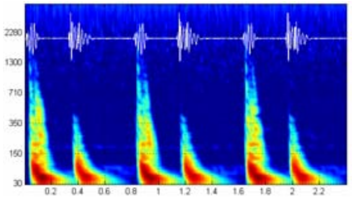
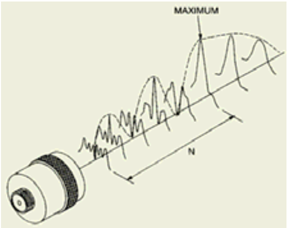
Physics of sound/ultrasound



Biomedical ultrasound



To become familiar with:

Sound / Ultrasound wave
Basics of psychoacoustics

Wave propagation and Scattering

Biomedical Ultrasound Transducers

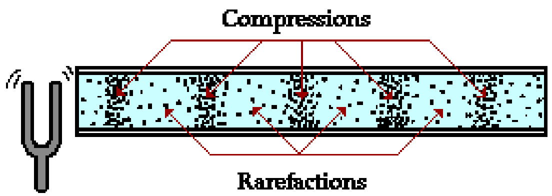
Ultrasound Imaging / Other Applications

Sound as a Mechanical Wave

- A Mechanical Wave: sound waves travel through air by way of particle interactions
- Mechanical waves require a medium through which to travel
- Cannot travel through a vacuum

Sound: mechanical wave (model)

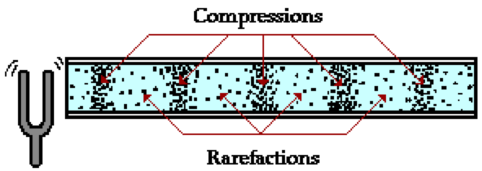
Ex. A vibrating string : The particles of the sound wave collect in “compressions” (high pressure areas) and “rarefactions” (low pressure areas)



Sound: pressure wave

spatial and temporal periodicity

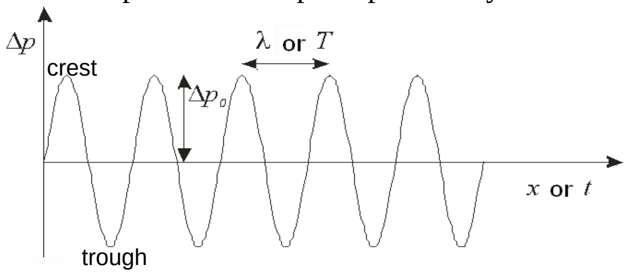
Sound: mechanical wave (model)

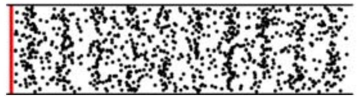


Sound: pressure wave

spatial and temporal periodicity

function

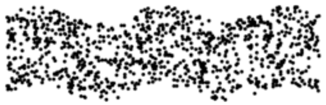




longitudinal wave

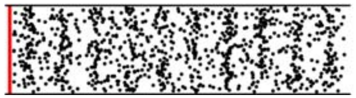
motion of particles is parallel to direction of energy movement

in the interior of liquids and gases only this type



transverse wave

motion of particles is perpendicular to direction of energy movement



longitudinal wave

transverse wave

hydrostatic pressure

pressure change, sound pressure

$p_{total} = p_{hydrostat} + \Delta p$

pressure DC + AC

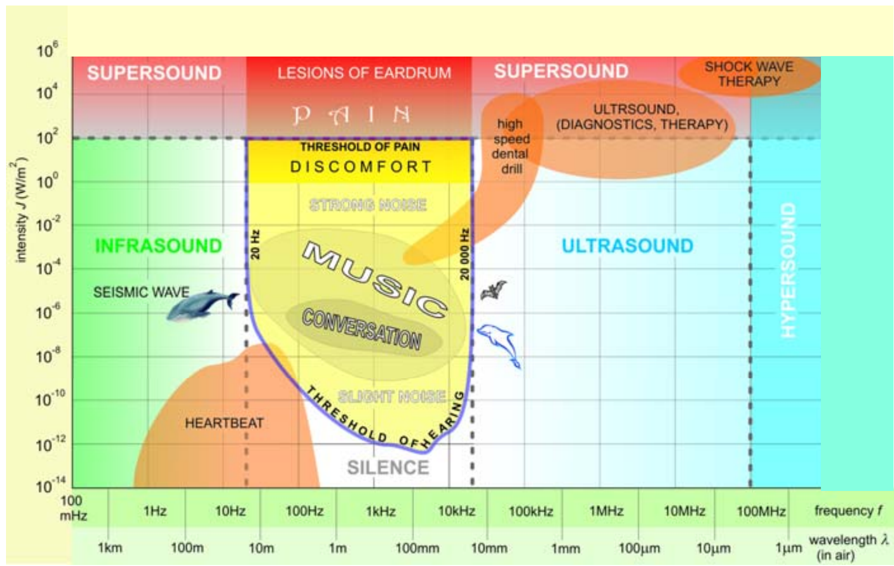
amplitude phase

$$\Delta p(t,x) = \Delta p_{max} \sin\left[2\pi\left(\frac{t}{T} - \frac{x}{\lambda}\right)\right]$$

$$c \cdot T = \lambda, \quad c = f \cdot \lambda$$

Textbook, Fig. II.46.

Frequency and intensity regions of sounds



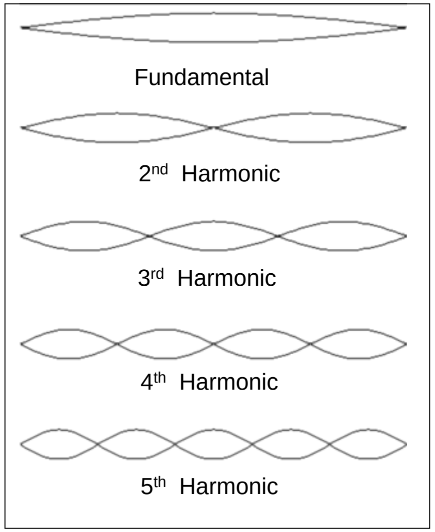
Lab. manual, Audiometry.

Frequency of Sound

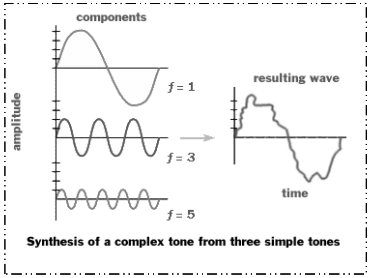
Harmonics

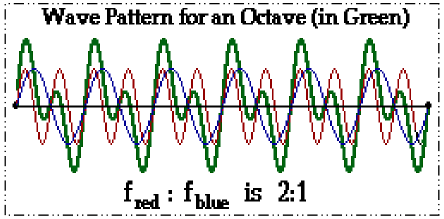
- Strings and air columns do not vibrate at one frequency; they also vibrate in smaller sections
- Additional frequencies are usually multiples of the **fundamental** vibration frequency – **"tones"**
- These **harmonics** are sometimes called **"overtones"**

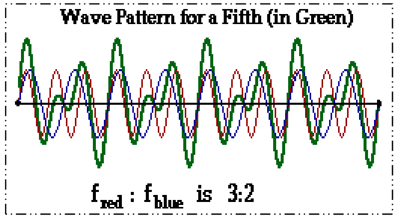
Overtone	Frequency (Hz)	Interval	Ratio
Fundamental	440	Tonic	2:1
2nd	880	Tonic	2:1
3rd	1320	Perfect 5th	3:2
4th	1760	Tonic	2:1
5th	2200	Major 3rd	5:4
6th	2640	Perfect 5th	3:2
7th	3080	Dominant 7th	9:5
8th	3520	Tonic	2:1



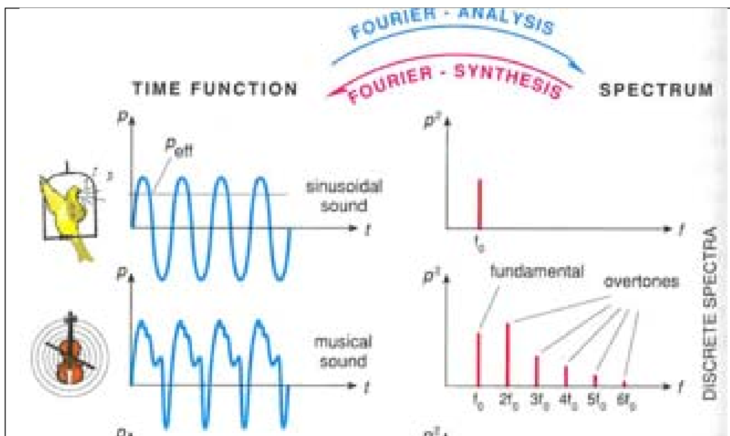
Wave Synthesis







Blue: Fundamental / Red: (Harmonic) / Green: Resultant sound



Periodic functions can be analysed into their constituent components (fundamentals and harmonics) by a process called **Fourier analysis**.

The **Fourier theorem states** that any waveform can be duplicated by the superposition of a series of *sine* and *cosine* waves.

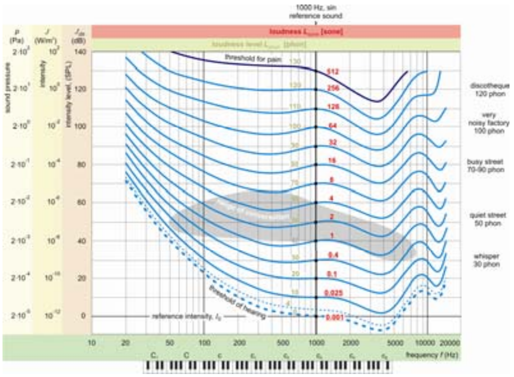
Textbook, Fig. IV.23.

Objective quantities	Psychoacoustical attributes
Frequency of tones	**Pitch**
Collection of frequencies, relative strengths of overtones/harmonics (spectrum)	**Timbre***
Intensity	**Loudness**

*Noise is a collection of frequencies that *are not* aligned in whole-number ratios

Equal loudness curves, measured using harmonic sound waves

$$L_{phon} = 10\lg\left(\frac{J}{J_0}\right)_{1000\text{Hz}}$$



Phon value of the sound of any frequency equals to the dB value of the reference sound (1000 Hz, sin) producing the same perception of loudness level.

Propagation and Scattering of Sound Wave

The role of elastic medium

$$\kappa = \frac{-\Delta V / V}{\Delta p}$$

compressibility
relative volume decrease over pressure

$$c = \frac{1}{\sqrt{\rho\kappa}}$$

speed of sound

$$Z = \frac{p}{v} = \frac{p_{max}}{v_{max}}$$

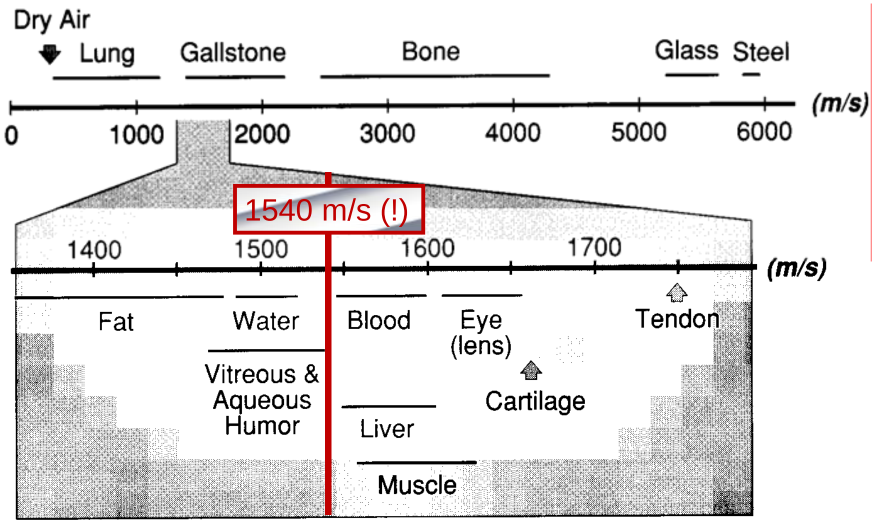
acoustic **impedance** (definition)

$$Z_{el} = \frac{U}{I}$$

$$\boxed{Z = c\rho} = \sqrt{\frac{\rho}{\kappa}}$$

acoustic **impedance** (useful form)

Speed of sound/US in different media



Density, Speed of Sounds and Acoustics Impedance of human tissue

Medium	Density (kg/m^3)	Speed of Sound (m/s)	Acoustic Impedance ($kg/m^2.s$) $x10^6$
Air	1.2	333	0.0004
Blood	1060	1566	1.66
Bone	1380-1810	2070-5350	3.75-7.38
Brain	1030	1505-1612	1.55-1.66
Fat	920	1446	1.33
Kidney	1040	1567	1.62
Lung	400	650	0.26
Liver	1060	1566	1.66
Muscle	1070	1542-1626	1.65-1.74
Water	1000	1480	1.48

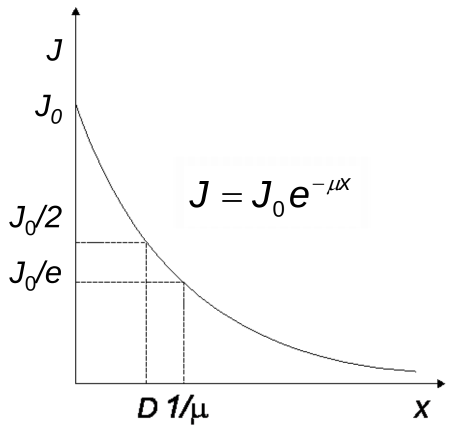
$$Z = c\rho = \sqrt{\frac{\rho}{\kappa}}$$

Intensity of US

$$J = \frac{1}{Z}\Delta p_{eff}^{\ 2}$$

intensity = energy-current density

Loss of energy during propagation (absorption)



attenuation:

$$\alpha = 10 \cdot \lg \frac{J_0}{J}\ \text{dB}$$

$$\alpha = 10 \cdot \mu \cdot x \cdot \lg e\ \text{dB}$$

μ is proportional to **frequency** in the diagnostic range

specific attenuation: $\boxed{\dfrac{\alpha}{f \cdot x}}$

μ is proportional to **frequency** in the diagnostic range

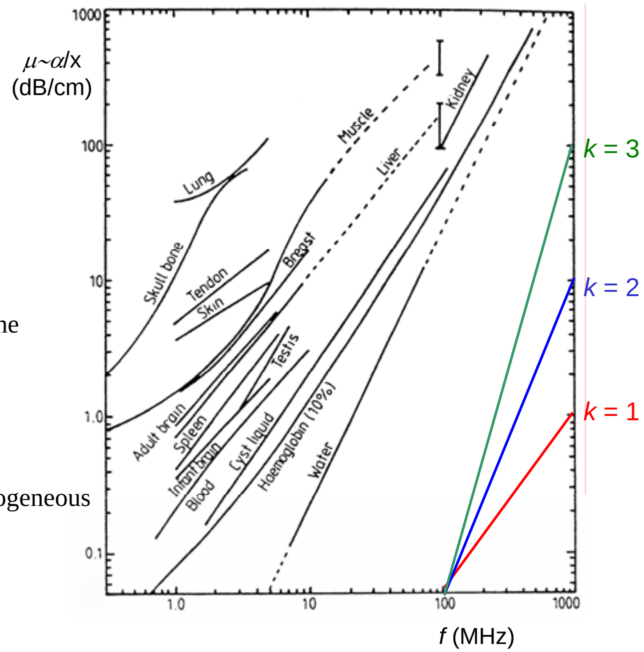
$$\mu \sim f^k, \quad k \sim 1(?)$$

$$\log \mu \sim k \log f$$

if the graph is a linear, the power function approximation is valid

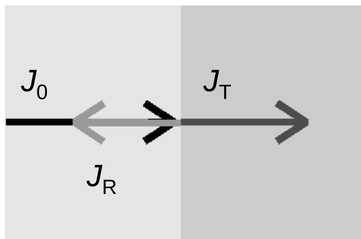
specific attenuation for soft tissues (homogeneous tissue model):

$$\boxed{\frac{\alpha}{f\,x} \sim 1 \frac{\text{dB}}{\text{cm MHz}}}$$



Phenomena at the boundary of different media

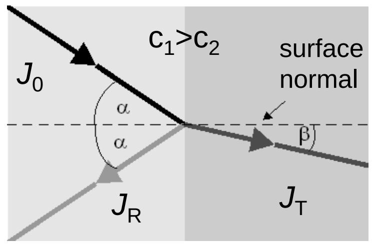
normal/perpendicular incidence



$J_0 = J_R + J_T$

reflection and transmission (penetration)

skew incidence



$$\frac{\sin\alpha}{\sin\beta} = \frac{c_1}{c_2}$$

Snellius-Descartes

Textbook, Fig. II.47.

Reflection (normal incidence)

reflectivity:

$$\boxed{R = \frac{J_{\text{reflected}}}{J_{\text{incident}}} = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2}\right)^2}$$

boundary surface	*R*
muscle/blood	0.001
fat/liver	0.006
fat/muscle	0.01
bone/muscle	0.41
bone/fat	0.48
soft tissue/air	0.99

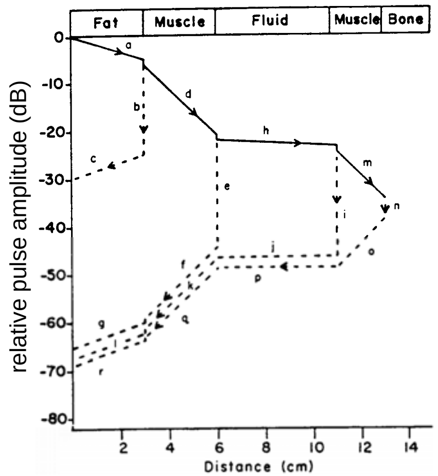
"full" reflection:

$$Z_1 << Z_2, \quad R \approx 1$$

optimal coupling:

$$Z_{\text{connecting}} \approx \sqrt{Z_{\text{source}} Z_{\text{skin}}}$$

Absorption and reflection



the later comes back the reflection, the deeper lays the reflecting surface and the weaker is the intensity

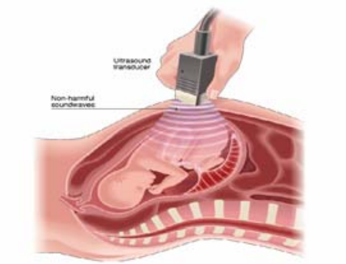
run time dependent amplification

TGC: time gain compensation

DGC: depth gain control

boundary surface	*R*	10lg*R* (dB)	*T*	10lg*T* (dB)
fat/muscle	0.01	-20.0	0.990	-0.044
muscle/blood	0.001	-30.0	0.999	-0.004
muscle/bone	0.41	-3.9	0.590	-2.291

Biomedical Ultrasound Transducers



Generation of US

source of electric signal (sine wave oscillator)

+

transducer (piezo-crystal)

Piezoelectric effect

piezo-crystal

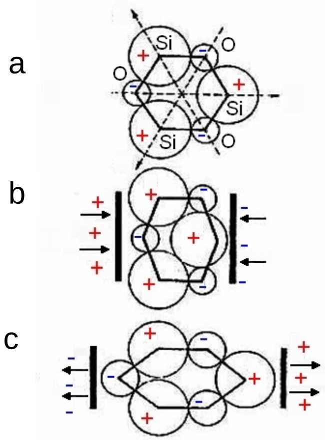
a) Center of charge of positive and negative charges coincides.

b) and c) ***inverse*** piezoelectric effect

As a result of potential difference, the charge centers are separated, i.e. crystal deformation.

$\downarrow\downarrow$

Production of US



Piezoelectric effect

piezo-crystal

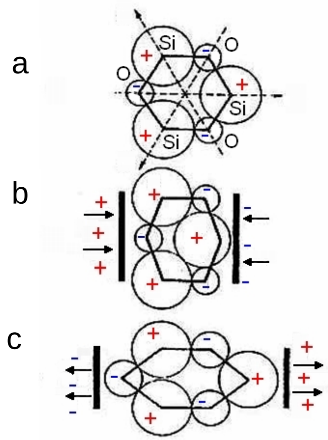
a) Center of charge of positive and negative charges coincides.

b) and c) ***direct*** piezoelectric effect

As a result of pressure , the charge centers are separated, i.e. a potential difference arises.

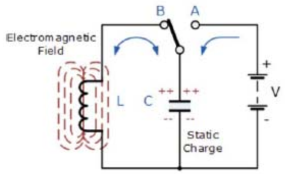
$\downarrow\downarrow$

Detection of US



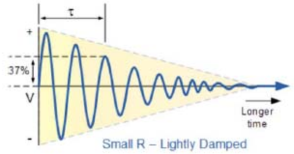
Source of electric signal : sine wave oscillator

LC circuit



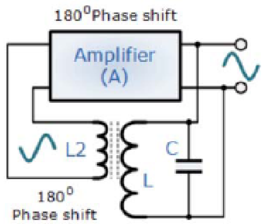
Damped Oscillations

$$f = \frac{1}{\sqrt{(2\Pi)^2 LC}}$$



Source of electric signal : sine wave oscillator

LC circuit with positive feedback amplifier

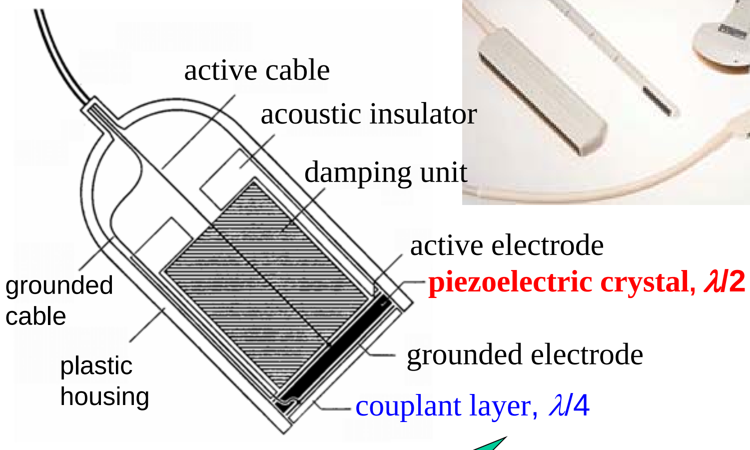


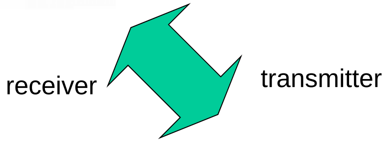
$$A_{U,\text{ feedback}} = \frac{A_U}{1-\beta A_U}$$

$A\tau\, \beta A_U$=1, amplification = „infinity“

output signal: sine voltage

Ultrasound transducer

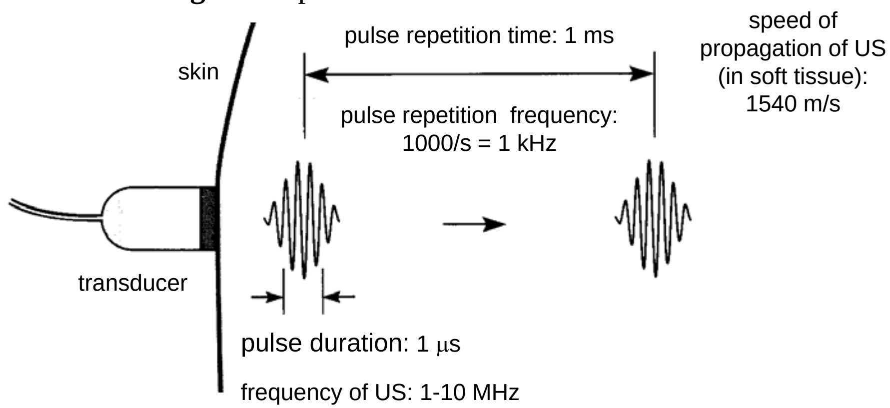




Characteristic of US pulses

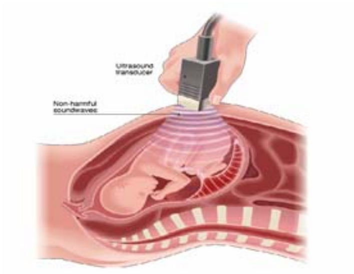
transducer: transmitter and receiver is the same unit

time sharing mode: pulses instead of continuous wave US



Textbook, Fig. VIII.32.

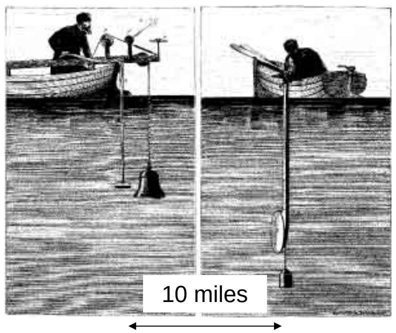
Ultrasound Imaging



Echo principle

1794 Spallanzani: bat's navigation

1822 Colladen measured the speed of sound in water



bottlenose dolphin

Reflection (normal incidence)

reflectivity:

$$\boxed{R = \frac{J_{\text{reflected}}}{J_{\text{incident}}} = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2}\right)^2}$$

boundary surface	*R*
muscle/blood	0.001
fat/liver	0.006
fat/muscle	0.01
bone/muscle	0.41
bone/fat	0.48
soft tissue/air	0.99

"full" reflection:

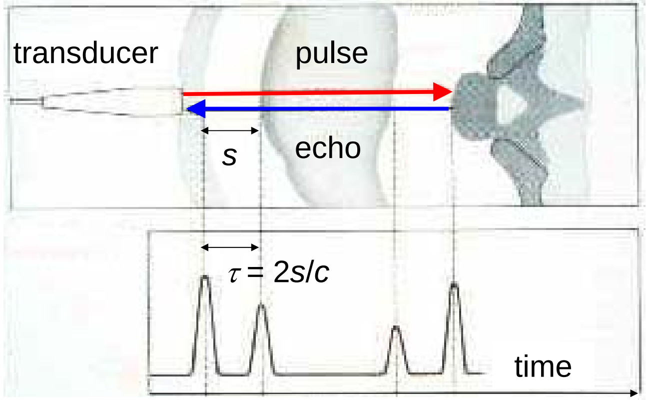
$$Z_1 << Z_2, \quad R \approx 1$$

optimal coupling:

$$Z_{\text{connecting}} \approx \sqrt{Z_{\text{source}} Z_{\text{skin}}}$$

Receiving the echos

A-mode
(*Amplitude*)
only 1-dimensional



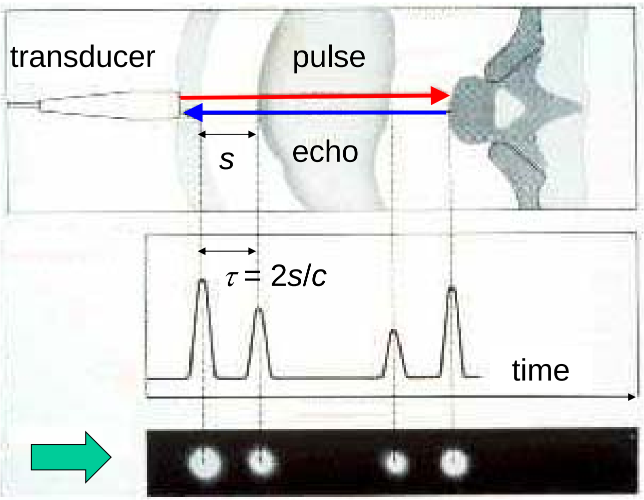
A display of ultrasonic echoes in which the horizontal axis of the cathode ray tube display represents the time required for the return of the echo and the vertical axis represents the strength of the echo.

cf. Textbook Fig. VIII.33

Receiving the echos

A-mode
(*Amplitude*)
only 1-dimensional

B-mode
(*Brightness*)
only 1-dimensional

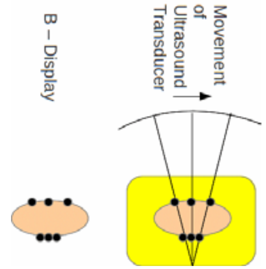


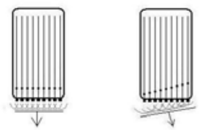
cf. Textbook Fig. VIII.33

2-dimensional B-mode – Ultrasound tomography

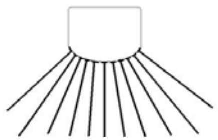
Several one-dimensional echo measurement

Multi unit arrays

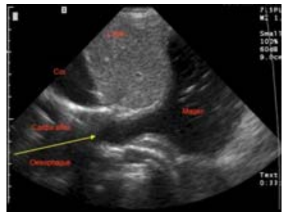




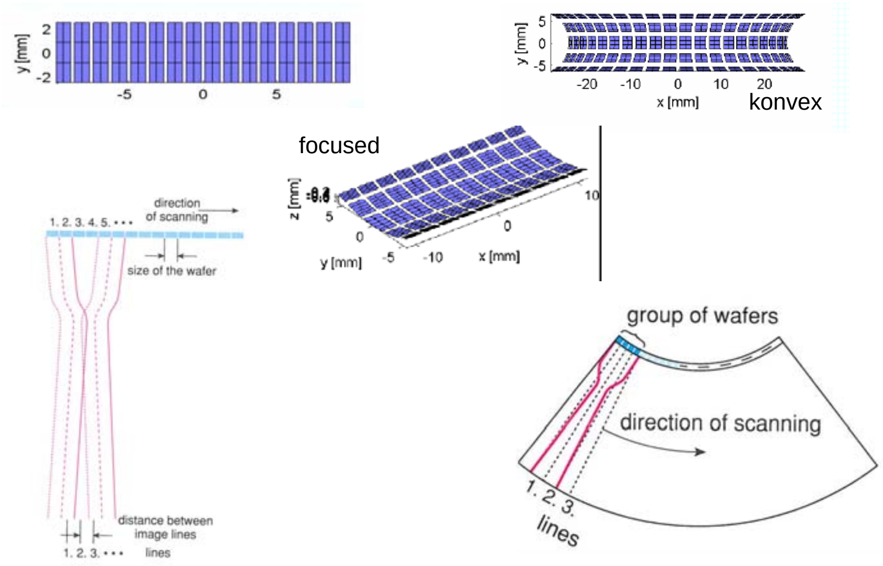
linear



curved

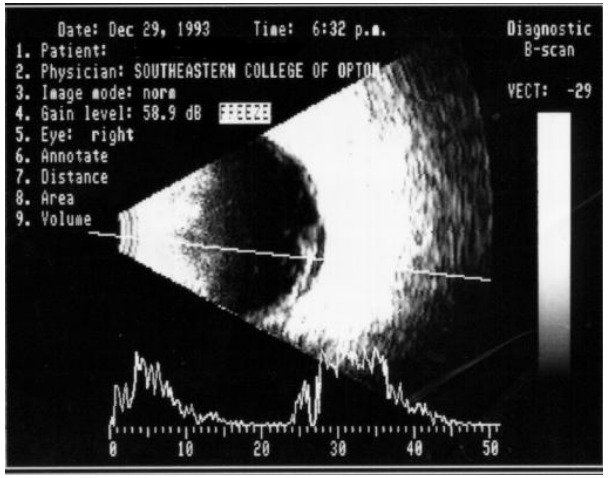


2-dimensional B-mode – Ultrasound tomography



2-dimensional B-mode and A-mode

(used in opthamology)



Real speed of propagation for the accurate determination of distances:

cornea: 1641 m/s

aqueos humour: 1532 m/s

crystalline lens: 1641 m/s

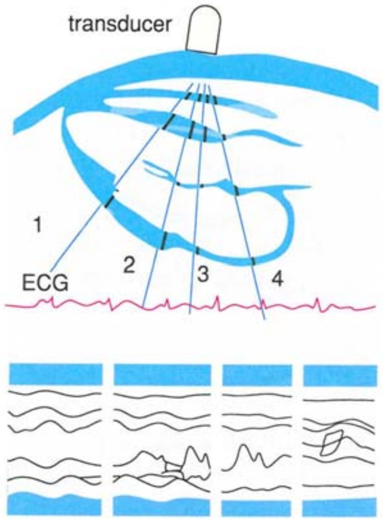
vitreous body: 1532 m/s

TM-mode

Motion of the reflecting surface is visualized

Typical in cardiology

ECG signal for reference



time

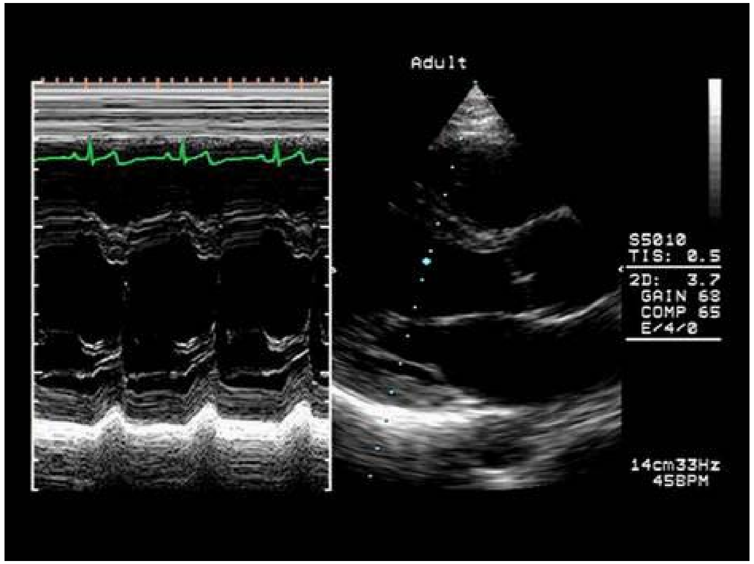
(vertical) time-dependent 1-dimensional B-mode

(T)M-mode

***T**ime–**M**otion*

Textbook Fig. VIII.34

TM-mode B-mode



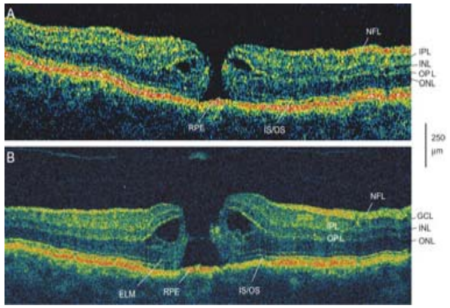
Resolving limit, resolution

Resolving limit is the distance between two object details which can be just resolved as distinct objects (the smaller the better).

Resolution (resolving power): the reciprocal of the resolving limit (the greater the better)

Axial resolving limit is the minimum separation between two interfaces located in a direction parallel to the beam so that they can be imaged as two different interfaces.

It depends on the pulse length. Pulse length is inversely proportional to the frequency.

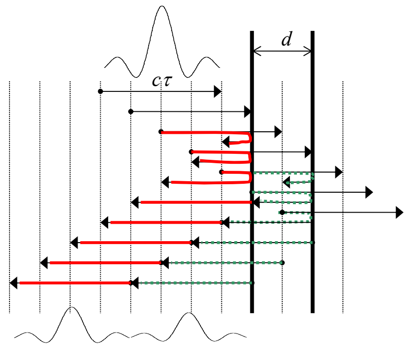


Axial resolving limit

τ : pulse duration

$c_1\tau \cong c_2\tau = c\tau$ pulse length

$\delta_{ax} = d = \frac{c\tau}{2}$ resolving limit

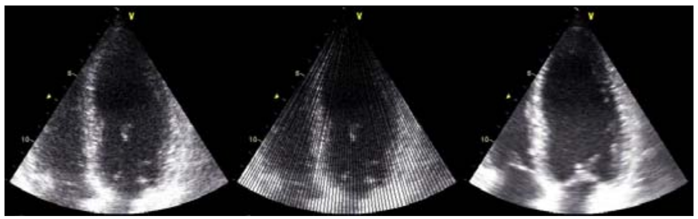


The axial resolving limit is the **half of the pulse length**. The echos from the adjacent surfaces in this case just hit another.

$$\tau \sim T = \frac{1}{f}$$

Resolving limit, resolution

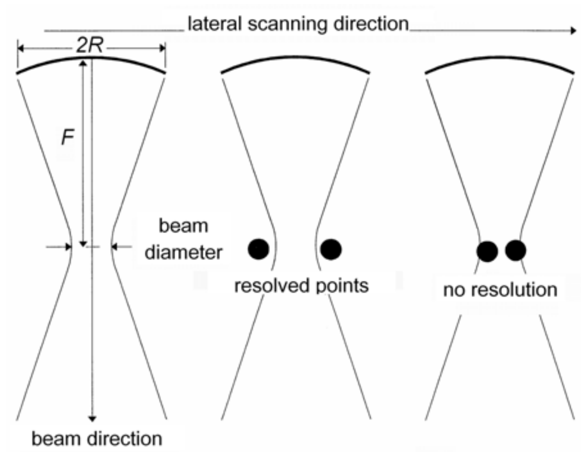
Lateral resolving limit is the minimum separation of two interfaces aligned along a direction perpendicular to the ultrasound beam. It depends on the beam width and beam density



Typical values

frequency (MHz):	2	15
wavelength (in muscle) (mm):	0.78	0.1
penetration depth (cm):	12	1.6
lateral resolving limit (mm):	3.0	0.4
axial resolving limit (mm):	0.8	0.15

Lateral resolving limit



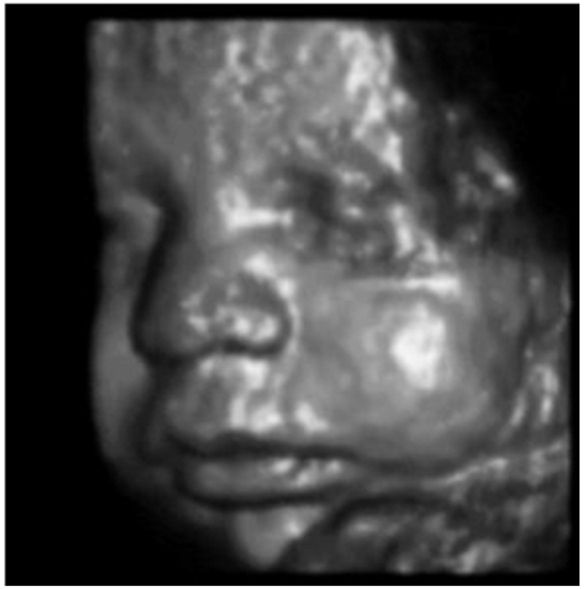
$$\left(\delta_{lat} \sim \frac{F}{2R} \cdot \lambda\right)$$

F: focal length

$2R$: diameter of the transducer

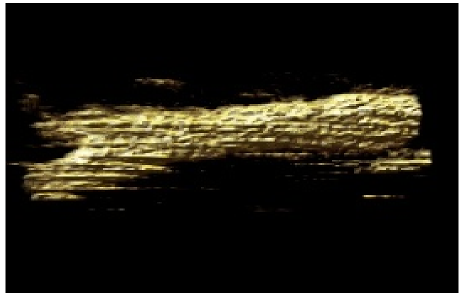
λ: wavelength

Reconstruction of the face of a fetus

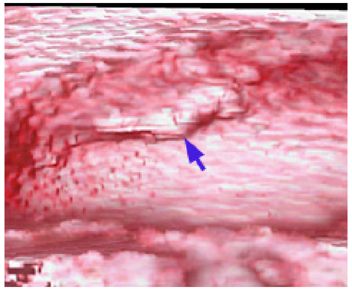


3D reconstruction

carotis

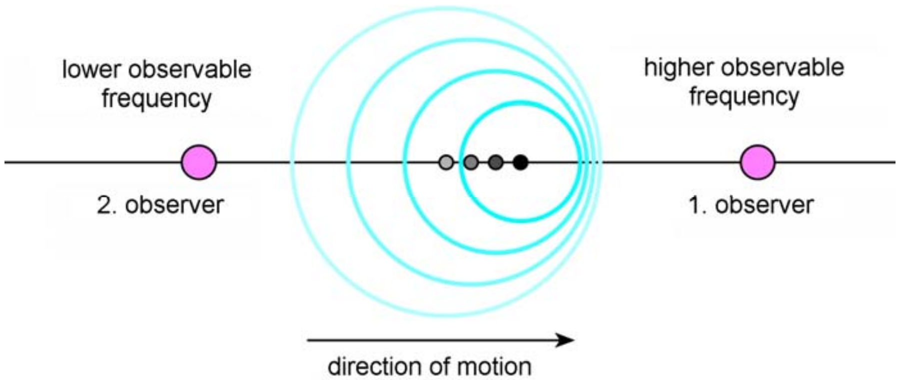


bladder



Doppler phenomenon

„The pitch of a train whistle seems to get higher as it approaches, then seems to lower as the train whistle moves away.” (C. Doppler, 1842)



Teextbook Fig. VIII.39

f': **observed frequency**, *f*: original frequency

(a) standing source and moving observer (v_O)
+: observer approaches the source
–: observer moves away from the source

$$\boxed{f' = f\left(1 \pm \frac{v_O}{c}\right)}$$

(b) moving source and standing observer
(if $v_S << c$, then „same” as (a))

$$f' = \frac{f}{1 \mp \frac{v_S}{c}}$$

(c) moving source and moving observer

$$f' = f\,\frac{1 \pm \frac{v_O}{c}}{1 \mp \frac{v_S}{c}}$$

(d) moving reflecting object (surface),
(if $v_R << c$)

$$\boxed{f' = f\left(1 \pm \frac{2v_R}{c}\right)}$$

Doppler frequency = frequency change = frequency shift

if v_i, v_R<<c (i= S or O)

rearranging equation (a)
moving source or observer:

$$\boxed{f' - f = \Delta f = f_D = \pm \frac{v_i}{c} f}$$

rearranging equation (d)
moving reflecting object or surface:

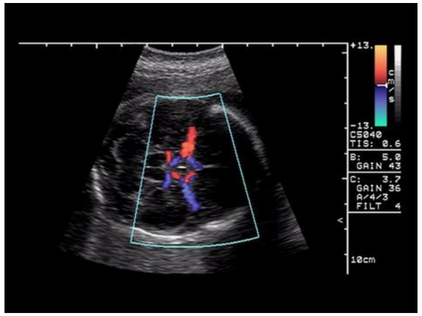
$$\boxed{\Delta f = f_D = \pm 2\frac{v_R}{c} f}$$

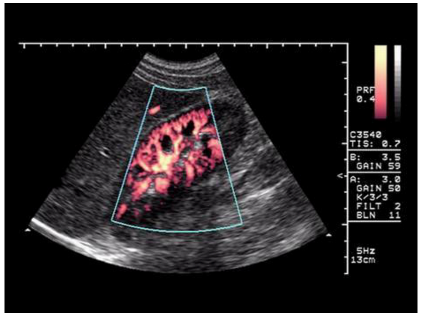
if v and c are not parallel, then $v \cos\Theta$ should be used instead of v (remark: if $\Theta = 90°$, $f_D = 0$)

Colour coding

Combination of two-dimensional B-image and Doppler

towards the transducer: warm colours
away from the transducer: cold colours





BART: Blue Away Red Towards

power Doppler

1-dimensional CW Doppler apparatus for measuring average flow velocity. Red blood cells as sound scatters

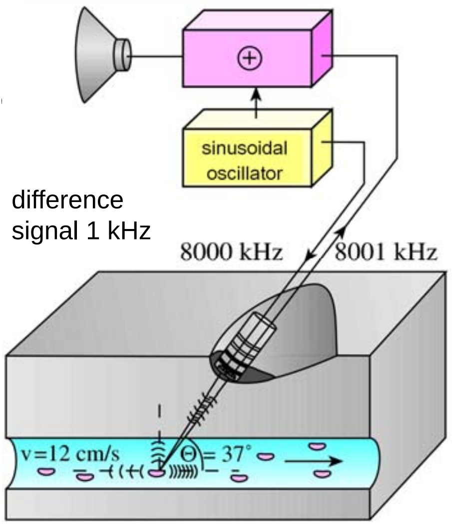
CW: continuous wave

source and detector are separate

$$|f_D| = 2\frac{v_R \cos\theta}{c} f$$

e.g. f=8000 kHz
v=12 cm/s
c=1600 m/s
Θ = 37°

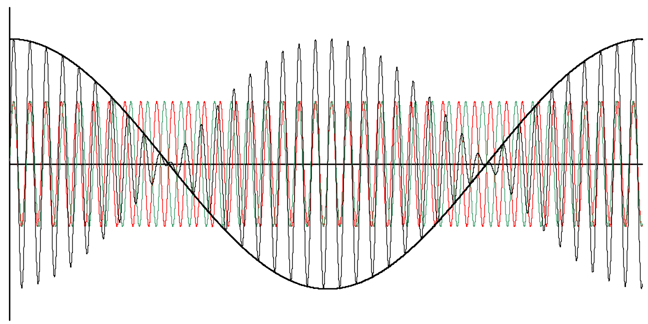
⇨f_D=1 kHz
(beating phenomenon)



Textbook Fig. VIII.41

Beating phenomenon

$f_{red} \geq f_{green}$ the beating frequency equals to the difference of the two interfering frequency



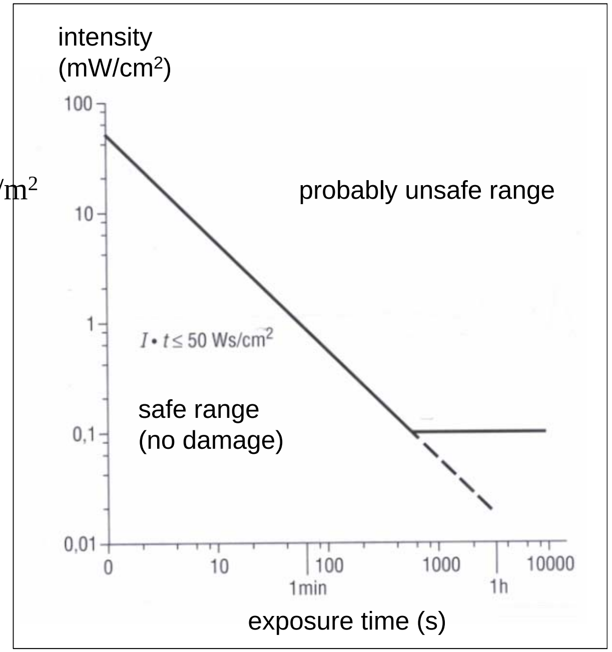
reminder: $\sin\alpha + \sin\beta = 2\sin\frac{\alpha+\beta}{2}\cos\frac{\alpha-\beta}{2}$

Safety

in the diagnostics:
10 mW/cm^2 = 100 W/m^2
cf. pain threshold: 10 W/m^2

in the therapy: 1 W/cm^2

spatial average temporal average (SATA) intensity;
spatial peak temporal peak (SPTP) intensity;
spatial peak temporal average (SPTA) intensity;
spatial peak pulse average (SPPA) intensity
spatial average pulse average (SAPA) intensity



Mechanisms of tissue damage

- Thermal Effect
  - Local heating produced by ultrasound wave is directly related to the intensity of ultrasound wave at any point in the medium.
  - The rate of increase in temperature has a direct relationship with ultrasound intensity and degree of absorption and is inversely proportional to tissue density and specific heat, *i.e.*

$$dT/dt = 2\alpha I/\rho C_m$$

  - Studies show that absorption are primarily related to the concentration of proteins.
  - In general, muscles (dense media) do not heat as fat.

Mechanisms of tissue damage

- Cavitation Effect:
  - Cavitation describe the formation, growth, and dynamic behavior of gas bubble irradiated by ultrasound.
  - In pure liquid, cavitation occur when the local pressure falls below the vapor pressure of a fluid and gas "boils"
  - Sound-induced oscillations of microbubbles causes gas to diffuse inward and outward during each cycle, because of pressure change inside the bubbles.
  - In water, a bubble resonating at 1 MHz with 100 mW/cm^2 can take 60 μW (90% of which convert to heat!)
  - It is estimated that 1 μm cavity collapsing in solid can create a local pressure of 1000 atm!
  - The internal temperature of a bubble could reach 1000 ^{0}C.
  - But, however, tissue viscosity is 100 times greater than water, and therefore, bubble motion is greatly limited.

Damjanovich, Fidy, Szöllősi: Medical Biophysics

II. 2.4.
VIII.4.2.