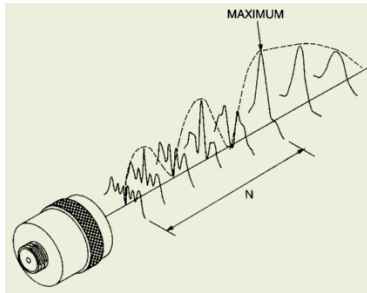


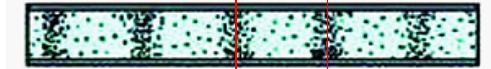
Physics of ultrasonography



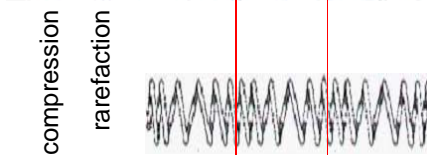
KAD 2018.02.22

Sound: mechanical wave (model)

whistle

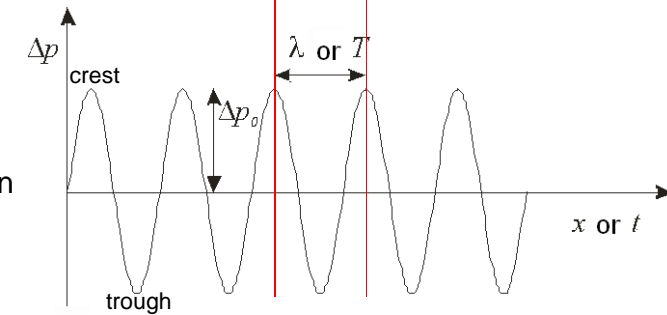


spring



spatial and
temporal
periodicity

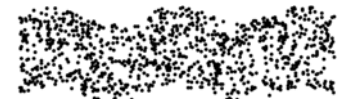
function



2



longitudinal wave
(in the interior of liquids and
gases only this type)



transverse wave

hydrostatic pressure pressure change,
pressure sound pressure

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

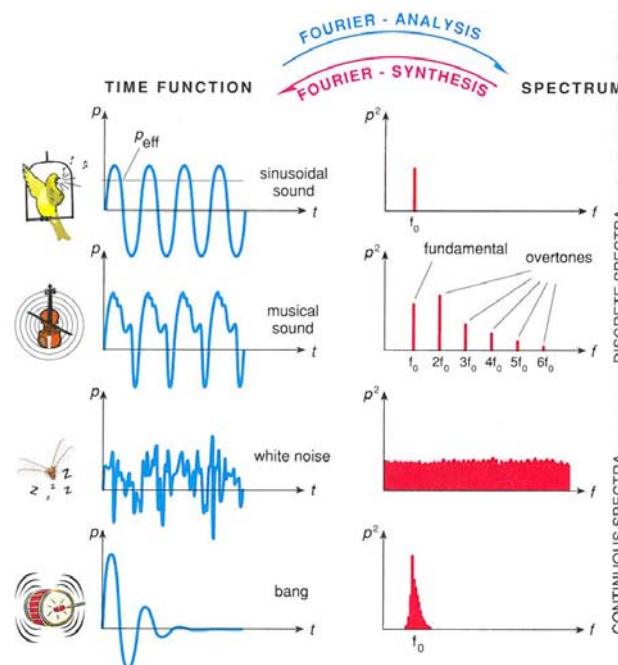
pressure DC + AC amplitude phase

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

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



3

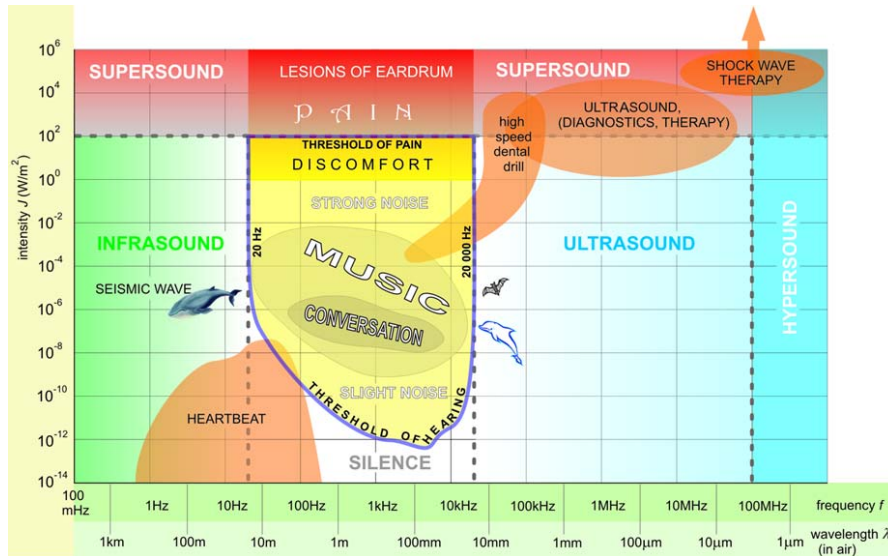


pitch:
frequency of the
fundamental

timbre (tone colour):
relative strengths of
overtones/harmonics
(spectrum)

4

Frequency and intensity regions of sounds



Laboratory manual, Audiometry

5

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)

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

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

acoustic impedance
(useful form)



6

Supplementary material

Acoustic impedance (way to the useful form)

$$y = y_{\max} \sin\left[\omega\left(t - \frac{x}{c}\right)\right]$$

$$p = p_{\max} \sin\left[\omega\left(t - \frac{x}{c}\right)\right]$$

$$\frac{\Delta y}{\Delta t} = v = y_{\max} \omega \cos\left[\omega\left(t - \frac{x}{c}\right)\right]$$

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

$$\frac{\Delta v}{\Delta t} = a = -y_{\max} \omega^2 \sin\left[\omega\left(t - \frac{x}{c}\right)\right]$$

$$\rho y_{\max} \omega^2 \sin\left[\omega\left(t - \frac{x}{c}\right)\right] = \frac{\Delta p}{\Delta x}$$

$$-\rho y_{\max} \omega c \cos\left[\omega\left(t - \frac{x}{c}\right)\right] = p$$

$$p_{\max} = \rho y_{\max} \omega c = \rho v_{\max} c$$

$$ma = F$$

$$\frac{1}{V} ma = \frac{1}{A \Delta x} F$$

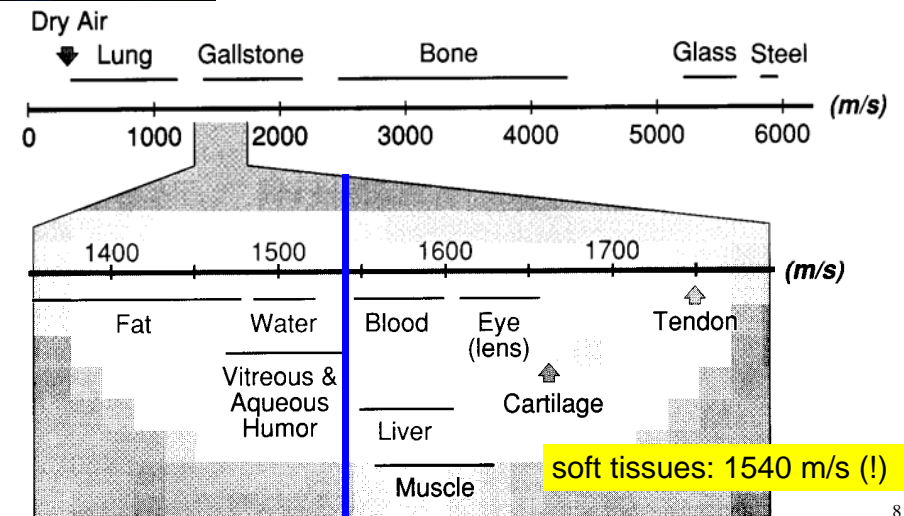
$$\rho a = -\frac{\Delta p}{\Delta x}$$

$$\frac{p_{\max}}{v_{\max}} = \rho c = Z$$

7

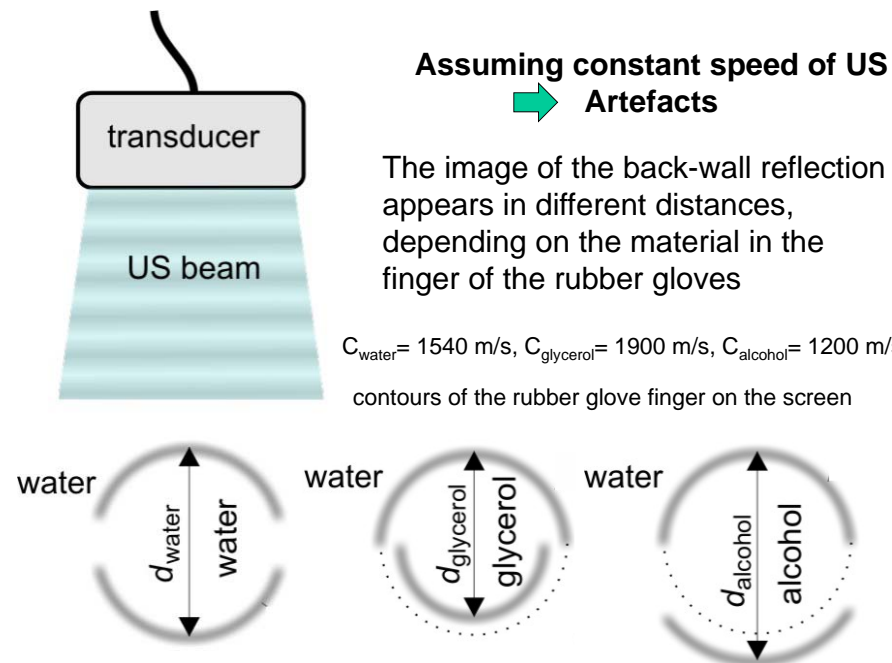


Speed of sound/US in different media



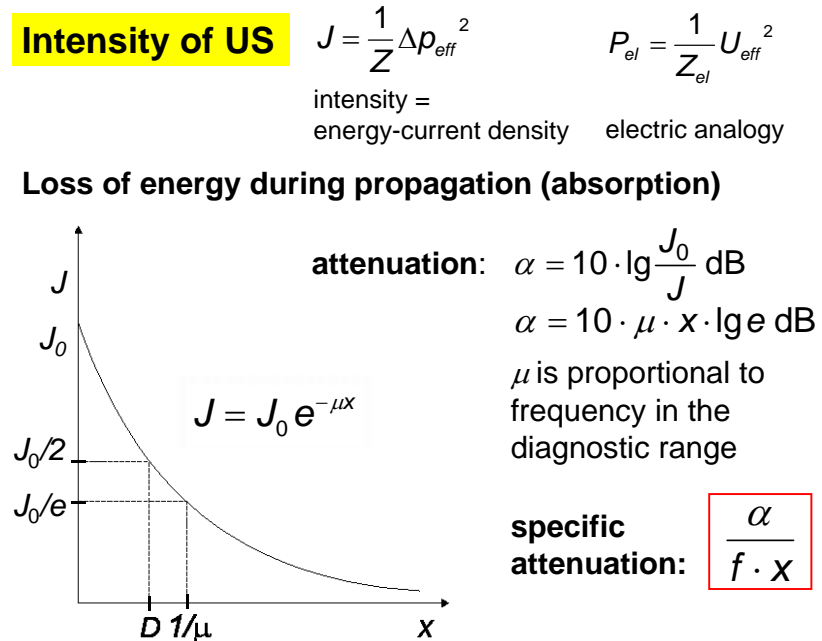
soft tissues: 1540 m/s (!)

8



Laboratory manual, US, Figure 15

9



10

μ is proportional to frequency in the diagnostic range

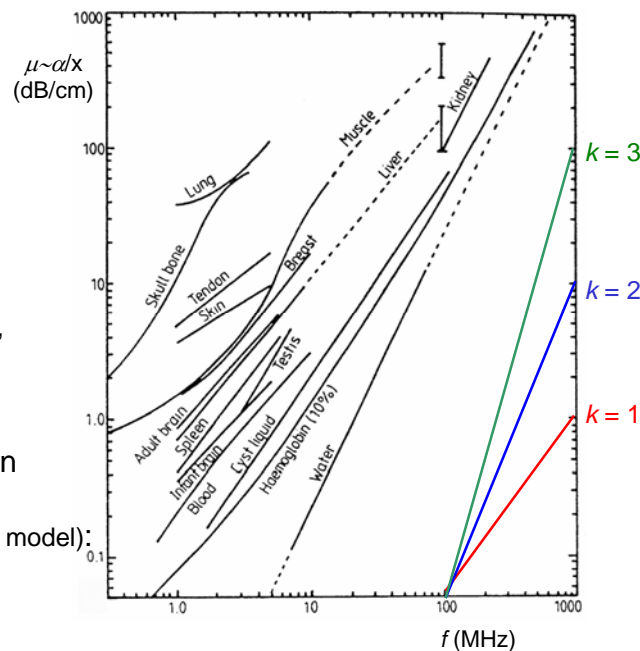
$$\mu \sim f^k, \quad k \sim 1(?)$$

$$\lg \mu \sim k \lg f$$

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

specific attenuation for soft tissues (homogeneous tissue model):

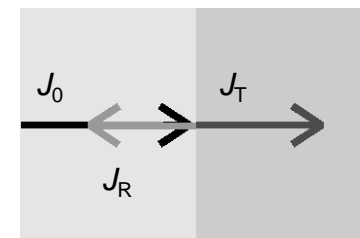
$$\frac{\alpha}{f \cdot x} \sim 1 \frac{\text{dB}}{\text{cm MHz}}$$



11

Phenomena at the boundary of different media

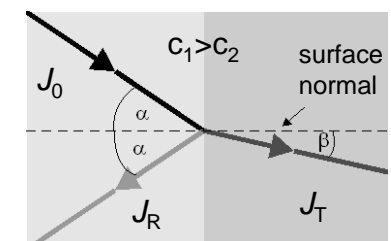
normal/perpendicular incidence



$$J_0 = J_R + J_T$$

reflection and transmission (penetration)

skew incidence



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

Snellius-Descartes

12

Reflection (normal incidence)

reflectivity:

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

“full” reflection:

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

optimal coupling:

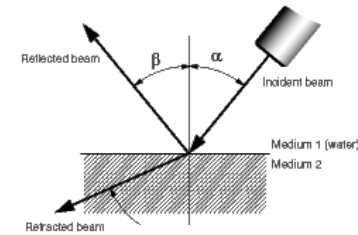
$$Z_{\text{connecting}} \approx \sqrt{Z_{\text{source}} Z_{\text{skin}}}$$



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

13

Supplementary material



Skew incidence

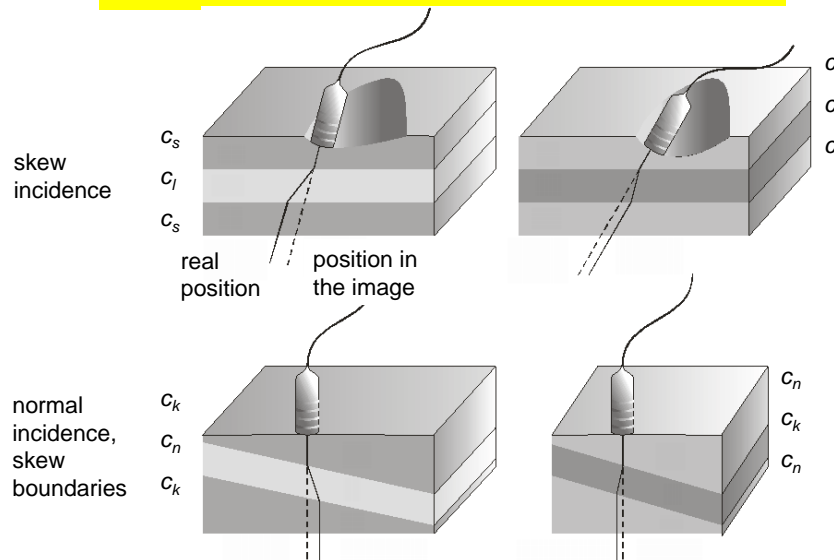
$$\text{Reflexion coefficient} = R = \left[\frac{Z_2 \cos \alpha - Z_1 \cos \gamma}{Z_2 \cos \alpha + Z_1 \cos \gamma} \right]^2$$

$$\text{Refraction coefficient} = D = \frac{4 Z_1 Z_2 \cos^2 \alpha}{(Z_2 \cos \alpha + Z_1 \cos \gamma)^2}$$

$$\text{Refraction angle} = \gamma = \text{asin} \left(\frac{c_2 \sin \alpha}{c_1} \right)$$

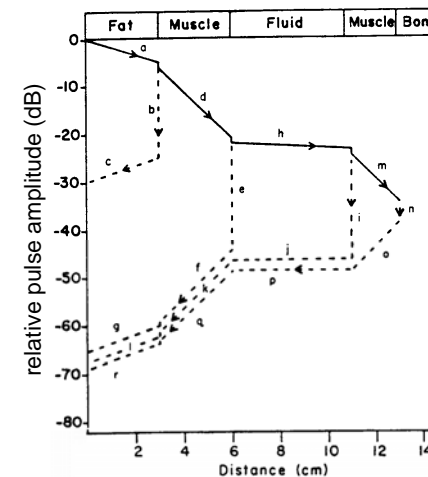
14

Phenomenon of skew incidence or normal incidence and skew boundaries



Biophysics textbook, Fig. on pg. 153

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

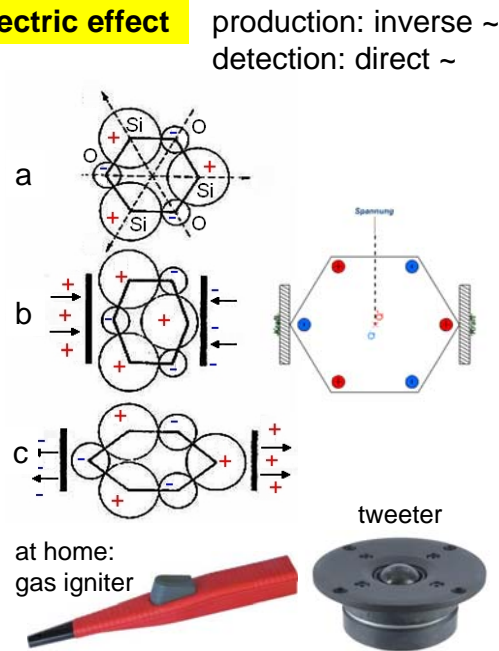
16

Generation of US. Piezoelectric effect

source of electric signal
(sine wave oscillator)+
transducer (piezo-crystal)

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

(b) and (c) As a result of
pressure, the charge
centers are separated, i.e.
a potential difference
arises (direct ~).
The crystal is deformed
when voltage is applied
(inverse ~).

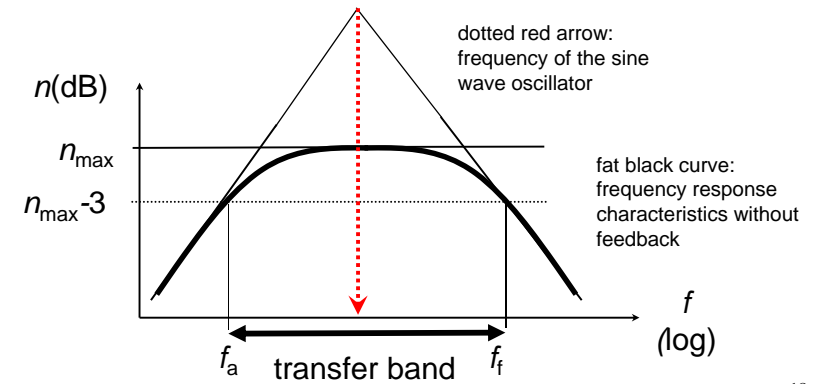


Source of electric signal : sine wave oscillator

amplifier with positive
feedback

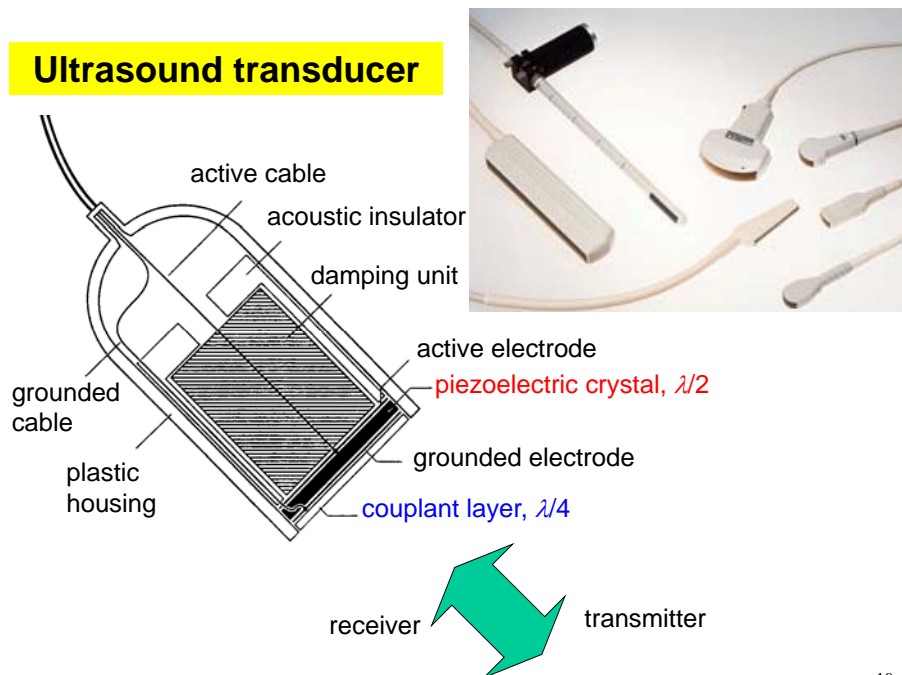
$$A_{U, \text{feedback}} = \frac{A_U}{1 - \beta A_U}$$

$\beta A_U = 1$, amplification = „infinity“ → sine wave oscillator
no input signal, output signal: sine voltage



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Ultrasound transducer

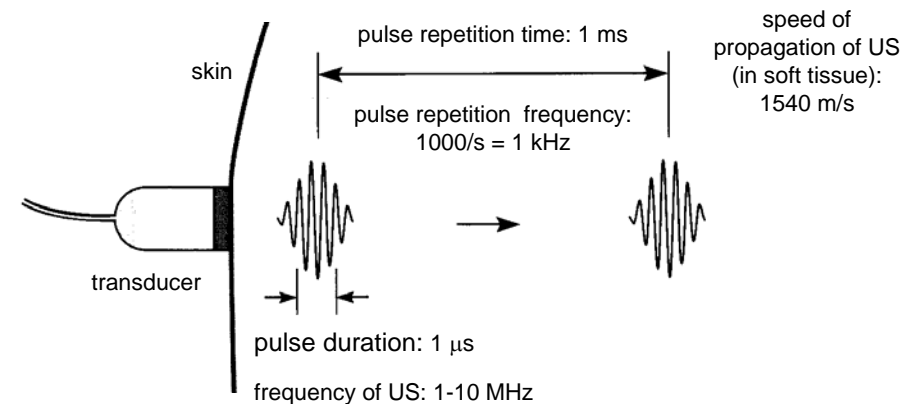


19

Characteristic of US pulses

transducer: transmitter and receiver is the same unit

time sharing mode: pulses instead of continuous wave US



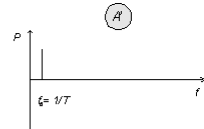
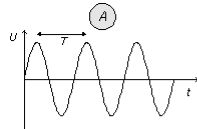
Biophysics textbook, Fig. VIII.32.

20

Time function

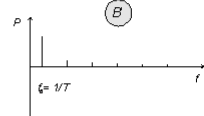
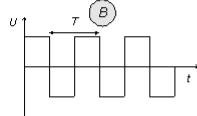
Spectrum

sine function



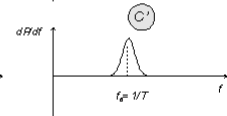
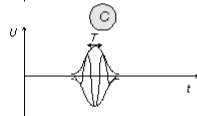
line spectrum (1 line)

square function



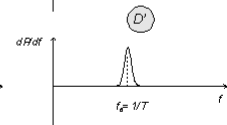
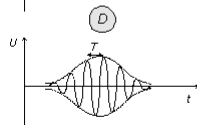
line spectrum

sine wave pocket (some „periods“)



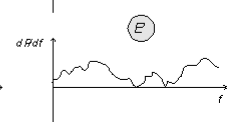
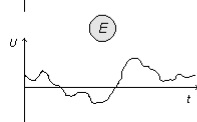
band spectrum

sine wave pocket (several „periods“)



band spectrum

aperiodic function

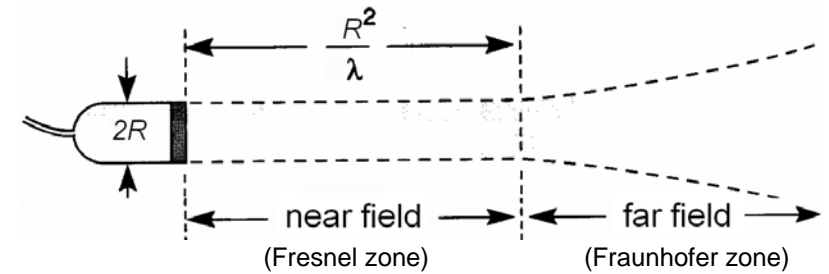


continuous spectrum

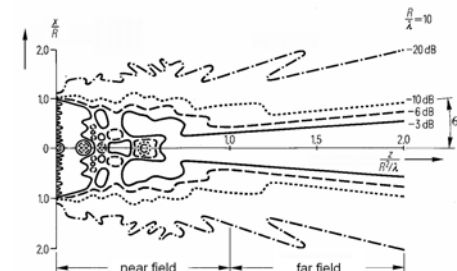
US pulse

21

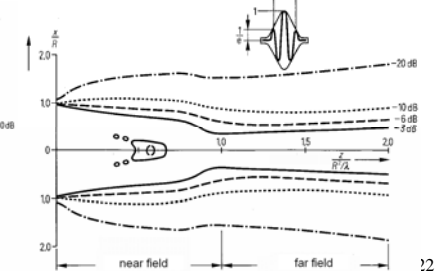
US beam shape (simplified version)



Beam shape, continuous wave US

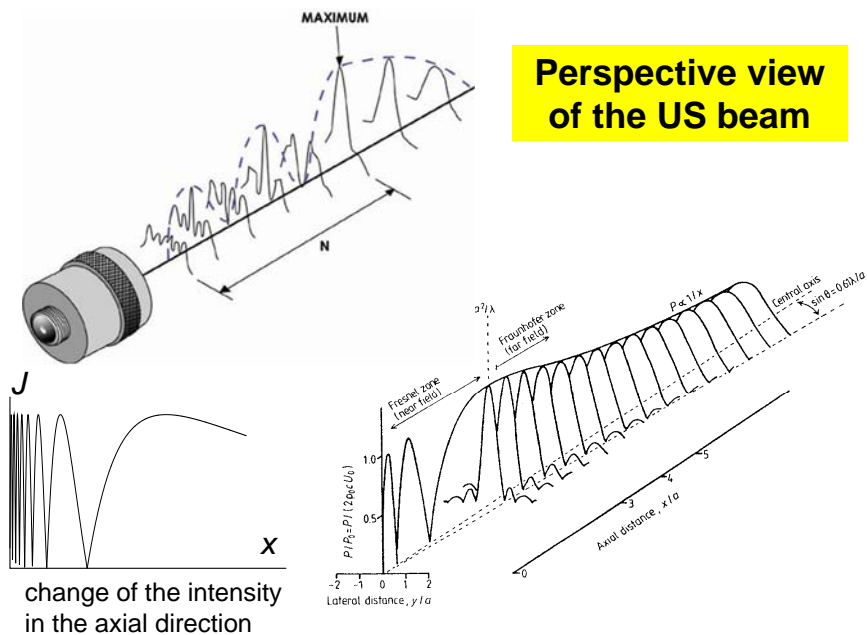


Beam shape, pulsed wave US



22

Perspective view of the US beam



23

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

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

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

24

Axial resolving limit

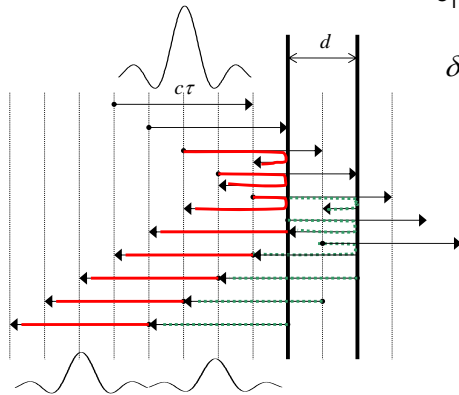
τ : pulse duration

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

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

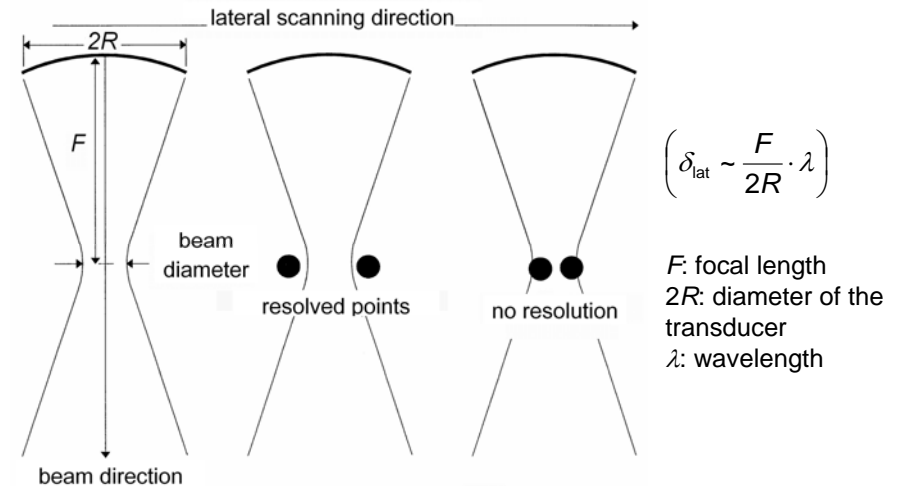
The axial resolving limit is the half of the pulse length. The echos from the adjacent surfaces in this case just hit another.

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



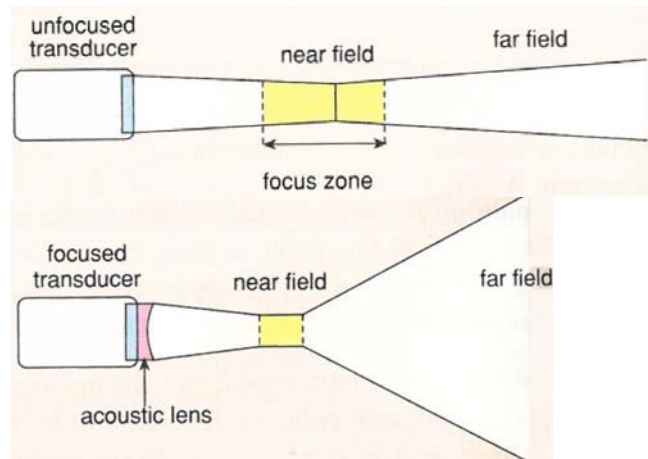
25

Lateral resolving limit



26

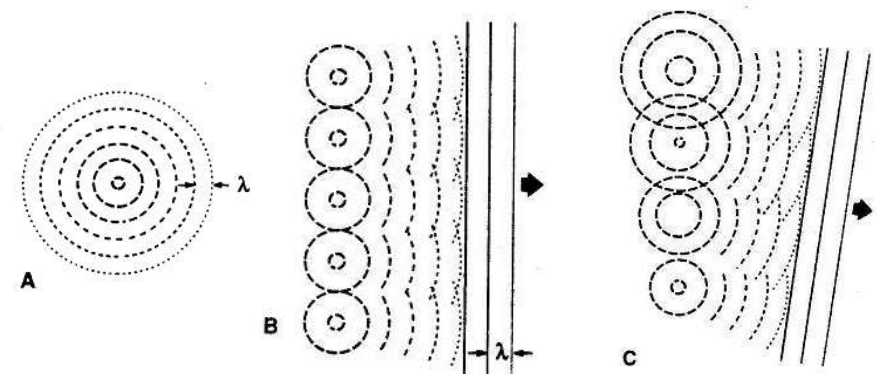
Focusing of the beam



Focusing increases the divergence of the beam in the far field regime and reduces the depth sharpness.

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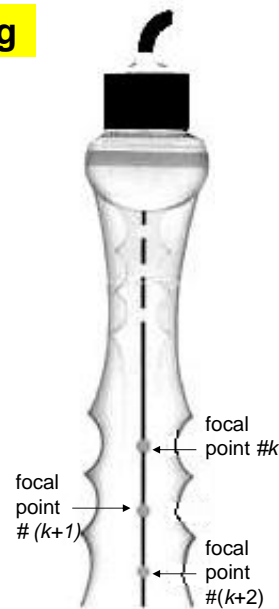
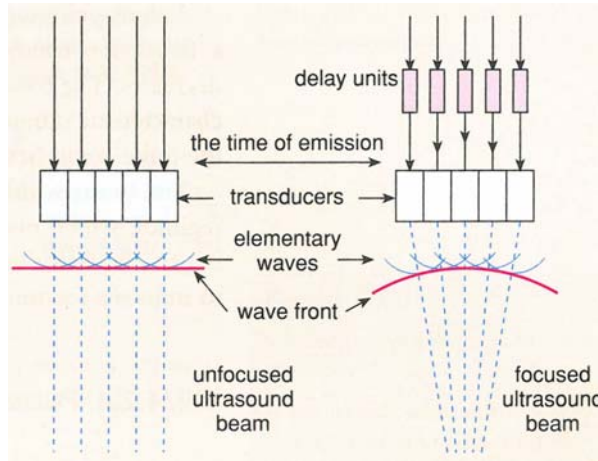
Huygens' principle



Any wave propagates so, that each point on a primary wavefront serves as the source of spherical secondary wavelets that advance with a speed and frequency equal to those of the primary wave. The primary wavefront at some later time is the envelope of these wavelets.

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Electronic focusing

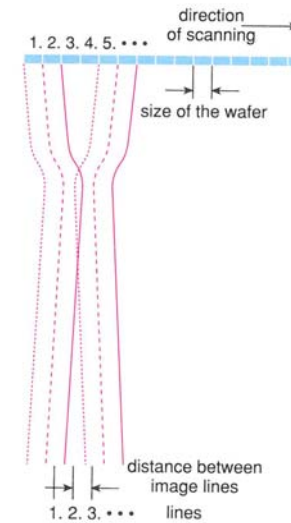


cf. Textbook Fig. on p.507

29

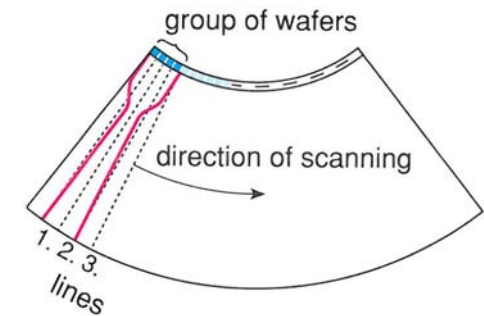
Scanning

multi unit linear array



cf. Textbook Fig. VII. 36-37

multi unit curved array

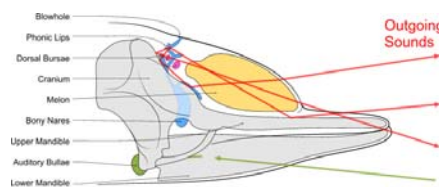
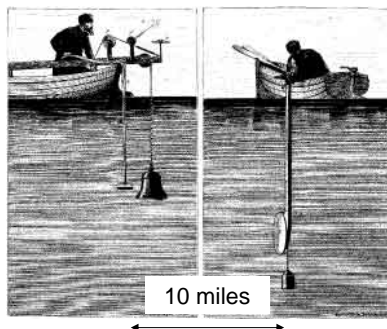


30

Echo principle

1794 Spallanzani:
bat's navigation

1822 Colladen
measured the speed of
sound in water

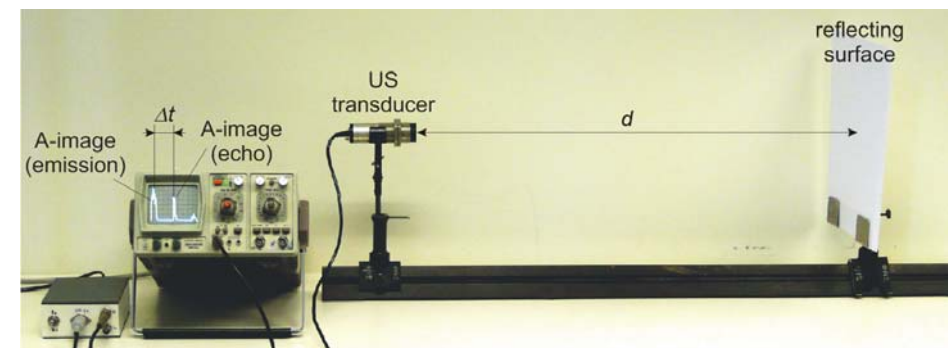


bottlenose dolphin

31

Echo principle

using a special US-head, short pulses are emitted in the air towards a reflecting surface, and the same US-head detects the echo signal

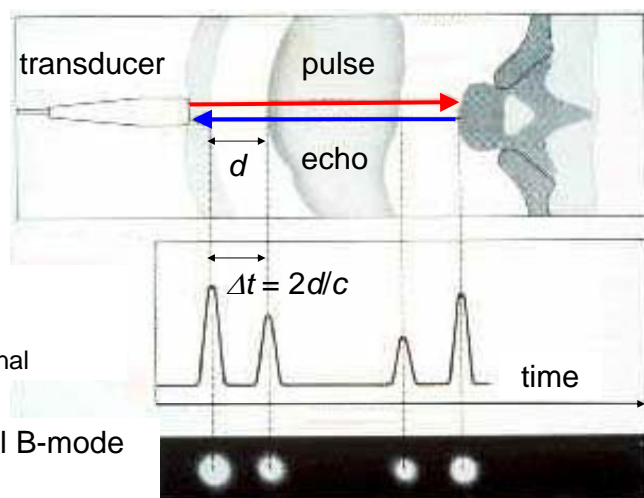


$$c\Delta t = d + d = 2d$$

32



Receiving the echos



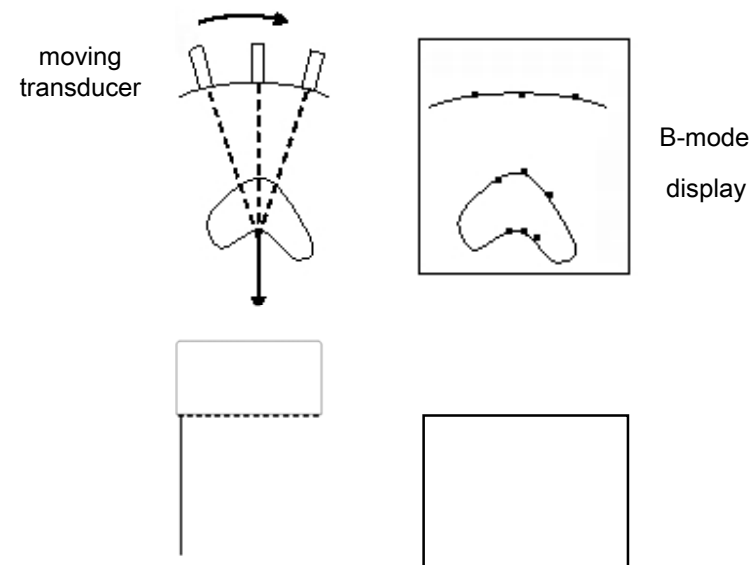
A-mode
(**A**mplitude)
only 1-dimensional

1-dimensional B-mode
(**B**rightness)

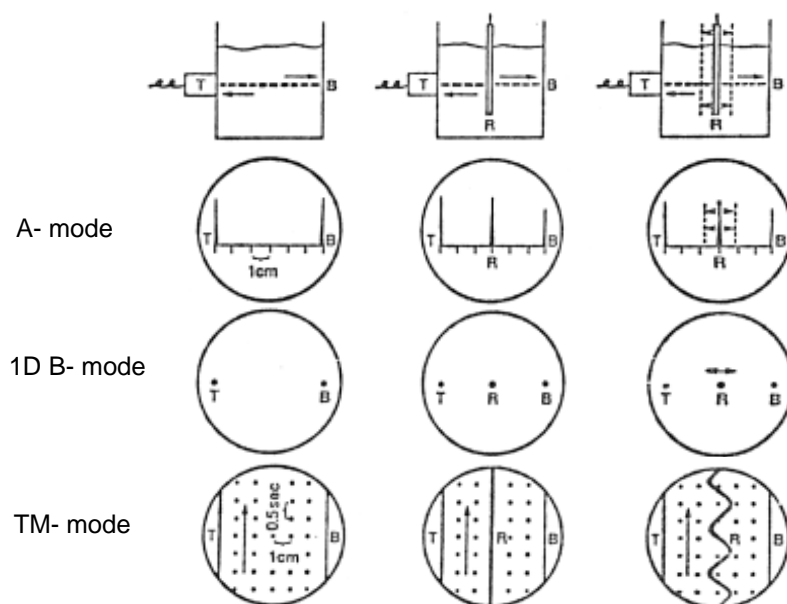
cf. Textbook Fig. VIII.33

33

2-dimensional B-mode

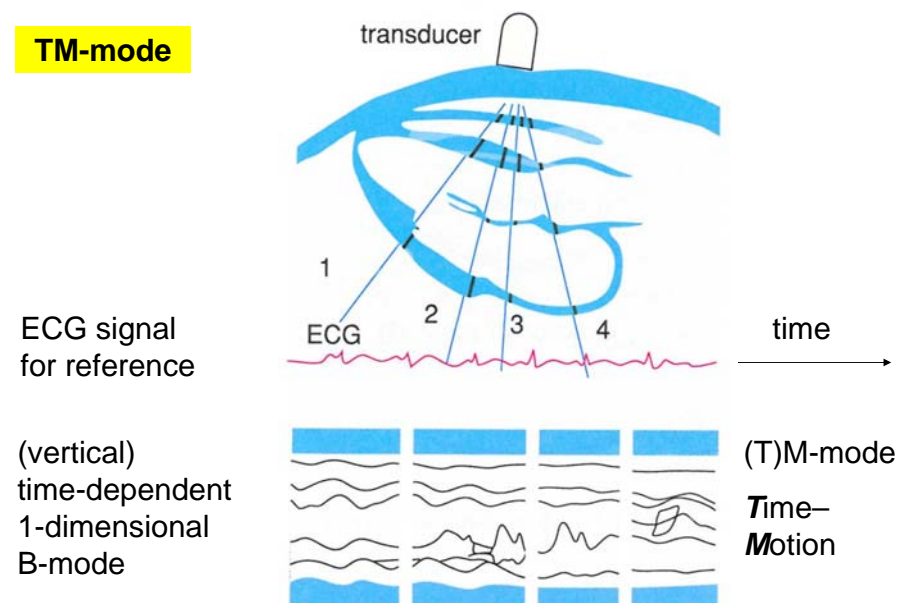


34



35

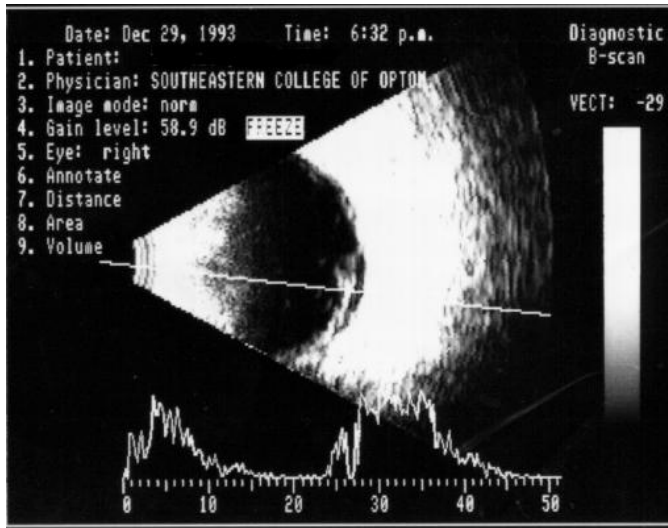
TM-mode



Textbook Fig. VIII.34

36

2-dimensional B-mode and A-mode (used in ophthalmology)



real speed of propagation for the accurate determination of distances:

cornea: 1641 m/s

aqueous humour: 1532 m/s

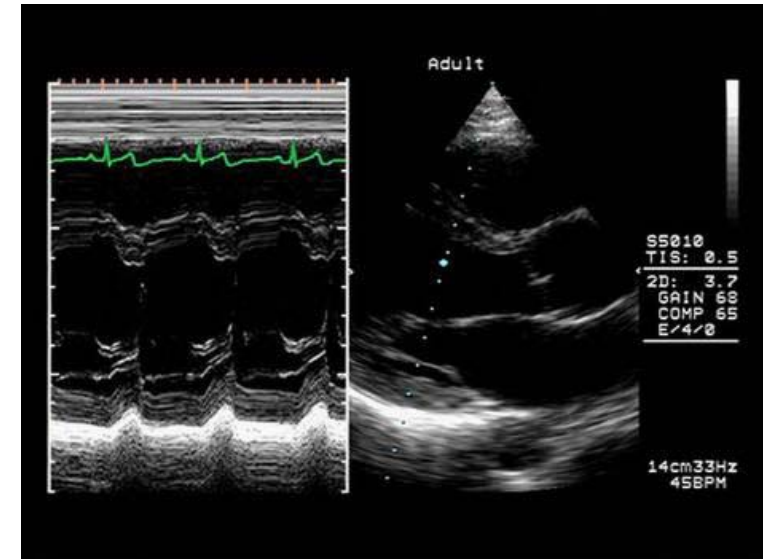
crystalline lens: 1641 m/s

vitreous body: 1532 m/s

37

TM-mode

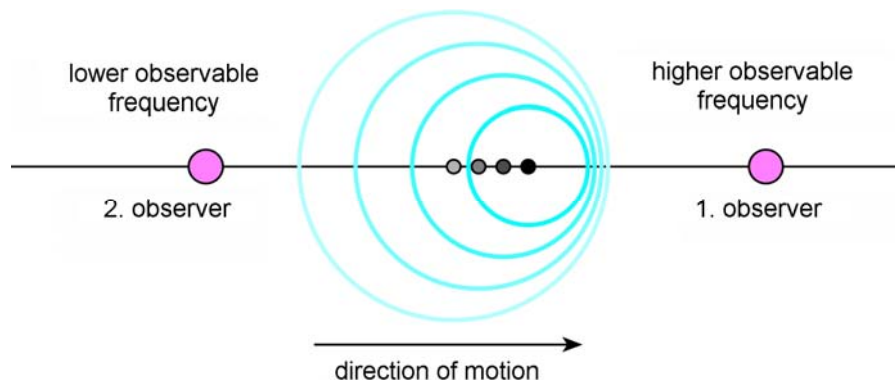
B-mode



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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)



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

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

- (b) moving source and standing observer
 (if $v_s \ll c$, 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 $v_R \ll c$)

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

40

Doppler frequency = frequency change = frequency shift

if $v_i, v_R \ll c$ (i= S or O)

rearranging equation (a)
moving source or observer:

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

rearranging equation (d)
**moving reflecting object
or surface:**

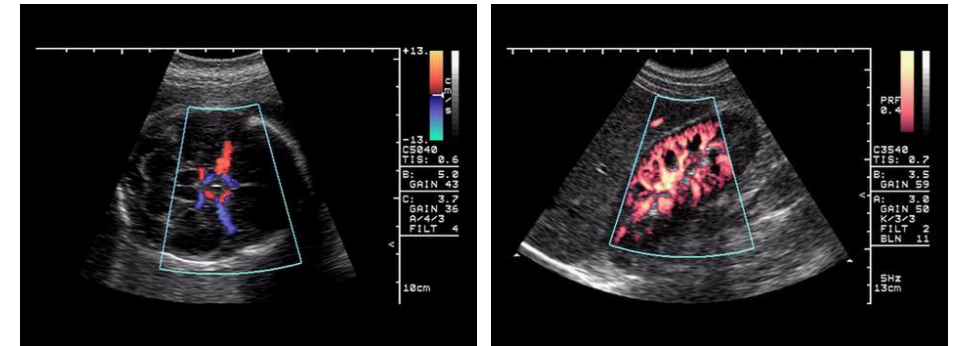
$$\Delta f = f_D = \pm 2 \frac{v_R}{c} f$$

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

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Colour coding

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



BART: **Blue** Away **Red** Towards

power Doppler

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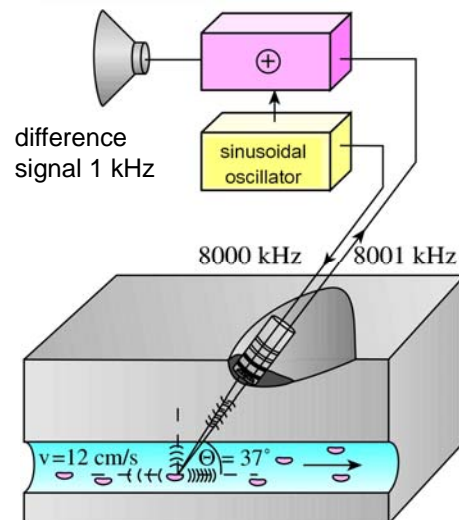
1-dimensional CW Doppler apparatus for measuring average flow velocity. Red blood cells as sound scatterers

CW: continuous wave
source and detector are separated

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

e.g. $f=8000$ kHz
 $v=12$ cm/s
 $c=1600$ m/s
 $\theta = 37^\circ$

$\Rightarrow f_D=1$ kHz
(beating phenomenon)

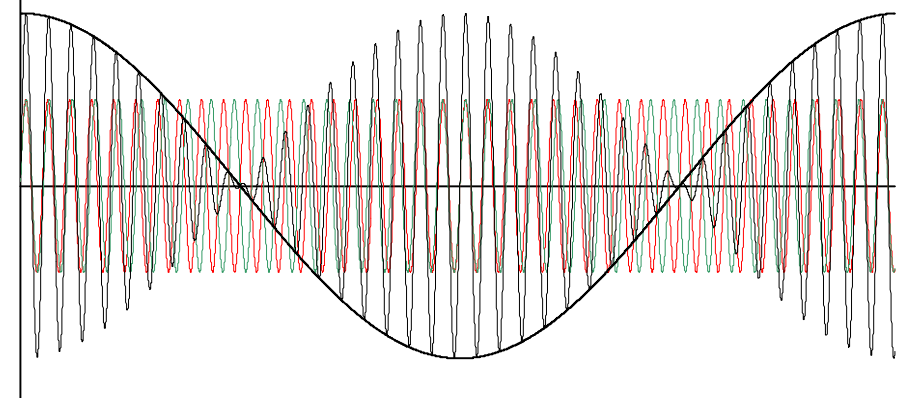


Textbook Fig. VIII. 41

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Beating phenomenon

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

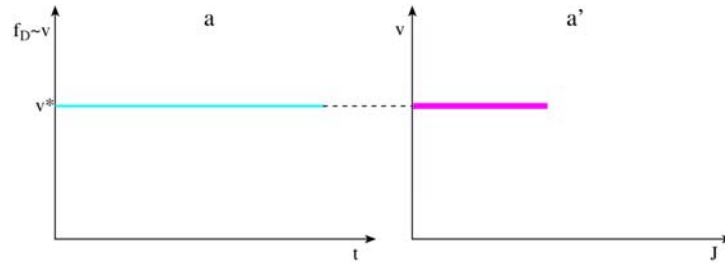


reminder: $\sin \alpha + \sin \beta = 2 \sin \frac{\alpha + \beta}{2} \cos \frac{\alpha - \beta}{2}$

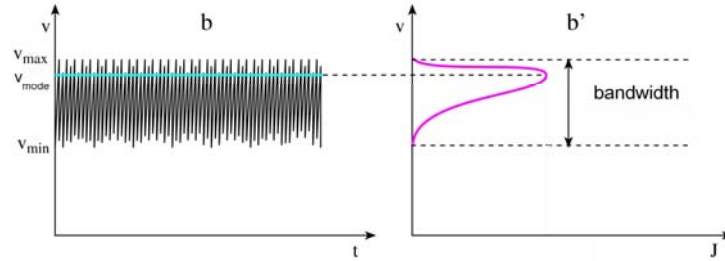
44

Doppler curves

one constant velocity (v^*)



frequency distribution (with v_{mode})



velocity distribution in TM-mode

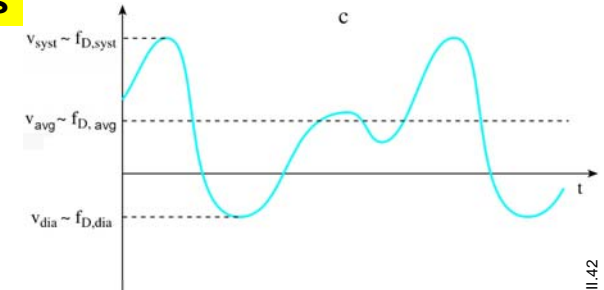
velocity distribution at a certain time

45

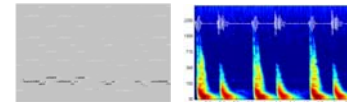
Textbook Fig. VIII.42

Doppler curves

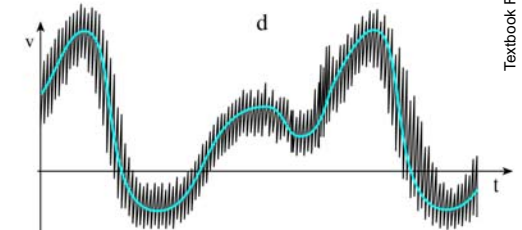
flow can be represented by one velocity in each moment



flow can be represented by a velocity distribution in each moment



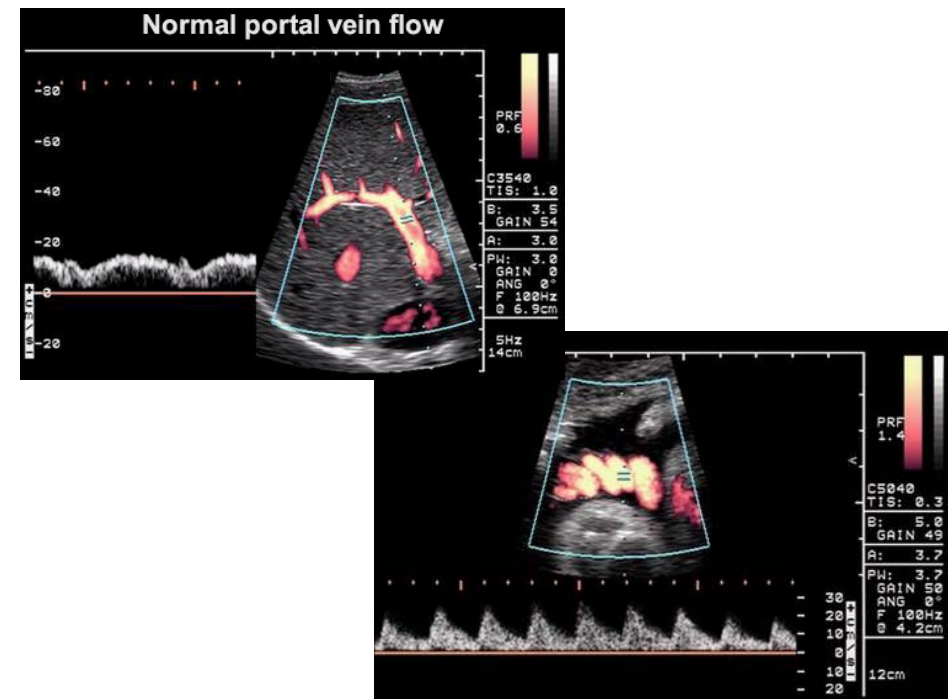
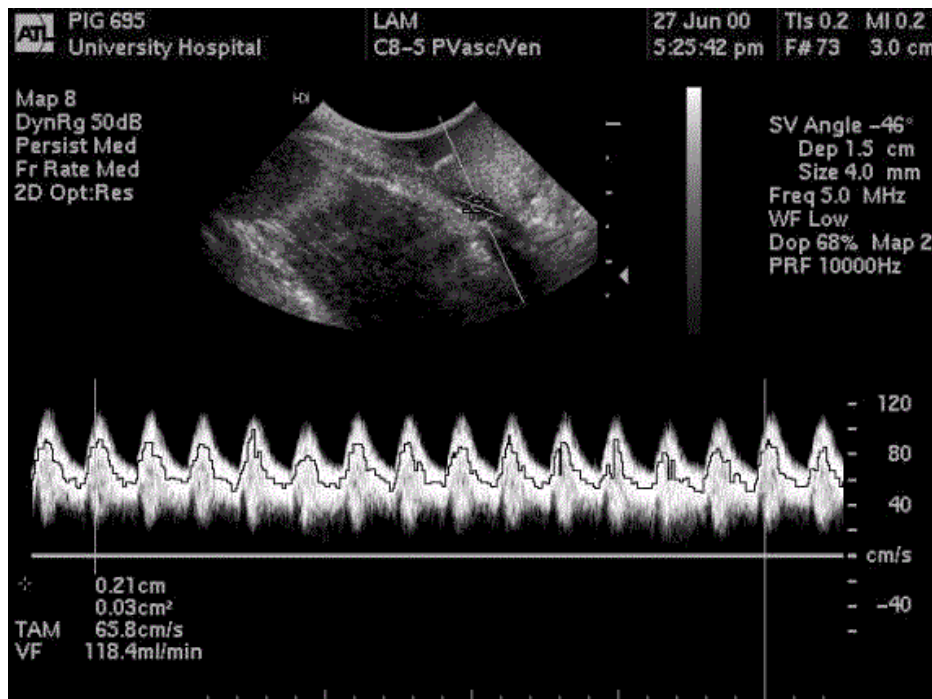
cf. voiceprint, music/heart beats in time-frequency representation



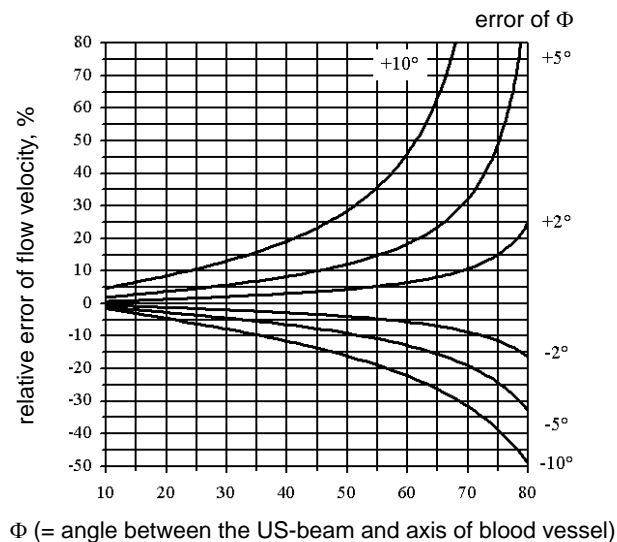
velocity distribution in TM-mode

Textbook Fig. VIII.42

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Error of measuring the angle between the US-beam and axis of blood vessel influences the error of the flow velocity



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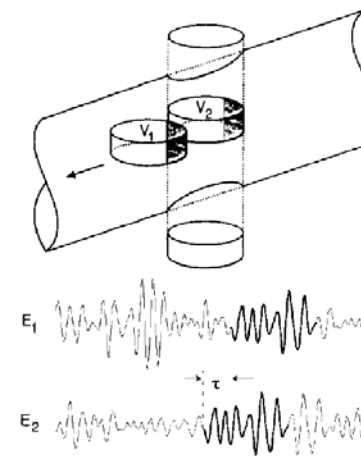
Speckle tracking/Time domain correlation method (CVI = color velocity imaging)

If the reflecting surface and/ or the scatterer are moving then the US signal at a fixed position depends on time.

Similar US pattern can be measured at a certain distance from the earlier position (where the reflecting surface/scatterer moved).

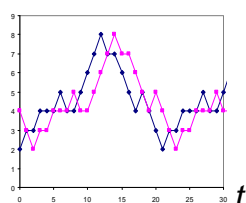
How can be compared the similarity of different functions?

The advantage of speckle tracking is the **angle independency**.

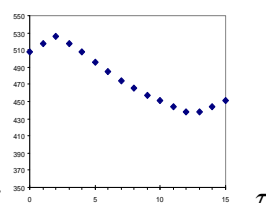


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time domain

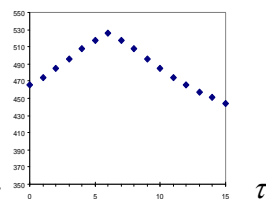
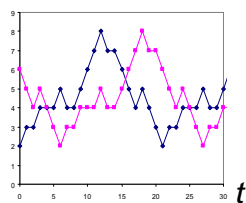


correlation function

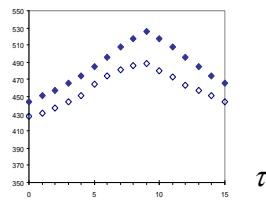
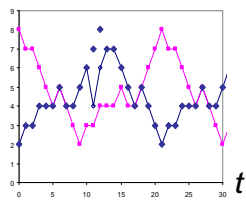


$$f_{\text{blue}}(t) = f_{\text{pink}}(t + \tau^*)$$

$$\tau^* = 2u$$



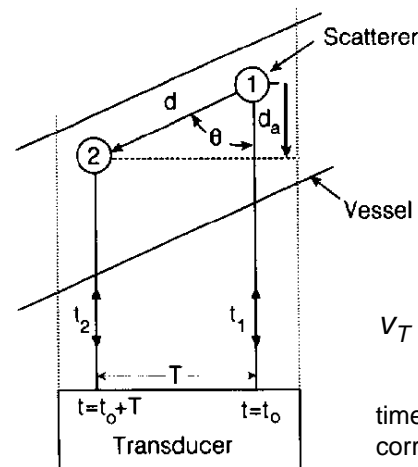
$$\tau^* = 6u$$



$$\tau^* = 9u$$

$$f_{\text{empty}}(t) \cong f_{\text{blue}}(t)$$

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$$d_a = \frac{(t_1 - t_2)c}{2}$$

$$d = \frac{(t_1 - t_2)c}{2\cos\theta}$$

$$v_T = \frac{(t_1 - t_2)c}{2T\cos\theta}$$

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

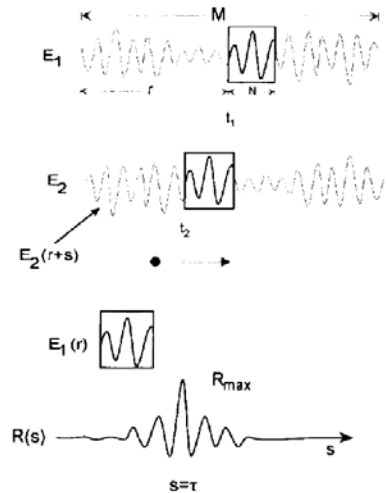
time domain
correlation method

cf.: Doppler
method

T : pulse repetition
time

f : frequency of US

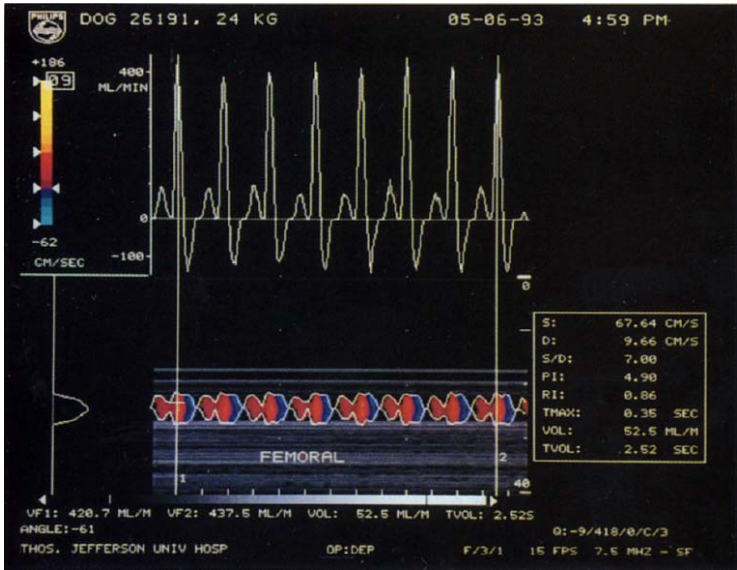
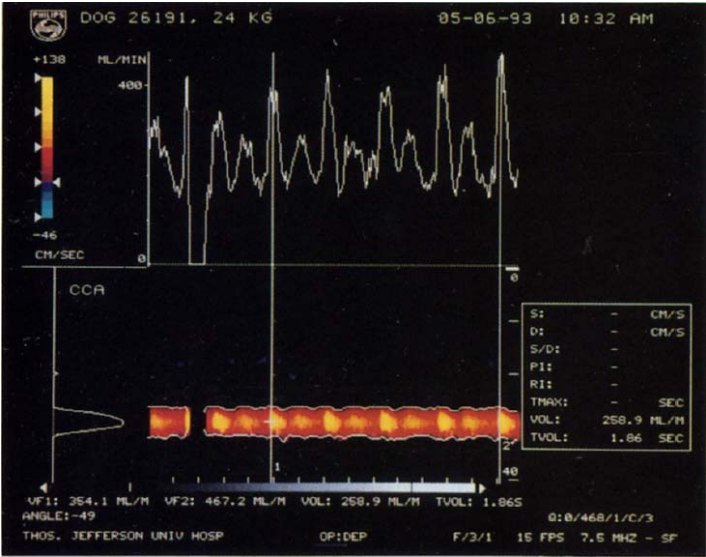
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correlation procedure consists of removing a window of width N at desired range from on echo signal E_1

E_1 is correlated at different locations along another echo signal E_2

the value of s producing the maximum corresponds to $s = \tau$



Sono-CT

image reconstruction from several multidirectional B-images

SonoCT yields better results than conventional B-mode sonography in terms of delineation, visualization of borders and artefact-free representation

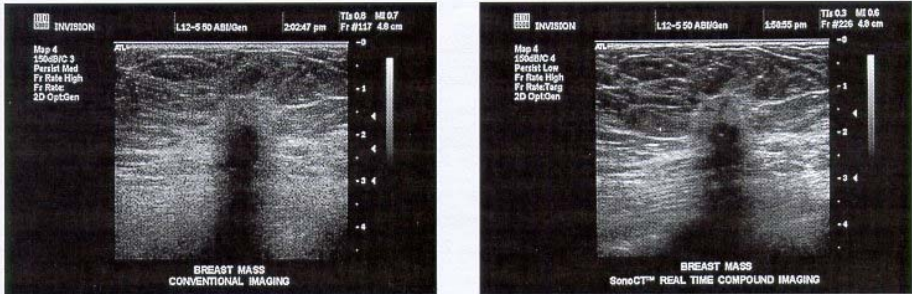
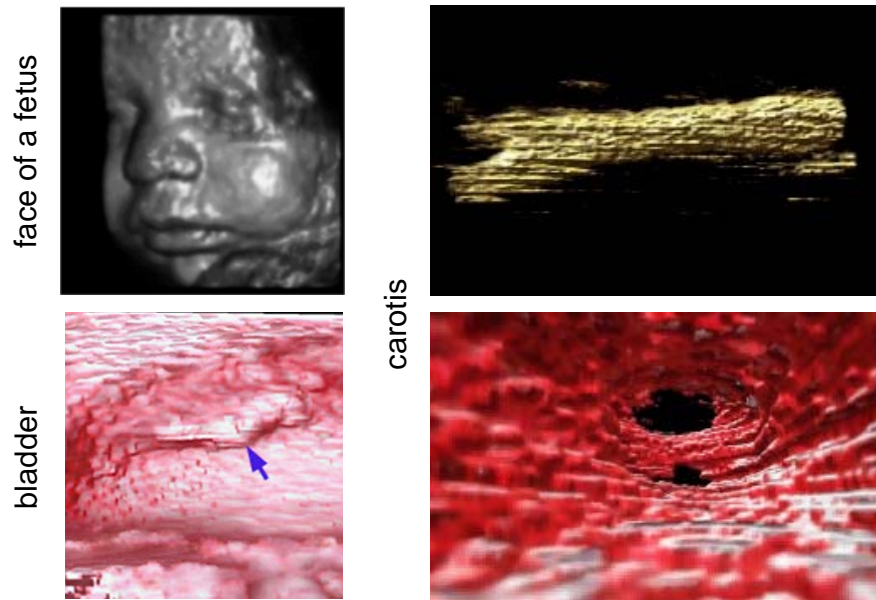


Figure 1. Breast fibroadenoma as shown on conventional ultrasound (left) and SonoCT ultrasound imaging (right).

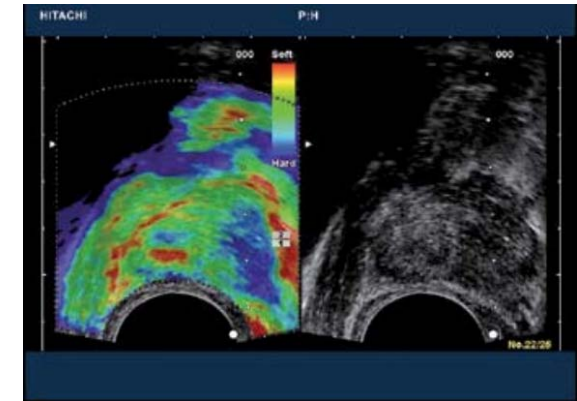
3D reconstruction



Sonoelastography

palpation:
one of the oldest
clinical skills,
tissue elasticity

strain:
a region of tissue is
compressed and the
**degree to which it
distorts** is assessed



Prostate cancer. The right panel shows the B-mode frame of a prostate ultrasound scan with the corresponding elastography frame on the left. The colour codes (here) blue for hard and red for soft. This hard lesion, which is not apparent on the B-mode, was a carcinoma on biopsy

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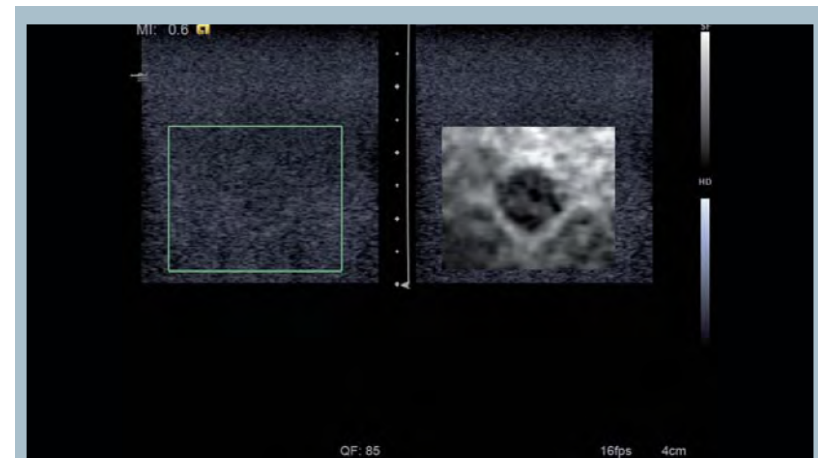
Cytoskeletal filaments as rigid bodies

In the first approximation, cytoskeletal filaments may be described as rigid bodies, which are deformed upon the action of external force. The deformation occurs in the direction of the applied force. The elasticity of rigid bodies is described by Hooke's law:

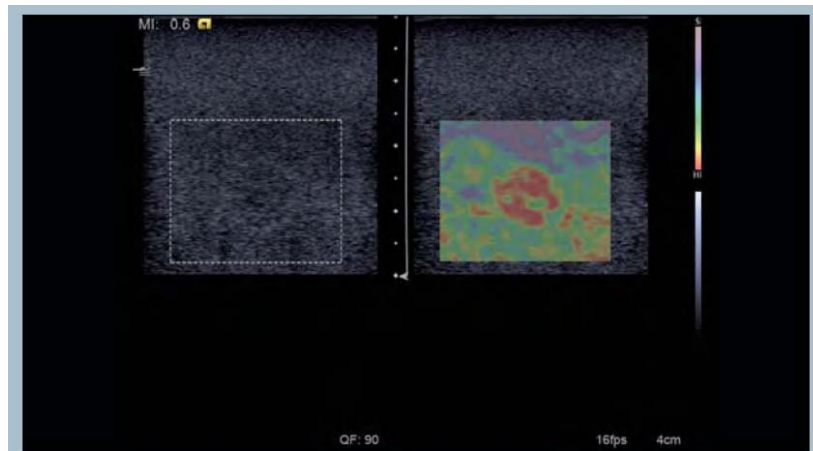
$$\frac{F}{A} = E \frac{\Delta L}{L}, \quad (\text{V.1})$$

where F is force, A is cross-sectional area, L is resting length, and ΔL is extension. The ratio F/A is stress (σ), E is the Young modulus and $\Delta L/L$ is relative extension or strain (ϵ). The ratio of stress and strain is known as the Young modulus (elastic modulus), the dimension of which is pressure (Pa). In the case of homogenous bodies the Young modulus is characteristic of the material properties only and is independent of shape and size.

Supplementary material

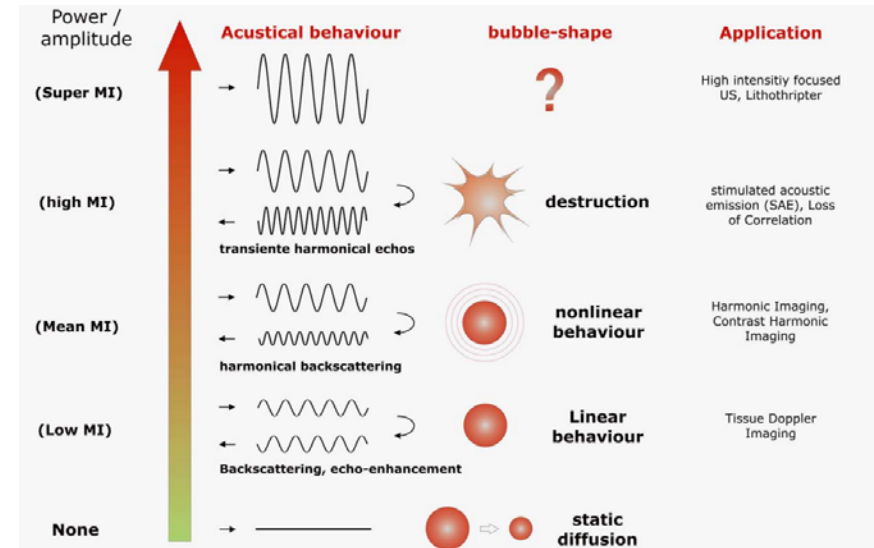


Grayscale eSie Touch Elasticity Imaging demonstrates a lesion which is more stiff (black) than the surrounding tissue. The conventional B-mode ultrasound is the same as displayed in the previous image.



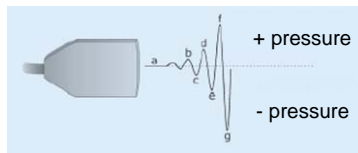
Color scale eSie Touch Elasticity Imaging demonstrates a lesion which is more stiff (red) than the surrounding tissue.

Contrast agents

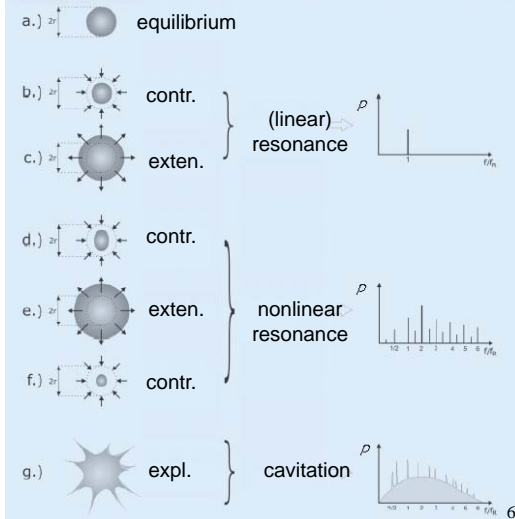


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Contrast harmonic imaging = CHI



Behaviour of microbubbles exposed to pulsed US



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C. Kollmann · M. Putzer
**Ultraschallkontrastmittel –
physikalische Grundlagen**
Radiologe 2005 · 45:503–
512

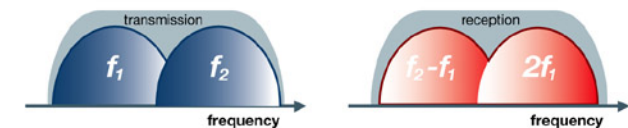
Tissue harmonic imaging = THI

Principle:

- Simultaneous transmission of 2 pulses at different frequencies
- Reception of signals at harmonic and differential frequencies
- Cancellation of fundamental signals using Pulse Subtraction

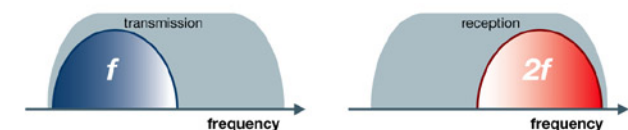
Toshiba

Differential Tissue Harmonic Imaging



Philips

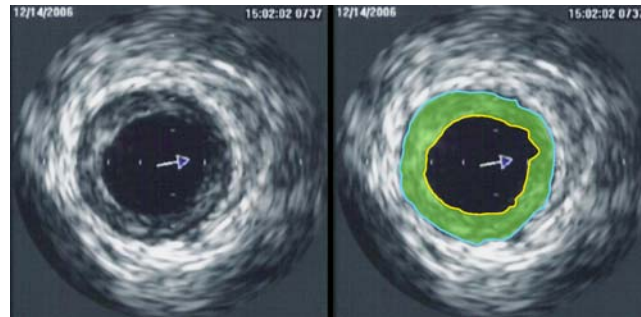
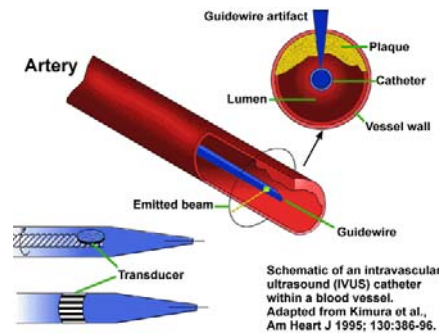
All Other Tissue Harmonic Imaging Methods



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Intravascular ultrasound (IVUS)

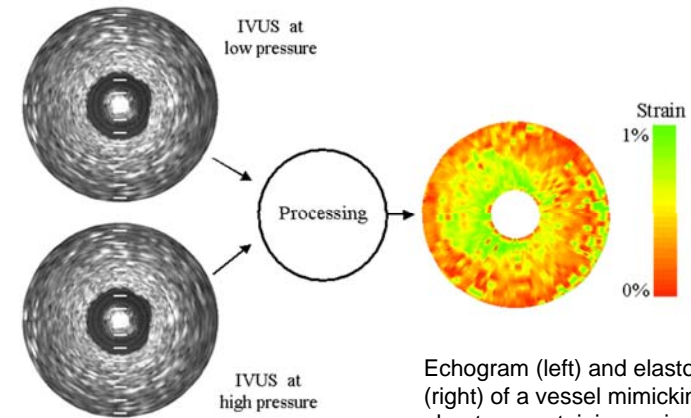
20-40 MHz, frame rate: 30 Hz



lumen of the coronary artery: yellow

outer elastic membrane: blue

Intravascular sonoelastography



Echogram (left) and elastogram (right) of a vessel mimicking phantom containing an isoechoic soft lesion between 7 and 11 o'clock. The lesion is invisible in the echogram, while it is clearly depicted in the elastogram

Safety

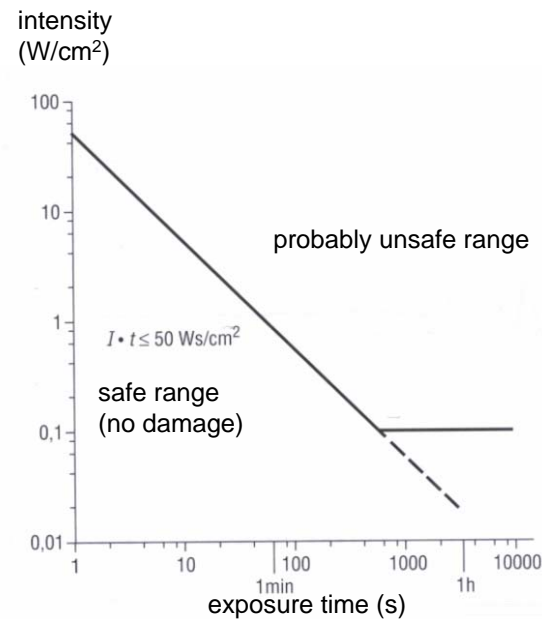
in the diagnostics:

$10 \text{ mW/cm}^2 = 100 \text{ W/m}^2$

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



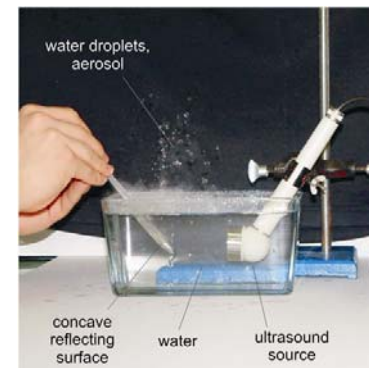
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more:

practices

in our practical laboratories

in the Skill Center



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