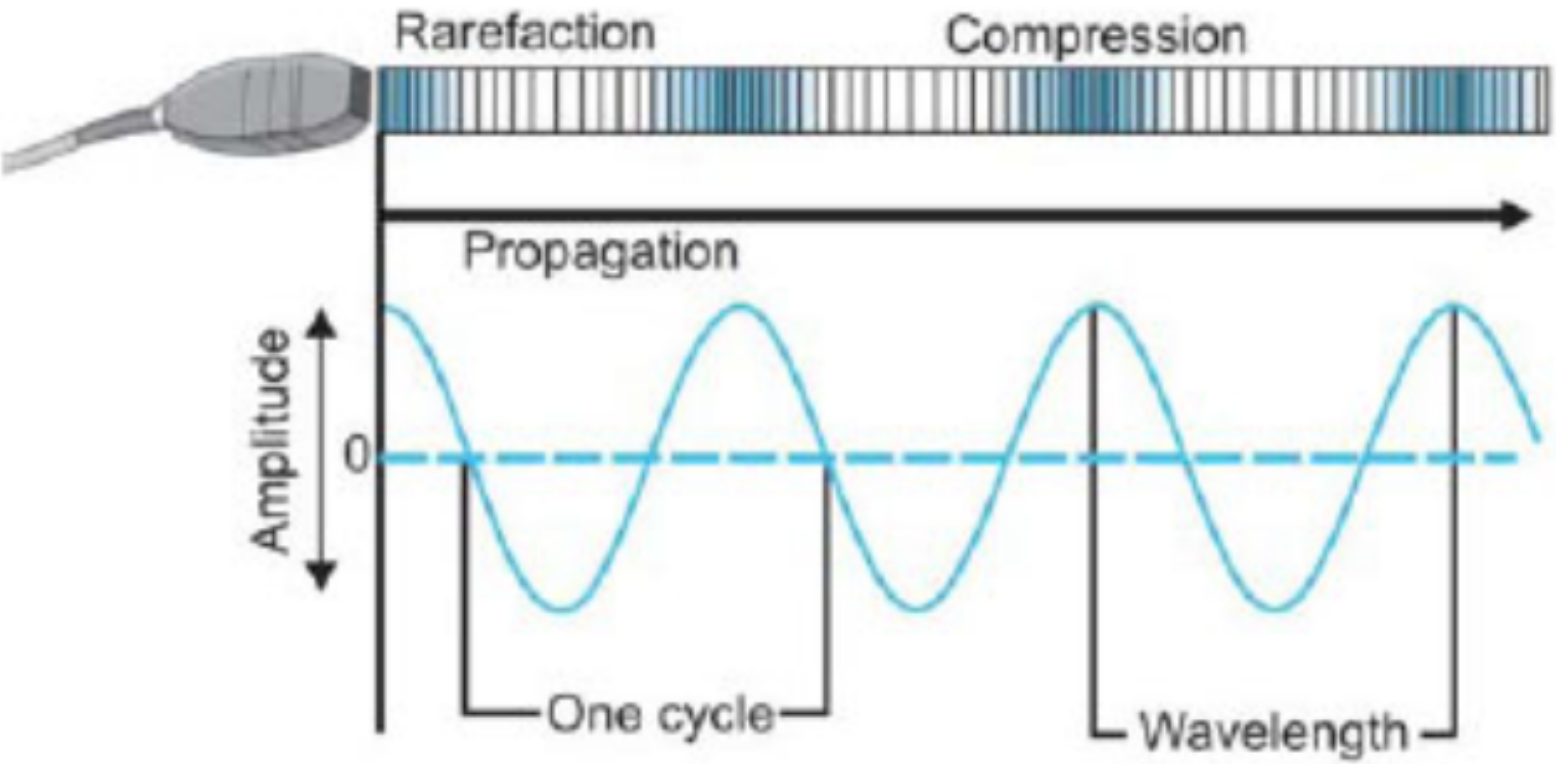




from Smith/Webb

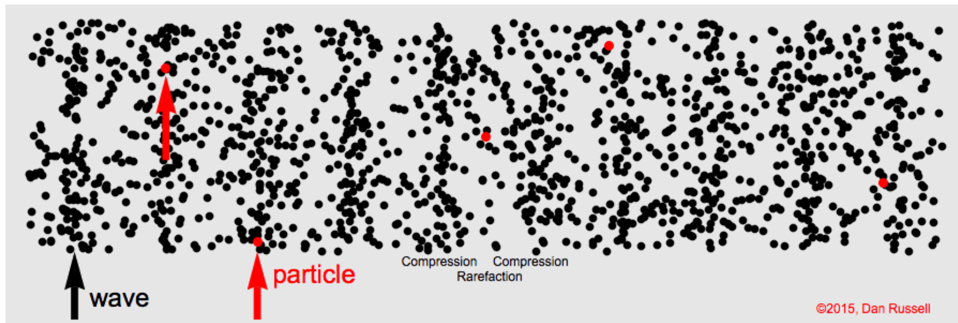
Ultrasonic Waves

Chapter 4.2

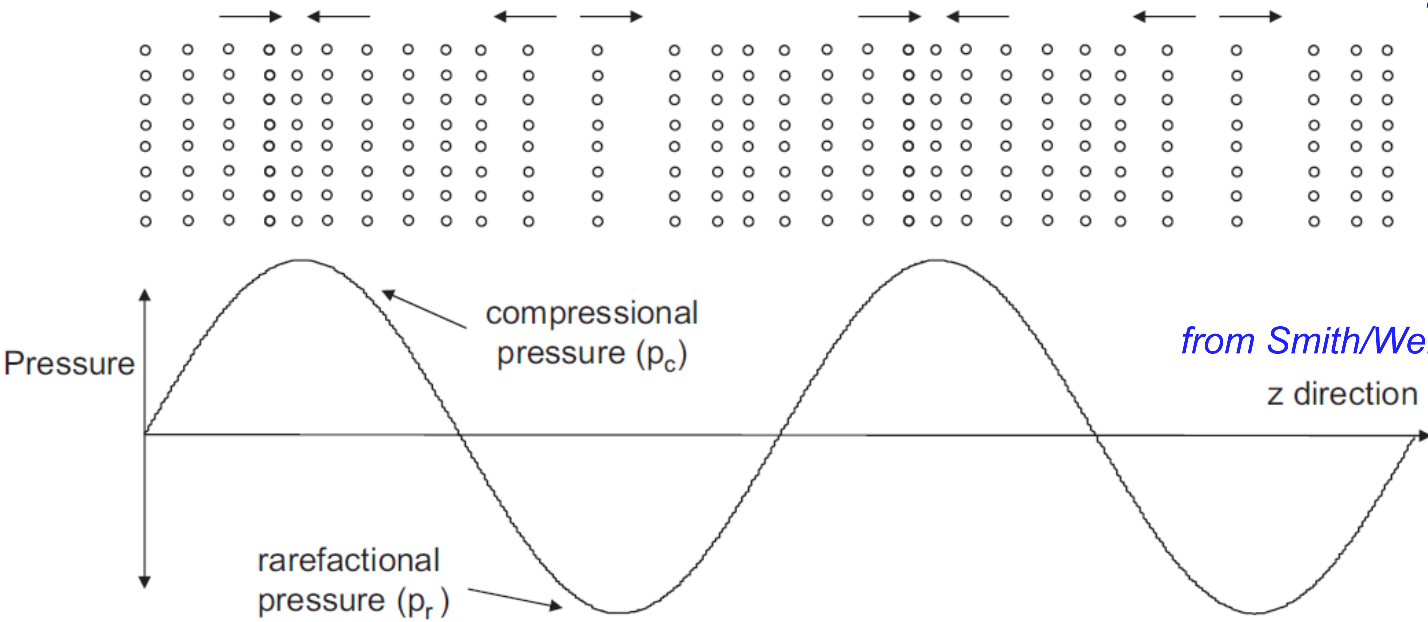


Ultrasound Propagation in Tissue

Longitudinal (pressure) wave $$\nabla^2 p - \frac{1}{c^2}\frac{\partial^2 p}{\partial t^2} = 0$$



particle displacement



from Smith/Webb

Key Quantities and Relationships

wave speed $$c = \frac{1}{\sqrt{\kappa \rho}}$$ κ = compressibility, ρ = mass density

(alternative notation $c = \sqrt{\frac{K}{\rho}}$, with stiffness K)

pressure $$p = \rho \; c \; u_z$$ u_z = particle velocity

wave impedance $$Z = \frac{p}{u_z} = \rho \; c = \sqrt{\frac{\rho}{\kappa}}$$

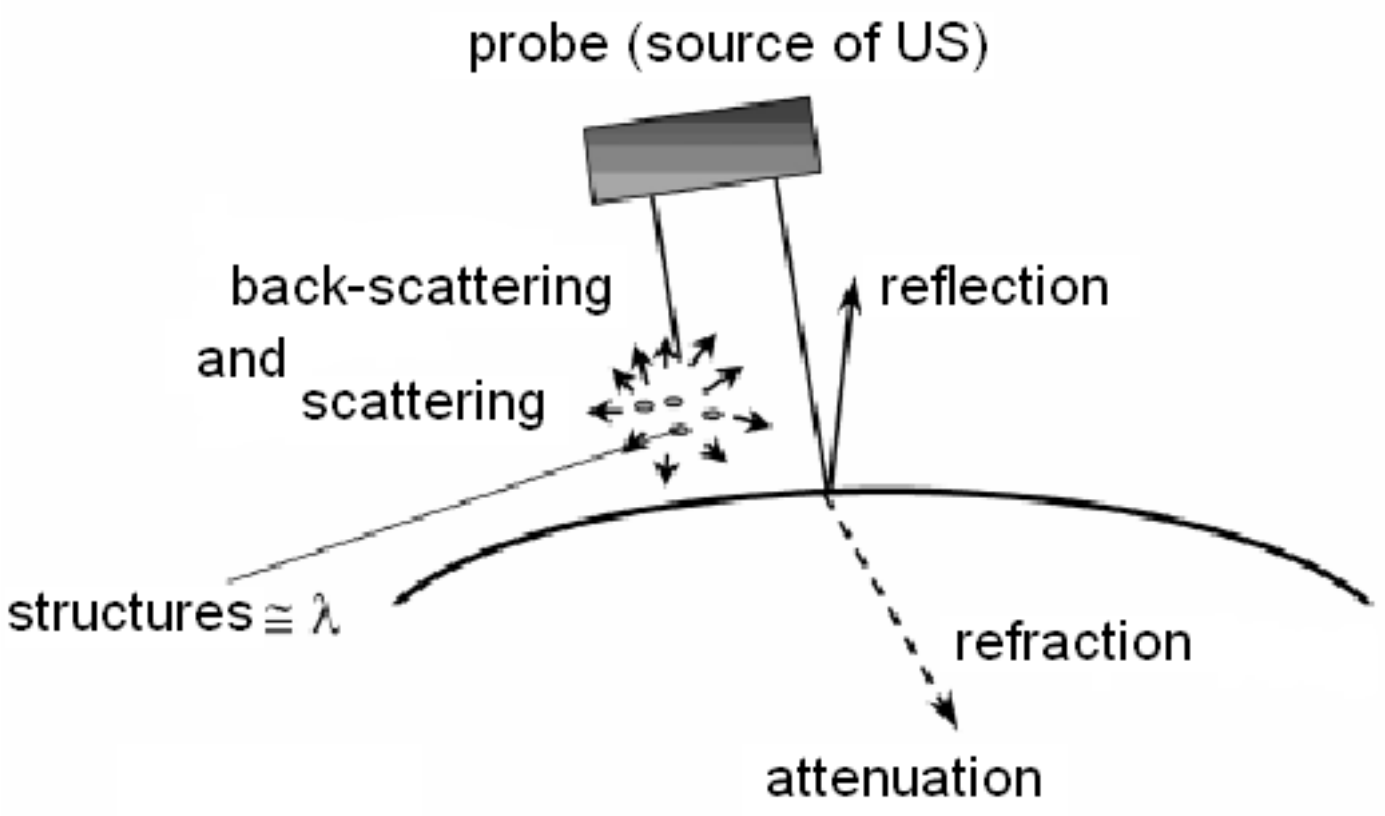
wave intensity $$I = \frac{1}{2} \, p \; u_z$$

Acoustic Properties of Biological Tissue

as well as materials used for the transducer construction

Substance	Z [kg/m^{2}s]	ρ [g/cm^3]	c [m/s]
Water	$1.48 \cdot 10^6$	1.0	1480
Blood	$1.65 \cdot 10^6$	1.06	1560
Brain	$1.58 \cdot 10^6$	1.03	1530
Fat	$1.36 \cdot 10^6$	0.92	1476
Muscle	$1.66 \cdot 10^6$	1.06	1568
Bone	$6 \cdot 10^6$	1.3 -1.8	2800-4100
Air	400 (!)	0.0012	330
Epoxy	$3 \cdot 10^6$	1.2	2500
Plexiglass	$3.2 \cdot 10^6$	1.18	2730
PZT Ceramic	$29 \cdot 10^6$	7.65	3790
Steel	$39 \cdot 10^6$	7.7	5000

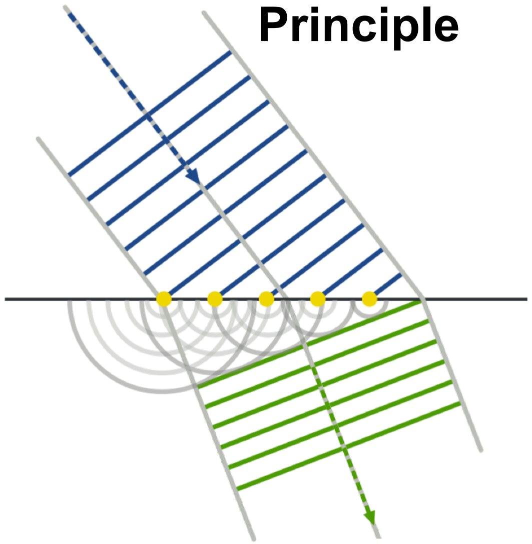
Acoustic interactions with tissue

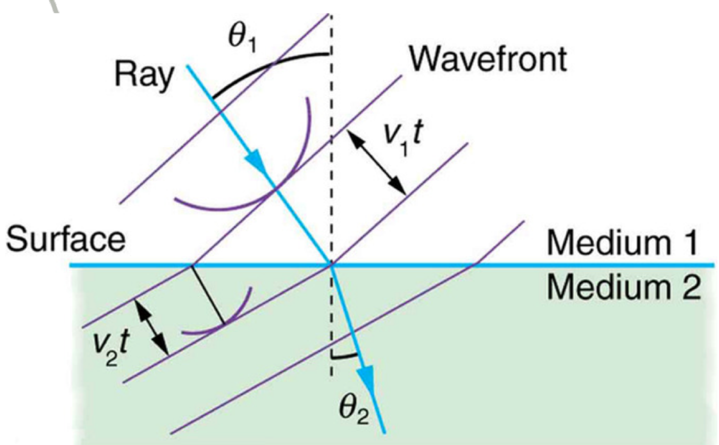


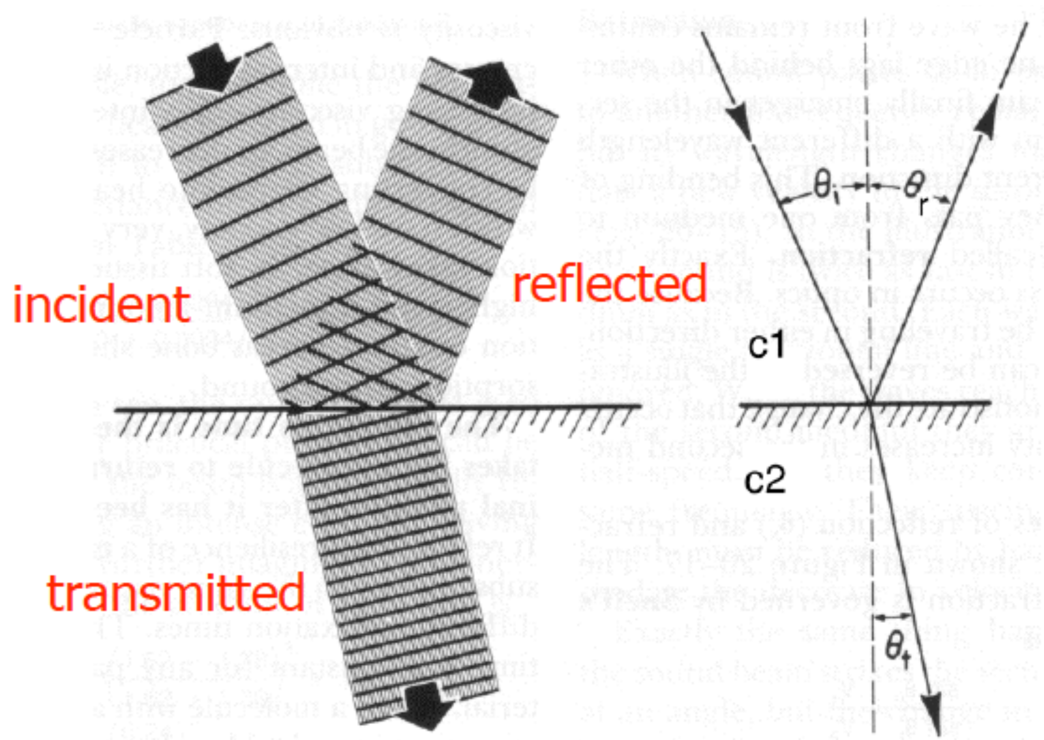
Refraction and the Snell's Law

<u>**Refraction**: Bending of waves from one medium to another</u>

Huygens (– Fresnel) Principle

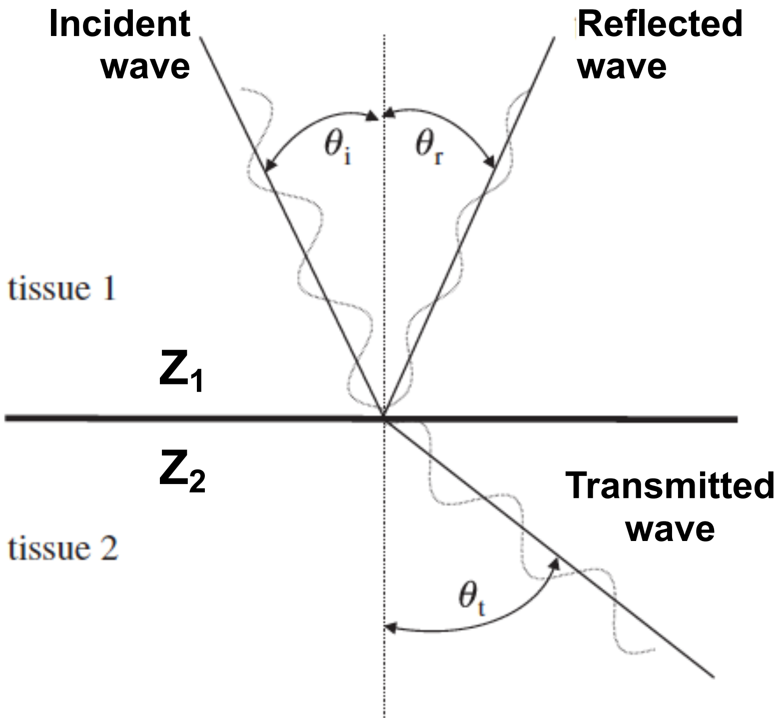






<u>Snell's law:</u> $$\frac{\sin\theta_i}{c_1} = \frac{\sin\theta_t}{c_2}$$

Reflected amount depends on acoustic impedance diff.



Pressure reflection coefficient:

$$\frac{p_r}{p_i} = \left(\frac{Z_2 \cos\theta_1 - Z_1 \cos\theta_2}{Z_2 \cos\theta_1 + Z_1 \cos\theta_2} \right) \stackrel{\text{def}}{=} R_p$$

For oblique incidence:

$$\text{For } \theta_1 = 0 \quad \Rightarrow \quad R_p = \left(\frac{Z_2 - Z_1}{Z_2 + Z_1} \right)$$

Intensity $\propto$ pressure-squared: $\frac{I_r}{I_i} = R_I = R_p^2$

Intensity transmission coefficient: $T_I = 1 - R_I$

- Larger the impedance |diff|, more the reflected intensity
- Larger the incidence angle, more the reflected intensity (less the transmission)

Slide added for exercise

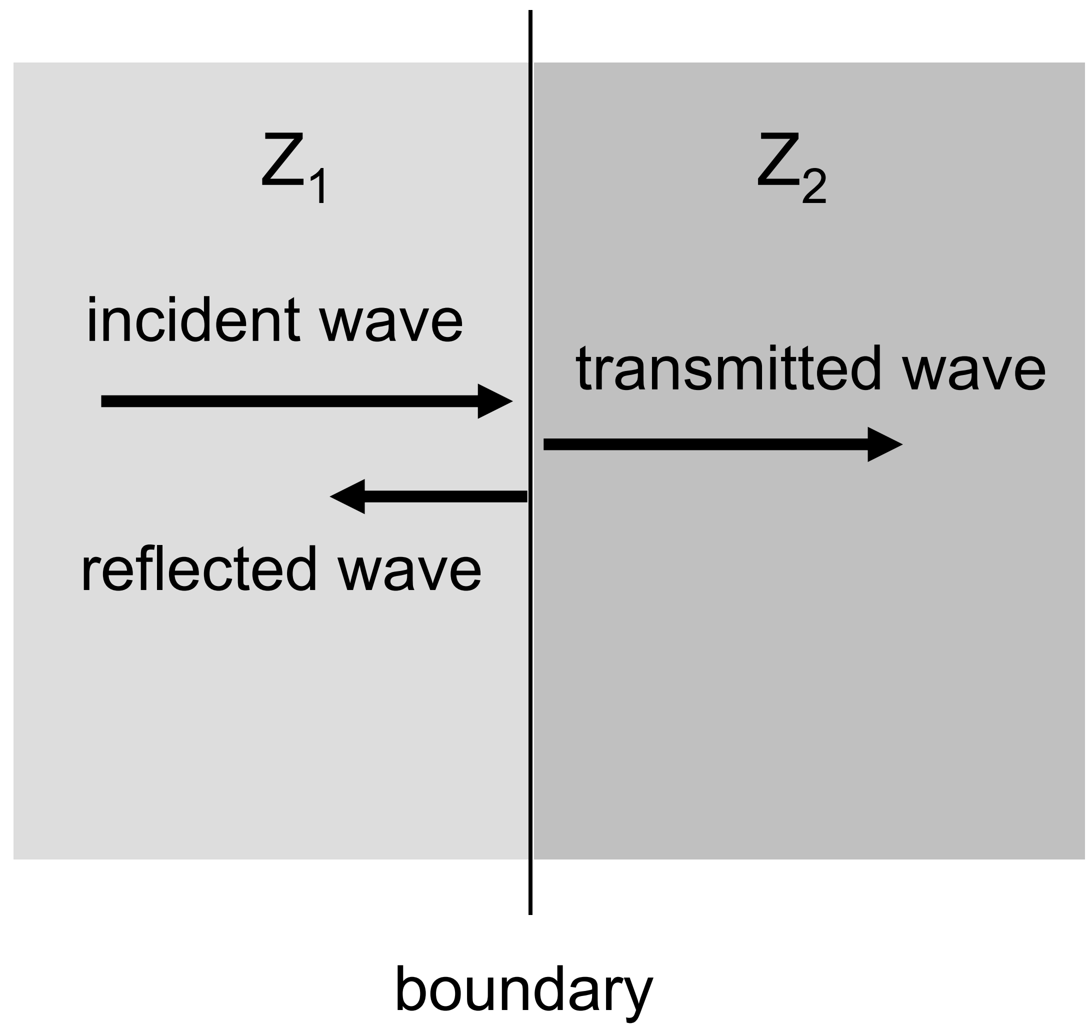
Perpendicular Incidence

Pressure reflection coefficient *r*

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

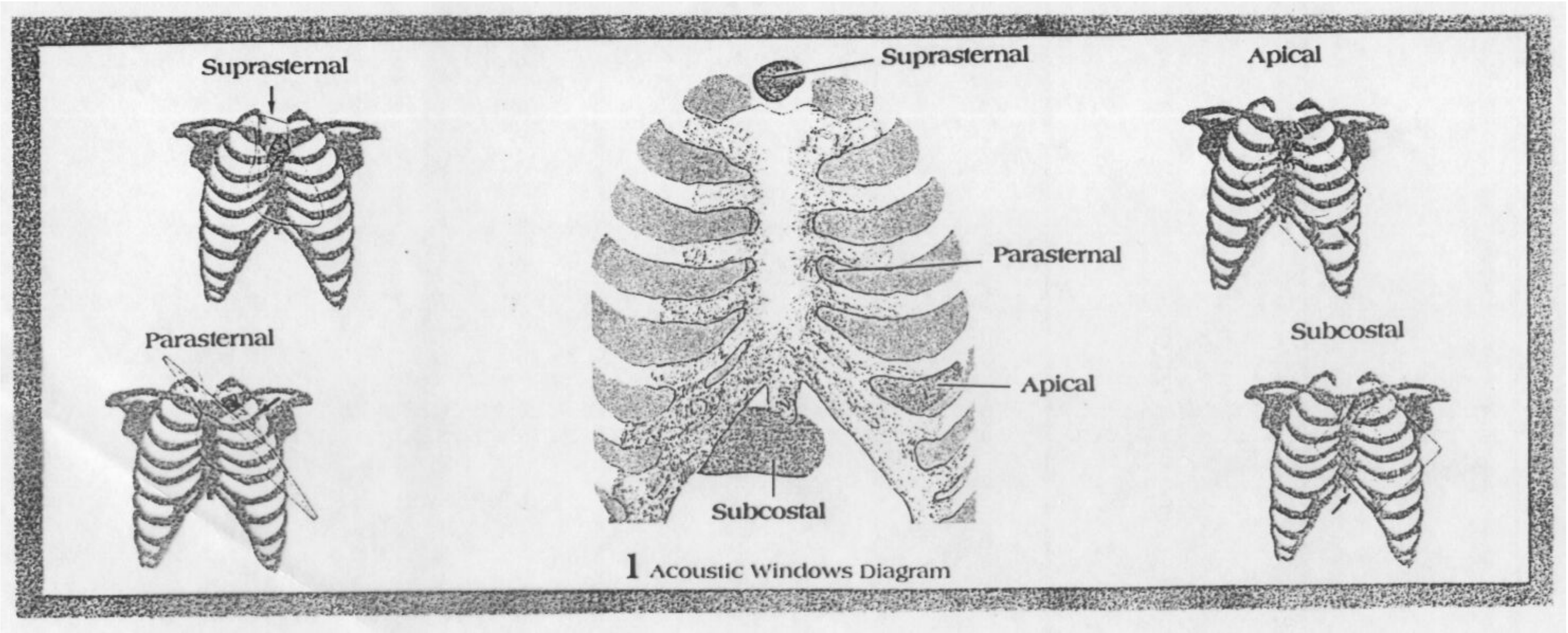
Pressure transmission coefficient *t*

$$t = \frac{2Z_2}{Z_2 + Z_1}$$



Acoustic Windows

Ultrasound is strongly absorbed in bones and is reflected nearly to 100% by tissue/air or air/tissue interfaces -> need for “acoustic window”

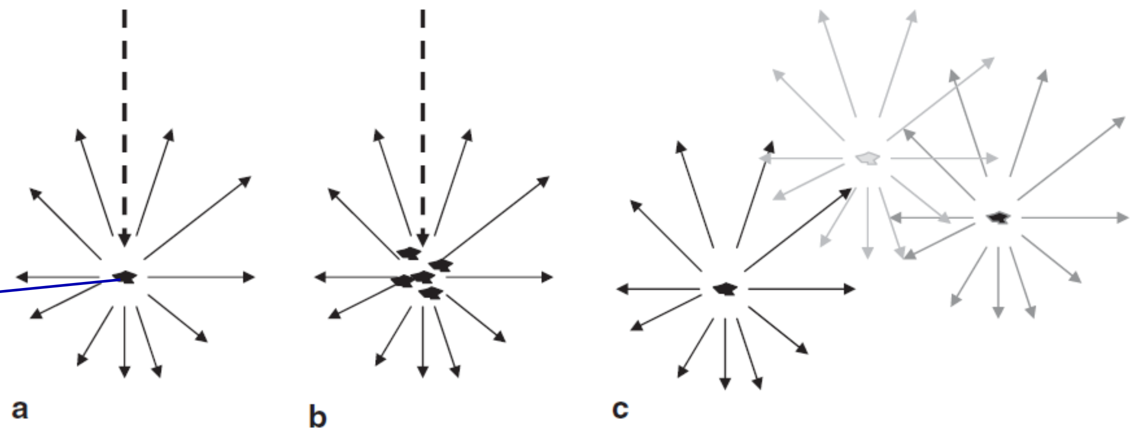


Acoustic windows for cardiac imaging

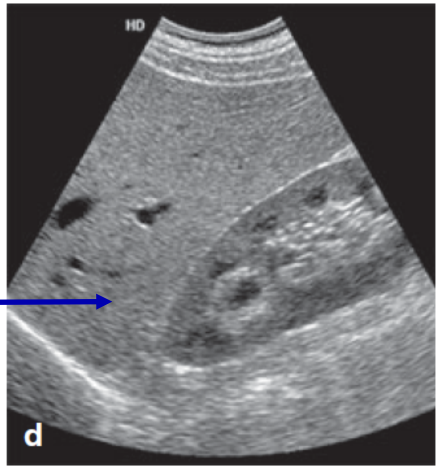
Rayleigh Scattering by Small Structures

- Trajectory deviation due to <u>local</u> inhomogenities
- In tissues, caused by "tiny" particles (cell nuclei, large proteins, inorganic particles, calcifications, …)
- Constructive and destructive interference between scattered waves cause typical "speckle" texture pattern in clinical ultrasound images

Rayleigh scattering



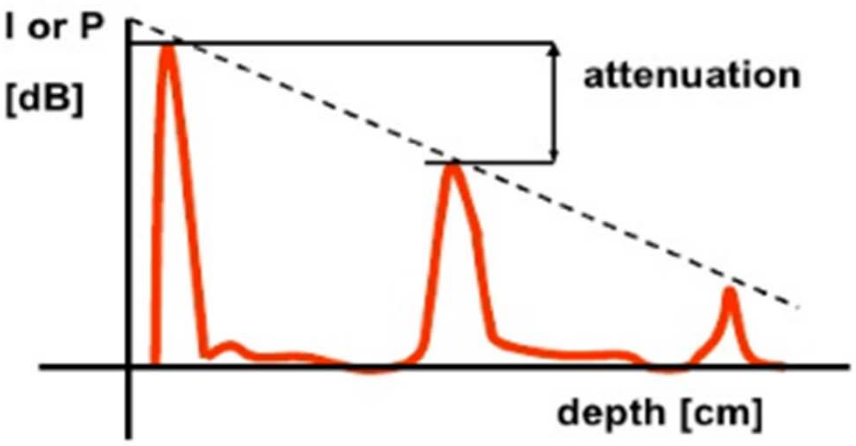
Typical ultrasound speckles (noise) texture



from Smith/Webb

Attenuation

- Energy loss of wave in material with distance
- due to **absorption**, scattering, mode conversion, etc, effects
- Logarithmic loss, as measured by phenomenological experiments:



$$A(z) = A_0\, e^{-\alpha z}$$
$$= A_0\, e^{-\alpha_0 f z}$$

z: depth
$A(z)$: amplitude
A_0: initial amplitude
α_0: attenuation coefficient
f: frequency

Attenuation in decibel: $20 \log_{10}\left(\frac{A_0}{A(z)}\right) dB = \underbrace{20 \log_{10}(e)\, dB\, \alpha_0}\, f z$

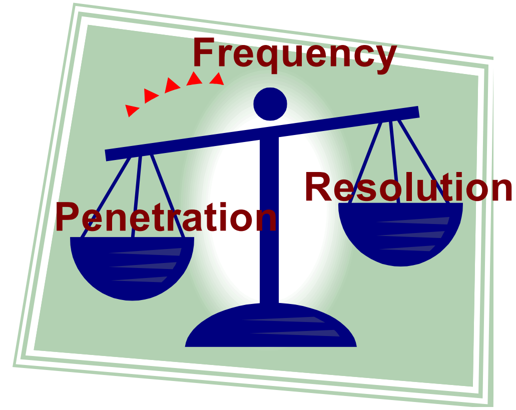
coefficient in decibel [dB/cm/MHz]

Acoustic properties of various materials

Material	Density, ρ [kg m^{-3}]	Speed, c [m s^{-1}]	Characteristic Impedance, Z [kg m^{-2} s^{-1}] ($\times 10^6$)	Absorption Coefficient, α [dB cm^{-1}] (at 1 MHz)	Approximate Frequency Dependence of α
Air at STP	1.2	330	0.0004	12	f^2
Aluminum	2700	6400	17	0.018	f
Brass	8500	4490	38	0.020	f
Castor oil	950	1500	1.4	0.95	f^2
Mercury	13,600	1450	20	0.00048	f^2
Polyethylene	920	2000	1.8	4.7	$f^{1.1}$
Polymethyl-methacrylate	1190	2680	3.2	2.0	f
Water	1000	1480	1.5	0.0022	f^2
Blood	1060	1570	1.62	[0.15]	
Bone	1380–1810	4080	3.75–7.38	[14.2–25.2]	
Brain	1030		1.55–1.66	[0.75]	
Fat	920	1450	1.35	[0.63]	
Kidney	1040	1560	1.62	—	
Liver	1060	1570	1.64–1.68	[1.2]	
Lung	400		0.26	[40]	
Muscle	1070		1.65–1.74	[0.96–1.4]	
Spleen	1060		1.65–1.67	—	
Water	1000	1484	1.52	[0.0022]	

Source: Data above the line are from P. N. T. Wells, *Biomedical Ultrasonics*, (New York: Academic Press, 1977). Data in square brackets below the line are taken from A. B. Wolbarst, *Physics of Radiology* (Norwark, CT: Appleton and Lange, 1993).

Tissue model	1000	**1540**	1.54	**0.5**



Tradeoff: The higher the frequency,

++ The better the imaging resolution,

-- But, the worse the penetration depth

Typical Depth of Penetration for Given Frequencies

Frequency (MHz)	Depth of Penetration (cm)
1	40
2	20
3	13
5	8
10	4
20	2

Slide added for exercise

Attenuation in Tissue

The attenuation increases exponentially with frequency, therefore the attenuation measure in ‘dB’ is roughly proportional to frequency:

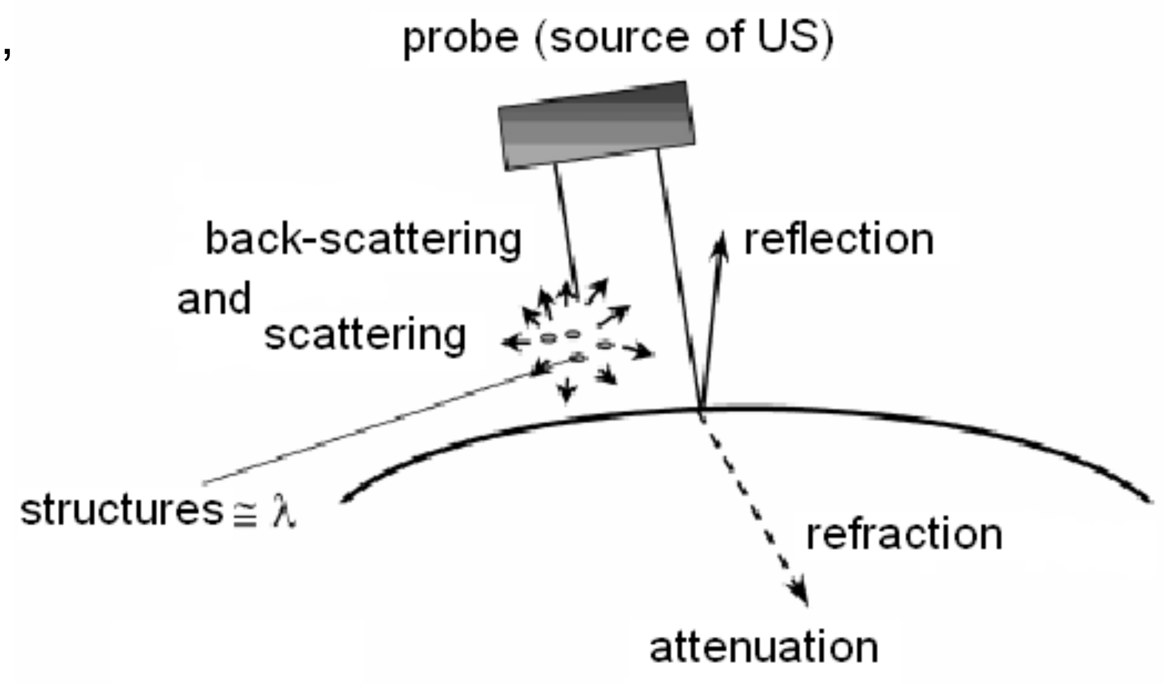
Attenuation measure / frequency	[dB/(cm MHz)]
kidney, brain	ca. 1.0
liver	0.8 - 1.3
skeletal muscle	1.5 - 2.2
fat	ca. 0.5
bone	ca. 13

For a given penetration depth, the sound frequency is limited due to strong increase of the attenuation with frequency.

Example: A 5 MHz sound wave is attenuated in kidney by 5 dB per cm.

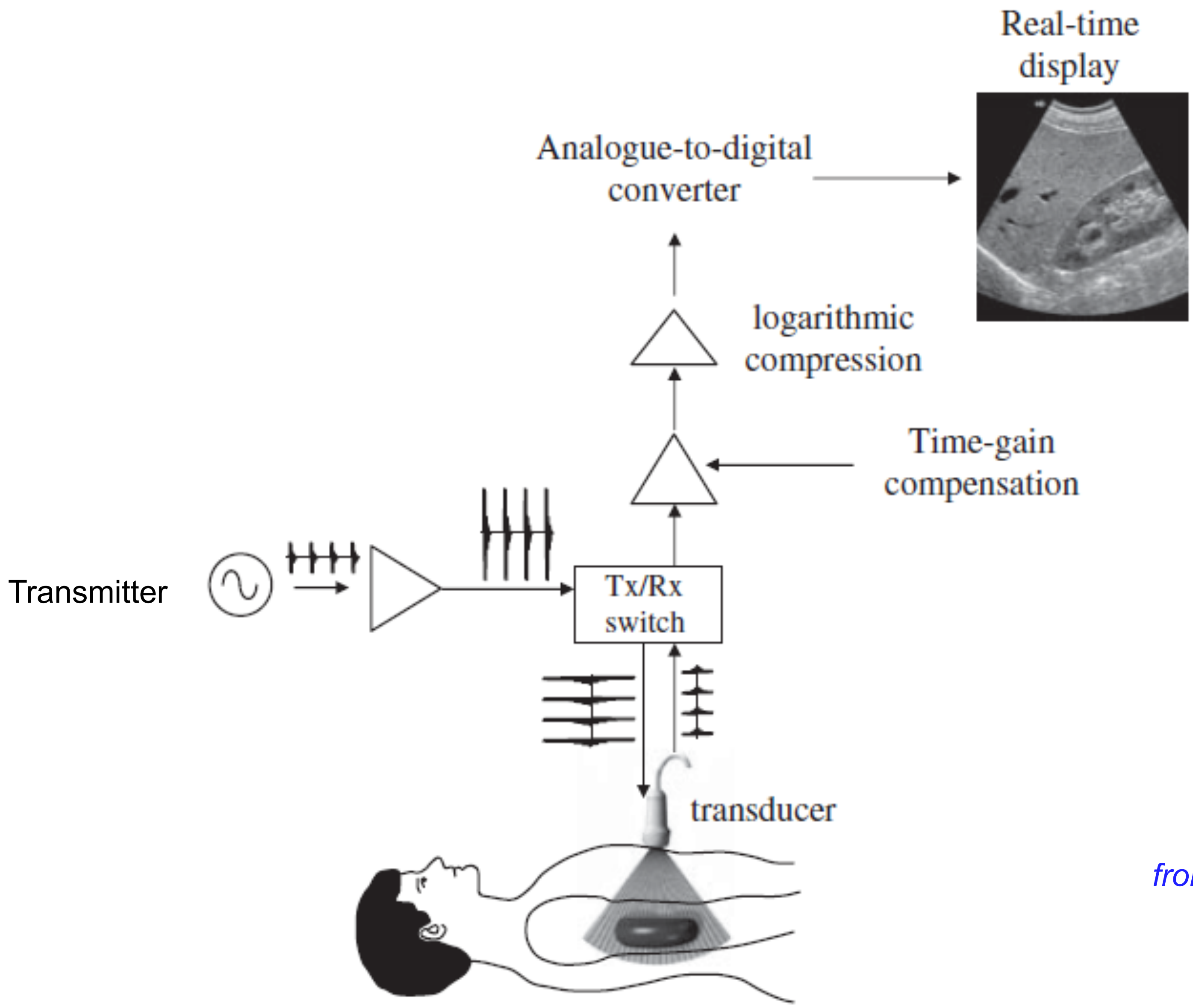
Acoustic interactions with tissue (summary)

- **Reflection**
  (at interfaces larger than wavelength, e.g. organ outlines)
- **Refraction**
  (beam direction change)
- **(Rayleigh) Scattering**
  (small structures; e.g., cells, nuclei, organelles, large proteins, ...)
- **Absorption**
  - Relaxation & classical absorption
  - Increases with frequency
- **Attenuation**
  degradation in wave energy with distance due to absorption, scattering, …



Instrumentation

Chapter 4.5

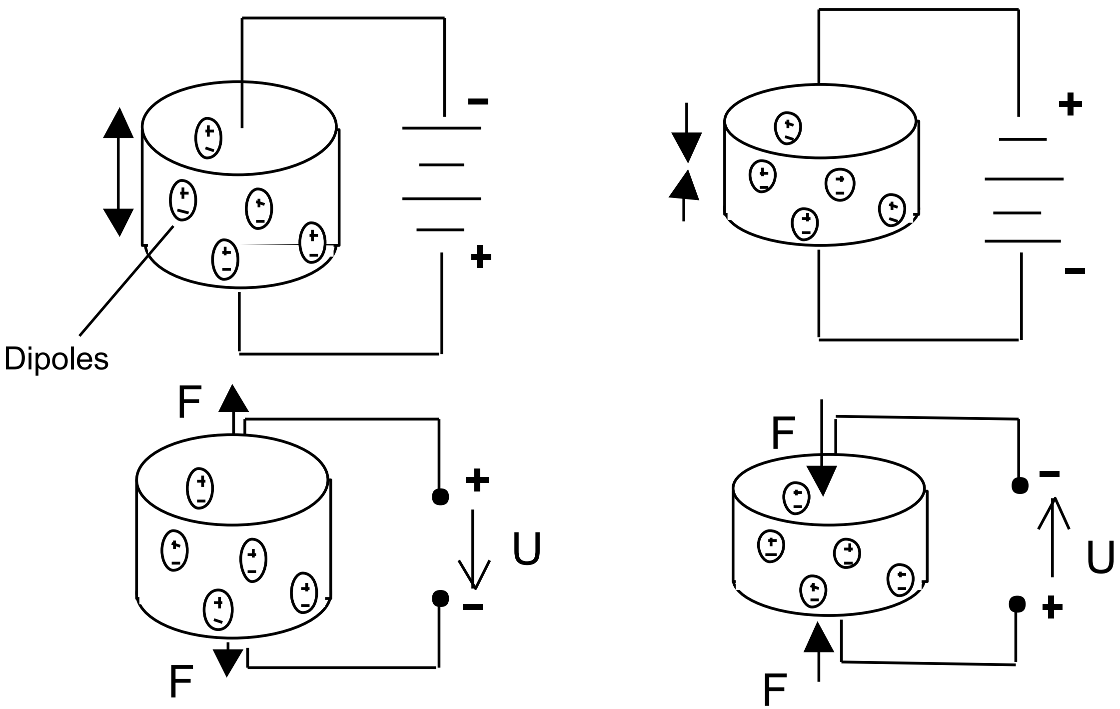


from Smith/Webb

Ultrasound Converter (Transducer)

Chapter 4.6

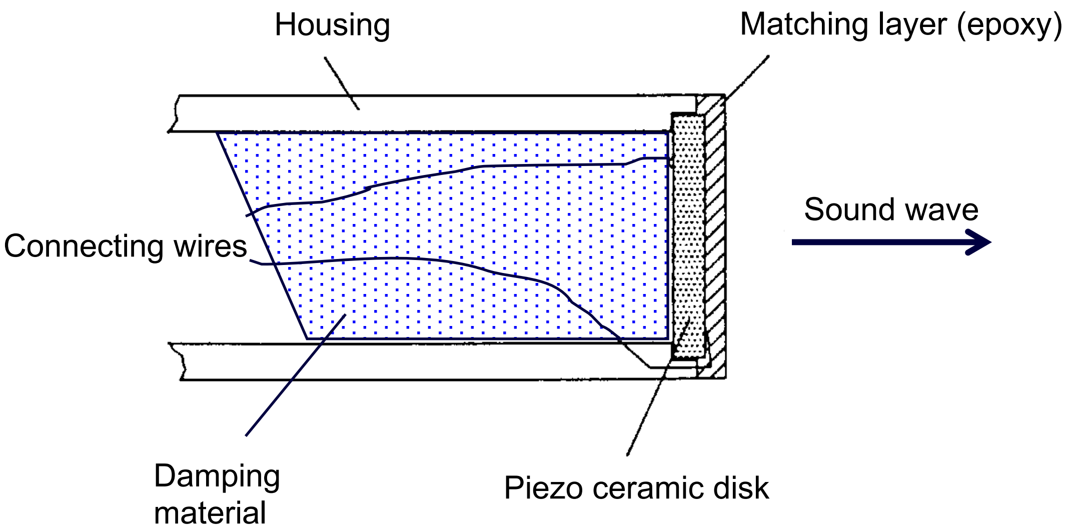
Piezoelectric effect:

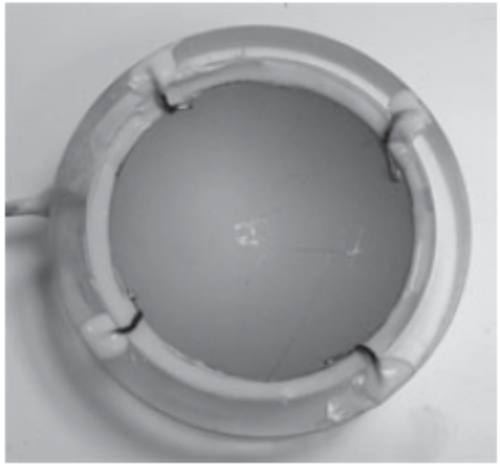


Applied voltage produces force (transmit)

Applied force produces voltage (receive)

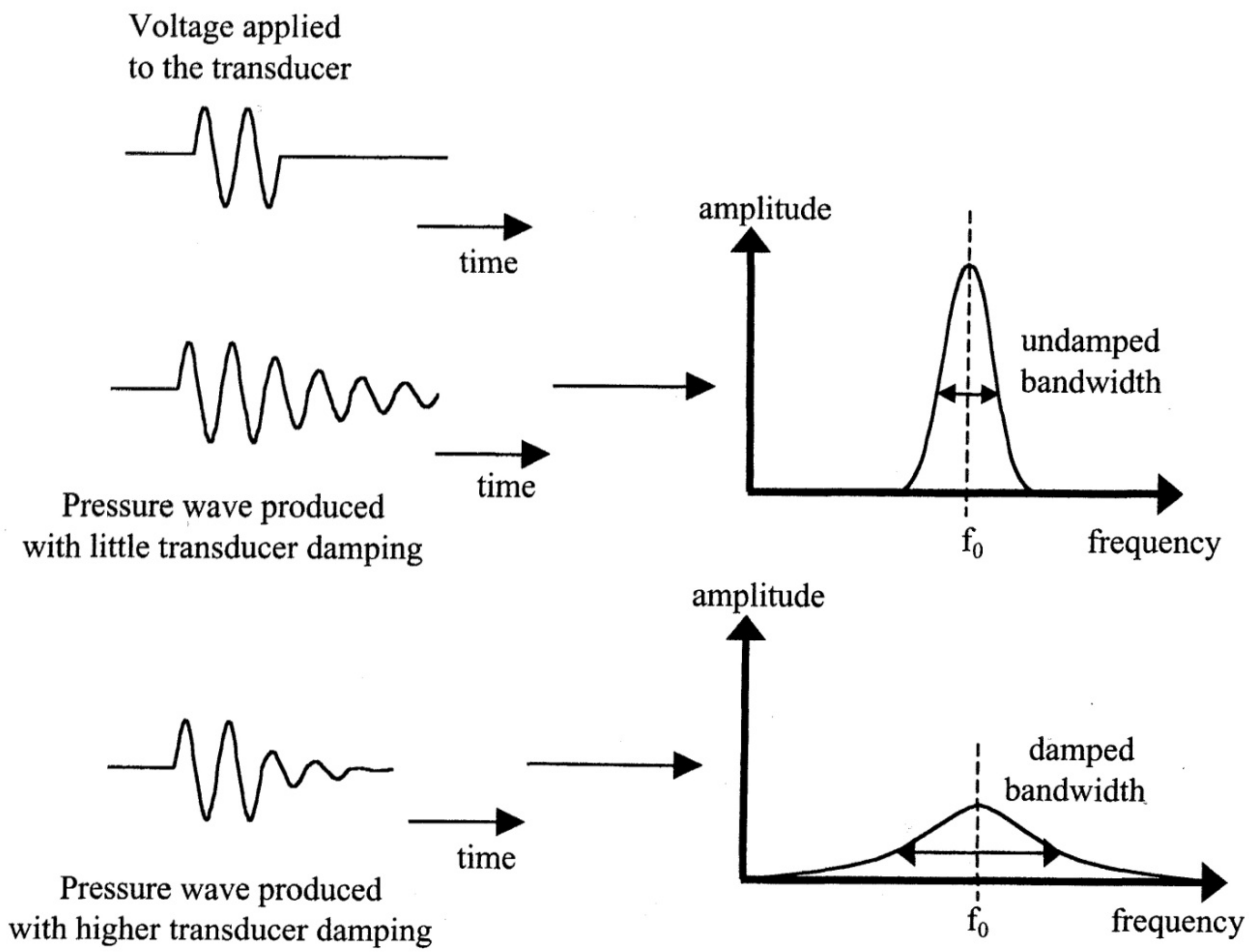
Single Element Transducer





from Smith/Webb

Damping

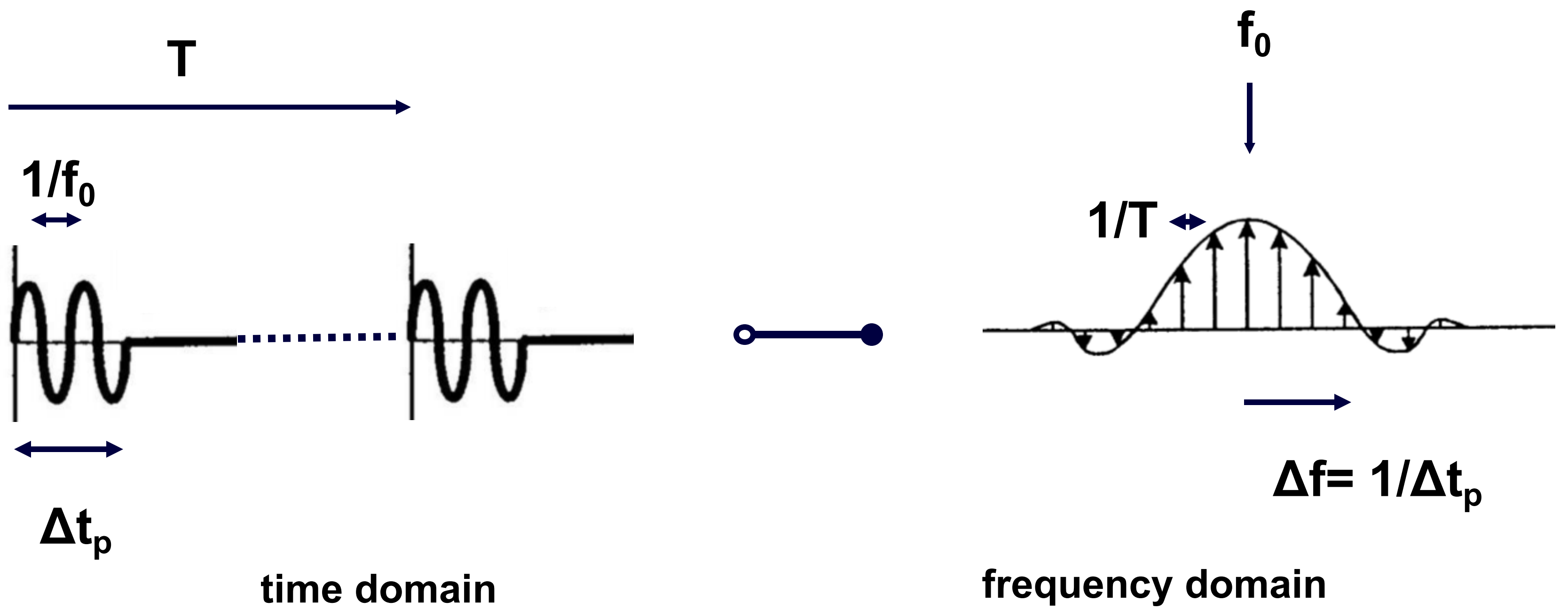


High transducer bandwidth required for short pulses

from Smith/Webb

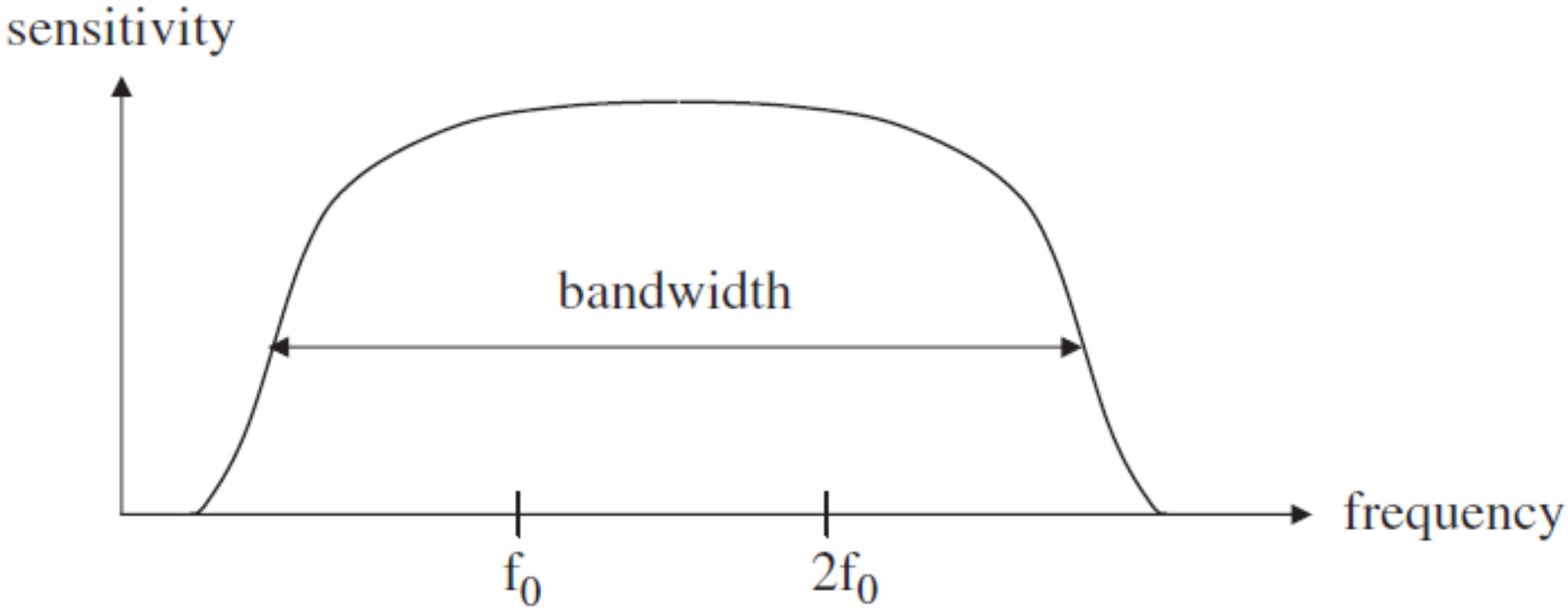
Pulsed Excitation

- A pulse contains not a single frequency, but a spectrum of width $\Delta f \sim 1/\Delta t_p$
- Pulsed excitation is what enables <u>resolution in axial direction</u>



from Smith/Webb

Broadband Transducer



Bandwidth can be larger than f_0 !

from Smith/Webb

Axial Resolution

Features must be at least half the pulse width apart:

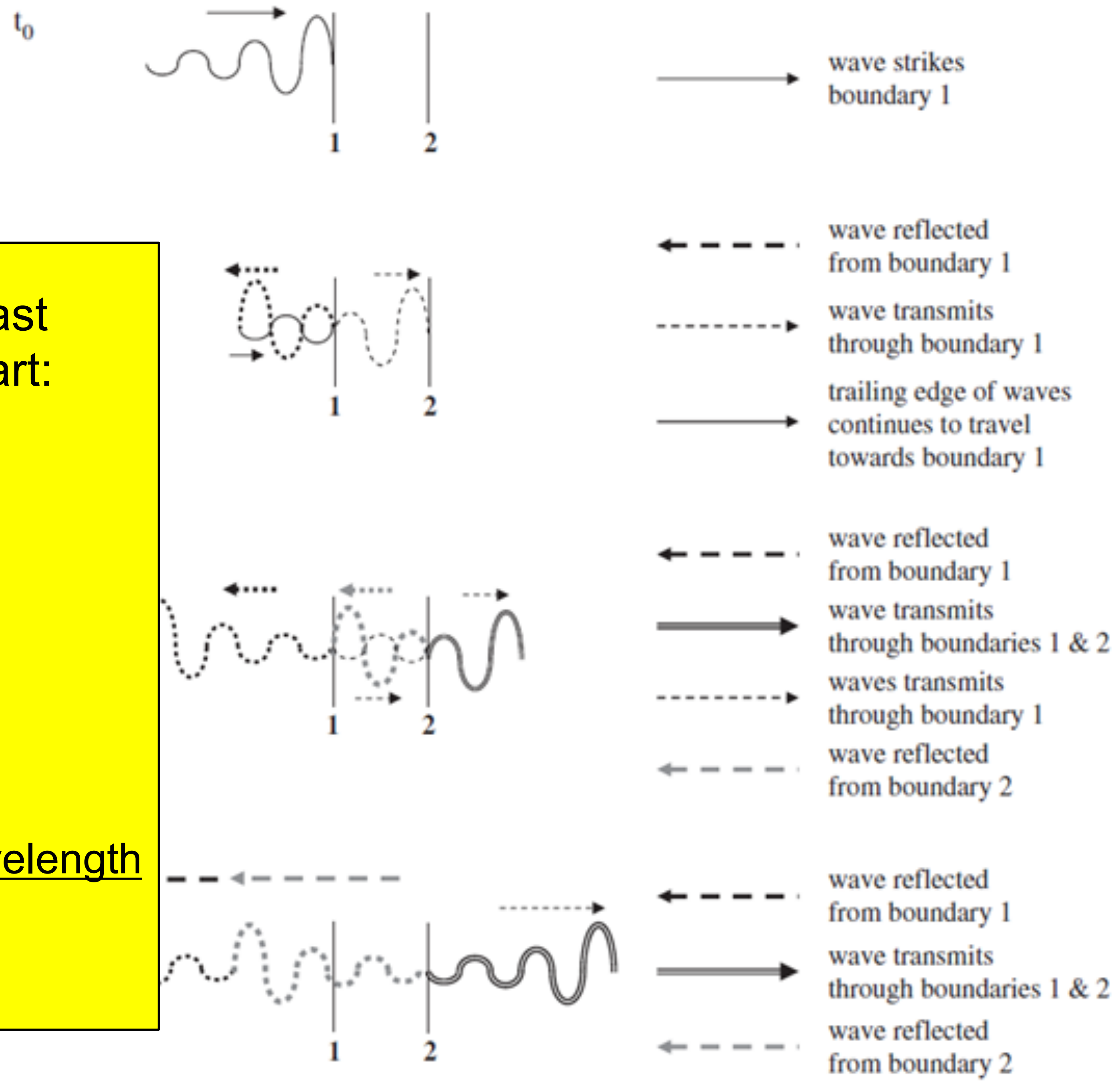
$$\Delta z = \frac{p_d\, c}{2}$$

p_d = pulse duration

c = wave speed

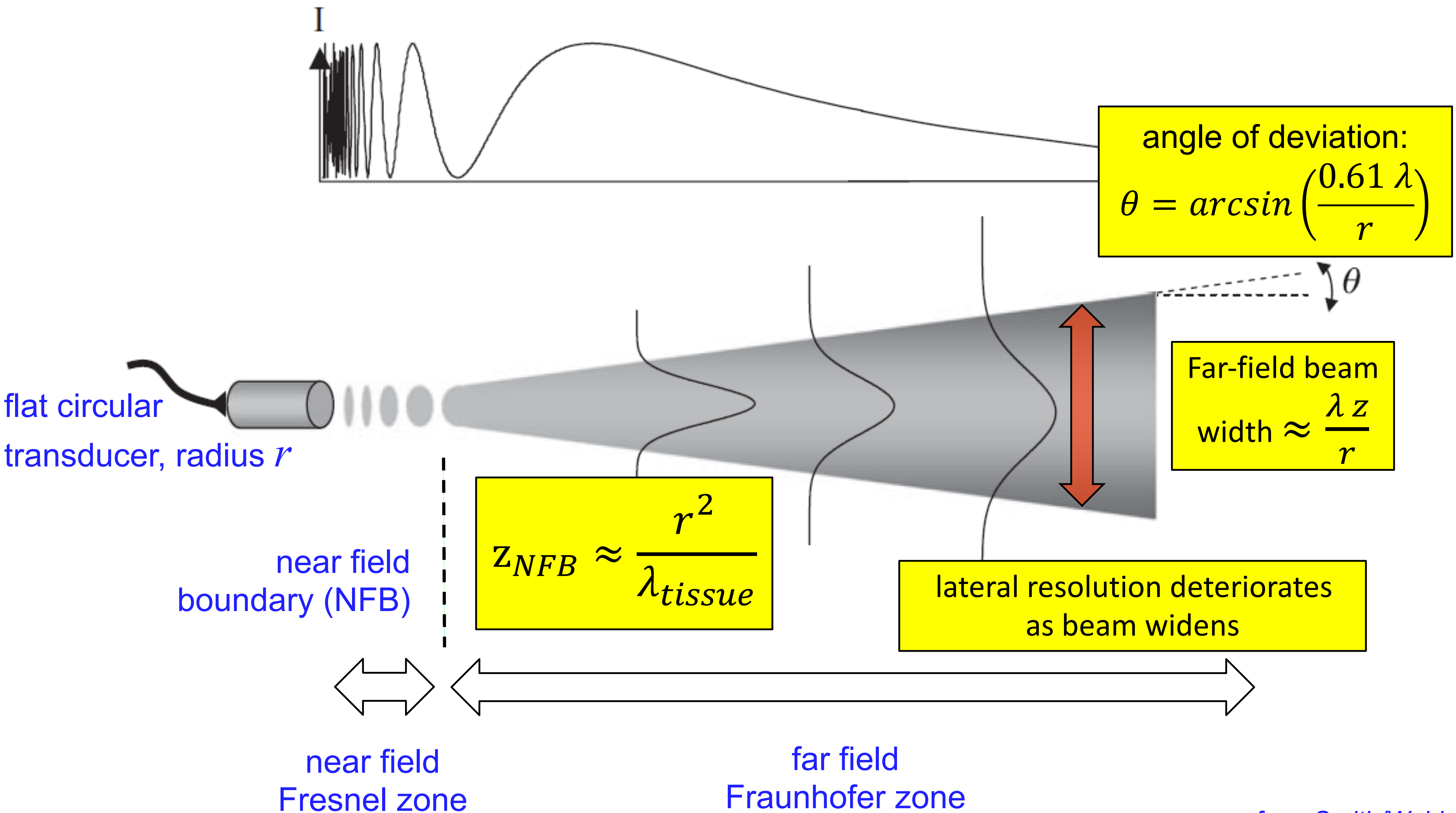
<u>For a single of single wavelength</u>

this implies $\Delta z \geq \frac{\lambda}{2}$



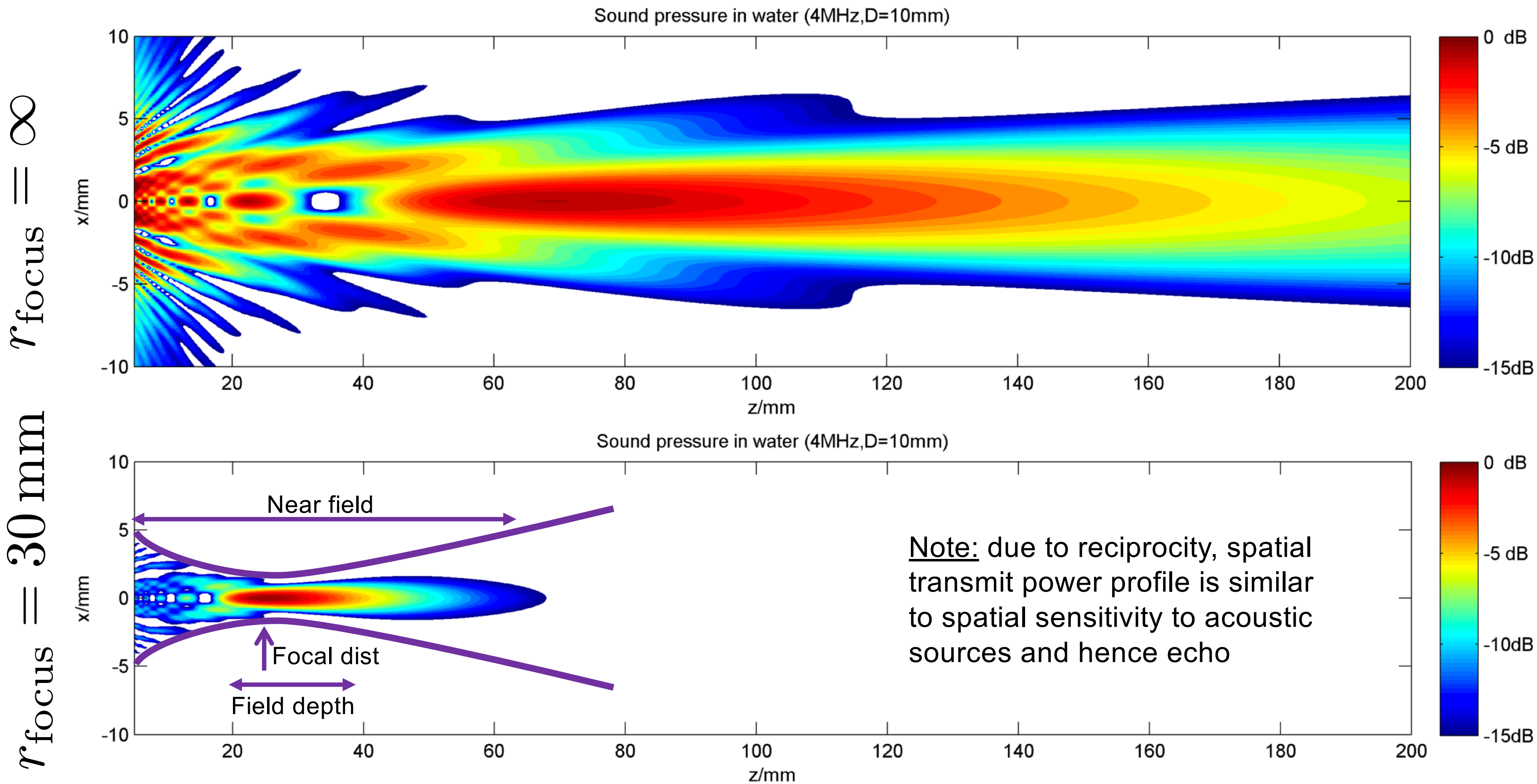
from Smith/Webb

Beam Geometry (natural focusing)



from Smith/Webb

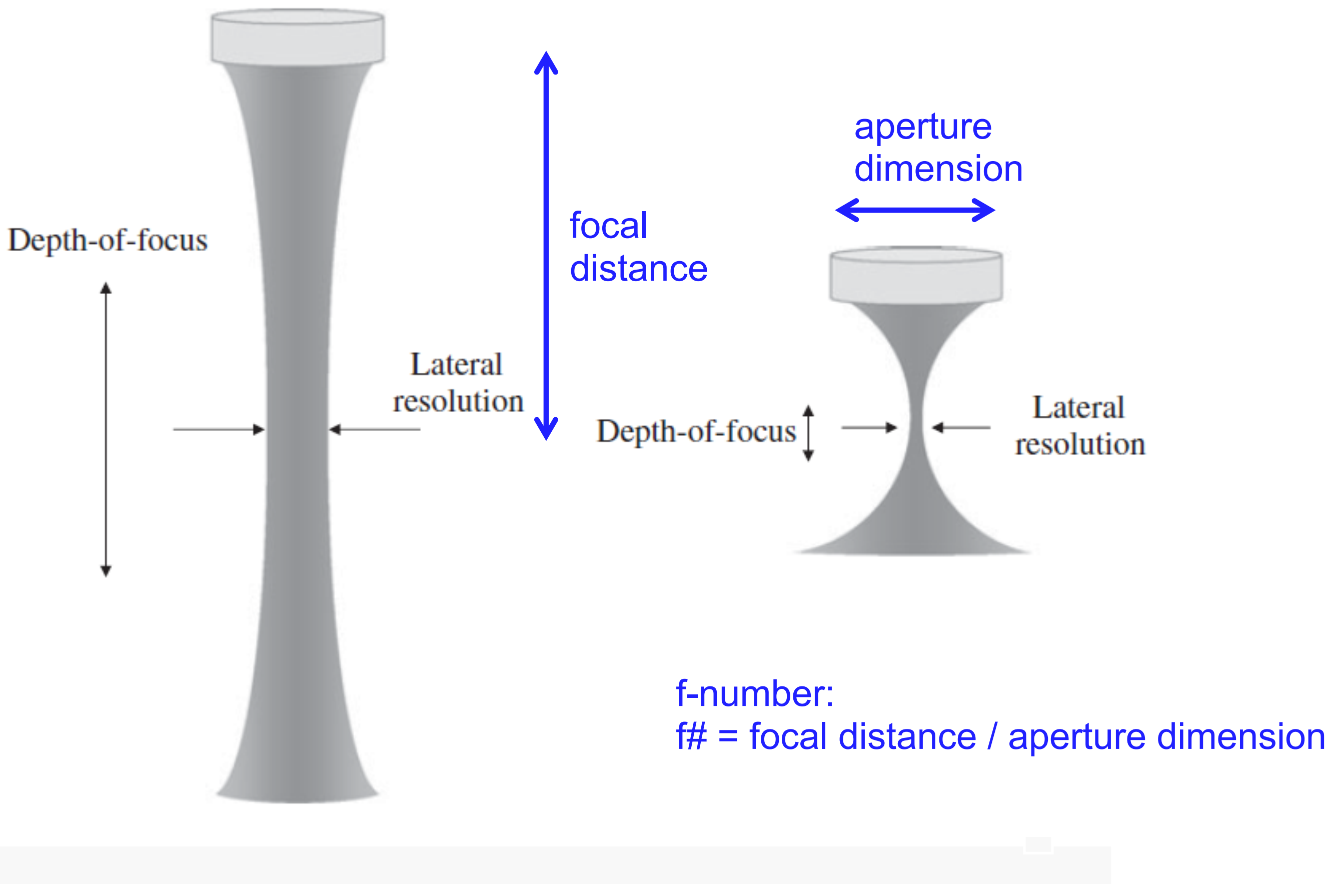
Focusing: Flat vs. curved transducer



Note: due to reciprocity, spatial transmit power profile is similar to spatial sensitivity to acoustic sources and hence echo

Applet for those interested: http://www.imasonic.com/Industry/IM_Design.php

Focusing with an Acoustic Lens



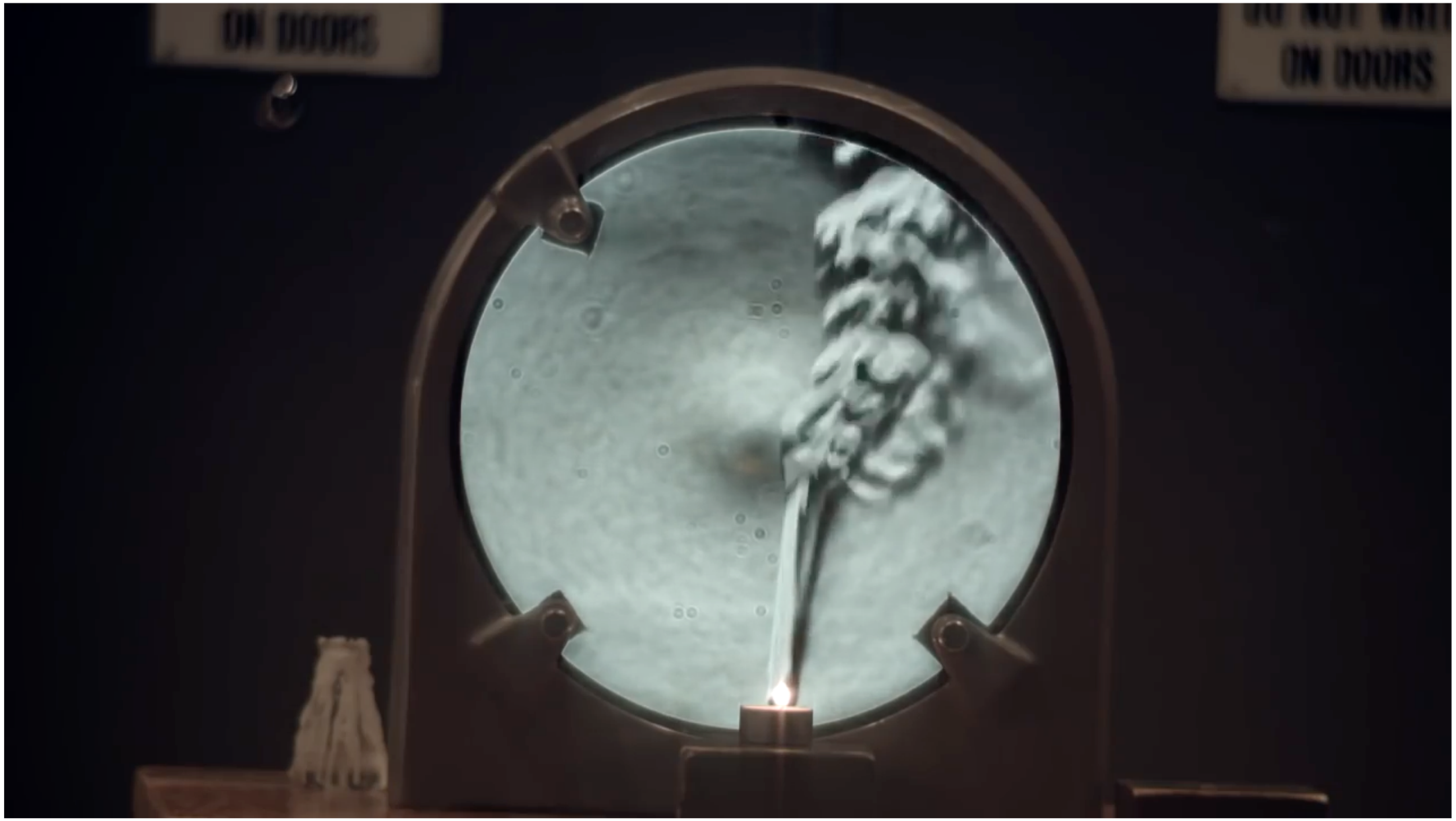
Strong focusing: good lateral resolution but short depth-of-focus and worse far-field resolution

Weak focusing: medium lateral resolution, longer depth-of-focus and medium far-field resolution

from Smith/Webb

Schlieren Imaging

can visualize density variations in transparent media



https://www.youtube.com/watch?v=mLp_rSBzteI

Harvard Natural Sciences Lecture Demonstrations

Beam Geometry

For ultrasonic waves:

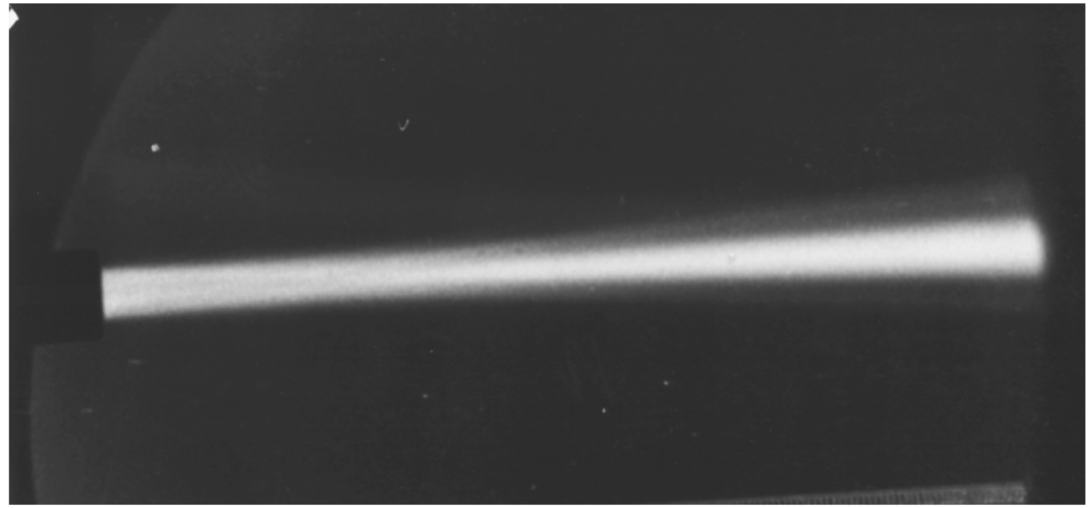
Pressure changes

local density variations

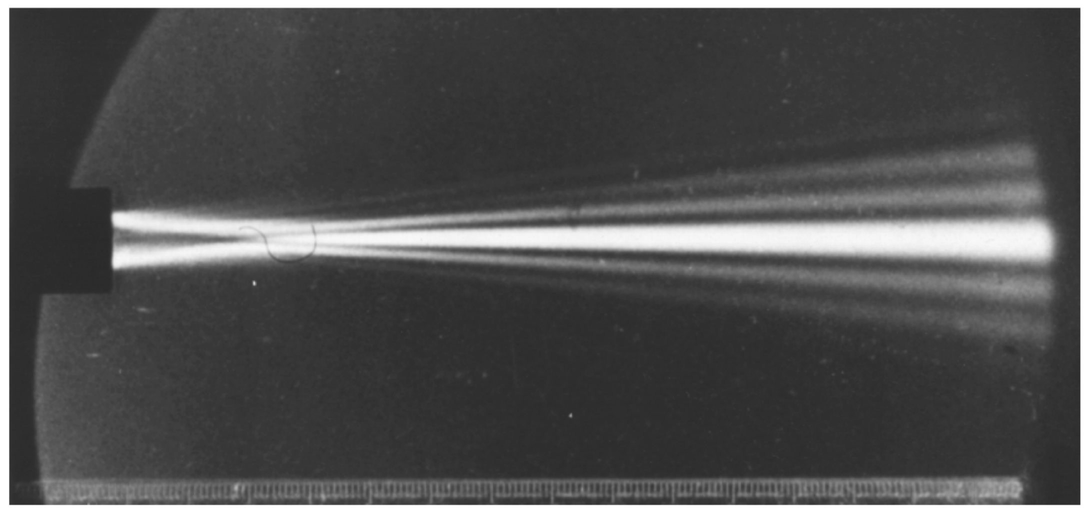
refractive index variations

can be imaged optically

Schlieren pictures of ultrasound in water tank

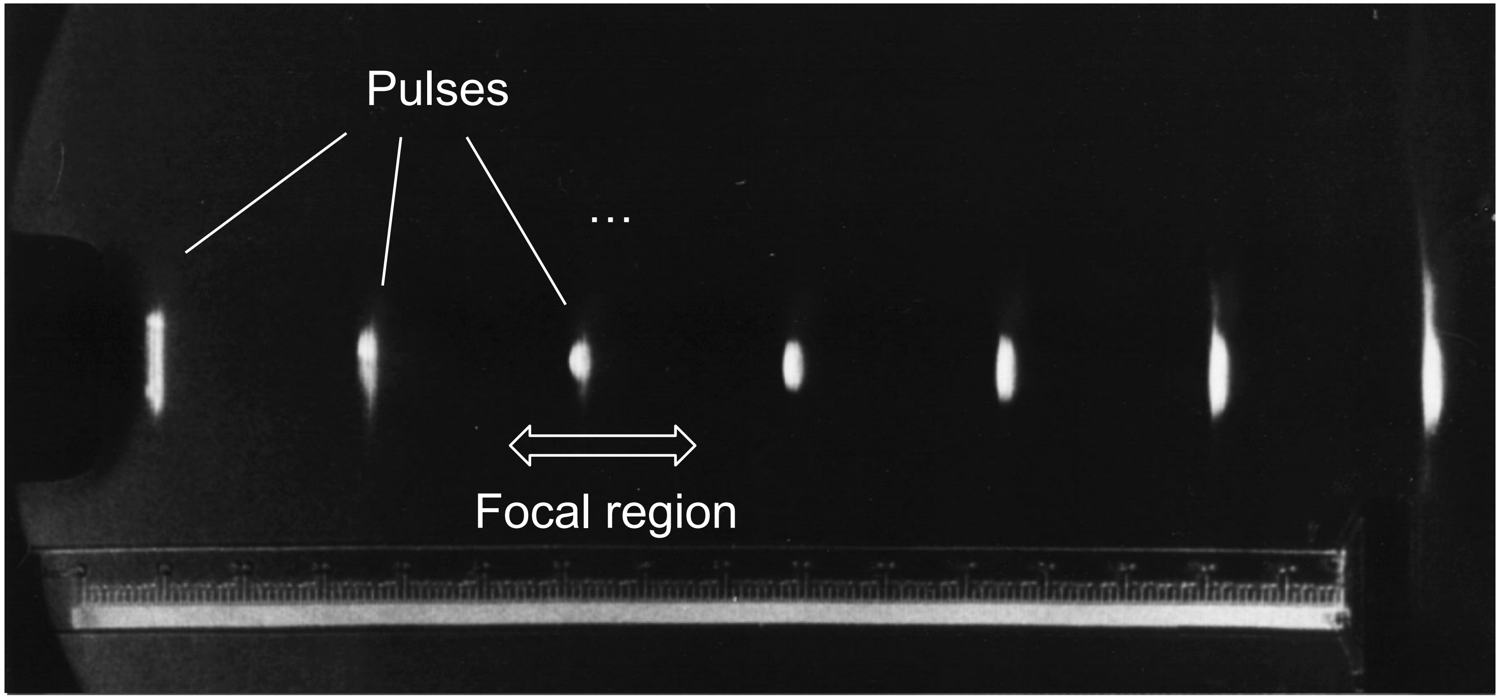


Unfocused ultrasound beam



Focused ultrasound beam

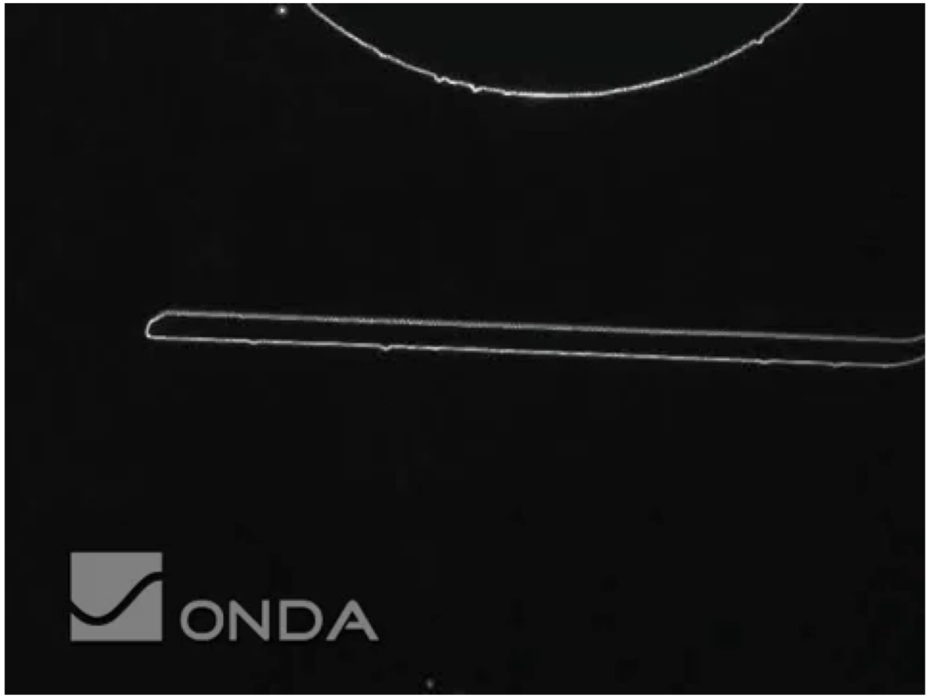
Beam Geometry



Focused and pulsed ultrasound beam in water tank

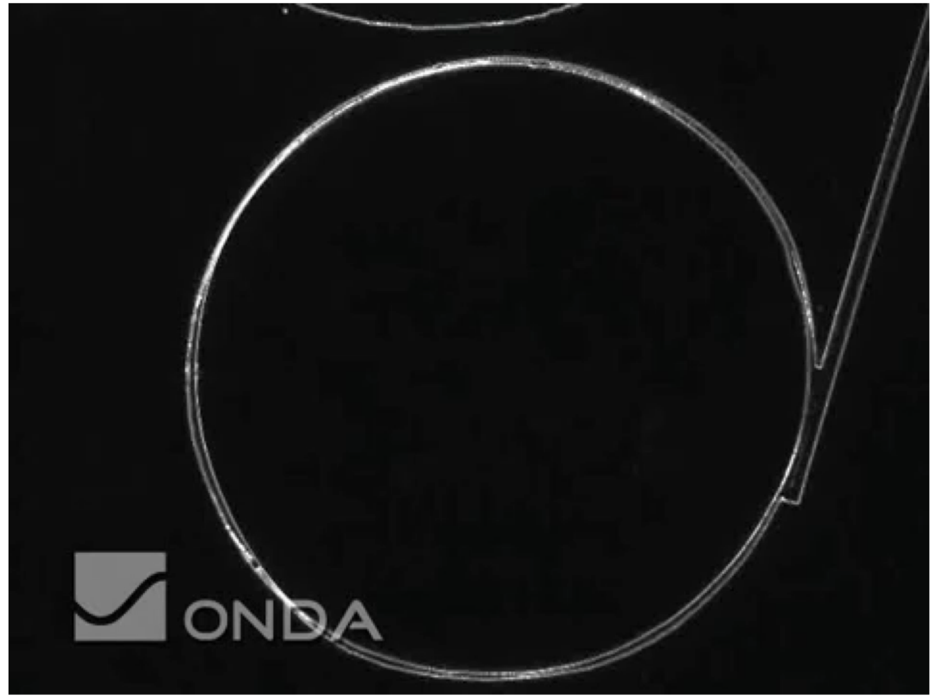
Schlieren picture

Propagation, Reflection, Focusing, Attenuation



https://www.youtube.com/watch?v=PPBKjSRTwqs

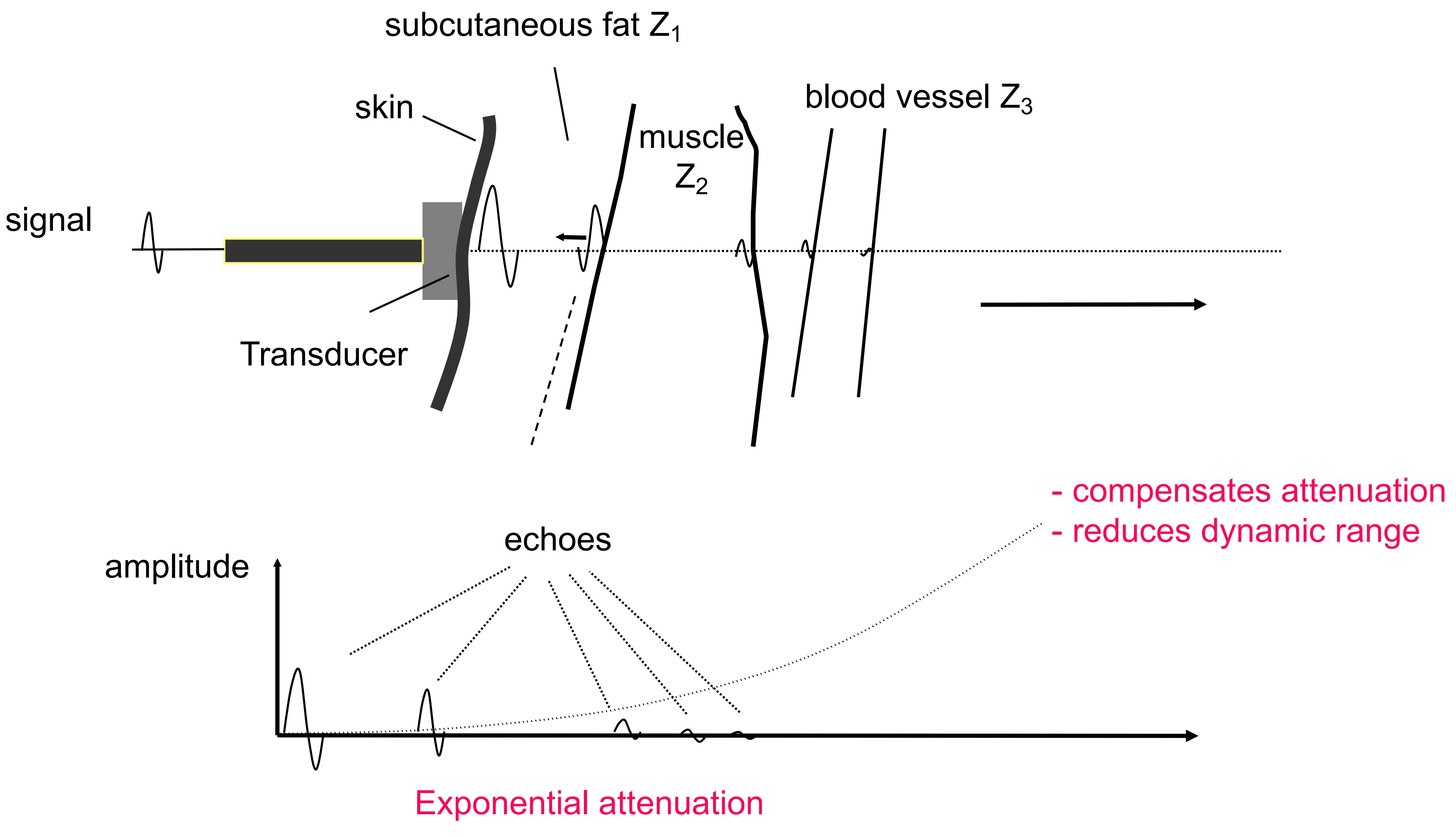
ONDA Corporation



https://www.youtube.com/watch?v=jBEV3lRPNJY

ONDA Corporation

Range Gain (time gain compensation)



SNR in Ultrasound Devices: Example

Center frequency: 3.5 MHz

Medium attenuation constant: 1 dB/cm/MHz

Imaging depth: 15 cm

Calculate Attenuation:

3.5 MHz x 1 dB/(cm MHz) x (2 x 15cm) = **105 dB**

SNR assuming no attenuation:

Transmitted peak pulse power: $P_{Tx} = 10$ W

Thermal noise of the transducer at bandwidth B = 1 MHz and T = 310°K

$$P_N = k\,T\,B = 4.3 \times 10^{-15}\ \mathrm{W}$$

with Boltzmann constant $k = 1.38 \cdot 10^{-23}$ J/K

➔ *SNR assuming no attenuation*: P_{Tx} / P_N → **154 dB**

SNR for imaging an object at 15 cm and reflecting 3% (i.e. 30 dB) of incident signal:

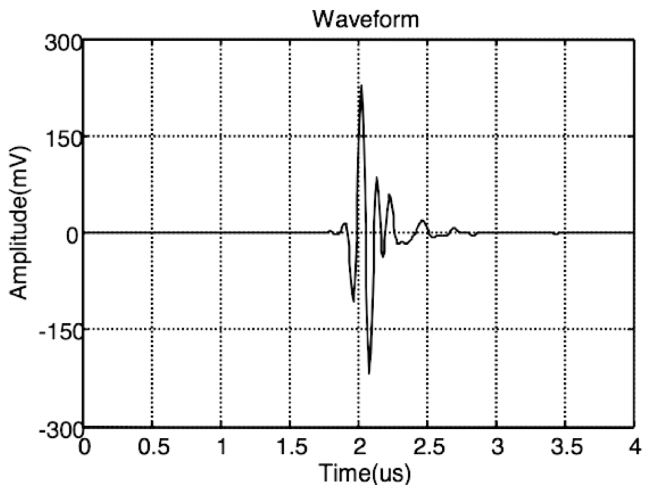
154 - 105 - 30 = **19 dB**

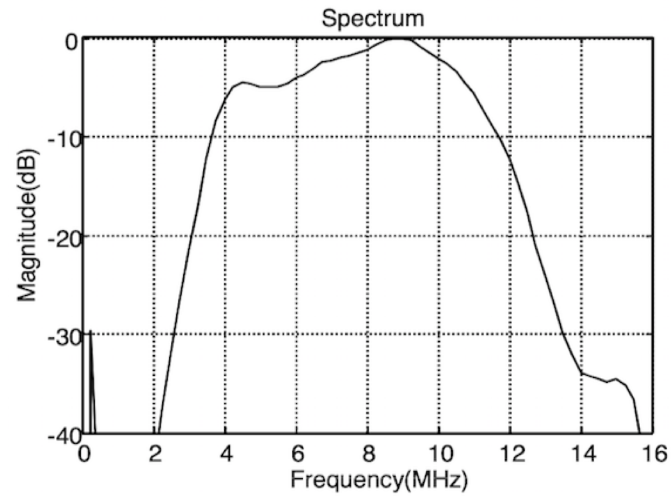
Sample transducer specs

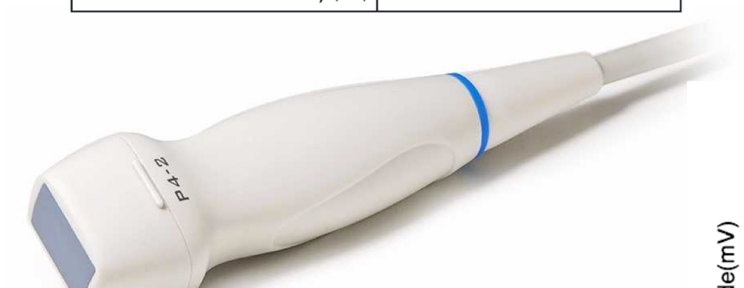
L 12-3v

192-element linear array transducer

Number of elements	192
Pitch (mm)	0.2
Elevation focus (mm)	20
Sensitivity (dB)	-58.4



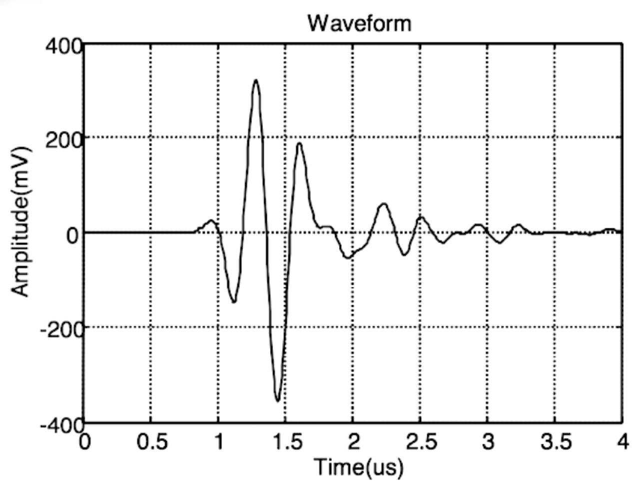


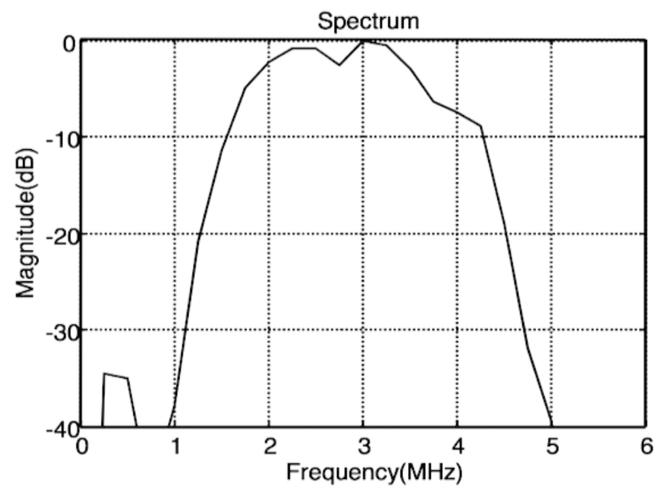


P 4-2v

64-element phased array transducer

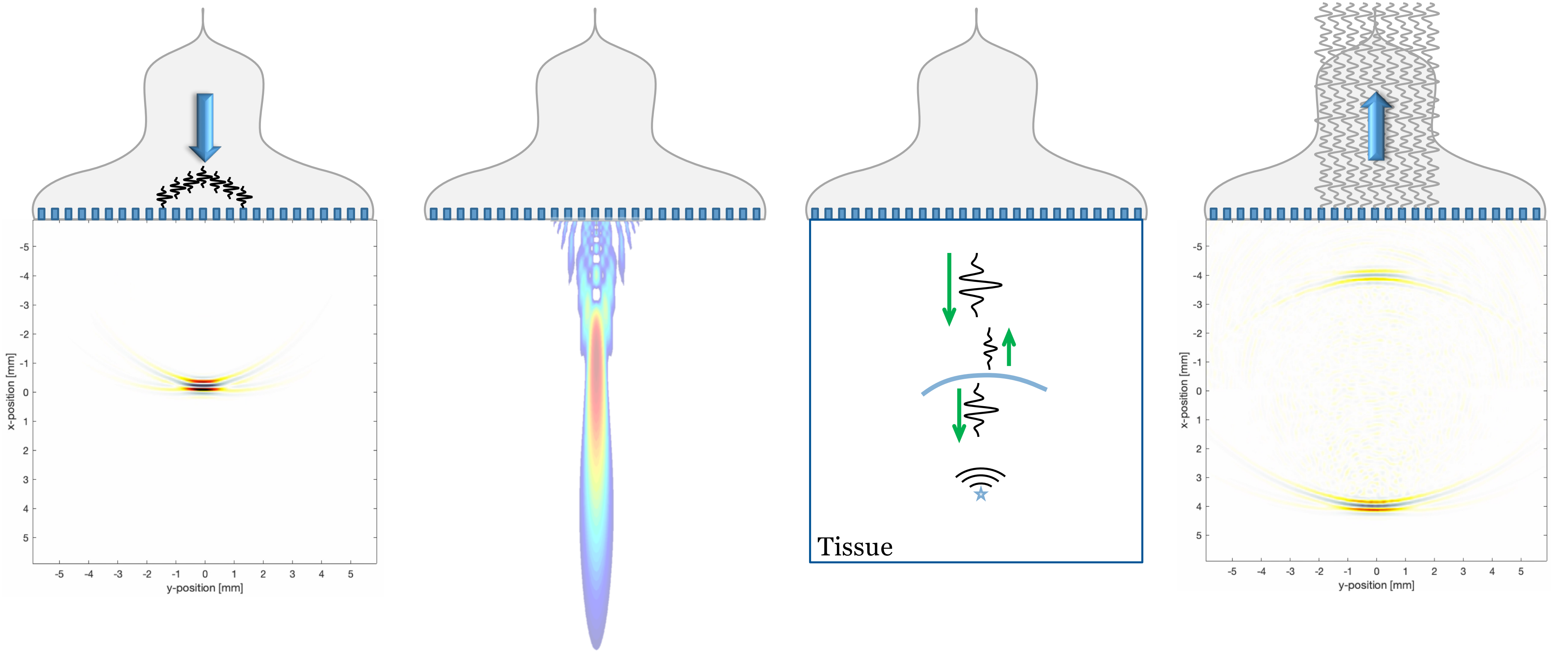
Number of elements	64
Pitch (mm)	0.3
Elevation focus (mm)	50-70
Sensitivity (dB)	-69 -95





http://verasonics.com/new-transducer-page/

Transmit (Tx) and Receive (Rx) : a sneak peak



Reflection/Refraction example

$$c_2 = 3c_1 \qquad \rho_1 = \rho_2$$