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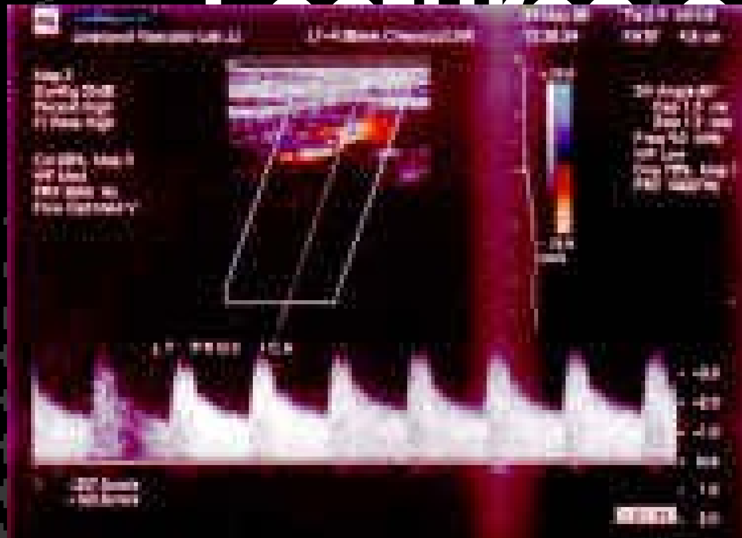
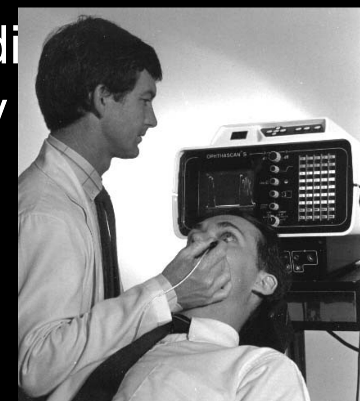
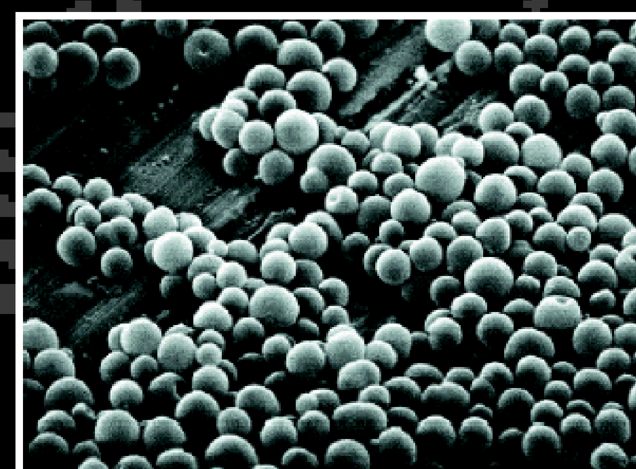
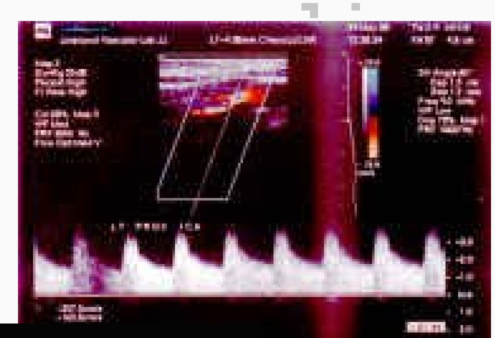


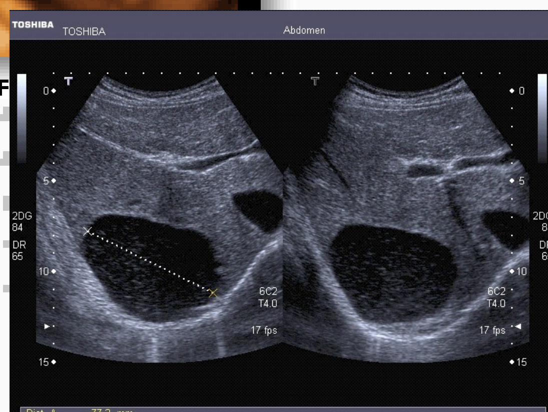


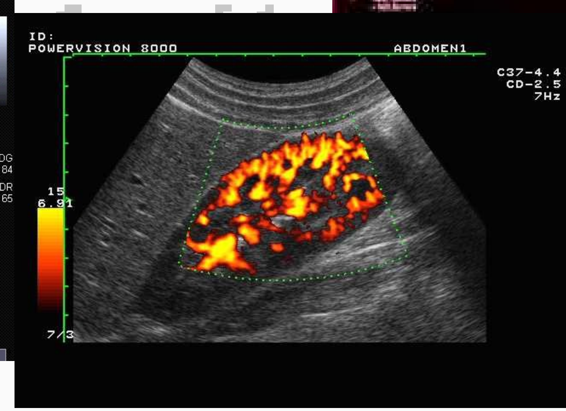


Lectures on Medical Biophysics

Department of Biophysics, Medical Faculty,
Masaryk University, Brno







Ultrasound diagnostics

Lecture outline

- Physical properties of ultrasound and acoustic parameters of medium
- Ultrasonography
    - Impulse reflection method
    - A-mode – one-dimensional
    - B-mode – two-dimensional
    - M-mode
    - Basic characteristics of US images
    - Interventional sonography
    - Echocontrast agents
    - Harmonic imaging
    - Principle of 3D imaging
- Doppler flow measurement
    - Principle of Doppler effect
    - Principle of blood flow measurement
    - CW Doppler system
    - Systems with pulsed wave – PW Doppler
    - Duplex and Triplex methods
    - Power Doppler method
    - Tissue Doppler Imaging (TDI)
- Ultrasonic densitometry
- Patient Safety: reducing Ultrasound ‘Doses’

Ultrasound diagnostics

- Ultrasound diagnostics started to develop in early 50‘ of 20th century. It allows to obtain cross-sectional images of the human body which can also include substantial information about its physiology and pathology.

- Ultrasound diagnostics is based mainly on reflection of ultrasound waves at acoustical interfaces

- We can distinguish:
  - Ultrasonography (A, B and M mode, 3D and 4D imaging)
  - Doppler flow measurement, including Duplex and Triplex methods (Duplex, Colour, Triplex, Power)
  - Tissue Doppler imaging
  - Ultrasound densitometry

Physical properties of ultrasound

Before we will deal with diagnostic devices, we need to understand what is ultrasound and what are the main acoustical properties of medium.

Ultrasound (US) is *mechanical oscillations* with frequency above 20 kHz which propagate through an elastic medium.

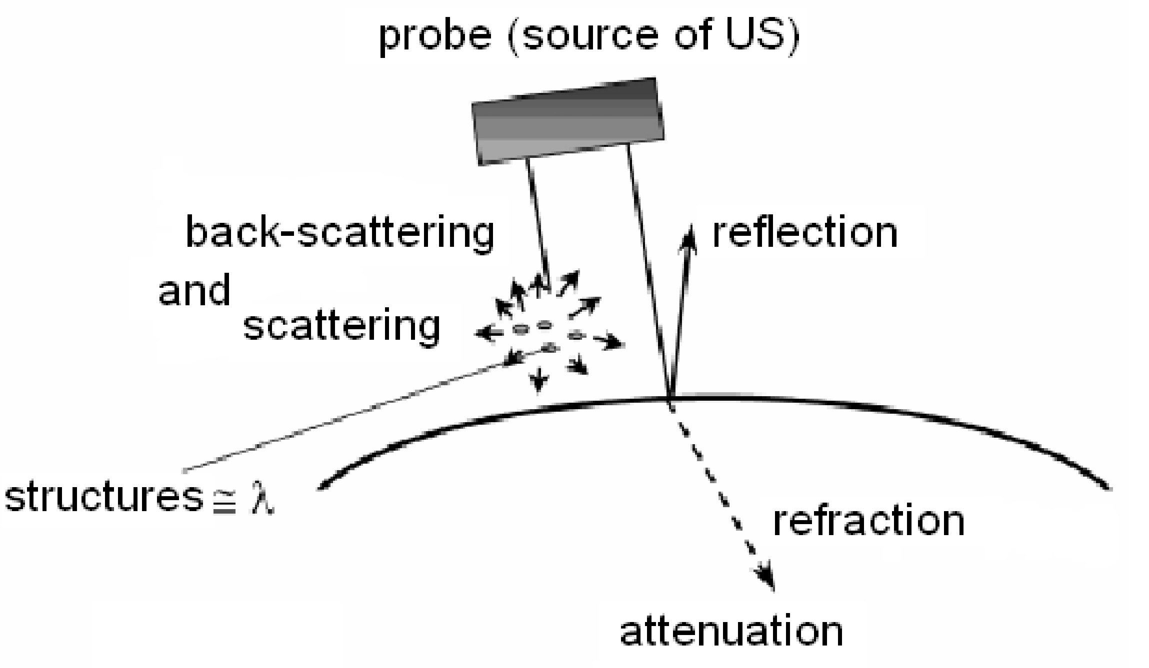
In liquids and gases, US propagates as longitudinal waves.

In solids, US propagates also as transversal waves.

Interactions of US with Tissue

- Reflection (smooth homogeneous interfaces of size greater than beam width, e.g. organ outlines)
- Rayleigh Scatter (small reflector sizes, e.g. blood cells, dominates in non-homogeneous media)
- Refraction (away from normal from less dense to denser medium, note opposite to light, sometimes produces distortion)
- Absorption (sound to heat)
  - absorption increases with f, note opposite to X-rays
  - absorption high in lungs, less in bone, least in soft tissue, again note opposite to x-rays
- Interference: ‘speckles’ in US image result of interference between Rayleigh scattered waves
- Diffraction

Acoustic parameters of medium:



Interaction of US with medium – reflection and back-scattering, refraction, attenuation (scattering and absorption)

Acoustic parameters of medium

$$c = \sqrt{\frac{K}{\rho}} \quad \left[ms^{-1}\right]$$

Speed of US *c* depends on elasticity and density ρ of the medium:

K - modulus of compression

in water and soft tissues c = 1500 - 1600 m.s^{-1}, in bone about 3600 m.s^{-1}

Acoustic parameters of medium

Attenuation of US expresses decrease of wave amplitude along its trajectory. **It depends on frequency**

$$I_x = I_o e^{-2\alpha x} \qquad \alpha = \alpha' . f^2$$

I_x – final intensity, I_o – initial intensity, 2x – medium layer thickness (reflected wave travels „to and fro“), α - linear attenuation coefficient (increases with frequency).

Since

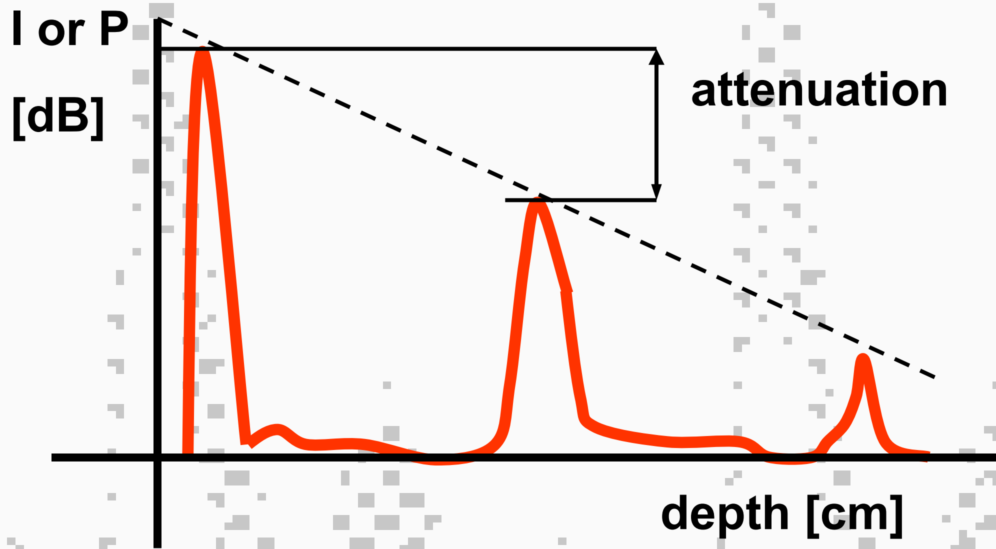
$$\alpha = \log_{10}(I_0/I_X)/2x$$

we can express α in units dB/cm. At 1 MHz: muscle 1.2, liver 0.5, brain 0.9, connective tissue 2.5, bone 8.0

Acoustic parameters of medium

Attenuation of ultrasound

When expressing intensity of ultrasound in decibels, we can see the amplitudes of echoes to decrease linearly.



$$\frac{I_x}{I_0} = {}^{-\alpha} \Rightarrow \quad = \alpha \Rightarrow \quad) =$$

Acoustic parameters of medium

Acoustic impedance: product of US speed *c* and medium density ρ

$$Z = \rho \cdot c \qquad (Pa.s/m)$$

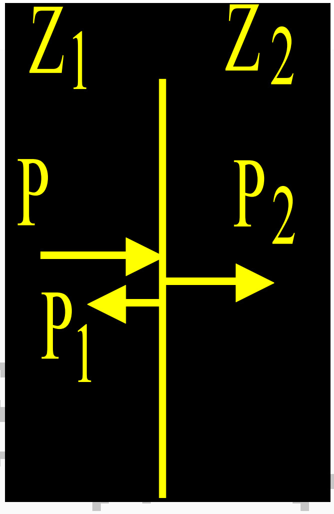
$Z.10^{-6}$: muscles 1.7, liver 1.65 brain 1.56, bone 6.1, water 1.48

Acoustic parameters of medium: US reflection and transmission on interfaces

We suppose perpendicular incidence of US on an interface between two media with different Z - a portion of waves will pass through and a portion will be reflected (the larger the difference in Z, the higher reflection).

$$R = \frac{P_1}{P} = \frac{Z_2 - Z_1}{Z_2 + Z_1}$$

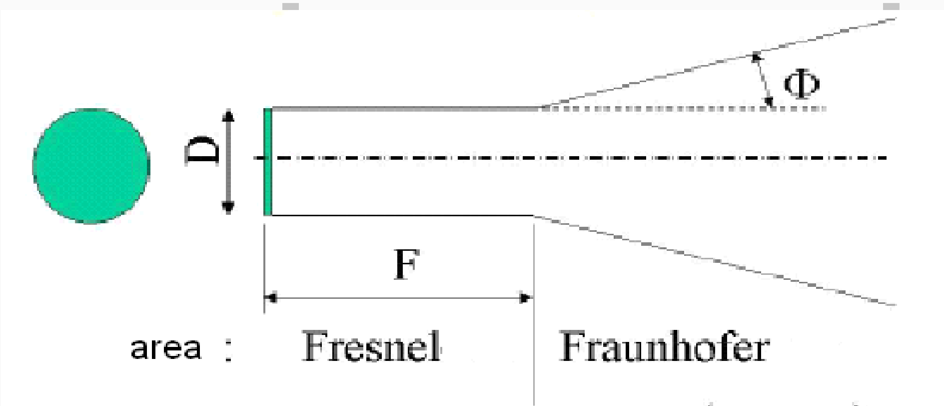
$$D = \frac{P_2}{P} = \frac{2\,Z_1}{Z_2 + Z_1}$$



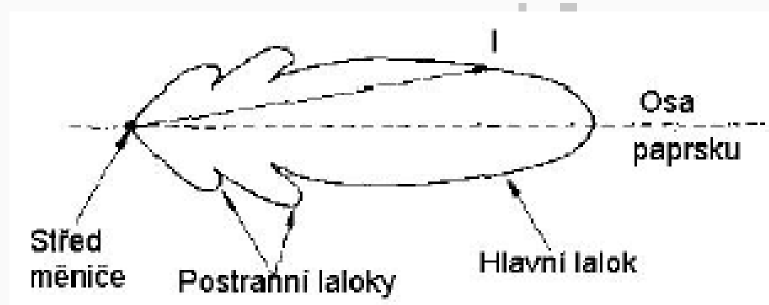
Coefficient of reflection R – ratio of acoustic pressures of reflected and incident waves

Coefficient of transmission D – ratio of acoustic pressures of transmitted and incident waves

Acoustic parameters of medium: Near field and far field



$$F = \frac{D^2}{4\lambda} [m] \qquad \Phi = \arcsin\left(1{,}22\,\frac{\lambda}{D}\right) [\circ]$$



- Near field (Fresnel area) – this part of US beam is cylindrical – there are big pressure differences in beam axis
- Far field (Fraunhofer area) – US beam is divergent – pressure distribution is more homogeneous
- Increase of frequency of US or smaller probe diameter cause shortening of near field - divergence of far field increases

Ultrasonography

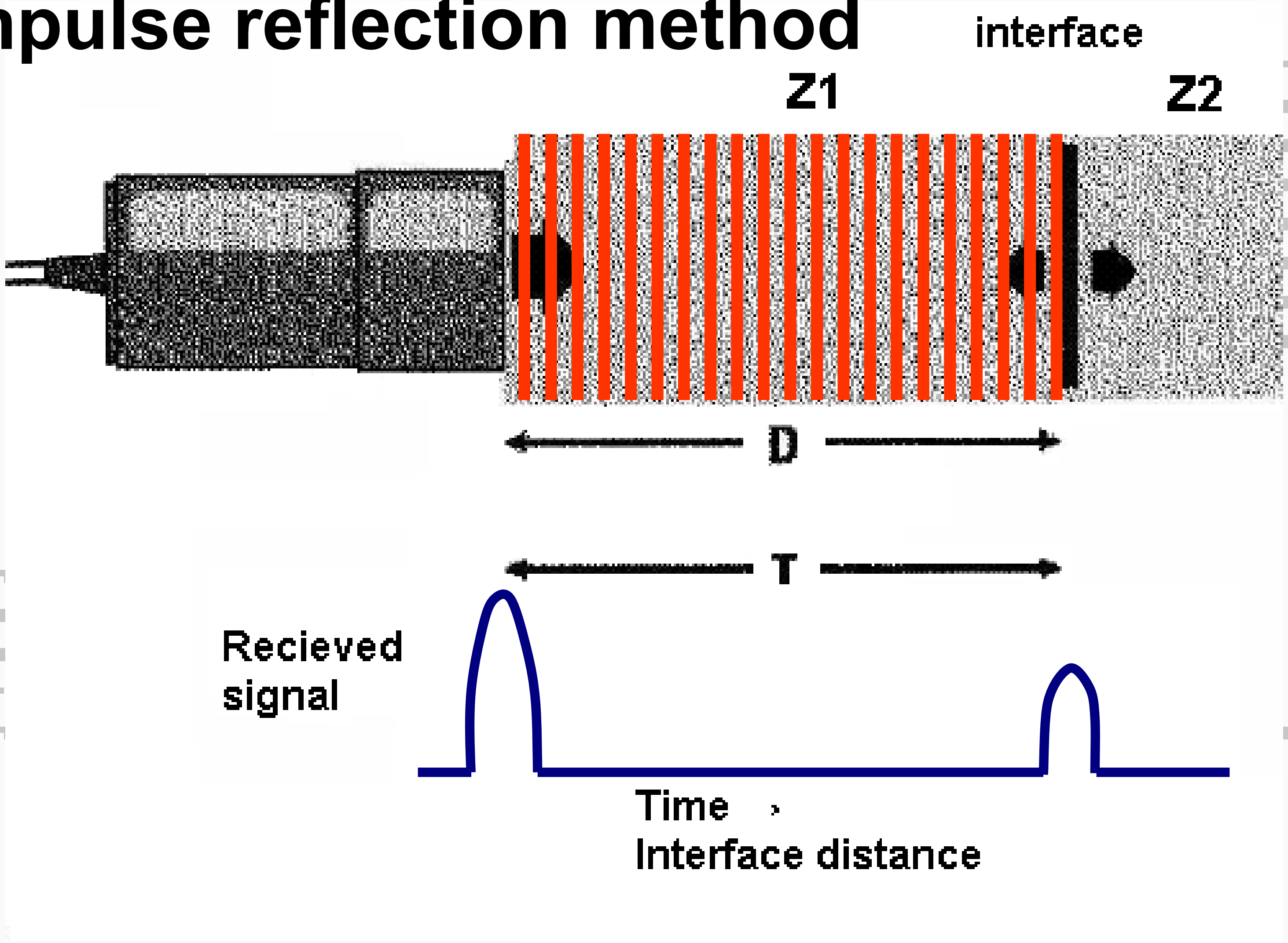
<u>Passive US</u> – low intensity waves which **cannot** cause substantial changes of medium.

In US diagnostics **(ultrasonography = sonography = echography) -** frequencies used are 2 - 40 MHz with (temporal average, spatial peak) intensity of about 1 kW/m^2

Impulse reflection method: a probe with one transducer which is *source as well as detector* of US impulses. A portion of emitted US energy is *reflected* on the acoustic interfaces and the same probe then receives reflected signal. After processing, the signal is displayed on a screen.

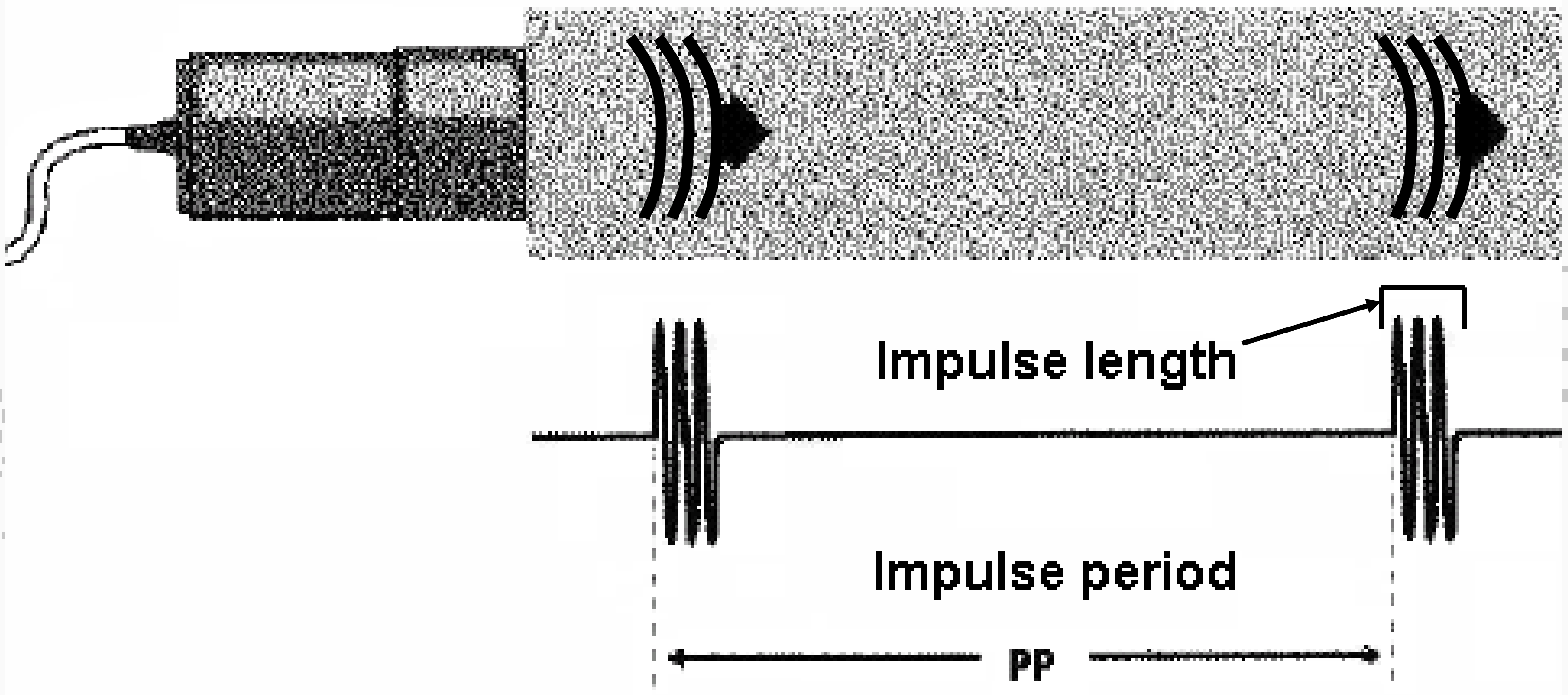
Ultrasonography

Impulse reflection method



Ultrasonography

Impulse reflection method



Ultrasonography

Impulse reflection method

Main parts of the US apparatus:

Common to diagnostics and therapy

- probe with electroacoustic transducer (transducers)
- generator of electric oscillations (continuous, pulsed)

Special parts of diagnostic apparatus

- electronic circuits for processing of reflected signal
- display unit
- recording unit

Ultrasonography

A-mode – one-dimensional

- Distances between reflecting interfaces and the probe are shown.

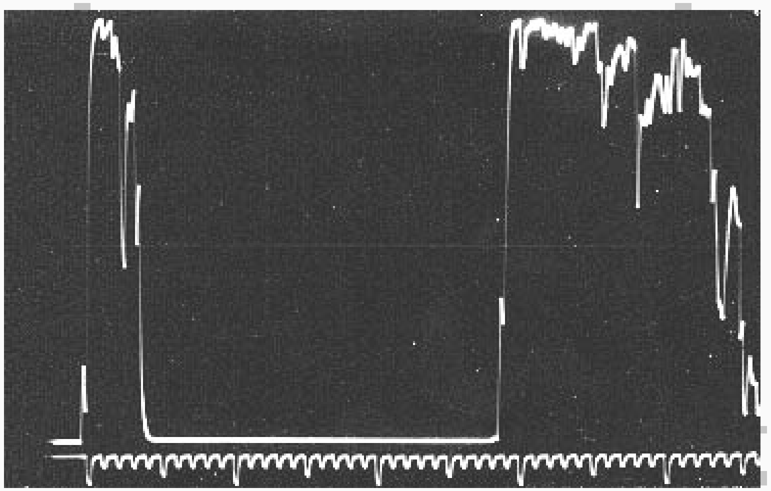
- Reflections from individual interfaces (boundaries of media with different acoustic impedances) are represented by ***vertical deflections*** of base line, i.e. the **echoes.**

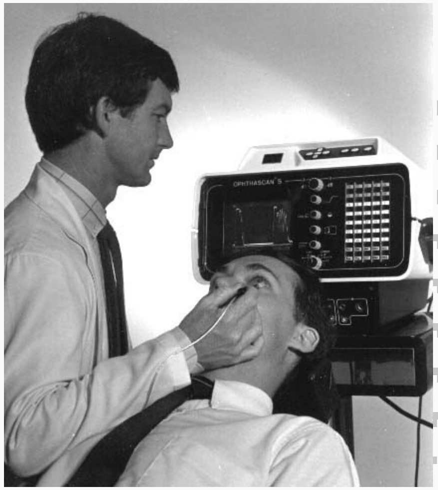
Echo amplitude is proportional to the *intensity of reflected waves* (amplitude modulation)

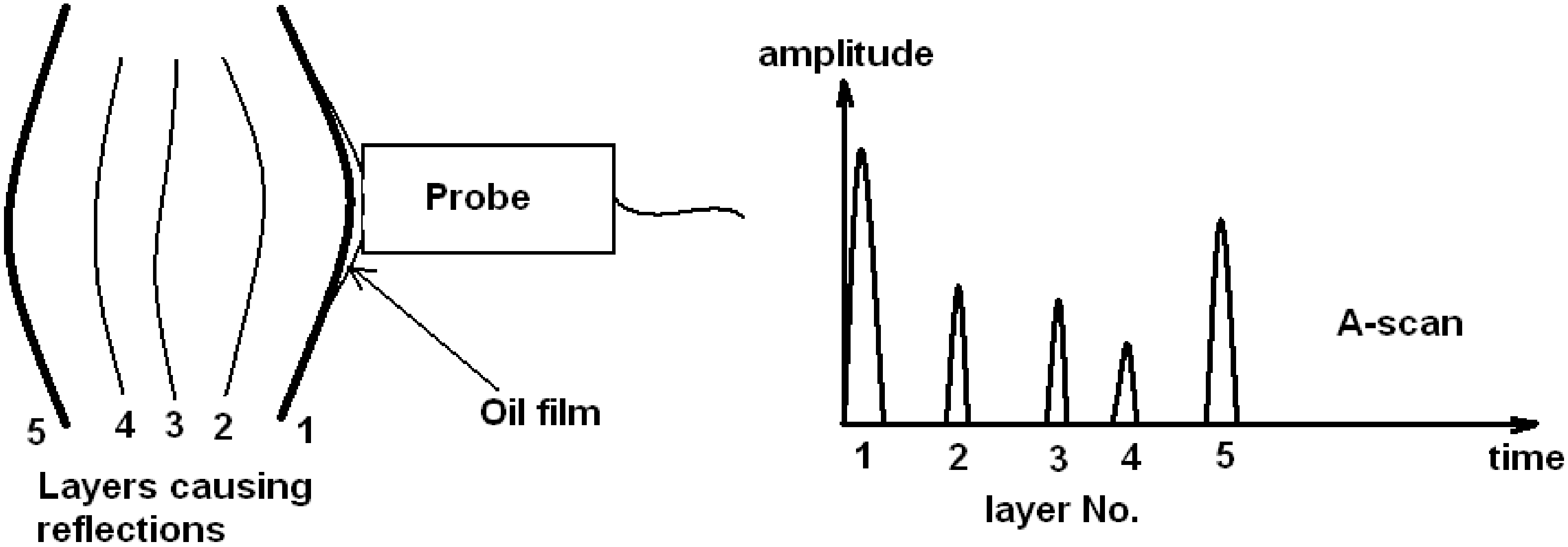
Distance between echoes shown on the screen is proportional to real distance between tissue interfaces.

Today used mainly in ophthalmology.

Ultrasonography
A-mode – one-dimensional







PRINCIPLE OF A-MODE SCAN

Ultrasonography

B-mode – two-dimensional

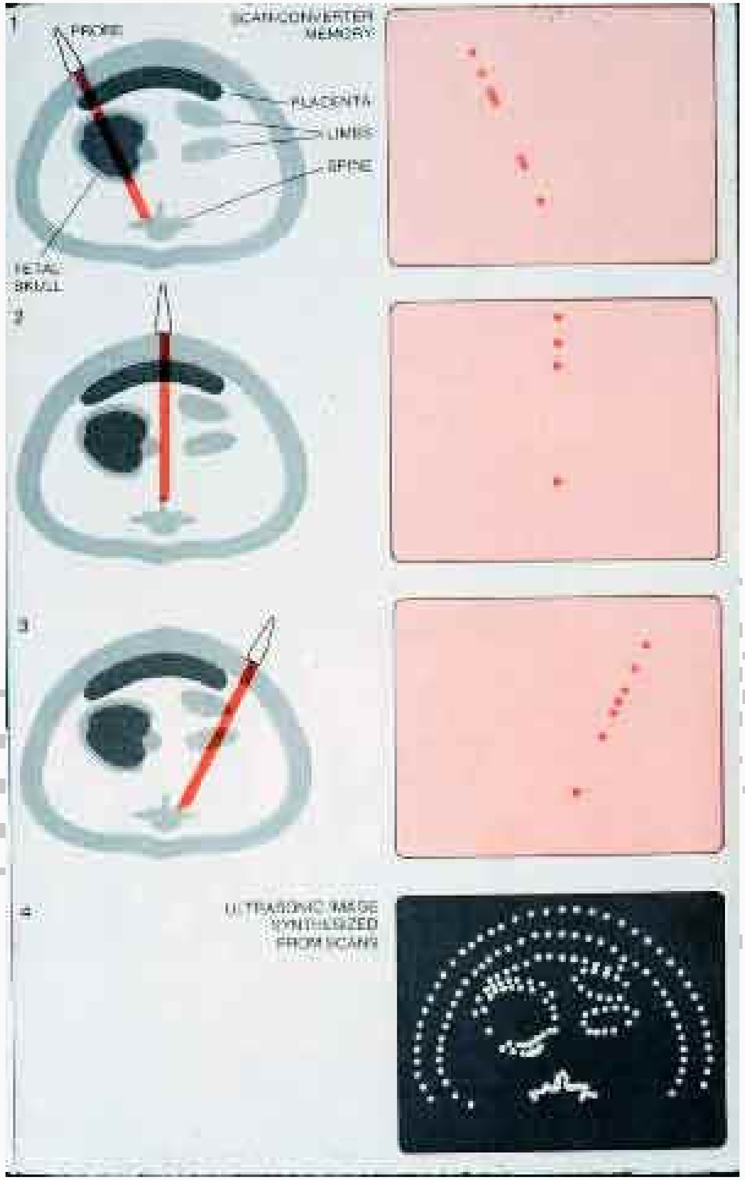
A tomogram is depicted.

Brightness of points on the screen represents intensity of reflected US waves (brightness modulation).

Static B-scan: a cross-section image of examined area in the plane given by the beam axis and direction of *manual* movement of the probe on body surface. The method was used in 50‘ and 60‘ of 20th century

Ultrasonography

B-mode – two-dimensional - static

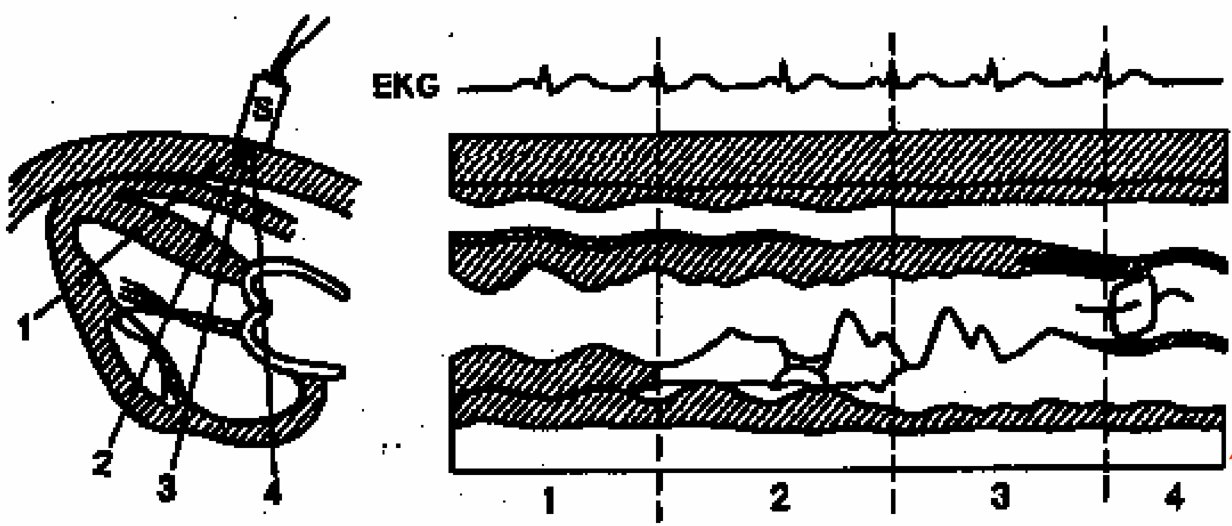


Ultrasonography **M-mode**

One-dimensional static B-scan shows movement of reflecting tissues. The second dimension is time in this method.

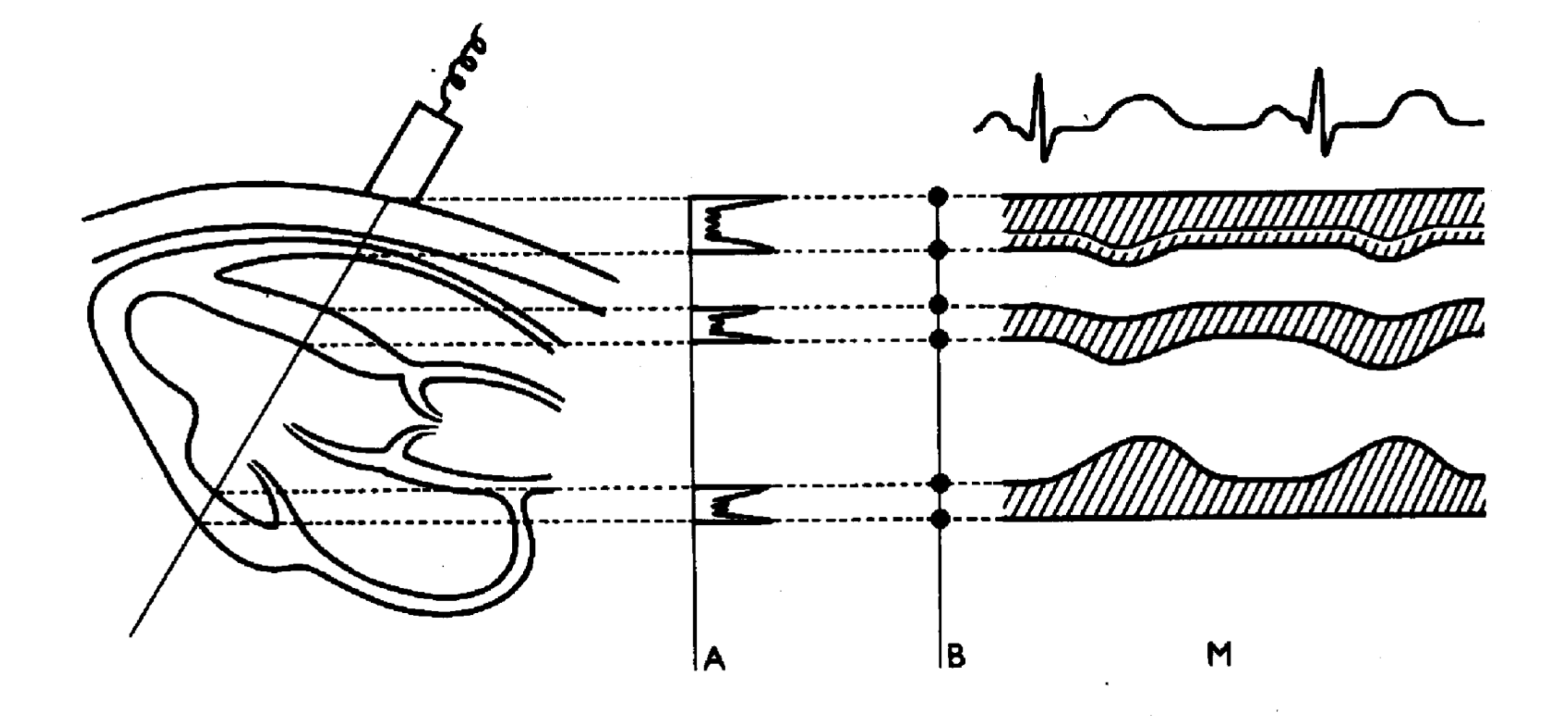
Static probe detects *reflections* from moving structures. The bright *points* move *vertically* on the screen, *horizontal shifting* of the record is given by slow time-base.

Displayed curves represent *movement* of tissue structures



Ultrasonography

Comparison of A-, B- and M-mode principle



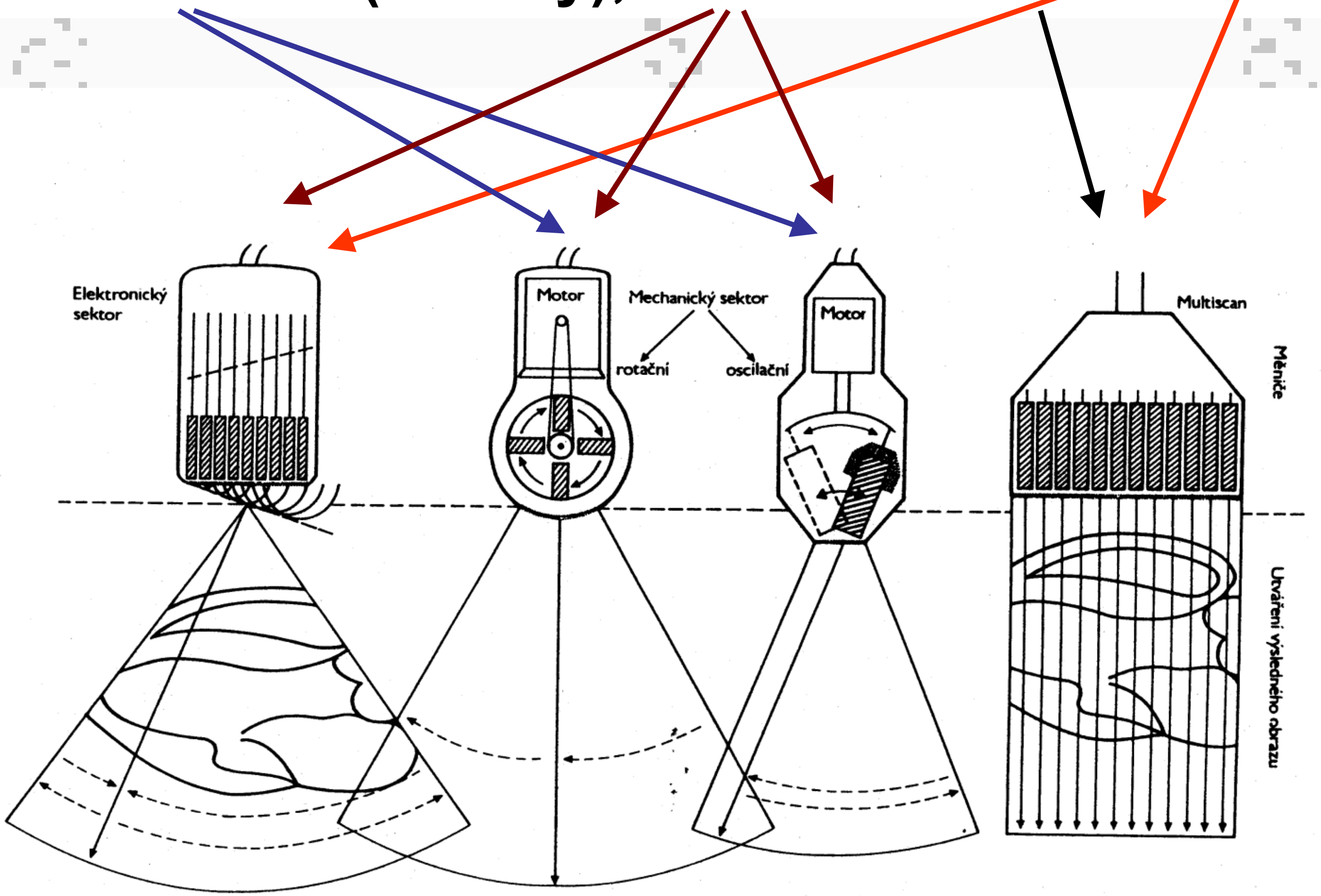
Ultrasonography

B-mode - dynamic

Repetitive formation of B-mode images of examined area by **fast deflection of US beam** mechanically (in the past) or electronically „in real time“ today.
Electronic probes consist of many piezoelectric transducers which are gradually activated.

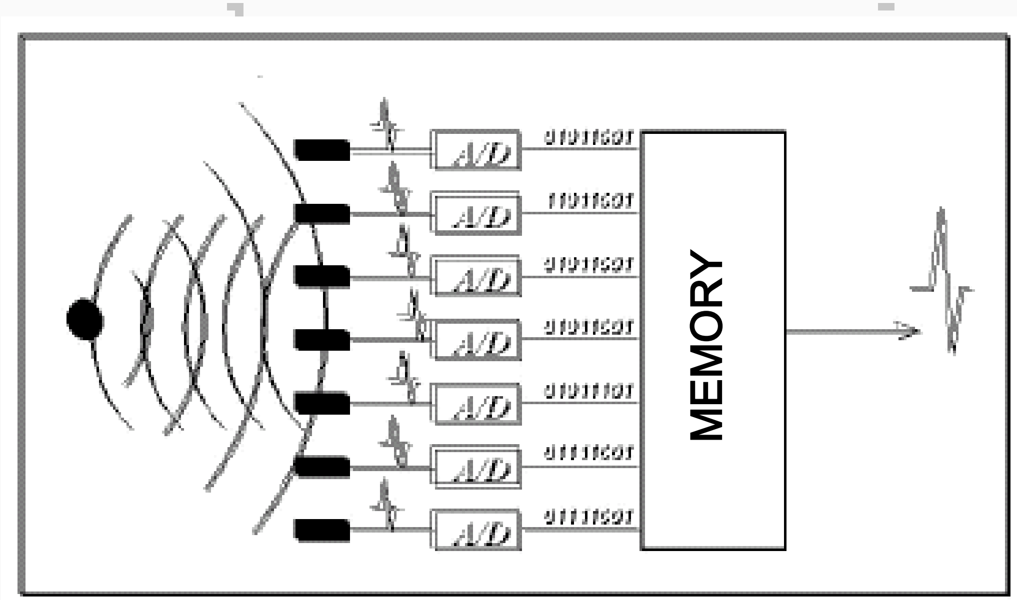
B-mode - dynamic

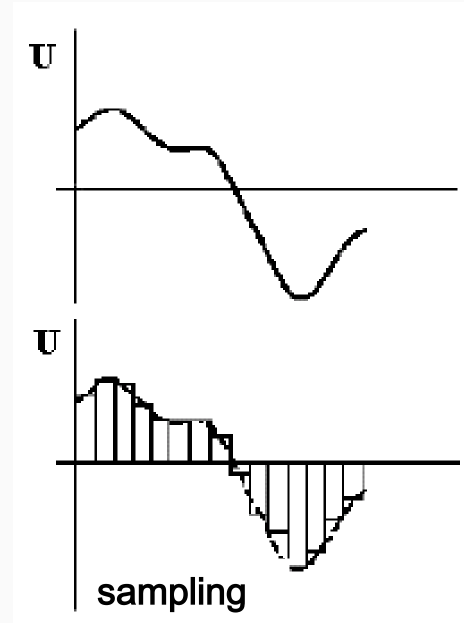
Ultrasound probes for dynamic B-mode: electronic and mechanical (history), sector and linear.



Abdominal cavity is often examined by **convex probe** – a combination of a sector and linear probe.

B-mode - dynamic

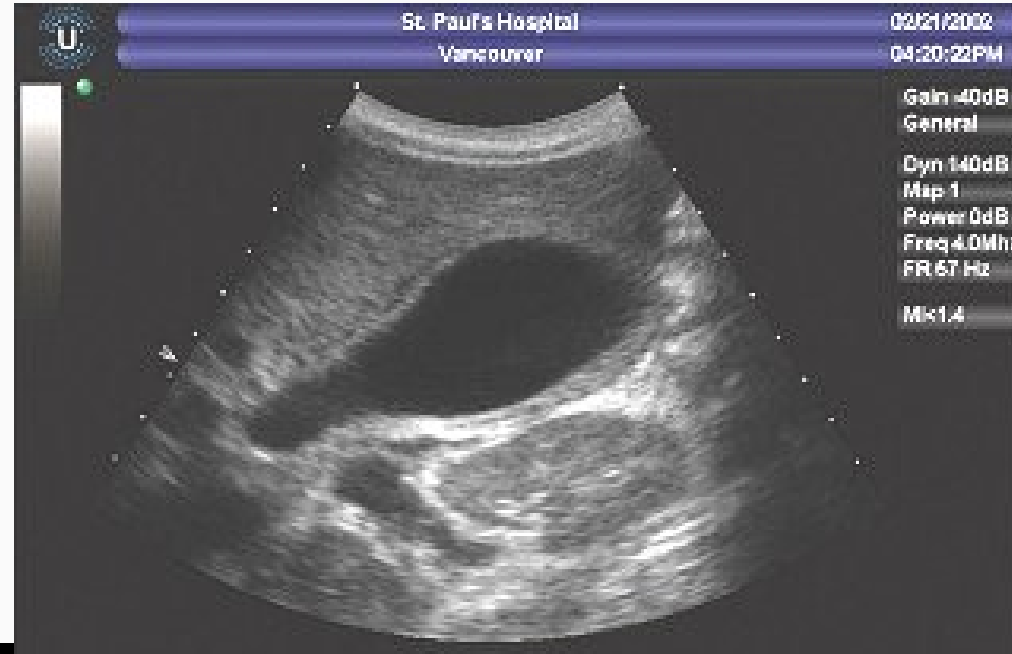


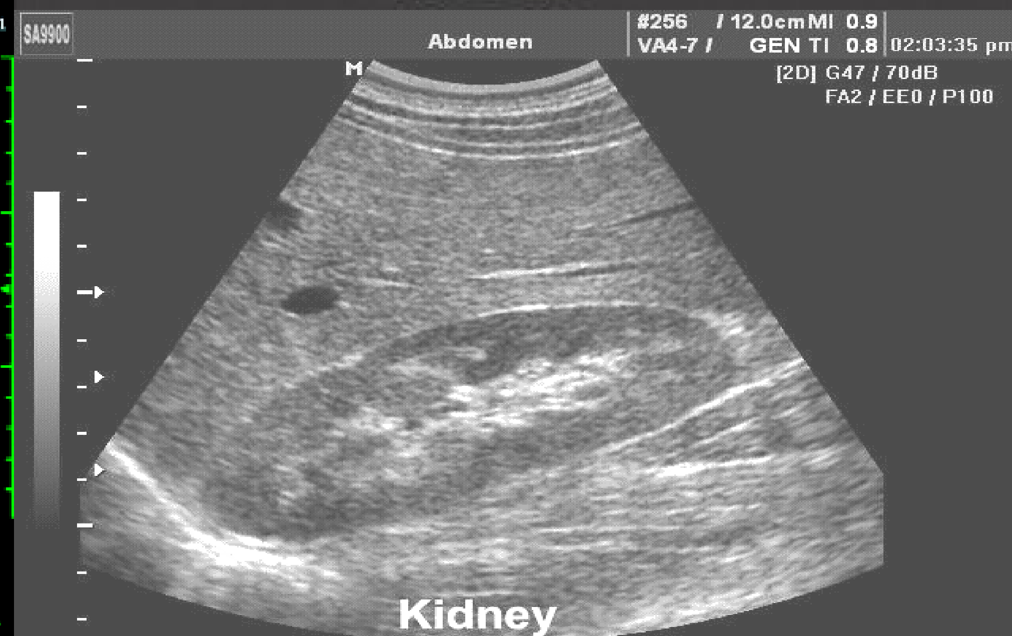


Modern ultrasonography - digital processing of image

- **Analogue part** – detection system
- **Analogue-digital converters (ADC)**
- **Digital processing of signal** – possibility of programming (preprocessing, postprocesssing), image storage (floppy discs, CD, flash cards etc.)

Ultrasonography B-mode - dynamic





Ultrasonography

Basic characteristics of US images

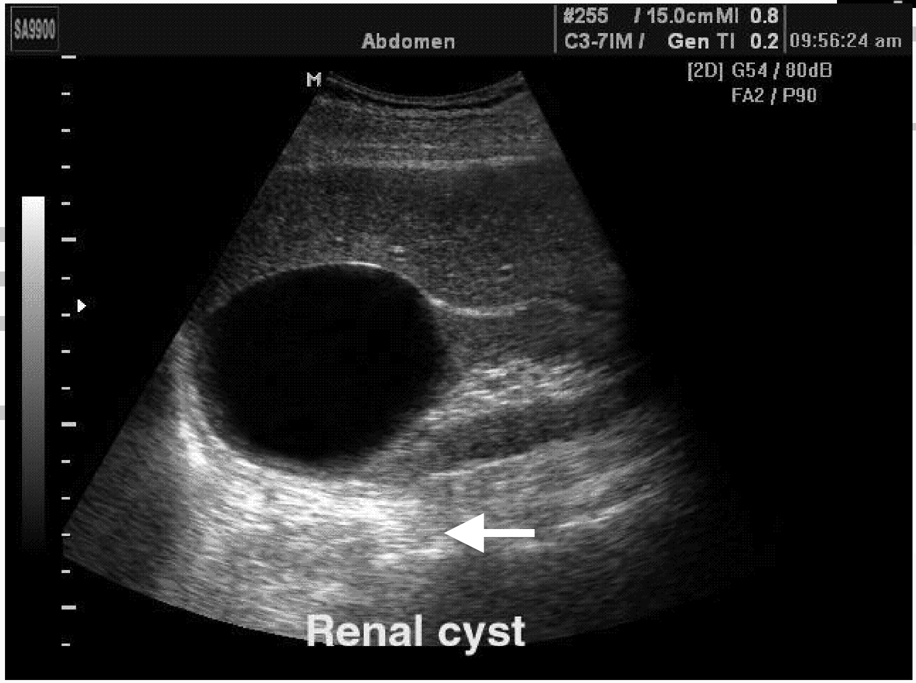
Degree of reflectivity – echogenity. The images of cystic (liquid-filled) and solid structures are different. According to the intensity of reflection we can distinguish structures:

hyperechogenic, izoechogenic, hypoechogenic, anechogenic.

- **Solid structures – acoustic shadow** (caused by absorption and reflection of US)
- **Air bubbles and other strongly reflecting interfaces cause repeating reflections** (reverberation, „comet tail“).

Ultrasonography

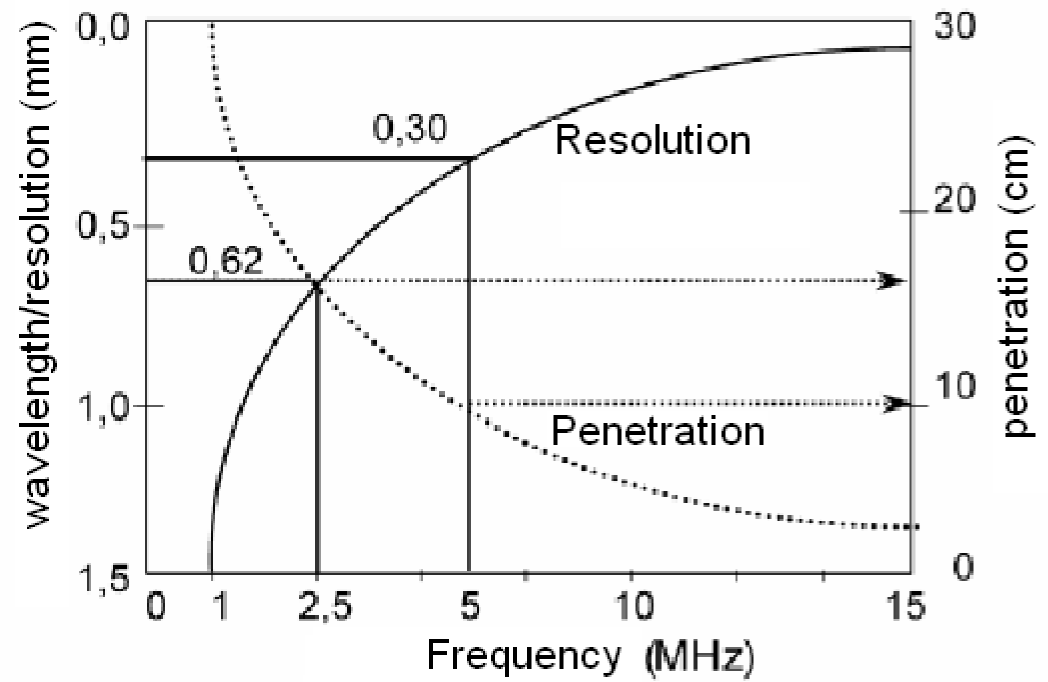
Acoustic shadow caused by absorption and reflection of US by a kidney stone (arrow)



Hyperechogenic area below a cyst (low attenuation of US during passage through the cyst compared with the surrounding tissues – arrow)

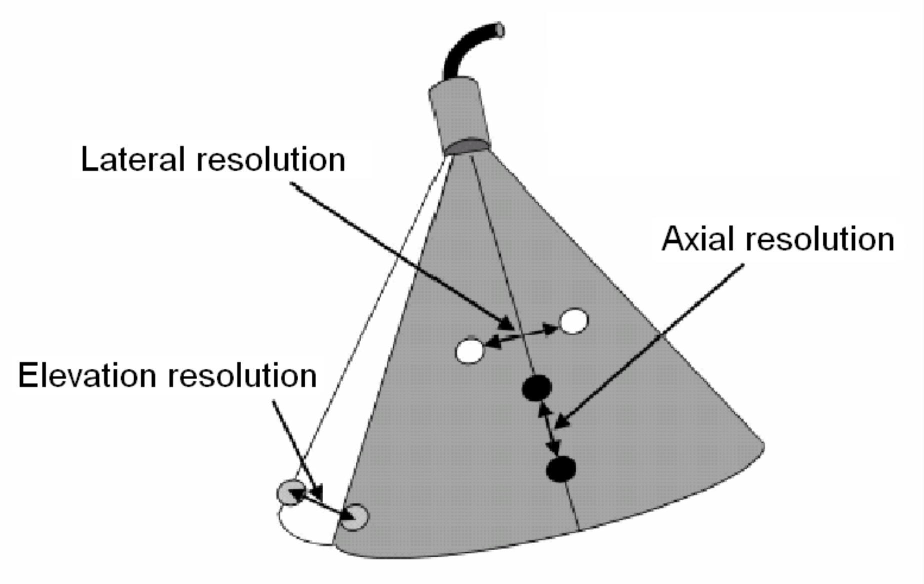
Ultrasonography

Spatial resolution of US imaging system is determined by the wavelength of the US**.** When the object dimension is smaller than this wavelength only **scattering** occurs. Hence higher spatial resolution requires higher frequencies



Limitation! – absorption of US increases with frequency of ultrasound = smaller penetration depth
Compromise frequency **3-5 MHz – penetration in depth of about 20 cm**

Spatial Resolution



- **Axial spatial resolution - it is given by the shortest distance of two distinguishable structures lying in the beam axis –** it depends mainly on frequency (at 3.5 MHz about 0.5 mm)
- **Lateral spatial resolution - it is given by the shortest distance of two distinguishable structures perpendicularly to the beam axis** – depends on the beam width
- **Elevation** – ability to distinguish two planes (sections) lying behind or in front of the depicted tomographic plane – it depends on frequency and beam geometry

Spatial Resolution

The best resolving power can be found in the narrowest part of the US beam profile.

Focusing – US beam is **converged** at the examined structure by means of acoustic lenses (shapes of the layer covering the transducer) or electronically.

- **The probes can be universal or specially designed** for different purposes with different focuses.
- **The position of focus can be changed** in most sector probes).

Ultrasonography

Interventional sonography

- **Interventional sonography is used mainly for guiding punctures**
- **diagnostic** – thin needle punctures to take tissue samples for histology
- **therapeutic** – for aspiration of a cyst or an abscess content or an exudate etc.
- Puncture can be done by „free hand“ – the probe is next to the puncture site – or the puncture needle is guided by a special probe attachment.

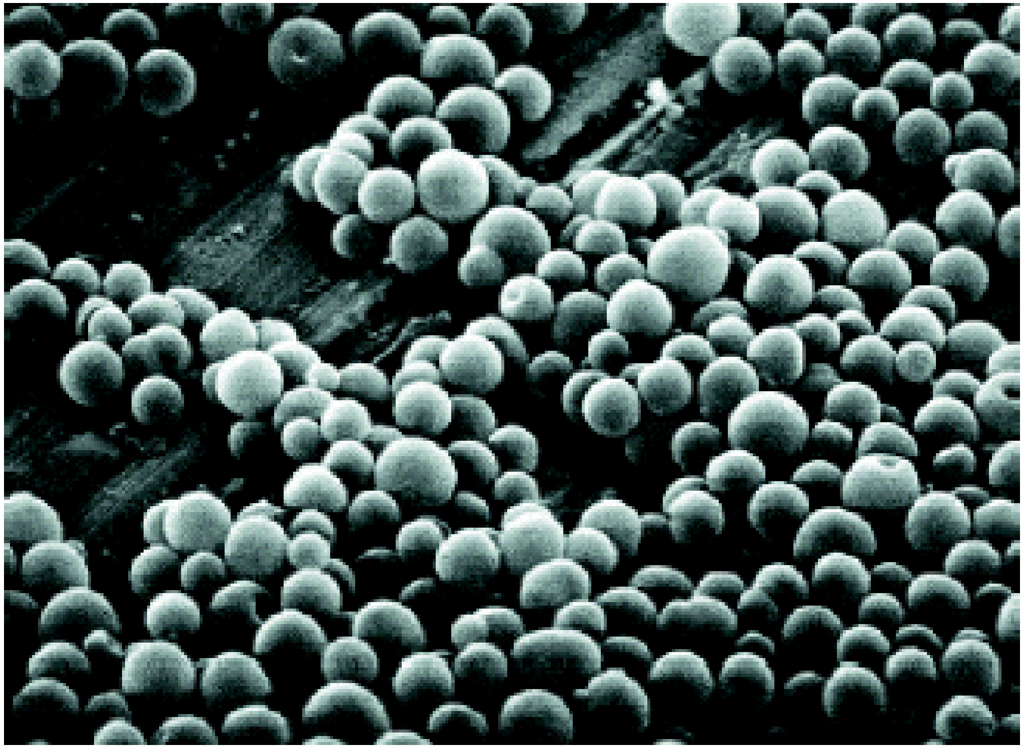
Ultrasonography

Echocontrast agents

- increase echogenity of streaming blood

Gas microbubbles
(mainly air or volatile hydrocarbons)

- free
- enclosed in biopolymer envelope

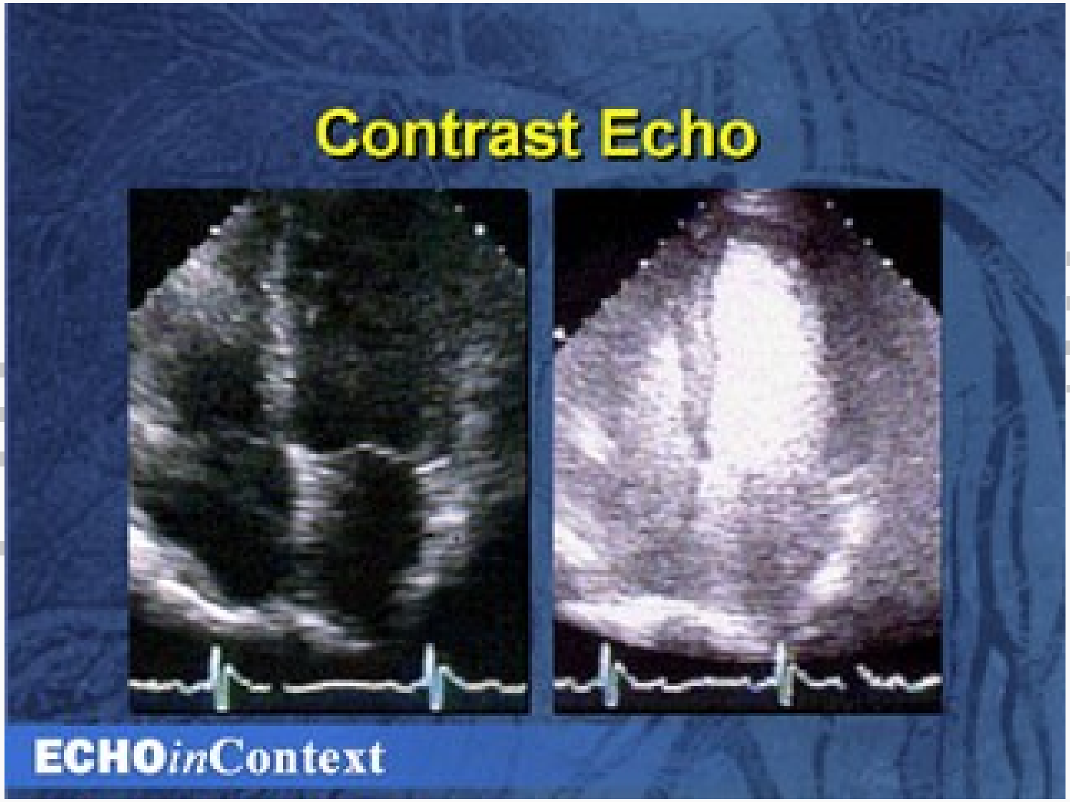


A SEM micrograph of encapsulated echocontrast agent

Ultrasonography

Echocontrast agents - application

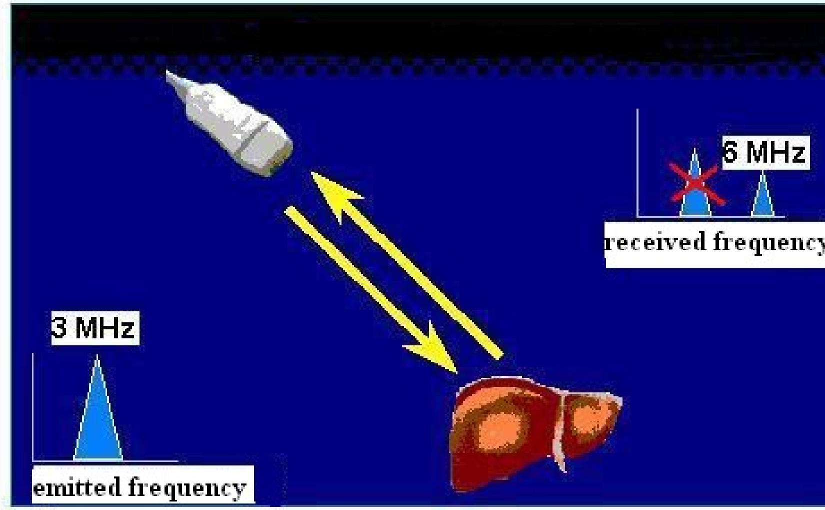
Enhanced demarcation of heart ventricle after application of the echocontrast agent

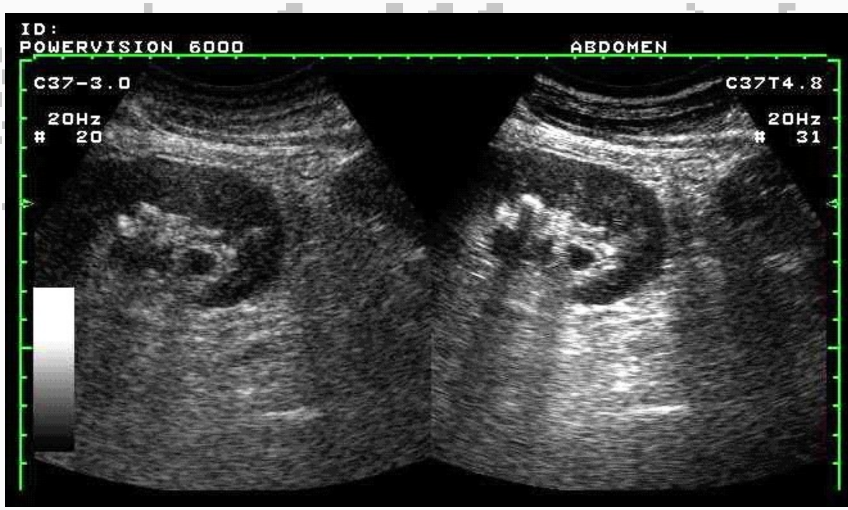


Ultrasonography

Harmonic imaging

An impulse with basic frequency f_0 is emitted into the tissue. The receiver, however, does not detect the reflected US with this same frequency but with the second harmonic frequency $2f_0$. Its source is tissue itself (advantage in patients „difficult to examine“). The method is **also** used with echocontrast agents – source of the second harmonic are oscillating bubbles. Advantageous when displaying blood supply of some lesions.

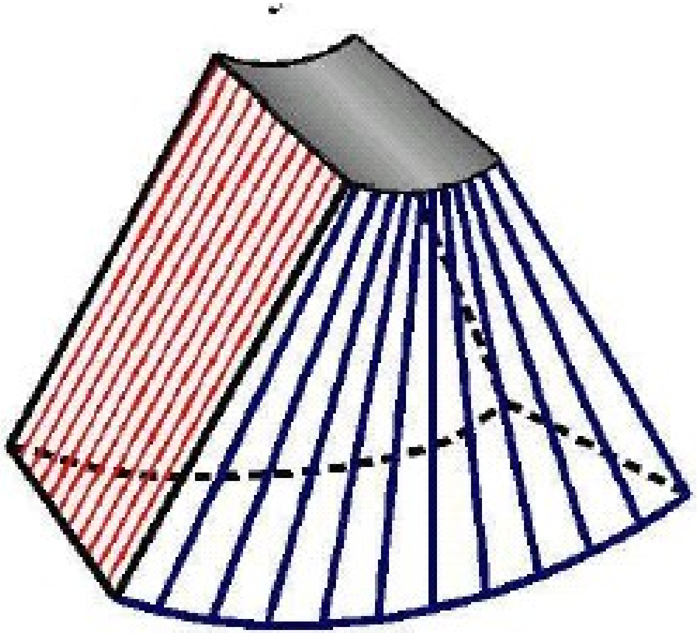




Conventional (left) and harmonic (right) images of a kidney with a stone.

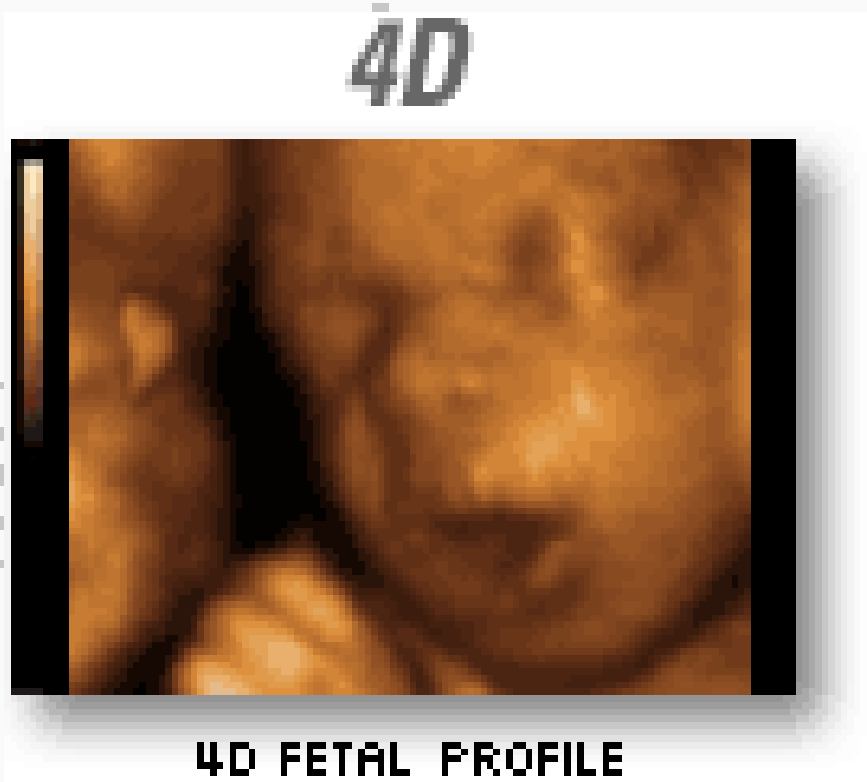
Ultrasonography
Principle of three-dimensional (3D) imaging

- The probe is linearly shifted, tilted or rotated.
The data about reflected signals in individual planes are stored in memory of a powerful PC which consequently performs **mathematical** reconstruction of the image.
Disadvantages of some 3D imaging systems: relatively **long time** needed for mathematical processing, **price.**



Four-dimensional (4D) image

The fourth dimension is time



Doppler flow measurement

Christian. A. Doppler (1803-1853), Austrian physicist and mathematician, formulated his theory in 1842 during his stay in Prague.

The Doppler effect (frequency shift of waves formed or reflected at a moving object) can be used for detection and measurement of blood flow, as well as, for detection and measurement of movements of some acoustical interfaces inside the body (foetal heart, blood vessel walls)

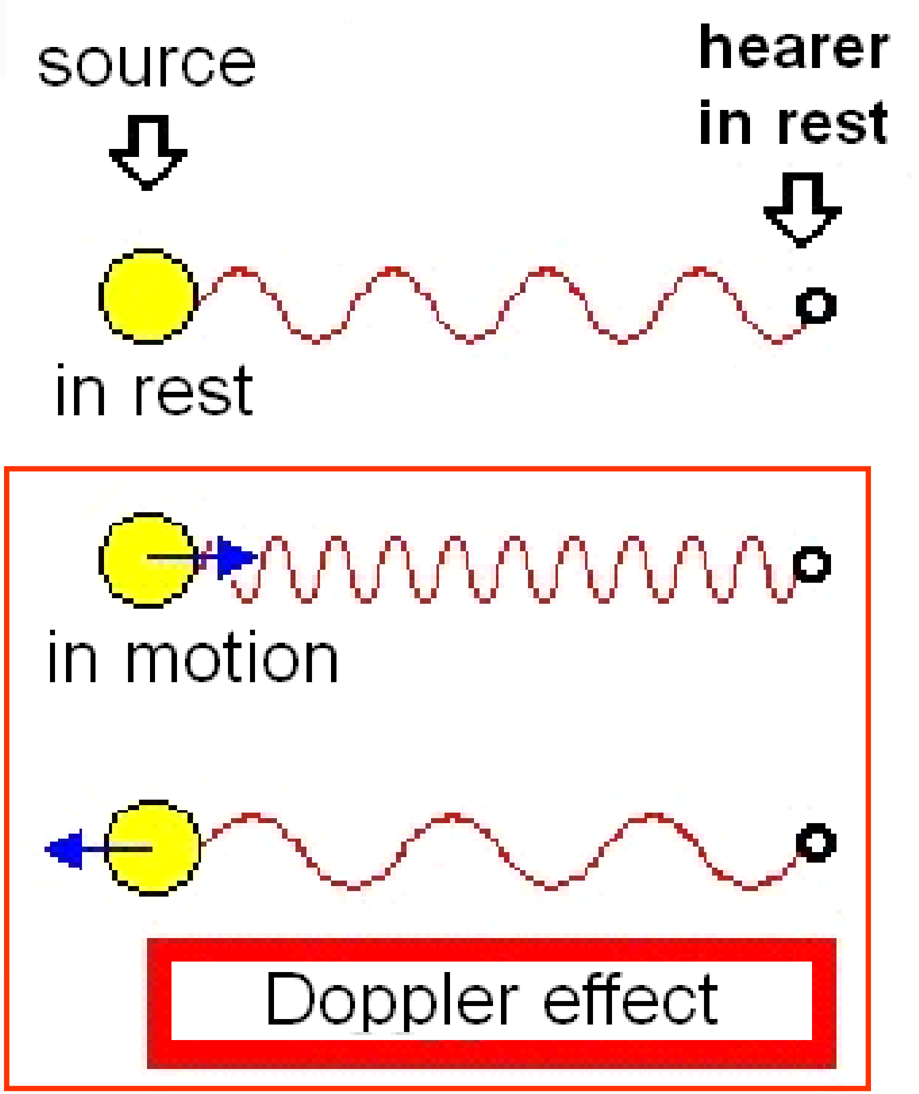
Doppler flow measurement

Principle of Doppler effect

perceived frequency corresponds with source frequency in rest

perceived frequency is higher when approaching

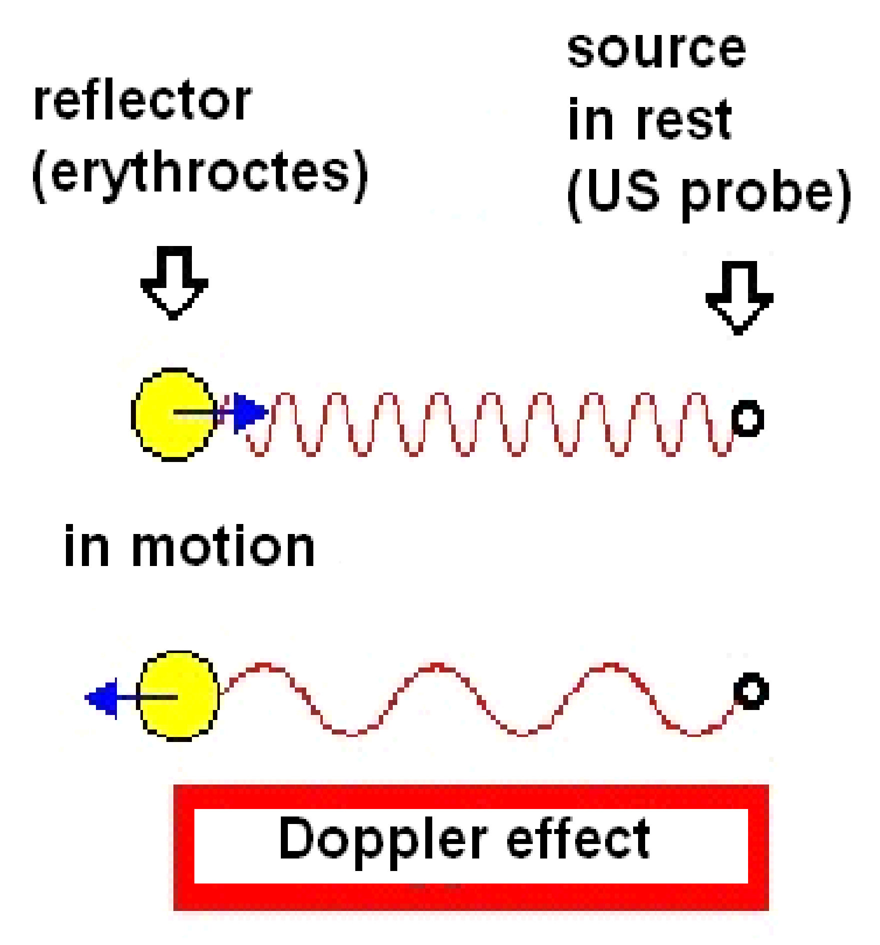
perceived frequency is lower when moving away



Principle of Doppler effect

Application of Doppler effect in blood flow velocity measurement

Moving reflector (back scatterer) = erythrocytes



Principle of blood flow measurement

US Doppler blood flow-meters

are based on the difference between the frequency of ultrasound (US) waves emitted by the probe and those reflected (back-scattered) by moving erythrocytes.

The frequency of reflected waves is (in comparison with the emitted waves)

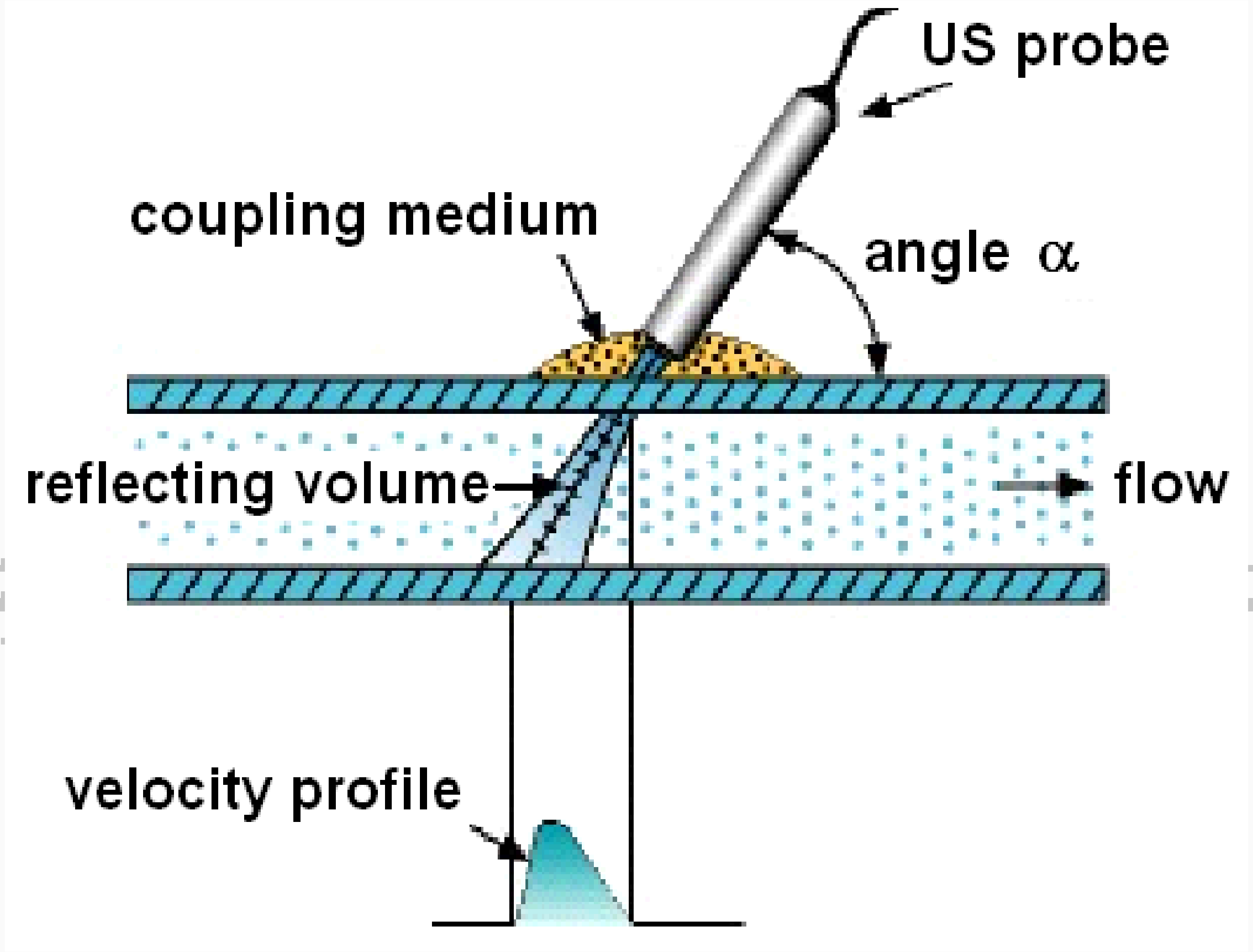
higher in forward blood flow (towards the probe)

lower in back blood flow (away from the probe)

The difference between the frequencies of emitted and reflected US waves **is proportional to** blood flow **velocity.**

Doppler flow measurement

General principle of blood flow measurement



Doppler flow measurement

1) Calculation of Doppler frequency change f_d
2) Calculation of „reflector“ (erythrocytes) velocity v

1) $$f_d = \frac{2 f_v \times v \times \cos\alpha}{c}$$

2) $$v = \frac{f_d \times c}{2 f_v \times \cos\alpha}$$

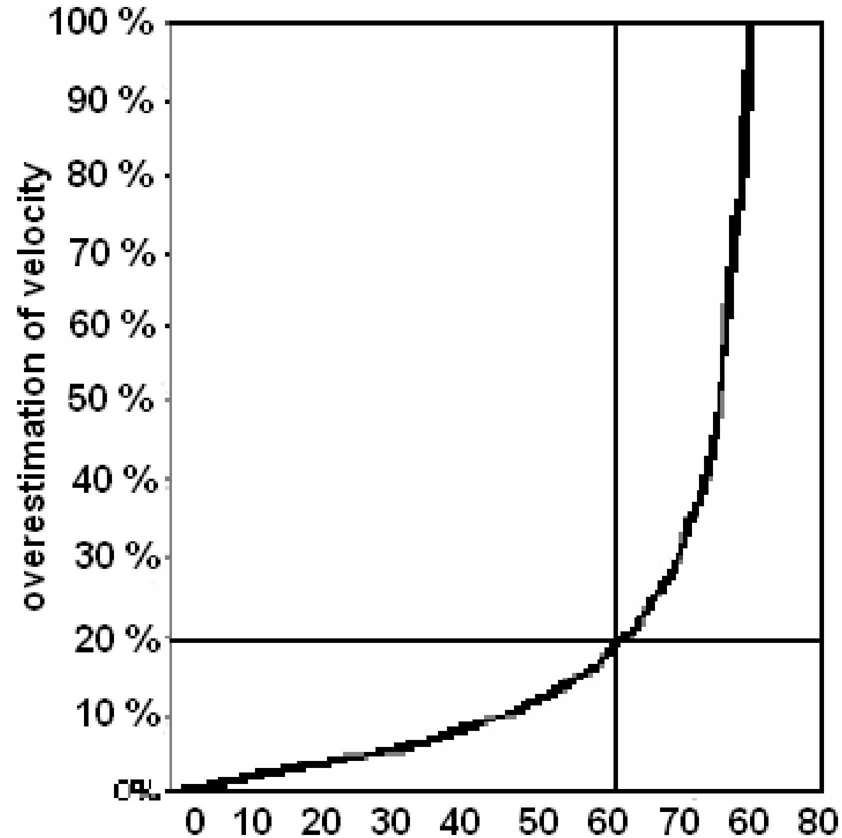
$\boldsymbol{f_v}$ - frequency of emitted US waves

α - angle made by axis of emitted US beam and the velocity vector of the reflector

c – US speed in the given medium (about 1540 m/s in blood)

Doppler flow measurement

Dependence of velocity overestimation on the incidence angle α (if the device is adjusted for $\alpha = 0$, i.e. $\cos\alpha = 1$)

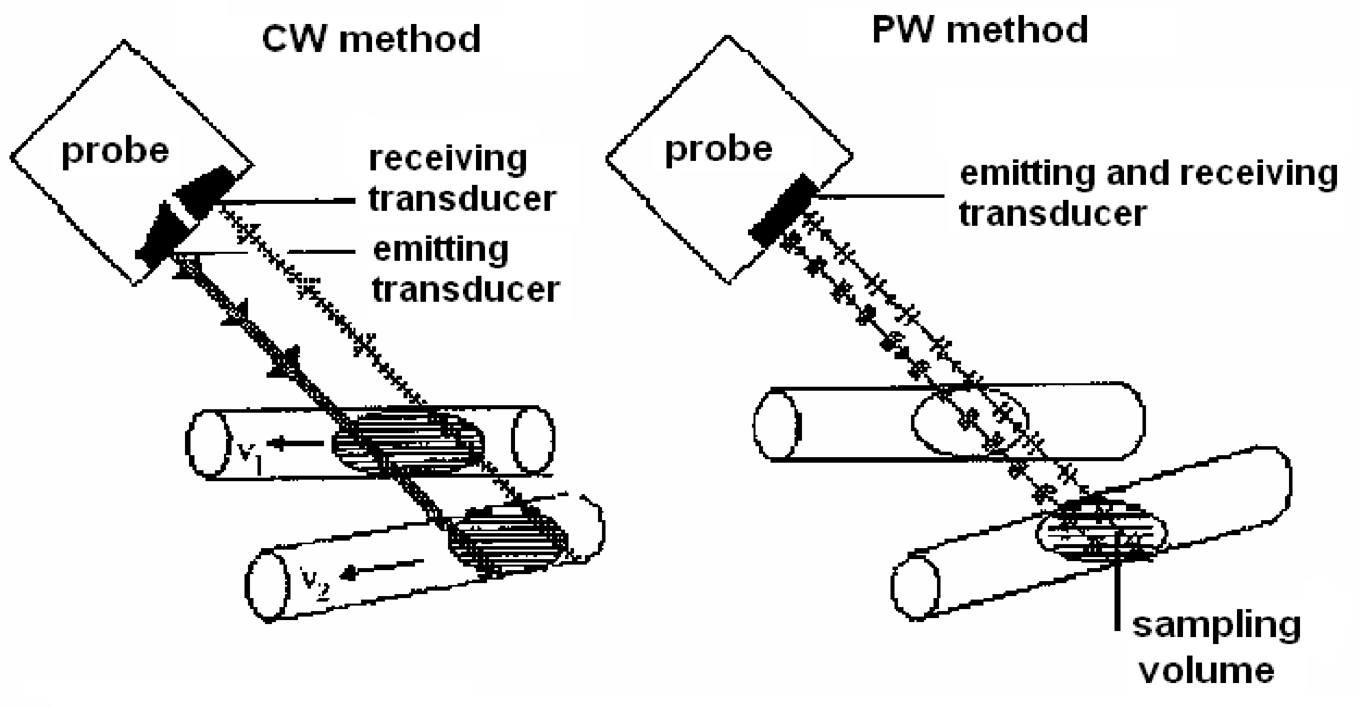


α - angle made by axis of emitted US beam and the velocity vector of the reflector

Doppler flow measurement

1) Systems with **continuous wave – CW**. They are used for **measurement on superficial blood vessels**. High velocities of flow can be measured, but without depth resolution. Used only occasionally.

2) Systems with **pulsed wave**. It is possible to measure blood flow with **accurate depth localisation.** Measurement of high velocities in depths is limited.



Doppler flow measurement

Systems with pulsed wave - PW

The probe has only **one transducer** which acts alternately as emitter and receiver.

The measurement of velocity and direction of blood flow in the vessel is evaluated in the so-called **sampling volume with adjustable size and depth.**

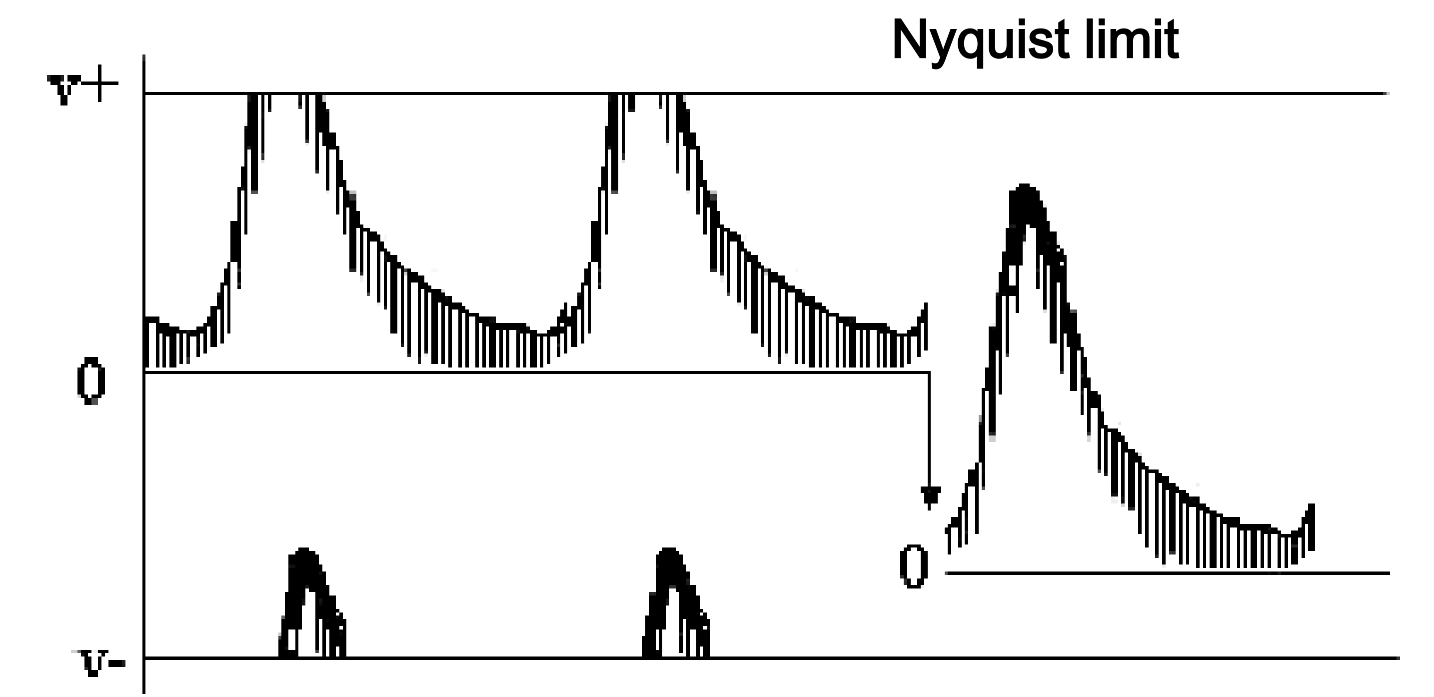
The pulse duration defines the size of the sampling volume (this volume should involve the whole diameter of the examined blood vessel).

Doppler methods

Pulse wave (PW) systems

Aliasing – at high repetition frequency of pulses the upper part of the spectral curve can appear in negative velocity range

- at velocity above 4m/s aliasing cannot be removed



Doppler methods

DUPLEX method

is a combination

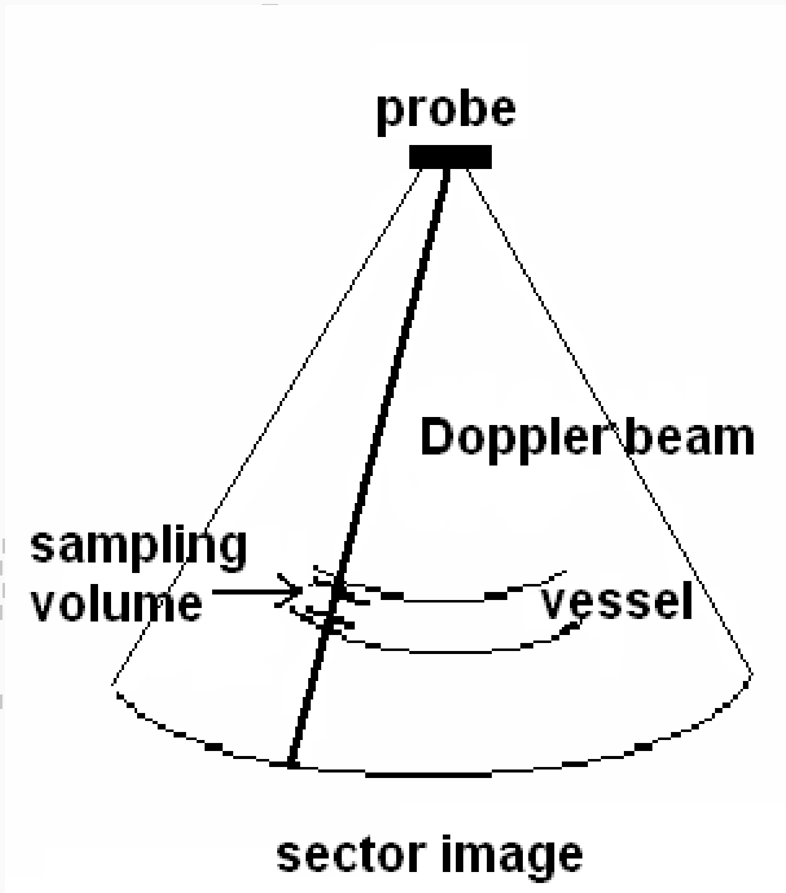
of dynamic B-mode imaging (the morphology of examined area with blood vessels is depicted)

and the PW Doppler system (measurement of velocity spectrum of blood flow).

It allows to examine blood flow inside heart or in deep blood vessels (flow velocity, direction and character)

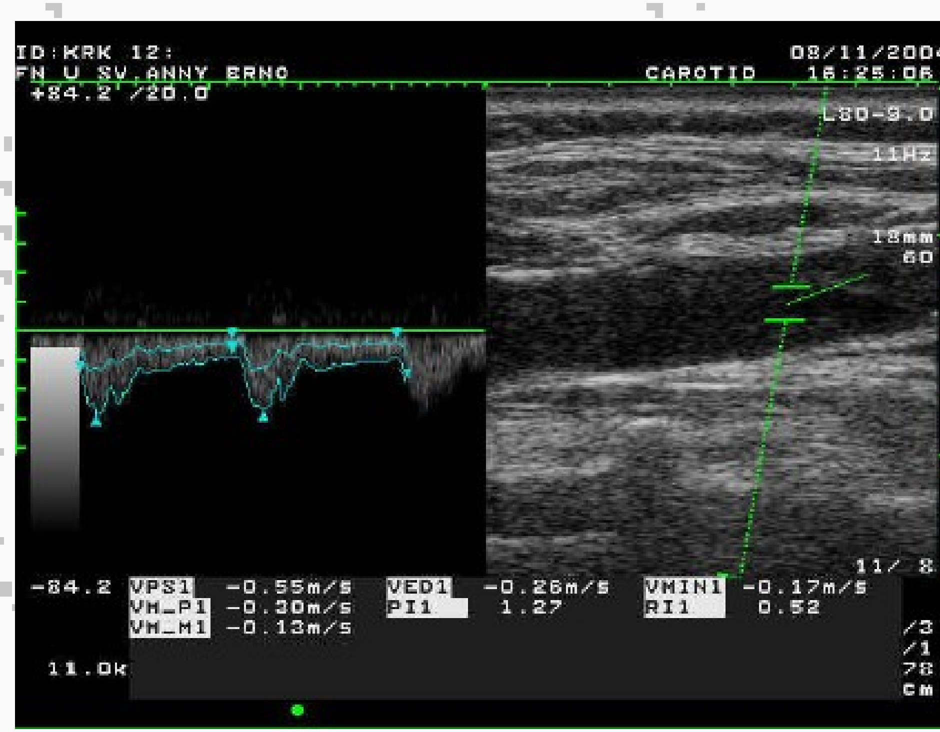
Doppler methods

Scheme: sector image with sampling volume



DUPLEX method

Image of carotid with spectral analysis of blood flow velocity



Doppler methods **DUPLEX method**

Placement of sampling volume (left) and the record of blood flow velocity spectrum in stenotic *a. carotis communis* (right)

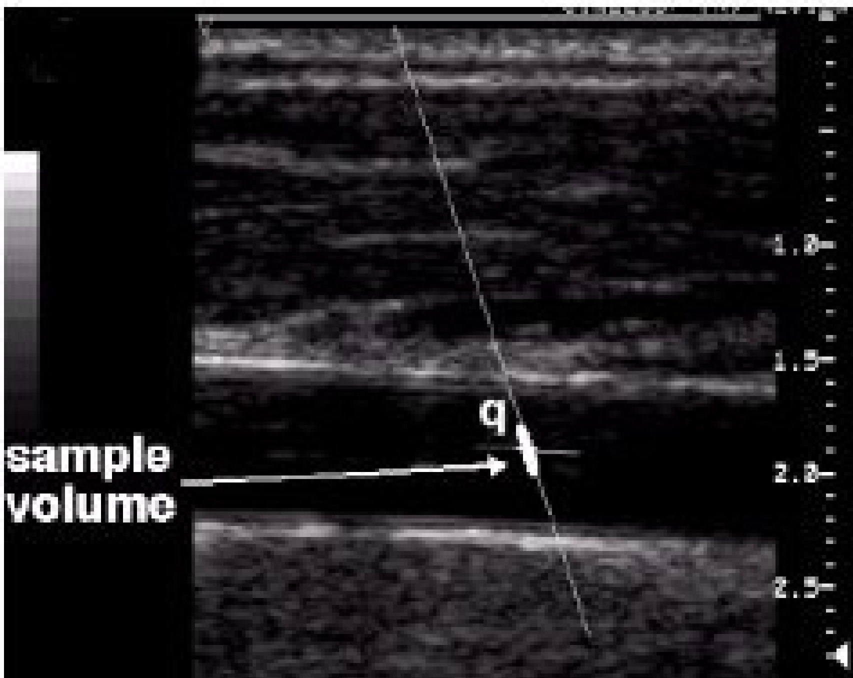


Figure 1

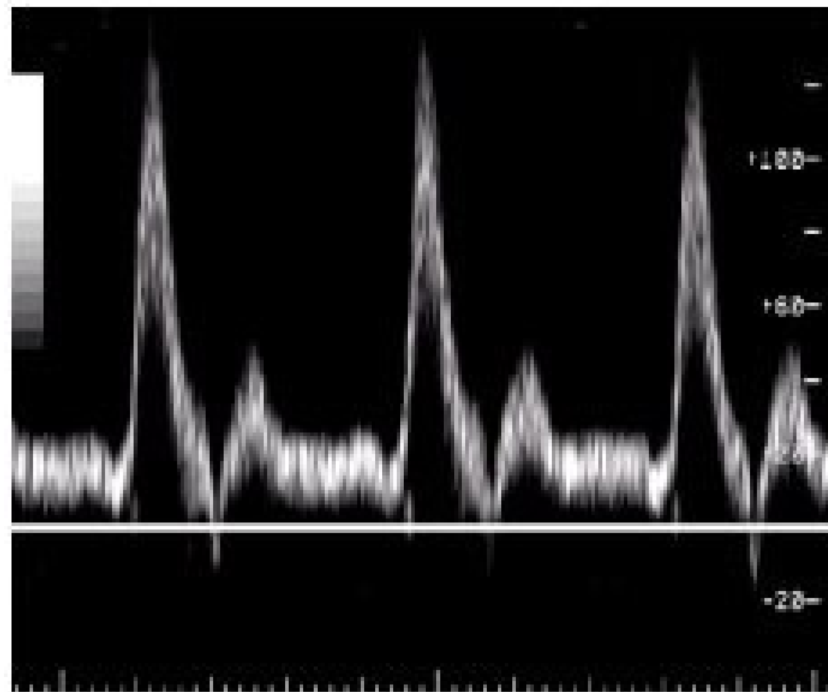


Figure 2

Doppler methods

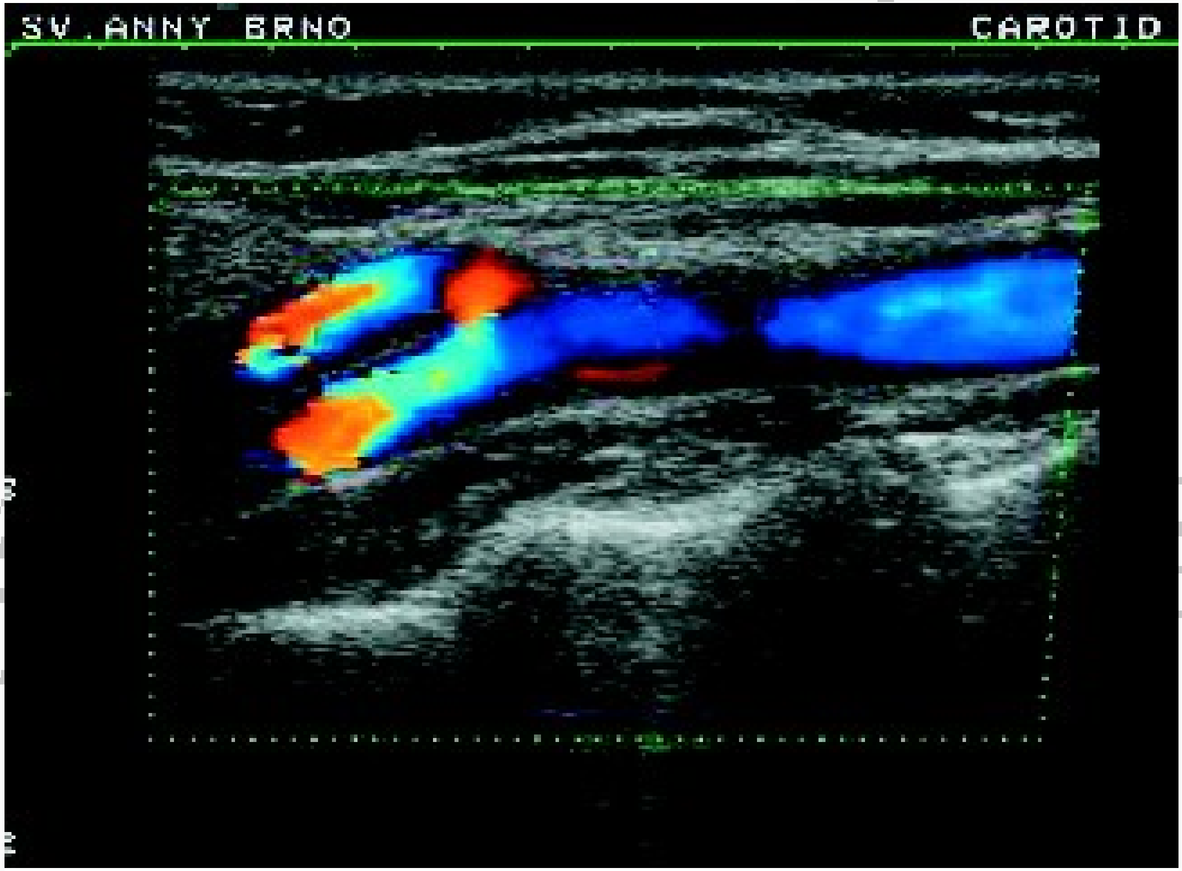
Colour Doppler imaging

The image consists of black-white and colour part.
The black-white part contains information about **reflectivity and structure** of tissues.
The colour part informs about **movements** in the examined section. (The colour is derived from average velocity of flow.)
The apparatus depicts distribution and direction of flowing blood as a two-dimensional image.
BART rule – blue away, red towards. The flow away from the probe is coded by blue colour, the flow towards the probe is coded by red colour. The brightness is proportional to the velocity, turbulences are depicted by green patterns.

Doppler methods

Colour Doppler imaging

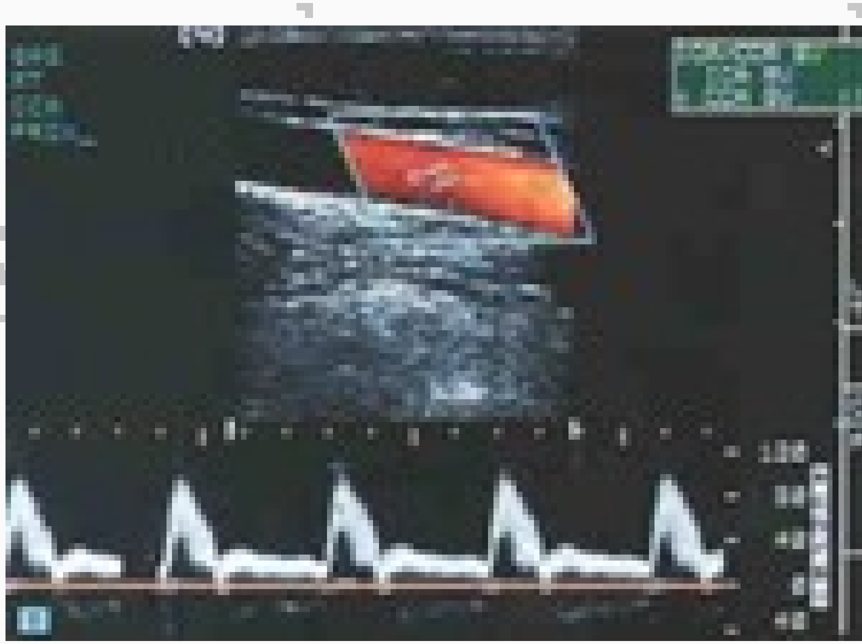
Carotid bifurcation

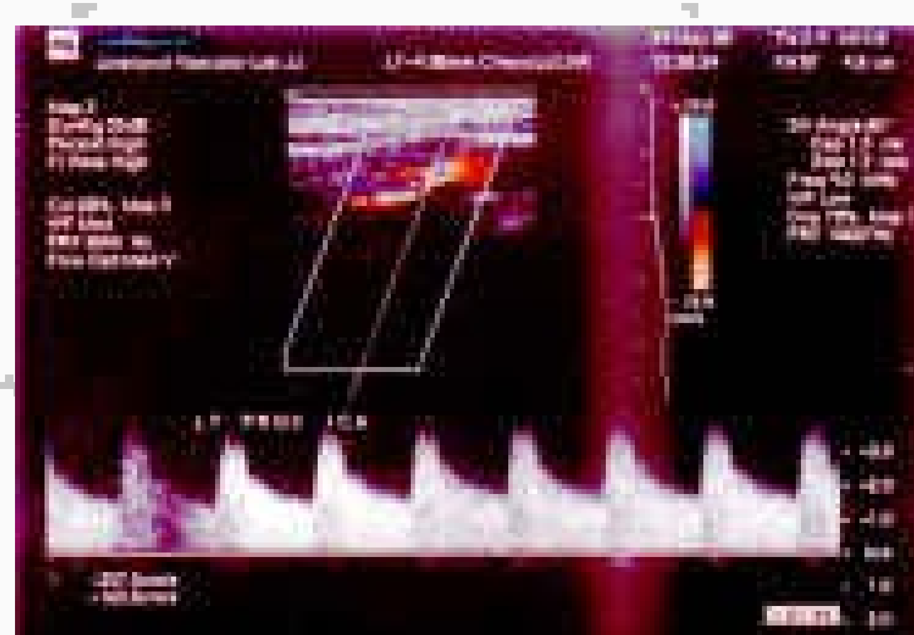


Doppler methods **TRIPLEX method**

A combination of duplex method (B-mode imaging with PW Doppler) and color flow mapping

Normal finding of blood flow in *a. carotis communis* (left) and about 90%-stenosis of *a. carotis interna* (right)





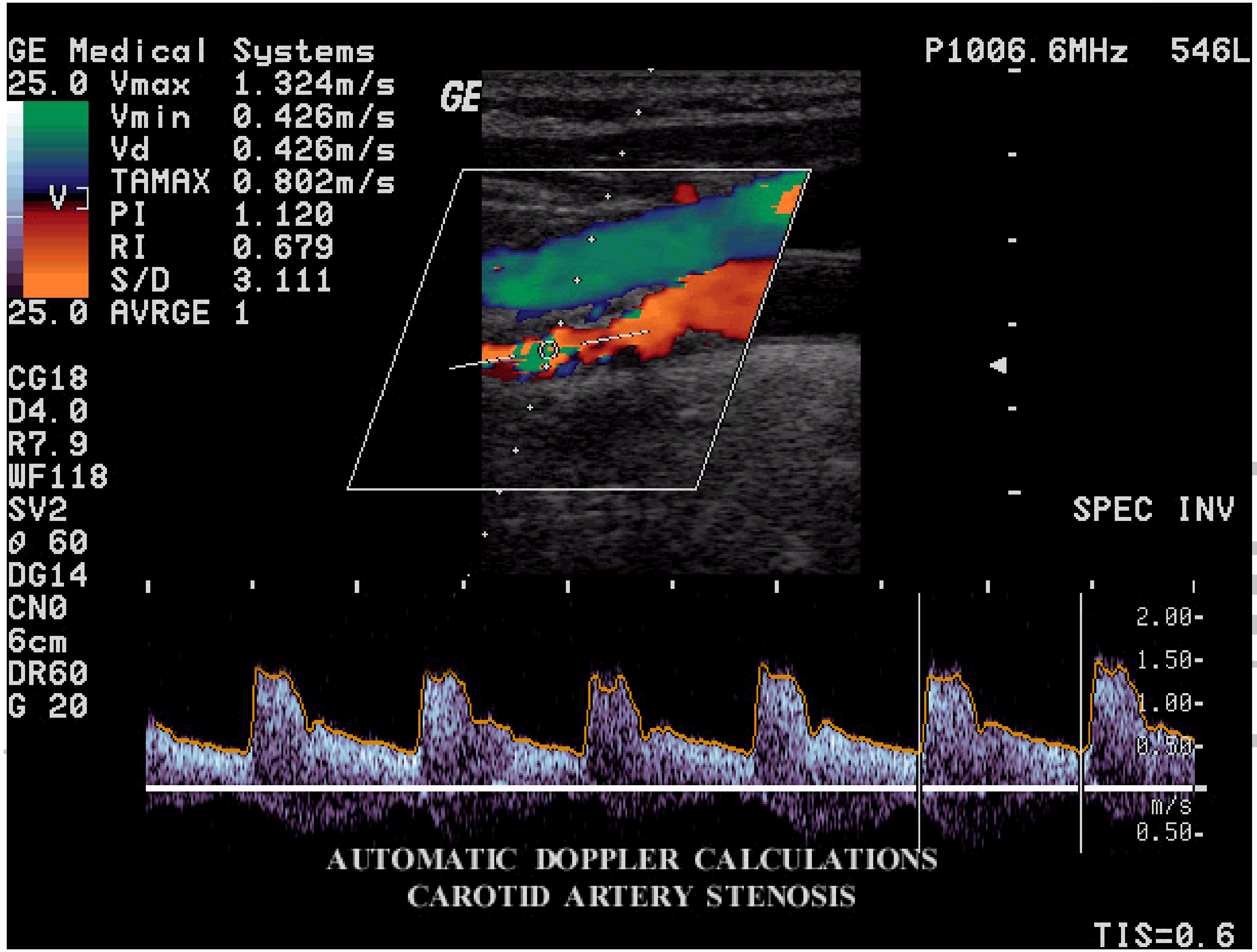
Doppler methods

TRIPLEX method

stenosis

of

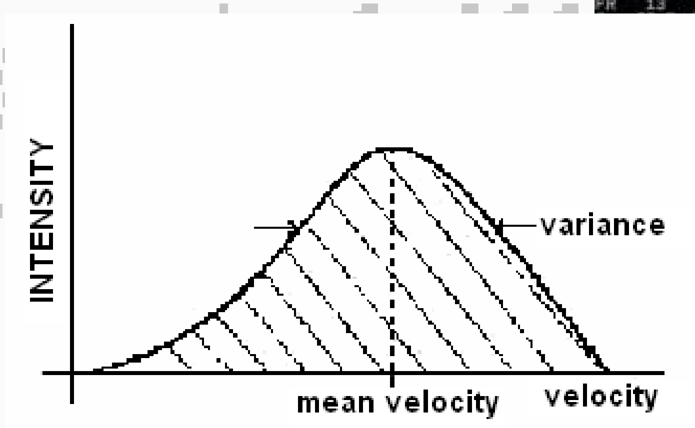
a. carotis



Doppler methods

POWER DOPPLER method

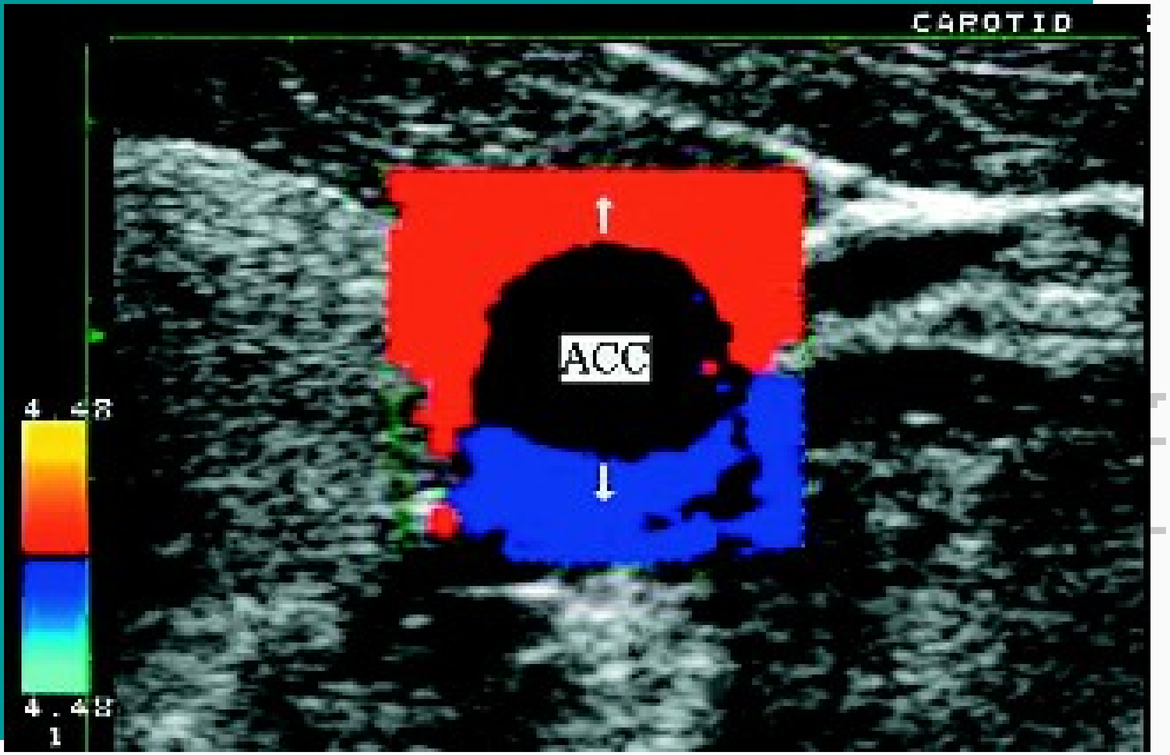
- the whole energy of the Doppler signal is utilised
- **mere detection of blood flow only little depends on the so-called Doppler incidence angle**
- **imaging of even very slow flows** (blood perfusion of tissues and organs)
- flow direction is not shown



Tissue Doppler Imaging (TDI)

Colour coding of information about velocity and direction of movements of tissues

Velocities 1-10 mm/s are depicted.



TDI of *a. carotis communis* during systole

Ultrasonic densitometry

It is based on both the measurement of speed of ultrasound in bone and the estimation of ultrasound attenuation in bone. In contrast to X-ray methods, ultrasound densitometry also provides information on the structure of bone and its elastic properties.

- The speed of ultrasound depends on the density and elasticity of the measured medium. The anterior area of the tibia and the posterior area of the calcaneus are frequently used as places of measurement. The speed of ultrasound is given by the quotient of measured distance and the transmission time.

- Ultrasound attenuation depends on the physical properties of the given medium and the frequency of the ultrasound applied. For the frequency range 0.1 - 1 MHz the frequency dependence is nearly linear. Attenuation is currently expressed in dB/MHz/cm.
- Clinical importance: diagnostics of osteoporosis

Ultrasonic densitometry

Ultrasound measurements used to assess bone density at the calcaneus

Patient Safety: reducing Ultrasound ‘Doses’

Prudent use of Ultrasound

- US is non-ionising BUT since many bioeffects of ultrasound have not yet been studied fully, ‘prudent’ use is recommended
- ALARA – as low as reasonably achievable (exposure)

- In practice ‘prudent’ = justification + optimisation

Biological Effects

- Possible bioeffects: inactivation of enzymes, altered cell morphology, internal haemorrhage, free radical formation …
- Mechanisms of bioeffects:
  - Mechanical effects
    - Displacement and acceleration of biomolecules
    - Gas bubble **cavitation** (stable and transient) – see the lecture on biological effects of ultrasound
  - Elevated tissue temperatures (absorption of ultrasound and therefore increase in temperature high in lungs, less in bone, least in soft tissue)
- All bioeffects are deterministic with a threshold (cavitation) or without it (heating).

Output Power from Transducer

- varies from one machine to another
- Increases as one moves from real-time imaging to colour flow Doppler
- M-mode output intensity is low but dose to tissue is high because beam is stationary

Risk Indicators

- To avoid potentially dangerous exposures, two indices were introduced. Their values (different for different organs) are often displayed on device screens and should not be exceeded.
- Thermal Index (TI): TI = possible tissue temperature rise if transducer is kept stationary
  - TIS: soft tissue path
  - TIB: bone near focus of beam
  - TIC: Cranium (near surface bone)
- Mechanical Index (MI): measure of possible mechanical bioeffects

More on the TI and MI

Thermal index – device power divided by the power that would increased the temperature by one degree under conditions of minimum heat loss (without perfusion).

Mechanical index (for assessment of cavitation-conditioned risk, increased danger when using echocontrast agents):

$$MI = I_{UZ} / \sqrt{f} \quad \left[W.cm^{-2}, MHz\right]$$

Justification

- No commercial demos on human subjects
- No training on students
- No ‘see baby just for fun’ or excessive screening in obstetrics

Optimisation of ‘Dose’ 1

- **Minimise TI and MI and use appropriate index (TIS, TIB, TIC), care in cases when these underestimate**
- **Check acoustic power outputs on manual**
- **Use high receiver gain when possible as opposed to high transmit power**
- **Start scan with low transmit power and increase gradually**

Optimisation of ‘Dose’ 2

- Avoid repeat scans and reduce exposure time
- Do not hold transducer stationary
- Greater care when using contrast agents as these increase the possibility of cavitation
- Exceptional care must be taken in applying pulsed Doppler in obstetrics
- Regular quality control of the ultrasound device

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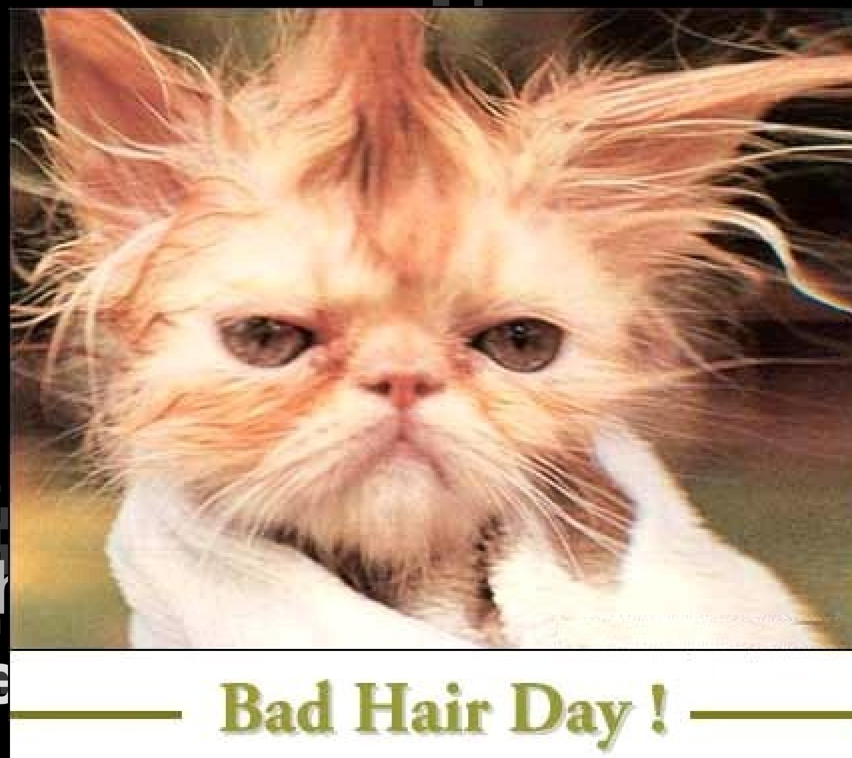
language

revision:



Graph

Lucie

Last revision: June 2009