Physical Properties of Sound, Ultrasound, Doppler-principle, Medical Applications of Ultrasound

The text under the slides was written by

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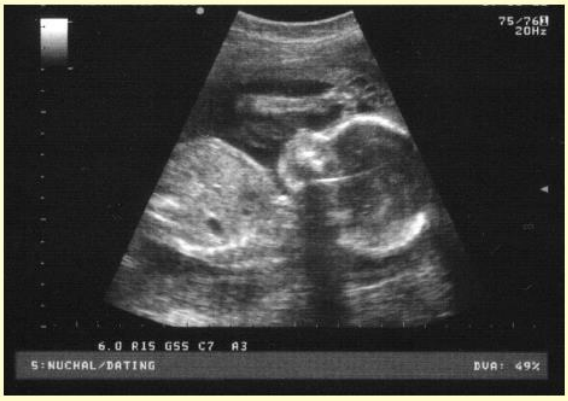


Related book chapters: Medical Biophysics, (Medicina, 3rd edition)

- II/2.4. Sound and Ultrasound (pp. 146-155.)
- VIII/4.2 Ultrasound Imaging – Direct Tomography 2. (pp. 504-521.)
- IX/5. Ultrasound-Therapy (pp. 554-555.)

How ultrasound (US) can be applied in medicine?

US imaging (ultrasonography):

- Application of low intensity US for diagnostic imaging
- Mostly soft tissues, muscles and the surface of bones can be visualized
- Determination of blood flow rate based on the Doppler-principle

US therapy:

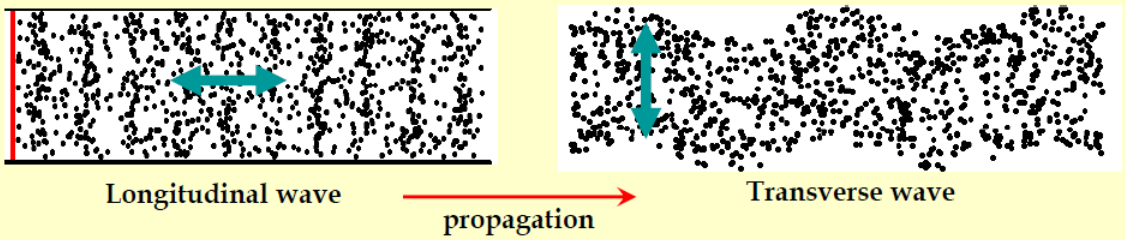
- Tumor ablation by local temperature increase induced by high intensity focused US
- Disintegration of kidney stones utilizing the mechanical effects of high intensity US

The aim of this lecture is to introduce the basic principles of the ***diagnostic*** and ***therapeutic*** use of ultrasound (US).

- Since both types of US applications are based on the interaction of US with the different tissues, first of all we have to understand the physical background of these interactions.
- The advantage of US diagnostics (ultrasonography) compared to other imaging techniques including conventional x-ray imaging or CT is that the low US intensities applied in ultrasonography do not cause physical, chemical or biological changes in the body tissues. Therefore, ultrasonography is considered a non-invasive medical imaging technique that can be applied repeatedly even for following the development of a foetus.
- In the majority of therapeutic applications several orders of magnitude higher sound intensities are used compared to ultrasonography. High intensity US may induce physical, chemical or biological changes in the tissues, thus it can be applied only locally and in a targeted manner to avoid the damage of the neighbouring healthy tissues.

Sound waves:

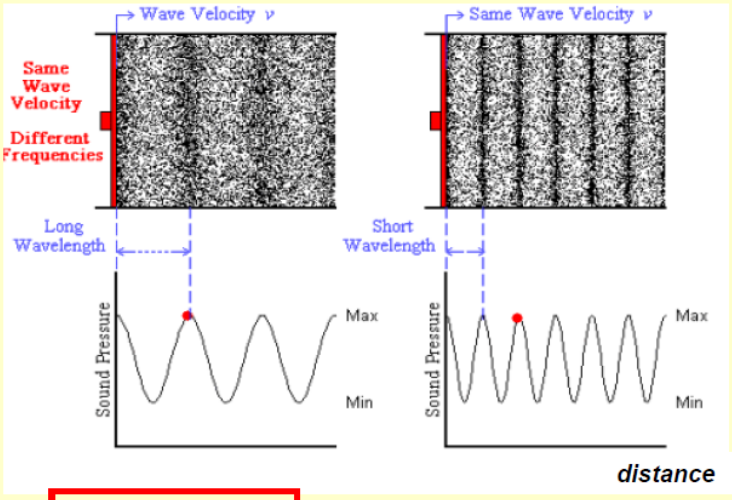
- mechanical vibration state of elastic media traveling as a wave ⇒ *mechanical wave*
- without medium ⇒ neither mechanical vibration nor its propagation is possible
- in gases and liquids – only *longitudinal sound waves* ***can occur***
  - oscillation of particles in the medium is parallel to the direction of propagation
  - density- and so pressure fluctuations along the direction of traveling
- in soft tissues sound propagates as a longitudinal wave
- in solids (and at the surface of liquids) – *transverse waves also can be formed*
  oscillation is perpendicular to the direction of propagation

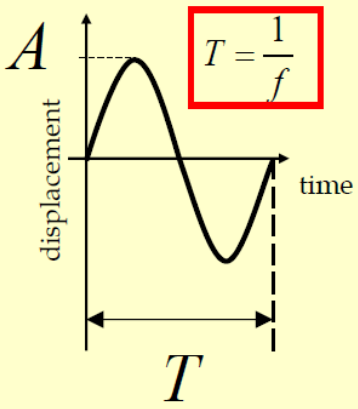


only the state of vibration propagates, no net particle transport!

- Sound is a ***mechanical wave*** that can propagate in elastic media as the mechanical vibrations of the particles. Only the state of vibration propagates, there is no net particle transport in the medium as a result of sound propagation. Thus it is important to realize that any individual particle in the medium is merely oscillating backwards and forwards even though the wave itself is actually moving through the medium away from the sound source.
- In accordance with the above, sound propagation requires a medium; in vacuum there is no sound propagation.
- Based on the phase state of a medium sound can propagate as a ***longitudinal*** or a ***transverse wave***. In gases and liquids only longitudinal sound waves can occur. In soft tissues sound propagates as a longitudinal wave. In solids and at the surface of liquids transverse waves also can be formed.
- As it is shown in the pictures in longitudinal waves the vibration of particles is parallel to the direction of wave propagation, while it is perpendicular to the direction of propagation in transverse waves.

Physical parameters of sound waves



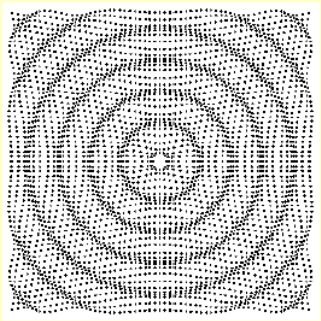


$$c = f\,\lambda$$

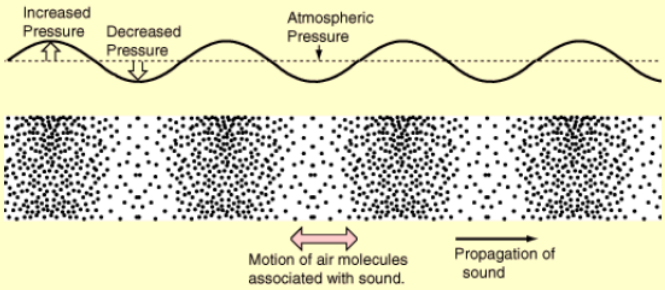
wavelength (λ): the distance measured in the medium between nearest points with identical values of particle displacement, pressure or density
frequency (f)**:** number of vibrations per second
speed of sound (c)
period (T): time required for **one complete cycle**
amplitude (A): maximum displacement of particles

- Waves can be described by various physical parameters as it was demonstrated for electromagnetic waves in the previous lectures.
- If we plot particle displacement, density or pressure (see the left figure) along the direction of sound propagation at some point as a function of time, it also forms a sine wave. ***Wavelength*** (***λ***) of sound wave is defined as the distance measured in the medium between nearest points with identical values of particle displacement, pressure or density. ***Frequency*** (***f***) is the number of vibrations per second. The ***speed of sound propagation (c)*** is determined by the composition of the medium. In a particular medium wavelength and frequency are inversely proportional to each other and their product is constant, and equal to the speed of sound propagation as it is expressed by the 1st equation.
- ***Period*** (***T***) is the time required for a complete vibration (complete cycle). The ***amplitude*** of the vibration (***A***) is the maximal displacement/range of the particles from their resting state upon the vibrations (right side figure).
- If the frequency of such a wave is ***f***, then it must be executing ***f*** complete cycles per second, therefore the period T of each will be given by ***1/ f***.

Point-like sound source (2D projection)



Propagation of a sound wave (e.g. in air)



<u>pressure differences as a periodic function of time and position:</u>
in the simplest case (harmonic oscillation)

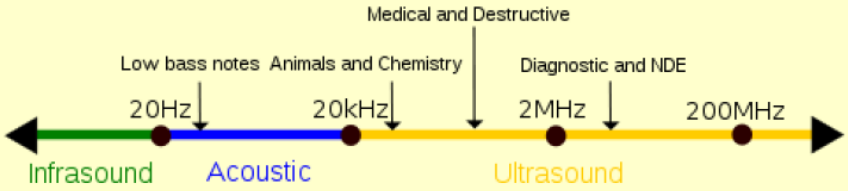
$$\Delta p(t,x) = \Delta p_{max}\, sin\left[2\pi\left(\frac{t}{T} - \frac{x}{\lambda}\right)\right]$$

Between displacement and pressure amplitudes – π/2 phase difference
presssure changes (Δp); pressure amplitude (Δp_{max})

- As it is demonstrated by the figures the disturbance or displacement introduced by the vibrating sound source is passed on through an elastic medium as a series of compressions and rarefactions. This means that both the density and the pressure fluctuate compared to their steady state values to which they will return when the wave has passed.
- For the description of sound-waves we can use the pressure difference ($\boldsymbol{\Delta p_{t,x}}$) as a periodic function of time (***t***) and position (***x***)**.** In case of harmonic oscillations $\boldsymbol{\Delta p_{t,x}}$ can be given by a simple sine function, where *T* is the period and $\boldsymbol{\lambda}$ is the wavelength of the sound wave, while $\boldsymbol{\Delta p_{max}}$ is the pressure amplitude, i.e. the maximal change of pressure.

Classification – by frequency

infrasound< audible (by humans) sound (acoustic region) < ultrasound < hypersound



hypersound: 10^{9}-10^{10} Hz ↔ 10^{12} – 10^{13} Hz

- medical imaging: typically 2-10 MHz ultrasound
- therapeutic ultrasound – lower frequencies (but higher intensities)

- Sound waves can be classified on the basis of their frequency (pitch). The frequency range of human hearing is between 20 Hz and 20000 Hz = 2×10^{4} Hz = 20 kHz and sounds of this frequency range are called ***audible sounds***. The higher frequency sound waves are called ***ultrasounds*** (***US)***, while sound waves having lower frequencies are called ***infrasounds***. The upper frequency limit of the US range is usually taken to be a few hundred MHz. Sounds having frequencies higher than this are called ***hipersounds***.
- It is worth noting that in diagnostic sonography typically the 2–10 MHz frequency US is applied, while in therapeutic applications lower frequencies, but much larger US intensities are used.

Speed of sound

- independent of its frequency
- **depends on density (ρ)** and **compressibility (κ)** of the medium

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

- *Compressibility*: relative volume decrease caused by a unit increase of pressure:

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

Soft tissues: ~1540 m/s
water: ~1500 m/s
air: ~330 m/s

Values of compressibility, density, and speed of sound in biological materials

Material	Compressibility (κ) $\times 10^9\ ms^2kg^{-1}$	Density (ρ) $10^3\ kgm^3$	Speed (c) ms^{-1}
Aluminum	0.009	2.70	6400
Bone	0.08-0.05	1.38-1.81	3050-3500
Liver	0.38	1.06	1570
Kidney	0.40	1.04	1560
Blood	0.38	1.06	1570
Fat	0.51	0.92	1460
Lung	5.92	0.40	650
Air	7650	1.2×10^{-3}	330

- The frequency (f) of sound is constant upon propagation, thus its wavelength (λ) also changes with c depending on the properties of the medium ($c=f\lambda$)
  - Medical ultrasound: f = 2-10MHz ⇒ in tissues: λ = 0.77-0.154mm

- The ***speed*** (***c,*** its unit is m/s) at which the sound wave propagates through a particular medium is determined by the ***density*** (***ρ;*** kg/m^3***)*** and ***compressibility*** (***κ;*** m×s^2×kg^{-1}**)** of the medium as it is expressed by the equation in a red rectangle.
- ***Compressibility*** is a measure of how easy it is to compress or deform the medium. It is given as a relative volume decrease caused by a unit increase of pressure.
- The density and compressibility values change with the phase state. Solids have about three orders of magnitude higher densities compared to gases, however they are relatively incompressible (***κ*** is more than three orders of magnitude lower), and therefore possess high values of ***c*** (~3000-6000 m/s). It can be seen from the table that the values for speed of sound in different soft tissues (e.g. liver, fat, kidney tissue) are very similar and quite close to the value measured in water (1500 m/s). ***For the speed of sound propagation in soft tissues we use the 1540 m/s value, which is a reasonable approximation in most cases.***
- The frequency does not change when a sound wave travels from one medium to another. The 2-10 MHz US frequencies generally applied in diagnostic sonography assuming that ***c*** is 1540 m/s in soft tissues imply that the wavelength values in soft tissues are in the range of 0.77-0.154 mm. The above wavelength values were calculated substituting the appropriate ***c*** and ***f*** values into the following equation: (***λ=c/f → 1540m/s : 2×10^{6}1/s=0.00077m=0.77mm***). The wavelengths of the diagnostic US are comparable to the size of larger cells or groups (islets) of cells.

Acoustic impedance (Z)

- unit: $kg \cdot m^{-2} \cdot s^{-1}$
- „resistance" of the medium against bringing into motion its particles
- ratio of sound pressure (relative pressure!) and the velocity of the particles (v)

$$Z = \frac{p}{v} \quad (1.)$$

$$p = vc\rho \quad (2.)$$

$$Z = c\rho \quad (3.)$$

$$c = \frac{1}{\sqrt{\rho\kappa}} \quad (4.)$$

- Z is a material constant

$$Z = \sqrt{\frac{\rho}{\kappa}} \quad (5.)$$

Values of Acoustic Impedance forBiological Materials

Material	Acoustic Impedance $10^6 kg\ m^{-2}\ s^{-1}$
Aluminum	17.28
Bone	7.80
Liver	1.65
Kidney	1.62
Blood	1.61
Fat	1.38
Lung	0.26
Air	0.00004

- ***Electrical impedance*** is the measure of the opposition that a circuit presents to a current (i.e. movement of electrons) when a voltage is applied. Analogously, ***acustic impedance*** (***Z***) is defined as the resistance of the medium against bringing into motion its particles. It is an important property of a medium from the point of sound propagation. Acustic impedance is given as the ratio of sound pressure **(*p*)** and the velocity of the particles **(*v*)** as it expressed by eq. 1.
- Substituting eq. **2** into eq. **1** acustic impedance can be expressed as a product of the sound speed (***c***) in a particular medium and the density (***ρ***) of this particular medium as it is shown by eq. 3.
- Summarizing eqs. 3 and 4 we can express ***Z*** as a function two parameters density (***ρ***) and compressibility (***κ***) that are characteristics of the medium. Since both parameters are material constants ***Z*** is indeed a material constant.
- We can conclude from the table shown here that the acustic impedance of solid materials is much larger compared to soft tissues, while the acustic impedance of air is five orders of magnitude lower.

Intensity of sound

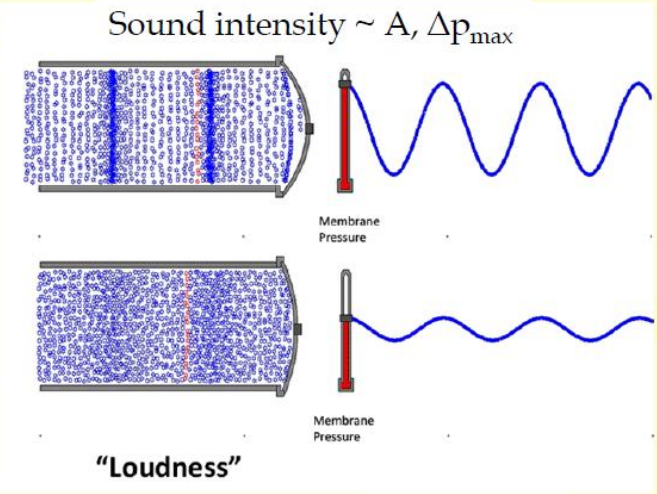
- **intensity (*J or I* [W/m^2]**

energy passing through a unit area perpendicular to the direction of sound propagation in unit time:

$$J = \frac{1}{2Z}\Delta p_{max}^2 \qquad J = \frac{1}{2}\rho v \omega^2 A^2$$

$$\Delta p_{max} = \sqrt{2ZJ}$$

A: displacement amplitude;
Δp_{max}: pressure amplitude



- **ultrasound intensities in medicine**

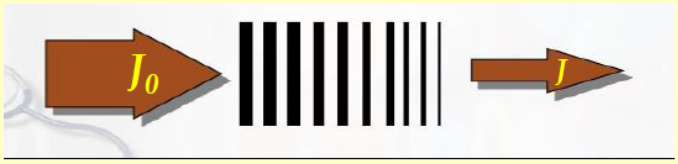
Diagnostic imaging: ~10^{-2} W/cm^2;
e.g. 10^{-1} W/cm^2 max. intensity for 1MHz diagnostic instruments (FDA)

Therapeutic applications: 0,1-10^5 W/cm^2

- Sound intensity ***(J)*** is defined as the amount of energy passing through a unit area perpendicular to the direction of sound propagation in unit time. Its unit is Watt/square metre (W/m^2). Sound intensity is proportional to the amplitude (exactly the square of the amplitude ~A^2) of the vibrating particles. Higher intensity sound induces larger amplitude vibrations of the particles upon its propagation and thus it induces larger pressure changes in the medium. This is shown in the animation/picture by the larger deformation of the membrane at the end of the tube in the direction of sound propagation.
- For US imaging (sonography) low US intensities are applied, which do not cause tissue damages. However, for therapeutic purposes (e.g. tumour ablation, disintegration of kidney stones) up to seven fold higher US intensities are used, therefore the neighbouring healthy tissues may also be damaged, if the targeting of the malformation is not appropriate.

Interaction of sound with the medium

I. Absorption



- dissipation of US energy primarily as heat (friction) ~ 90% of attenuation
- for sound/US travelling in a parallel beam: $J = J_0 e^{-\mu x}$
  absorption coefficient (μ); thickness (x)
- absorption capacity depends on
  → the quality of the medium
  → the frequency (f) of sound – in the diagnostic US-range: $\mu \sim f$
- half-value thickness: $x_{1/2} \rightarrow J = J_0/2$

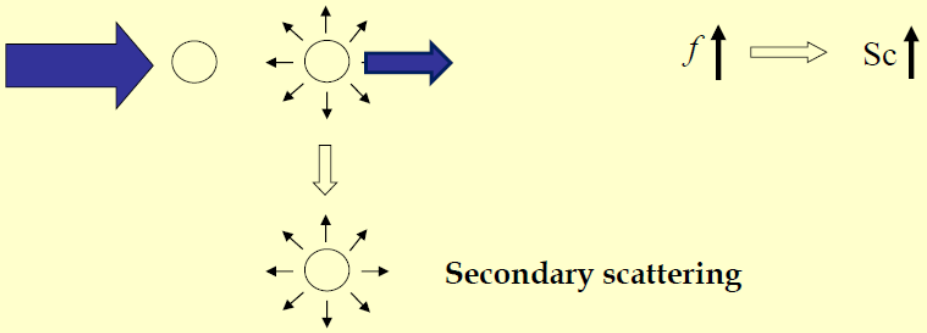
Thickness of various materials required reducing the intensity by half (half-value thickness).

Material	Half-value thickness (cm)	
	2 MHz	5 MHz
Air	0.06	0.01
Bone	0.1	0.04
Liver	1.5	0.5
Blood	8.5	3.0
Water	340	54

- Propagation of sound brings the particles of the media into motion causing friction between the vibrating particles and resulting in dissipation of a certain fraction of the energy in the form of heat.
- In the case of sound travelling as a parallel beam the general exponential law describing the attenuation of radiations observed in media holds with a good accuracy. We will detect ***J*** sound intensity after a sound beam of J_0 initial intensity passes through an ***x*** thickness of a medium with ***μ*** attenuation coefficient.
- Besides the material properties of the medium, the absorption capacity also depends on the frequency of the sound wave. In the frequency range applied in US diagnostics the absorption coefficient is proportional to the frequency (**μ~*f***), i.e. absorption increases with the increase of sound frequency. It is obvious, that the larger the frequency of the vibrations is, the larger the energy loss is within a given distance in a medium, since more oscillations are taken by the particles in a unit time leading to more frictional energy-loss. The table demonstrates that increasing the frequency of US results in a decrease of the ***half value thickness*** (***$x_{1/2}$***: the width of the medium which decreases the sound intensity by half).
- Although, increase of the US frequency improves the axial resolution in US diagnostics as we will see later, increasing the frequency decreases the intensity of the echo signal.

II. Scattering

→ alteration of the direction of sound waves
→ intensity-decrease along the direction of propagation



In practical applications: correction term is added to the absorption coeffcient

$$\mu = \mu_{abs} + \mu_{sc}$$

- ***Scattering*** is the other mechanism that can be responsible for the decrease of sound intensity in the direction of its propagation. This happens when a sound wave interacts with objects that are small or rough on a scale comparable to the wavelength**.** Most of the energy passes beyond the single scatterer without change, while a small fraction of the energy is redirected (scattered) in almost all directions. Thus scattering results in an intensity decrease in the direction of sound propagation.
- Therefore the attenuation coefficient (***μ***) is given by the sum of the absorption coefficient (***μ$_{abs}$***) and the scattering coefficient (***μ$_{scattering}$***).
- Scattering also contributes to the image formation in ultrasonography. Typical images from liver, kidney and thyroid tissues are composed largely of scattered echoes produced by the islets of cells in these organs.

III. Phenomena at the boundary of different media

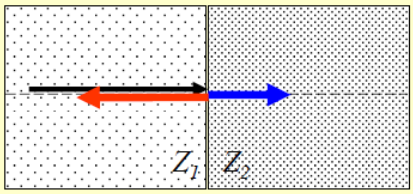
- at the boundary of media with different acoustic impedances → reflection
- different speed (usually Z is also different)→ refraction

Reflection

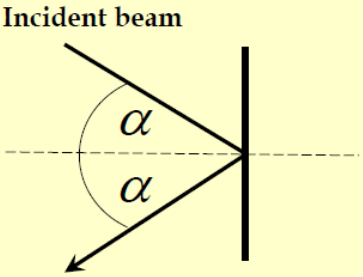
- reflexivity (R): ratio of reflected (J_R) and incident intensity (J_0)

$$R = \frac{J_R}{J_0} \qquad R = \left(\frac{Z_1 - Z_2}{Z_1 + Z_2} \right)^2$$

- if the difference between Z_1 and Z_2 is large → $R \approx 1 \Rightarrow$ full reflection



Perpendicular incidence



Skew incidence

- When a sound wave reaches the interface of two media it can be ***reflected*** (partially or completely) or ***transmitted*** to the second medium.
- ***Reflection*** of sound waves at a surface occurs when the acustic impedances of the two neighbouring media are different. If a surface (interface) is large compared to the wavelength and relatively smooth it can reflect sound waves in a manner identical to that of light striking a mirror (specular reflection). The extent of reflection is described by the reflection coefficient (reflexivity, ***R***) which is the ratio of the reflected (J_R) and the incoming (J_0) sound intensities. The reflexivity of an interface is determined by the change of the acustic impedance (***Z***) across the interface, thus ***R*** is expressed as the difference of the acustic impedances of the two media (Z_1 - Z_2) normalized to the sum of the acustic impedances (Z_1 + Z_2).
- The angle of incidence equals the angle of reflection, thus in case of perpendicular incidence the reflected beam returns to the source as it is shown in the left figure (*red arrow*).

III. Phenomena at the boundary of different media: Reflection

Values of intensity reflection coeffitient (R) for biological interfaces.	
Muscle/Blood	0.0009
Fat/Liver	0.006
Fat/Muscle	0.01
Bone/Muscle	0.41
Bone/Fat	0.48
Soft tissue/Air	0.99 !!!

Medical US applications:

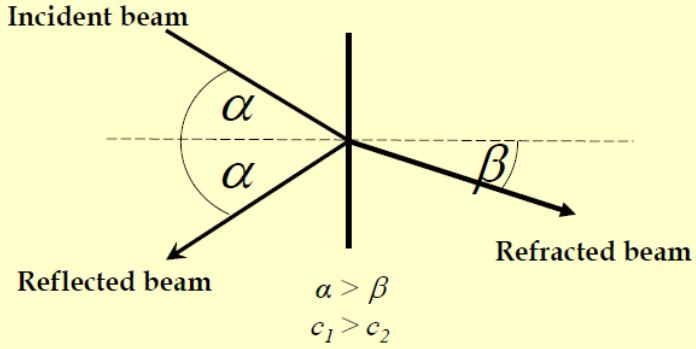
– Connecting medium should be used between the US source and the body (gel; water)

– Bone shadow, stone shadow, etc.

– Echo signal generally is rather weak ← small differences in Z values

- Acustic impedances of the different soft tissues (e.g. liver, kidney, fat etc.) are very close to each other resulting in weak reflections (echoes) from their interfaces. However, the weak echoes formed at their interfaces may be sufficient for their visualization in US imaging.
- If Z largely differs between the two sides of the interface such as in case of the soft tissue/bone interface a large fraction of the sound intensity is reflected, therefore we obtain a very strong echo signal from the surface of bones, while we cannot get echoes from the tissues behind them. This phenomenon is called ***„ultrasonic shadowing”***, see also later.
- We have practically complete reflection from the air/soft tissue interface. Consequently a connecting medium (e.g. gel or water) should be used between the US source and the body surface. The acustic impedance of the connecting medium is close to the acustic impedance of the body tissues facilitating the entry of US into the body.

III. Phenomena at the boundary of different media: Refraction



Snellius-Descartes law:

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

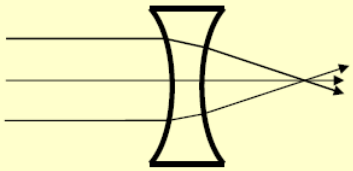
$Z = \rho c$, so if $\alpha > \beta$

⇩

$c_1 > c_2$

⇩

$Z_1 > Z_2$ (if ρ is similar)



- „acoustic lenses" for focusing the US

  acoustic lenses are made of solid materials, where the sound propagates at higher speed compared to the surrounding medium

- At an interface the sound wave entering the second medium may change direction, which is called ***„beam bending"*** or ***refraction***. The amount of bending is described by the Snell's Law similarly to optics: the ratio of the angles of incidence and refraction is equal to the ratio of the sound speeds in the two neighbouring media (***c_1/c_2***). Relatively large changes in the sound velocity are required for significant refraction; therefore refraction is likely to become important where there are also large changes in the acustic impedance.
- In US diagnostics refraction may contribute to image artefacts.
- The phenomenon of refraction can also be utilized for focusing the US. The so-called ***acustic lenses*** are used for focusing the sound waves. They are made from solid materials wherein sound propagates at higher speed compared to the surrounding medium, consequently entering into the lens ***c_1*** is less than ***c_2*** and thus the sound beam is bended away from the normal line. Conversely, leaving the acustic lens ***c_1*** is larger than ***c_2*** thus the sound beam is bended towards the normal line leading to focusing.

Formation and propagation of sound waves

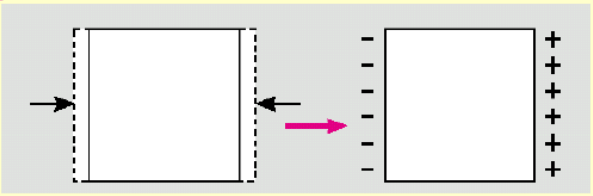
• **source** – vibrating object, which is capable to bring the particles of the **medium** into motion; frequency of vibration ⇒ frequency of sound
• propagation of sound wave – via the interactions among the particles in the medium (deformation of the medium)
• „back and forth" conversion of kinetic and potential energy into each other

Generation of ultrasound:
⇒ inverse piezoelectric effect
⇒ electrostriction
⇒ magnetostriction

- Vibrating objects are used for generation of sound waves. The frequency of the vibrating object determines the frequency of the emitted sound.
- For generation of high frequency US suitable for medical applications we can utilize the ***inverse piezoelectric effect***, ***electrostriction*** or ***magnetostriction***.

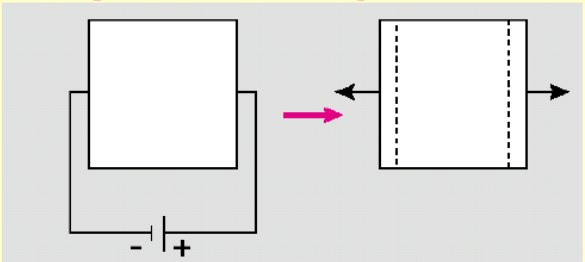
Generation and detection of US: Piezoelectricity

piezoelectric effect – detection of US



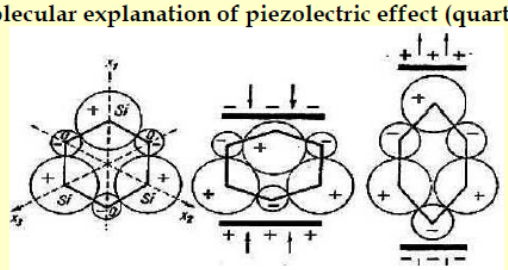
Applied Stress Induced Voltage

inverse piezoelectric effect – generation of US

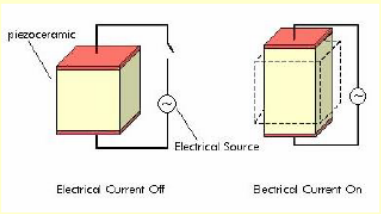


Applied Voltage Induced Stress

Molecular explanation of piezolectric effect (quartz)



resting state compression extension



• piezoelectric materials: typically crystals (quartz, topaz, cane sugar etc.), certain ceramics (e.g. Lead Zirconate Titanate - PZT), biological materials (DNA, bones, some proteins)
• direct piezoelectric effect: conversion of mechanical stress into voltage (charge separation!)
periodic deformation of the piezoelectric material due to pressure fluctuations carried by US ⇒ detection of US
• inverse piezoelectric effect: conversion of voltage into mechanical stress
alternating voltage ⇒ mechanical vibration at the same frequency ⇒ generation of US

- In ***piezoelectric materials*** (e.g. quartz, Lead Zirconate Titanate (PZT) etc.) mechanical deformation induces charge separation and thus voltage difference. This phenomenon (***direct piezoelectric effect***) can be utilized for the conversion of mechanical deformation into electric voltage and thus for the detection of US.
- The above phenomenon is reversible (***inverse piezoelectric effect***): electric field causes mechanical deformation (shape change) of the crystal. If we apply alternating voltage with a certain frequency to the crystal we will experience periodic change in the shape of the crystal with exactly the same frequency. If the frequency of the applied voltage is in the US frequency range the crystal emits US into the neighbouring medium.
- ***Molecular explanation for the piezoelectric property of quartz:*** In resting state the centres of positive and negative charges coincide with each other in the quartz crystal consequently the surfaces of the crystal are electrically neutral. Compressing the crystal along the axis of the hexagon brings the Si and O atoms closer to each other. As a result of compression the charge centres become separated, therefore on one side of the crystal the negative charges of the oxygen atoms dominate, while on the opposing side the positive charges of the silicon atoms show up arising a potential difference. Taken together, mechanical deformation of the quartz crystal induces charge separation i.e. electric voltage difference.

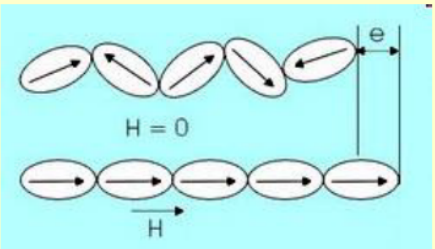
Generation and detection of US

Electrostriction

• elastic deformation of dielectric materials in an electric field
• arrangement of electric dipoles ⇒ increases the thicknes in the direction of the applied electric field and reduces the thickness perpendicular to it
• direction of deformation is independent of the polarity of the field, but the extent of it depends on the strength of the field ⇒ alternating electric field causes fluctuations in size
• *non-reversible* – mechanical deformation doesn't induce electric field

Magnetostriction (Joule-effect)

• change in the dimensions of ferromagnetic materials (e.g. iron, nickel) in a magnetic field due to the changes in the direction and extent of its magnetization
• alternating magnetic field ⇒ fluctuation in size
• *can be reversed* - inverse magnetostriction (mechanical deformation → change in magnetic properties)
• magnetostriction transducers could transmit and receive high-frequency sound vibrations



- ***Electrostriction*** is a property of all dielectric materials. Dielectrics are dipole molecules that do not conduct electricity. Electrostriction is defined as the elastic deformation of a dielectric material under the force exerted by an external electric field. It is caused by displacement of dipoles in the crystal lattice upon being exposed to an external electric field. This displacement will accumulate throughout the bulk material and result in an overall strain (elongation) in the direction of the field, while the thickness will be reduced in the orthogonal directions. Reversal of the electric field does not reverse the direction of the deformation. However, the extent of deformation depends on the strength of the applied electric field. Thus applying an electric field with oscillating strength can be used for US generation if the frequency of the oscillations is in the range of the US frequency.
- ***Magnetostriction* (*"Joule-effect"*)** is a change in the dimensions (shape) of ferromagnetic materials (e.g. iron, nickel) in a magnetic field due to the changes in the direction and extent of its magnetization. Similarly to electrostriction we can experience elongation in the direction of the applied magnetic field and decrease in the thickness perpendicular to it, as you can see in the figure. The reciprocal effect, the change of the magnetic properties of a material when subjected to a mechanical stress, is called ***„Villari effect"***. Magnetostriction is applied for US generation, while the inverse magnetostriction is used for detection of US.

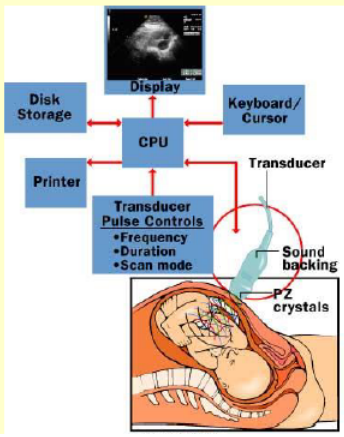
Diagnostic sonography

1. **Generation of ultrasound**
   piezoelectric transducers – inverse piezoelectric effect
   pulse *vs.* **continuous** techniques

2. **Interaction of US with human tissues**
   „The good, the bad and the ugly …"
   reflection, scattering → signal (echo)
   absorption, reflection, scattering, refraction → attenuation of US
   refraction → artifact production

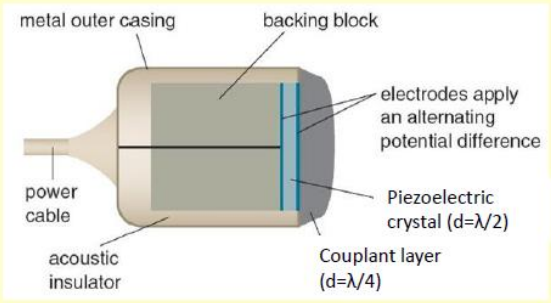
3. **Detection of the echos**
   piezoelectric transducers – piezoelectric effect
   intensity, arrival time and frequency of the echos

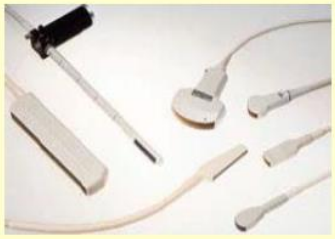
4. **Data processing → image formation**
   amplification
   displaying the signal/image formation
   spatial/temporal resolution of the image

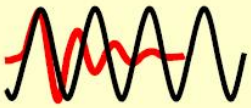


1. In ultrasonography (US imaging) we often apply the so-called ***pulse-echo methods***, when a short US pulse is emitted into the body and subsequently the reflected echoes are detected to gain information about the position and sometimes also the structure of the echoing surfaces. To allow echoes from closely spaced interfaces to be resolved separately, the US pulse must be short. A typical ultrasound pulse consists of a few cycles (2-3 cycles) of oscillations. In pulse-echo methods very often the same transducer is used for generating US pulses and detecting echoes in an alternating manner.
2. In pulse-echo methods the most important interactions between the US and the medium from the point of image formation are reflection and in some soft tissues scattering (see earlier). Absorption decreases the intensity of the echo signals, while scattering may also contribute to artefacts related to the false positioning of certain tissue interfaces.
3. In most cases the piezoelectric effect is used to generate the US pulses.
4. As a result of data processing an image is formed. The time required for receiving the echo signal can provide information about the distances of the echoing surfaces from the transducer, since the speed of US propagation in the soft tissues is known (1540 m/s). The intensities (amplitudes) of the echo signals depend on the absorption coefficient of the tissues in the pass of the echoes and the distances of the echoing surfaces from the transducer. Upon image formation US signals from deeper tissues are amplified in a greater extent (this method is called Time Gain Compensation) in order to compensate for the gradual decrease of echo intensity with increasing depth.

Ultrasound generation







Continous mode
pulse mode

US sources=transducers: convert one type of energy into another one
- Continous or pulse mode
- Pulse mode: the amplitude and the frequency values are not constant

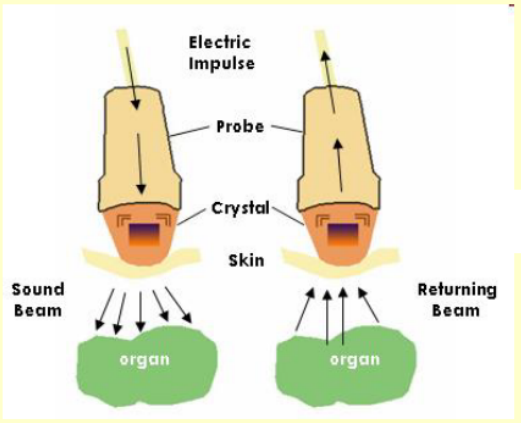
The major components of transducers:

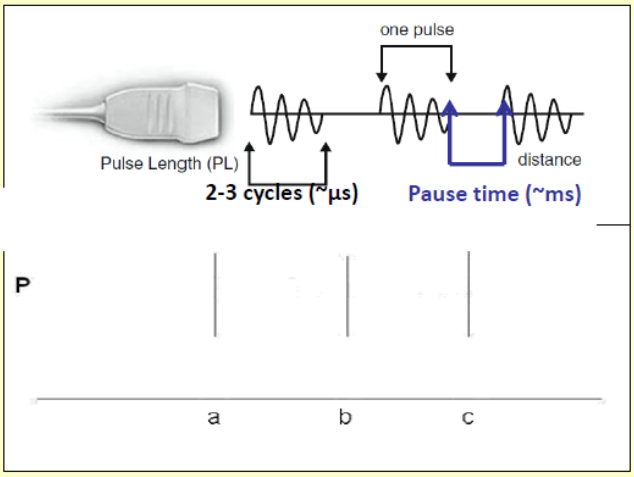
- **Piezoelectric wafer (crystal):**
  - thickness of the piezoelectric wafer = half of the US vawelength (few hundreds of μm)
- **Backing block/damping unit**:
  - large absorption coefficient → efficient absorption of US in the reverse direction
- **Couplant layer:**
  - facilitates the transmission of US from the transducer to the human tissues: ($Z_{couplant}=\sqrt{Z_{source} \times Z_{skin}}$)
  - Sound vawes reflected from its inner surface should be in phase with the original vawe (d=λ/4)

- US sources are called transducers, because they can convert one type of energy to another type (electric energy↔mechanical energy).
- The actual sound-generating and detecting component of the transducers is a thin ***piezoelectric plate/crystal.*** In ideal cases the thickness of this piezoelectric plate is equal to the half of the wavelength of the US pulse to be generated, which is in the order of some hundreds of micrometres in case of diagnostic US.
- Behind the piezoelectric plate there is a ***damping unit*** made from a large absorption coefficient material. Its function is to absorb the vibrations propagating in the reverse direction. In addition, after switching off the alternating voltage the damping unit stops the vibrations of the piezoelectric plate and thus it helps us to produce short US pulses (you can see the shape of a short pulse compared to a continuous wave in the right figure).
- The ***couplant layer*** is part of the housing of the transducer and protects the thin piezoelectric plate from mechanical damage. On the other hand, it helps the transmission of the US pulses into the human body. The ideal thickness of the couplant layer is one fourth of the emitted US wavelength, thus the sound wave reflected back from its inner surface will be in the same phase with the original wave (constructive interference). In ideal cases the acustic impedance of the couplant layer ($Z_{couplant}$) is equal to the geometric mean of the acustic impedances of the piezoelectric crystal (Z_p) and the body tissues (Z_t): $Z_{couplant}=\sqrt{Z_p \times Z_t}$

Diagnostic sonography

Pulse-echo methods I





• emission of a short US pulse → transmission and reflection in the body → return and detection of the echo (amplitude and time lag, frequency shift) → signal amplification and processing → forming and displaying the image (e.g. on a cathode ray tube)

• *necessary pause time between 2 pulses – in the order of milliseconds (determined by the speed of sound, distance of reflecting surfaces)*

• *pulse length – in the microsecond range (2-3 cycles long)*

- In ***pulse-echo methods*** the same transducer is used alternately for generation and detection of US pulses. The emitted US pulses are only 2-3 cycles long and thus the duration of a pulse is only some microseconds (considering the high frequency of the US used in imaging).
- The echo signals are detected in the time interval between the emissions of two subsequent US pulses (called ***pulse repetition time (PRT)*** or ***pause time***). The length of PRT can be changed between 100 µs - 1 ms. Longer PRT ensures longer period of time for decting the echo signals, thus echoes arriving back from tissues situated deeper inside the body can also be detected.

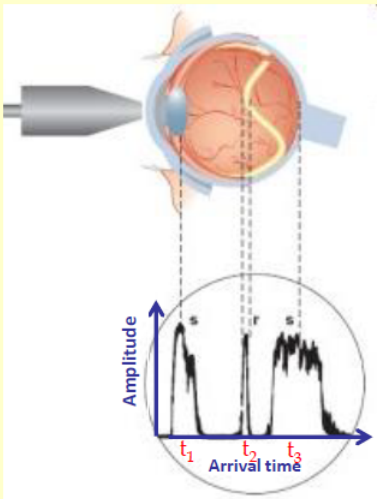
Diagnostic sonography
Pulse-echo methods II

One-dimensional A (amplitude) – images:

- fixed position US emitter producing a narrow beam
  ⇒ echoes are from a single direction
- echoes from different depths are displayed
  in the same order on the x-axis
- arrival time of the echo signals (t)
  → **distance (d)** of reflecting surfaces from the transducer:
  $ct=2d \Rightarrow d=ct/2$
  → **distance between two echoing surfaces:**
  $d_{12}=(ct_1-ct_2)/2$

- signal amplitudes – rarely used for diagnostic purposes

- signals arriving from deeper tissues are weaker – time gain compensation (TGC)



- The simplest pulse-echo method is the so-called one-dimensional ***A-mode (amplitude) imaging.*** The A-mode scanner sends a narrow US beam into the body and detecting the echo signals it is able to detect the reflecting surfaces in one line. The intensities (amplitudes) of the echo signals from different depths of the body are displayed in the same order on the x-axis representing the ***arrival time*** (***t***).
- From the arrival time (***t***) knowing the speed of sound propagation (***c=1540 m/s***) we can determine the distance of the echoing/reflecting surface from the transducer ***(d=ct/2***) or the distance of two reflecting surfaces from each other (***$d_{12}=(ct_1-ct_2)/2$***). Both equations contain a division-by-two, because the arrival time is measured from the emission of the US signal from the transducer, thus it involves the time while the US pulse reaches the reflecting surface and then the echo returns to the transducer.

Diagnostic sonography

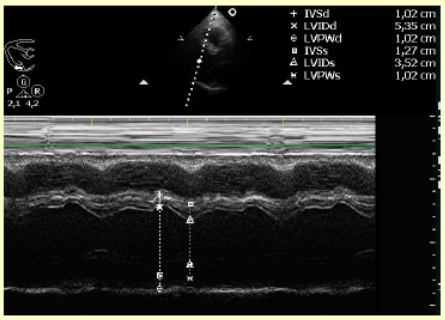
Pulse-echo methods III

One-dimensional B (brightness) – images

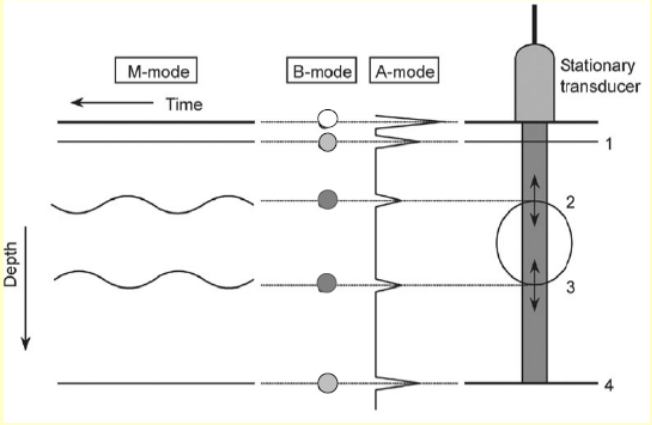
- echo intensity is represented by the brightness of the dots/pixels of the screen
- not used in itself, it is the basis of more advanced procedures

TM- (or M-) mode (time and motion)

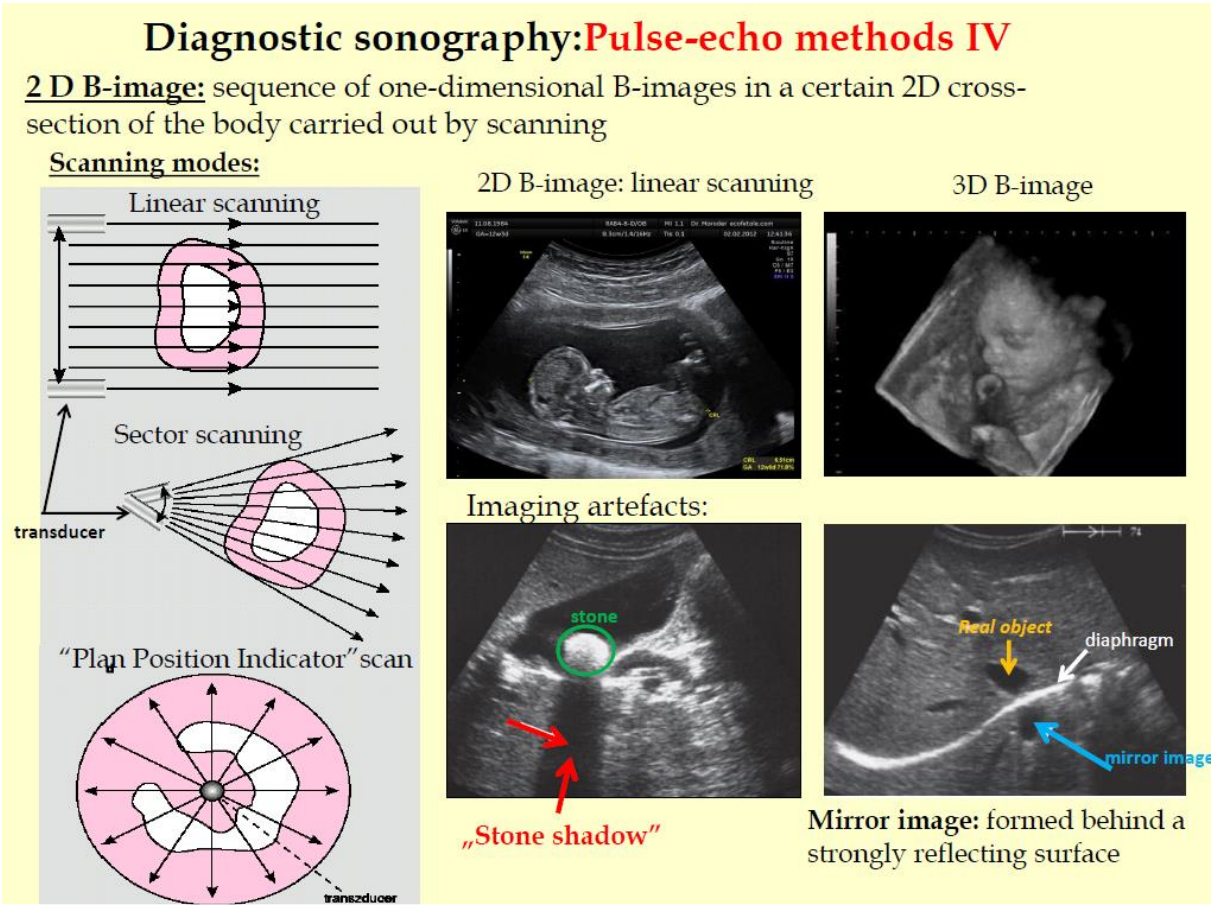
- used for detection of movement of reflecting surfaces (e.g. organ boundaries) in the direction of measurement
- B-images created by subsequent US pulses are plotted next to each other
- real timescale
- e.g. cardiological applications



Echocardiography (M-mode)



- ***One-dimensional B-mode (brightness) image***: The intensity (amplitude) of the echo signal can be easily visualized by the brightness of the appropriate pixel (dot) of the screen (as it is shown in the right figure). Thus the second axis of the coordinate system can be used for visualizing other parameters. This method is not used independently; however it can serve as the basis of more advanced procedures.
- In ***M (motion)-mode images*** one-dimensional B-images created by subsequent US pulses are plotted next to each other as a function of time. Using this method we can visualize the movement of the reflecting surfaces such as heart valves (see the left image). M-mode images are very important in the diagnostics of heart diseases.



- ***Two dimensional B-images:*** Two dimensional B-mode images are cross-sectional images representing tissues and organ boundaries within the body. The 2D images are formed as a sum of many one-dimensional B-mode images moving the transducer (scanning) on the surface of the body.
- The simplest method is called ***linear scanning.*** In this case a 2D image is generated by moving the scanner (transducer) in one line on the surface of the body.
- In case of ***sector scanning*** the angle of the scanner is changed. This method is most favourable when the transmission of US pulses into the body cavities is limited because of the surrounding bones. For instance the thoracic organs are examined with placing transducer between two ribs. In case of neonates cranial US imaging can be carried out through the relatively thin temporal window of the skull.
- We can also introduce the scanner into certain body cavities (such as the oesophagus, rectum or vagina). The advantage of this technique is that the transducer is close to the organs, tissues to be examined and thus the US signals are not absorbed.
- Using many 2D images taken from different directions we can make 3D images.
- The lower US images demonstrate two typical ***imaging artefacts***. Imaging artefacts are reflections that do not represent real anatomical structures. For example we readily observe a ***shadow*** (dark space) behind kidney or gall bladder stones, because stones reflect most of the US intensity, thus the remaining intensity is insufficient to produce detectable echoes from distal structures. Another imaging artefact is the ***mirror-image artefact*** (image duplication): The image of an object situated in front of a strongly reflecting surface (e.g. diaphragm) can be duplicated virtually, because the echoes returning back from the strongly reflecting diaphragm interact again with the object.

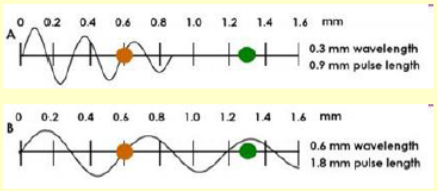
Diagnostic sonography - Resolution

Spatial resolution

- the minimal distance between two points that may be distinguished by their US images
- axial and lateral resolution

Axial resolution

- the ability to distinguish two structures which lie along the axis of the US beam (~ 0.5 mm)
- higher frequency (10-15 MHz) US ⇒ short pulse length ⇒ better resolution, because of its greater attenuation it is applied for imaging of superficial structures
- lower frequency (2-5 MHz) US ⇒ lower resolution, lower attenuation, for imaging deeper structures

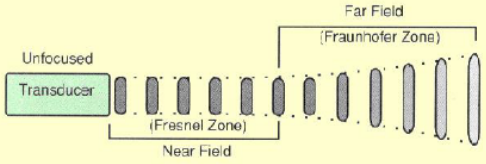


5 MHz transzducer, 3 cycles/pulse

2,5 MHz transzducer, 3 cycles/pulse

Lateral resolution

- it is determined by the beam width (D)
- $D^2/4\lambda$ determines the length of the Fresnel-zone with parallel beam



- ***Spatial resolution*** is an important property of every imaging technics. It is defined as the minimal distance between two points of the object that may be distinguished by their US images. In US imaging we discriminate between ***axial*** and ***lateral resolutions***.
- ***Axial resolution*** describes the ability to distinguish two structures which lie along the axis of the US beam. Its limit is ~ 0.5 mm. The pulse length itself is 2-3 cycles in modern scanners. Since ***c*** is constant in a certain medium, higher frequencies mean shorter wavelengths and hence shorter pulses. Thus, all other parameters being constant, higher frequencies lead to better axial resolution. In case of larger frequency the length of the US pulse is shorter, thus the orange and green objects can be discriminated in the upper image. Conversely lower US frequency results in longer pulse length and thus lower resolution. However, unfortunately higher frequency US is attenuated more in the tissues (***μ ~ f***) and thus it can only be applied for imaging of superficial tissues.
- ***Lateral resolution*** is determined by the beam width (***D***). The US beam has two distinguishable zones which are called the ***near field (Fresnel-zone)*** and the ***far field (Fraunhofer-zone)*** as it is shown in the lower figure. In the near-field US propagates as a parallel beam (pencil beam) and thus it serves good lateral resolution, while in the far field it starts to diverge deteriorating the lateral resolution. Generally, the near field extends to a distance $D^2/4\lambda$ from the transducer, where D is the beam width and $\boldsymbol{\lambda}$ is the wavelength of the US. It is important to realize that the tissues or organs we want to examine should be in the near field, where the lateral resolution is good. Unfortunately, smaller beam width also produces shorter near fields, so we have to come to a compromise to choose a beam width that ensures that the structures to be examined localize in the near field and also provides reasonably good lateral resolution.

Doppler-effect I.

A  B

$f_{observed} > f$ $f_{observed} < f$

- Christian Doppler (1824)
- the source and the observer are in motion relative to each other ⇒ observed frequency differs from the emitted one
- standing source – moving observer, or moving source – standing observer different formulas
- if v (relative speed)<<c ⇒ difference is negligible, both of them can be used
- standing source – moving observer:

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

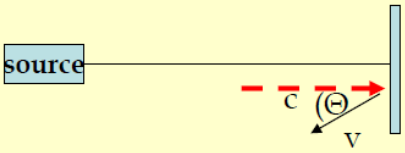
- Doppler-shift (f_D):

$$f_D = f' - f = \frac{\pm v}{c} f$$

- The ***Doppler phenomenon*** is often exploited in US diagnostics. If the source and the observer are in relative motion to each other the observed frequency differs from the emitted one. This phenomenon can be observed in cases of both mechanical and electromagnetic waves. A common example of the Doppler-effect is the decrease in pitch of a siren that is heard as an ambulance passes an observer. Conversely, an increase in pitch of a siren is heard when an ambulance approaches an observer.
- In case of standing source and moving observer the relationship between the observed frequency (***f'***) and the emitted frequency (***f***) is given by the 1st equation, where ***v*** is the velocity of the moving observer and ***c*** is the speed of sound propagation in a given medium.
- Reorganizing the 1st equation we can express the Doppler-shift (f_D), which is the difference between the observed and emitted frequencies.

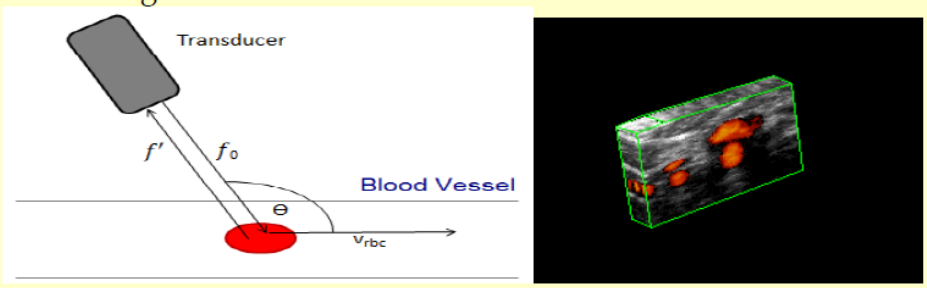
Doppler-effect II.

- moving reflecting body ⇒ relative velocity is 2v
- $f_D=(2v/c)f$ (absolute value)
- if v and c are not parallel only the c direction component of v does matter



$$f_D = \frac{2v\cos\Theta}{c} f \qquad v = \frac{c}{2f\cos\Theta} f_D$$

- US frequency shift ⇒ velocity of moving structures can be determined
- **Doppler mode**: measurement of blood flow (red blood cells are US scattering centers)
- **Colour Doppler**: velocity information is presented as a color-coded overlay on top of a B-mode image

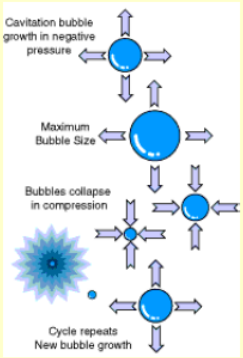


- US can be applied for measuring the blood flow rate in larger blood vessels utilizing the Doppler phenomenon. US pulses are scattered by the red blood cells, since their size is comparable to wavelength of the applied US. The frequency of the US signal scattered by the blood cells differs from the original US signal emitted by the transducer (***Doppler-shift***), since the scattering objects (blood cells) are in motion relative to the transducer.
- When studying the blood flow rate blood cells are considered as targets moving with a ***v*** velocity compared to the stationary US source, while upon formation of the echo blood cells behave as secondary sound sources moving with a ***v*** velocity. Since the above effects are additive in generating a Doppler-shift we use 2***v*** relative velocity and thus the Doppler-shift is given by the following formula, if the US beam is parallel with the direction of blood flow: $f_D=(2v/c)f$. If the US beam is not parallel with the direction of blood flow only the ***c*** direction component of ***v*** matters (***2v × cosΘ***). In case of perpendicular incidence the Doppler-shift is zero, since ***cos90°= 0***. In practice we always obtain a weak Doppler-shift, because the direction of the movement of some blood cells may slightly deviate from the 90°.
- It is very convenient that the value of Doppler-shift (f_D) tends to lie in the audiofrequency range (20 Hz – 20 kHz) allowing the operator to monitor the Doppler signal simply by listening to it. Since ***c*** is known, measuring the Doppler-shift (f_D) we can determine the rate and the direction of the blood flow. In Colour-Doppler technique the above data are displayed by red and blue colours superimposed on 2D B-mode images. Traditionally red colours represent blood flow in the direction of the transducer, while blue colours show movement away from the transducer.

Effects of high intensity US

Primary effects:

- **cavitation (in liquids)**
  - microscopic voids (<100 µm) lacking liquid, short lifetime
  - expansion phase of US → formation of bubbles, compression phase → shrinking and collapse
  - cohesion forces are broken due to rapid alterations of the pressure
  - threshold intensity depends: frequency of US, viscosity of the medium
  - *enormous temperature and pressure changes upon formation and collapse*
- **absorption** (see earlier)
- **sound pressure:** The US wave exerts pressure on objects in its direction of propagation. The pressure is directly proportional to the US intensity.

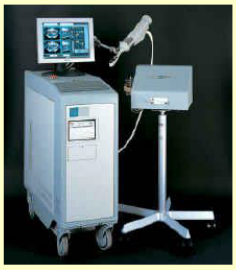


Secondary effects:

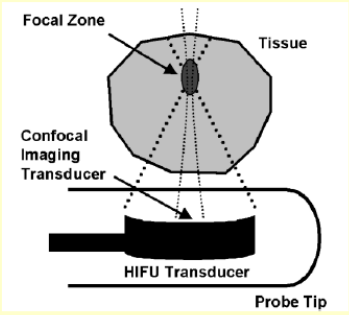
- mechanical
  - mechanical „rubbing" (← speed difference between particles with different sizes)
  - dispergation, cleaning (← cavitation)
- mechanical effect + absorption → thermal effect
- chemical effect (absorption → excitation → chemical processes)
- biological effect (e.g. disintegration of bacteria)

- The primary effects of US (***sound pressure***, ***absorption***, ***cavitation*** and ***mechanical rubbing effect***) are the consequences of the high frequency and large amplitude vibrations of the particles of the medium induced by the propagation of US.
- The US wave exerts pressure on objects in its direction of propagation. It is called ***sound pressure*** and it is directly proportional to the US intensity.
- If the sizes of particles differ in the medium, speed differences arise between them upon sound propagation. Smaller particles move faster, while larger ones move slower or stay motionless and consequently friction occurs between them, which causes the typical ***mechanical rubbing effect*** of US (also called micro massage).
- Upon propagation in a medium the energy of US is gradually ***absorbed***. In cases of larger US intensities absorption results in a significant temperature increase in the medium.
- In a liquid subjected to high intensity US the cohesive bonds between the molecules can be broken because of the rapid alterations in pressure, and thus small (microscopic) voids free of liquid (***cavitation bubbles***) can be formed. These small bubbles (around 100 µm in size) are formed when the pressure decreases below its resting level (expansion phase) and collapse when the pressure increases above it (compression phase). Formation and collapse of the cavitation bubbles occur very fast, leading to enormous temperature changes in the medium.
- The above mentioned primary effects of US may induce ***secondary effects*** such as ***dispergation, chemical*** and ***biological effects***. For example mechanical rubbing effect can be used for dispergation of solid materials. Thermal effects caused by high intensity US can induce chemical reactions that cannot occur otherwise. Thermal and mechanical effects can induce biological effects such as damaging of cells (e.g. bactericide and fungicide effects).

Therapeutic applications of high intensity US



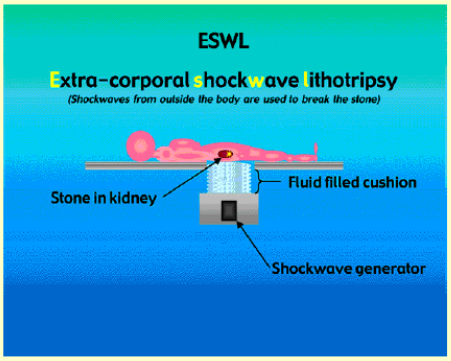
High Intensity Focused Ultrasound (HIFU)



- cancer tissue destruction by localized temperature increase induced by US („hyperthermia therapy")
- for treatment of tumors not shaded by bones or air

Extracorporeal Shockwave Lithotripsy (ESWL)

- stone breaking – e.g., kidney stones
- 100 kHz to 1 MHz frequency, (50 MPa acoustic pressure waves)
- direct shearing forces, as well as cavitation bubbles formed in the liquid surrounding the stone contribute to its fragmentation



- ***HIFU (High Intensity Focused Ultrasound)*** exploits local temperature increase (***hyperthermia***) and cavitation brought about by the absorption of high intensity US for eradicating tumours. Similarly to gamma knife tumours can be irradiated from different directions to obtain high US intensity in the targeted tumour sufficient for its elimination, while only lower intensities pass through the neighbouring healthy tissues to avoid their damages. This method is suitable for treating tumours of organs, which are not shadowed by air or bones.

- ***Extracorporeal Shockwave Lithotripsy (ESWL):*** High intensity shock waves are generated outside the patient's body and focused on the kidney stone. The length of the individual US pulses is very short (about 5 μs), and the maximum value of the sound pressure is around 40 MPa. The exact physical background of stone breaking is still unknown, although it seems likely that mechanical shear forces as well as cavitation in the surrounding liquid can contribute to it. The high intensity US pulses partly absorbed in the stone and partly reflected from its surfaces may generate forces acting in opposite directions leading to disintegration of the stone. On the other hand, the formation and collapse of cavitation bubbles in the surrounding liquid may damage the surface of the stone contributing to its erosion.