Medical Imaging

Ultrasound Imaging

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Overview

Advantages

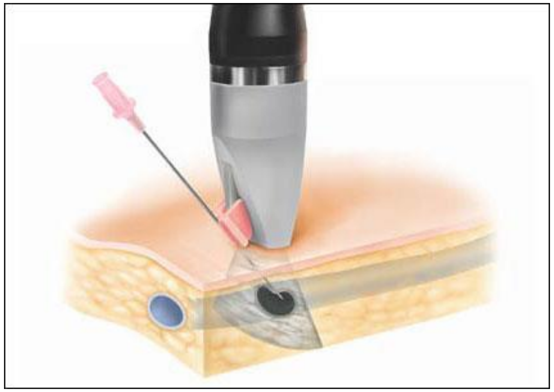
- non-invasive
- inexpensive
- portable
- excellent temporal resolution

Disadvantages

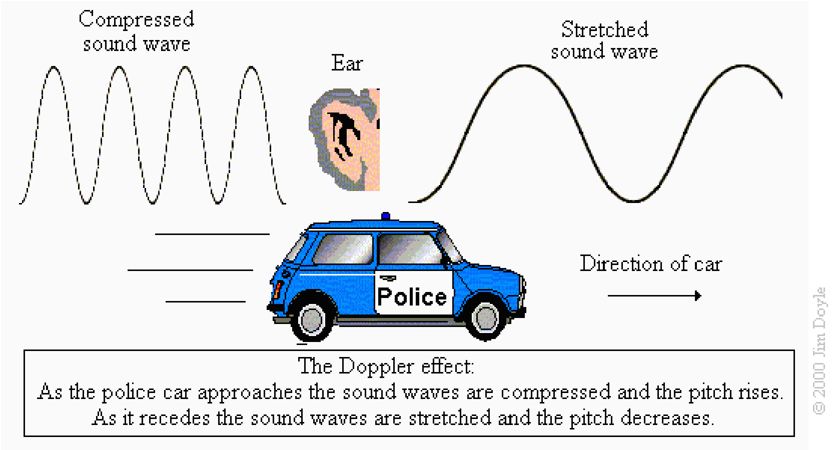
- noisy
- low spatial resolution

Samples of clinical applications

- echo ultrasound
  - cardiac
  - fetal monitoring
- Doppler ultrasound
  - blood flow
- ultrasound CT
  - mammography



US guided biopsy



Doppler effect

History

Milestone applications:

- publication of *The Theory of Sound* (Lord Rayleigh, 1877)
- discovery of piezo-electric effect (Pierre Curie, 1880)
  - enabled generation and detection of ultrasonic waves
- first practical use in World War One for detecting submarines
- followed by
  - non-destructive testing of metals (airplane wings, bridges)
  - seismology

Pierre Curie
1859 - 1906

- first clinical use for locating brain tumors (Karl Dussik, Friederich Dussik, 1942)

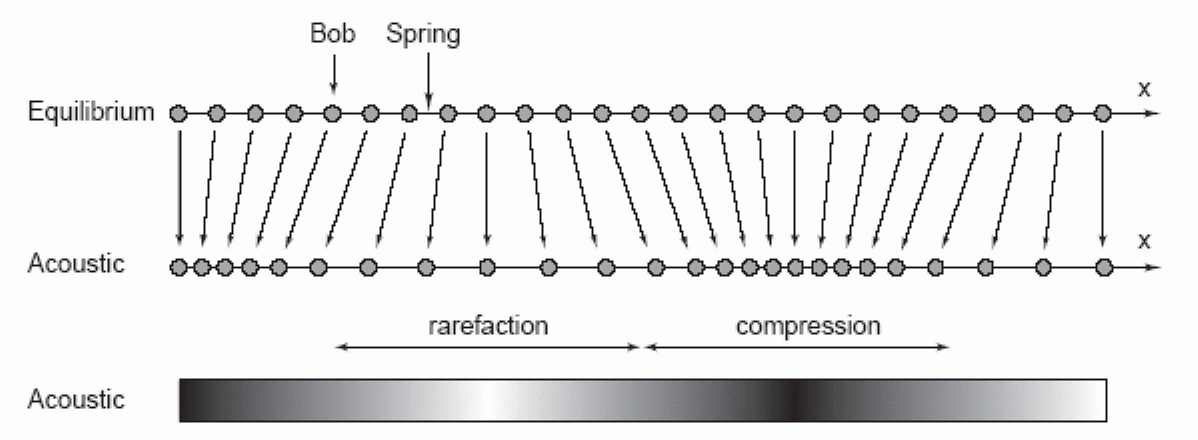
Karl Theodore Dussik

- the first greyscale images were produced in 1950
  - in real time by Siemens device in 1965
- electronic beam-steering using phased-array technology in 1968
- popular technique since mid-70s
- substantial enhancements since mid-1990

Ultrasonic Waves

US waves are longitudinal compression waves

- particles never move far
- transducer emits a sound pulse which compresses the material
- elasticity limits compression and extends it into a *rarefaction*
- rarefaction returns to a compression
- this continues until damping gradually ends this oscillation

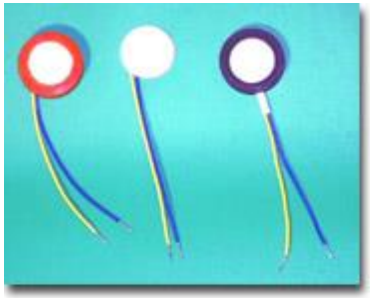


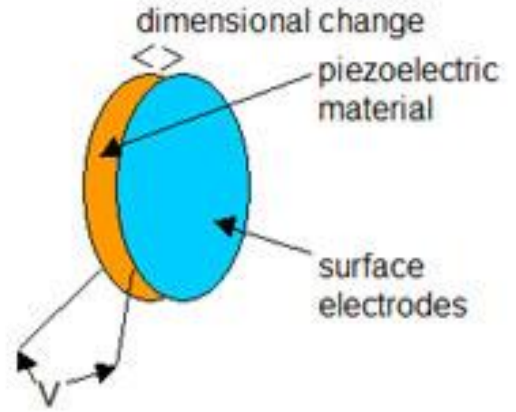
- ultrasound waves in medicine > 2.5 MHz
- humans can hear between 20 Hz and 20 kHz (animals more)

Generation of Ultrasonic Waves

Via piezoelectric crystal

- deforms on application of electric field → generates a pressure wave
- induces an electric field upon deformation ← detects a pressure wave
- such a device is called *transducer*





Wave Propagation

Two equations

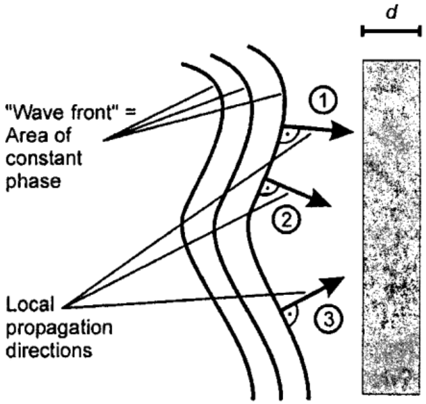
- wave equation:

$$\nabla^2 \Delta p = \frac{1}{c_0} \frac{\partial^2 \Delta p}{\partial^2 t^2} \qquad c_0 = \sqrt{\frac{1}{\rho_0 \beta_{s0}}}$$

Δp: acoustic pressure, ρ_0: acoustic density , β_{s0}: adiabatic compressibility

- Eikonal equation:

$$\frac{\partial^2 t}{\partial^2 x} + \frac{\partial^2 t}{\partial^2 y} + \frac{\partial^2 t}{\partial^2 z} = \frac{1}{F^2(x,y,z)}$$

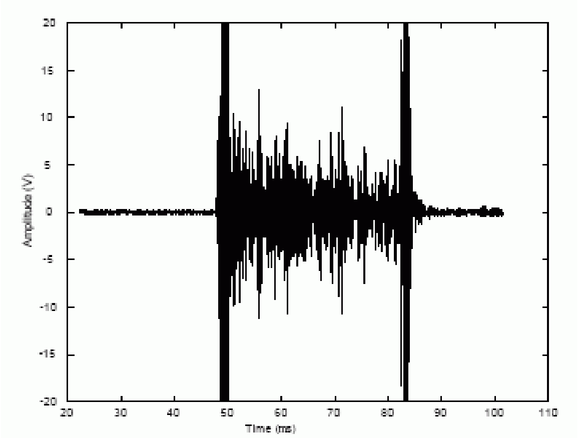


1/F: "slowness vector", inversely related to acoustic velocity *v*

- models a surface of constant phase called the *wave front*
- sound rays propagate normal to the wave fronts and define the direction of energy propagation.

Effects in Homogeneous Media

Attenuation



- models the loss of energy in tissue

$$H(f,z) = e^{-a_0 f^n z}$$

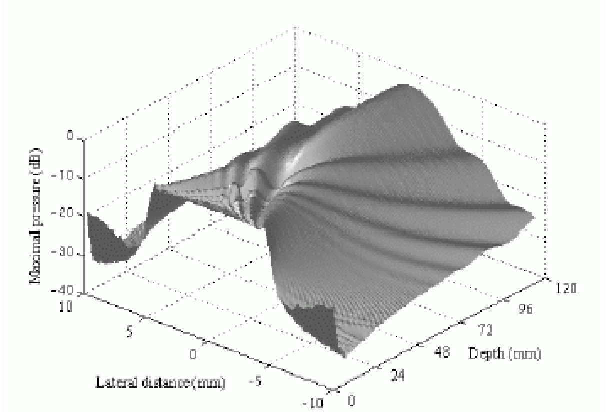
- f: frequency, typically n=1, z: depth, a_0: attenuation coefficient of medium,

Non-linearity

- wave equation was derived assuming that Δp was only a tiny disturbance of the static pressure
- however, with increasing acoustic pressure, the wave changes shape and the assumption is violated

Diffraction

- complex interference pattern greatest close to the source
- further away point sources add constructively



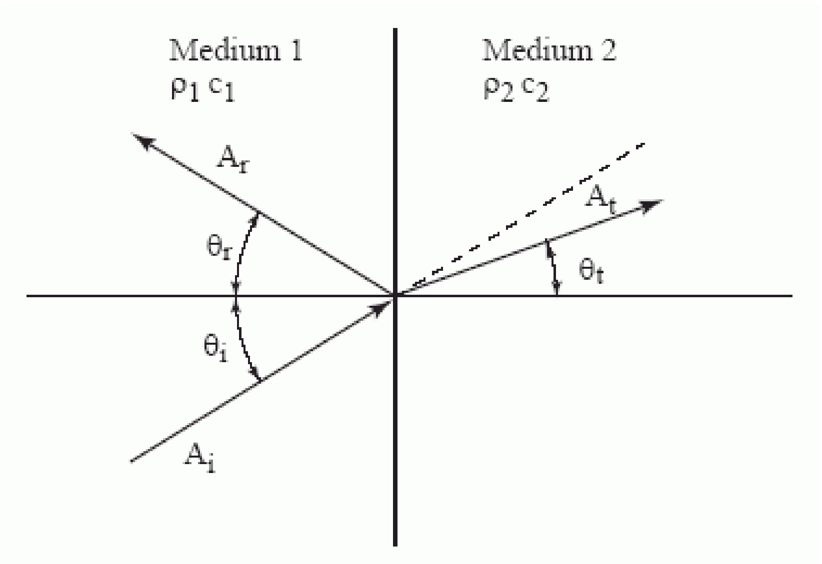
simulation with a circular planar source

Effects in Non-Homogeneous Media (1)

Reflection and refraction

- at a locally planar interface the wave's frequency will not change, only its speed and angle

$$\frac{\sin\theta_i}{c_1} = \frac{\sin\theta_r}{c_1} = \frac{\sin\theta_t}{c_2}$$



- for $c_2 > c_1$ and $\theta_i > \sin^{-1}(c_1/c_2)$ the reflected wave will not be in phase when

$$\cos\theta_t = \sqrt{1-(\frac{c_2}{c_1}\sin\theta_i)^2}$$

is complex

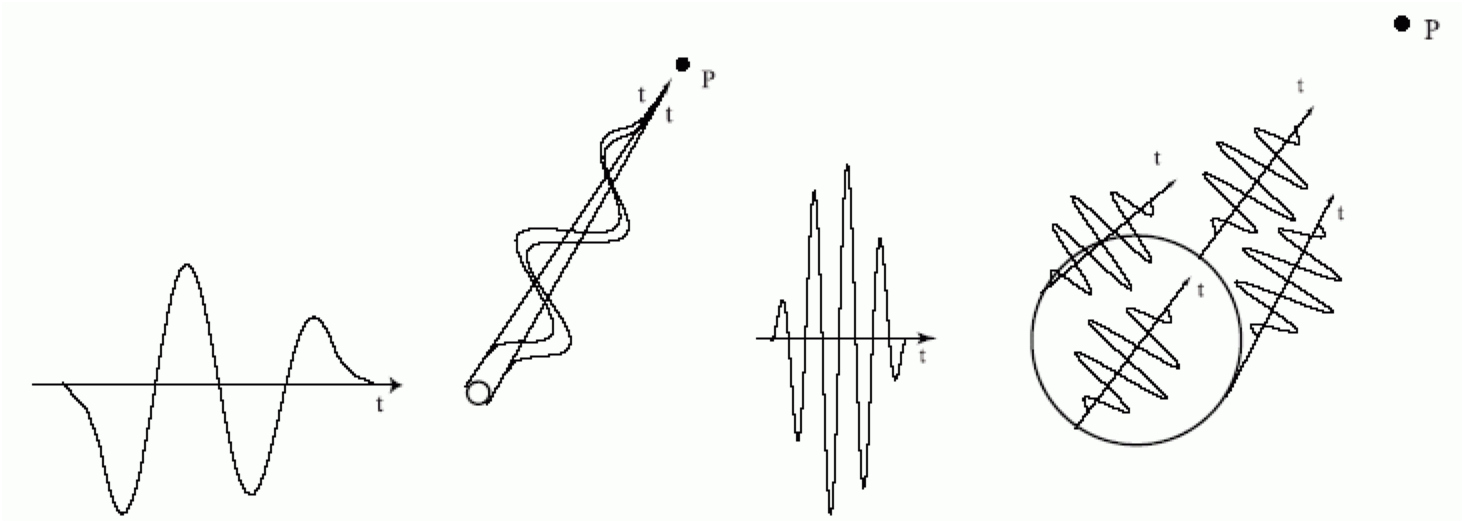
- the amplitude changes as well: $T+R=1$, $Z=\rho v$

$$T = \frac{A_t}{A_i} = \frac{2Z_1\cos\theta_t}{Z_2\cos\theta_i + Z_1\cos\theta_t} \qquad \mathrm{R} = \frac{A_r}{A_i} = \frac{Z_2\cos\theta_i - Z_1\cos\theta_t}{Z_2\cos\theta_i + Z_1\cos\theta_t}$$

Effects in Non-Homogeneous Media (2)

Scattering

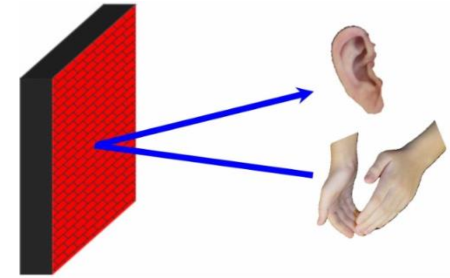
- if the size of the scattering object is << λ then get constructive interference at a far-enough receiver P
- if not, then need to model scattering as many point scatterers for a complex interference pattern



small object << λ

large object

Data Acquisition: A-Mode

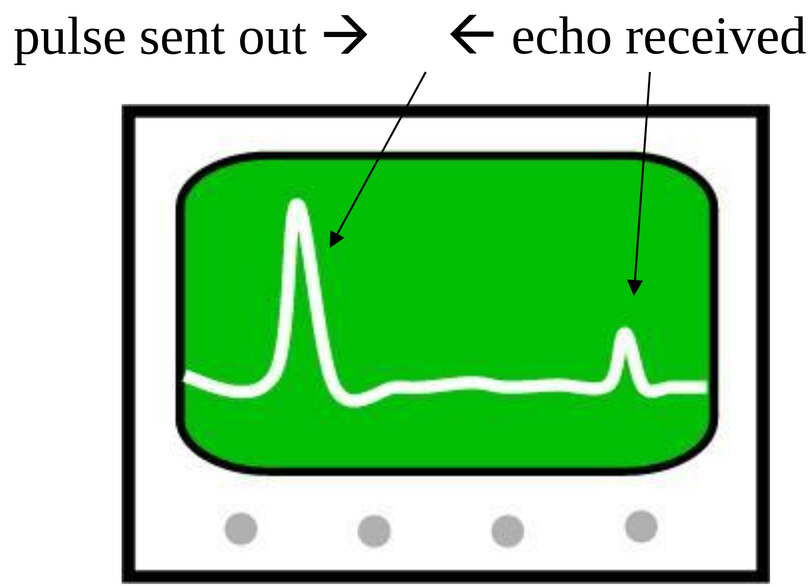


‘A’ for Amplitude

Simplest mode (no longer in use), basically:

- clap hands and listen for echo:

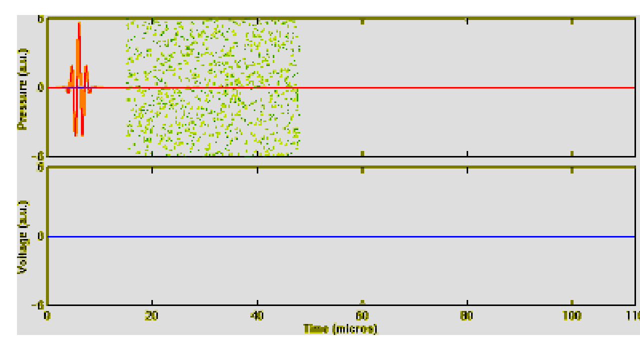
$$\text{distance} = \frac{\text{time expired} \cdot \text{speed of sound}}{2}$$



- time and amplitude are almost equivalent since sound velocity is about constant in tissue

Problem: don’t know where sound bounced off from

- direction unclear
- shape of object unclear
- just get a single line



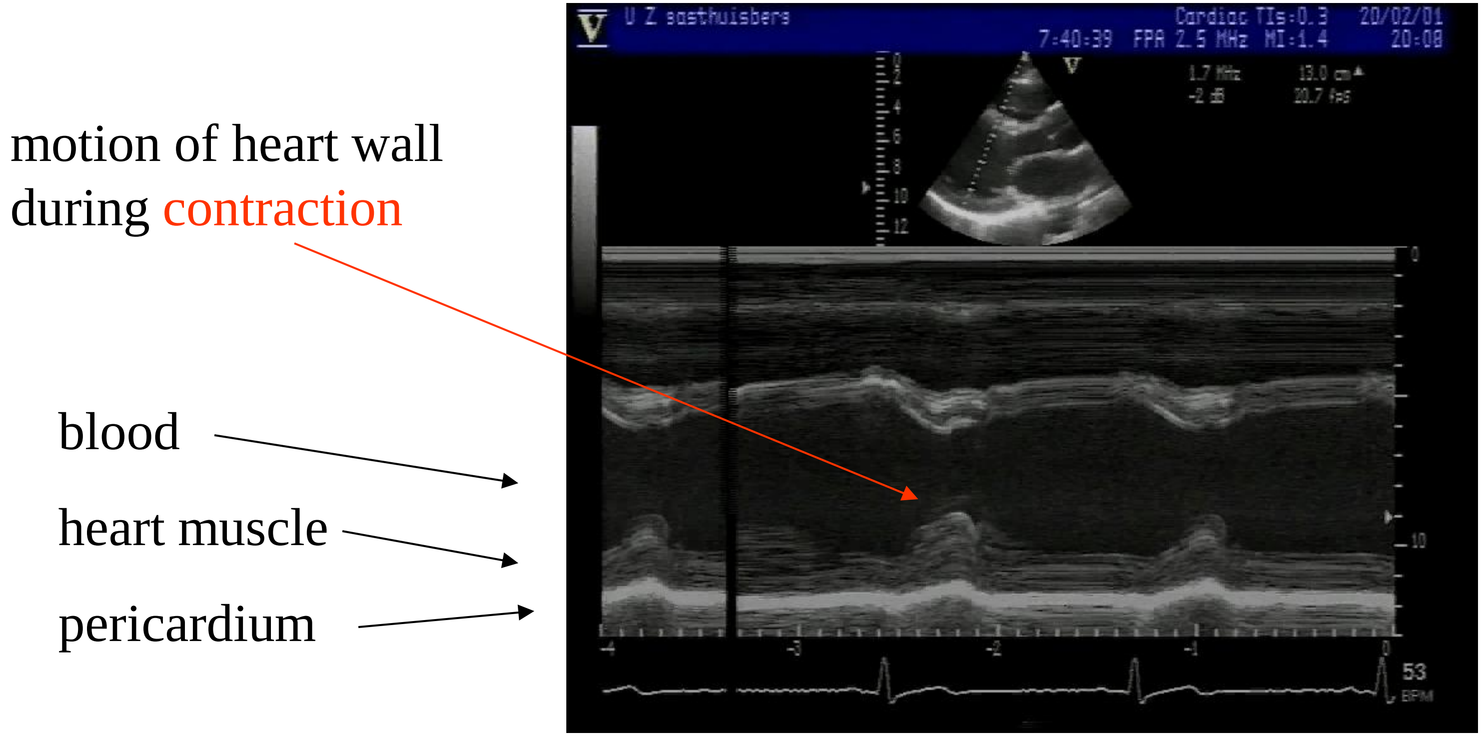
Data Acquisition: M-Mode

‘M’ for Motion

Repeated A-mode measurement

Very high sampling frequency: up to 1000 pulses per second

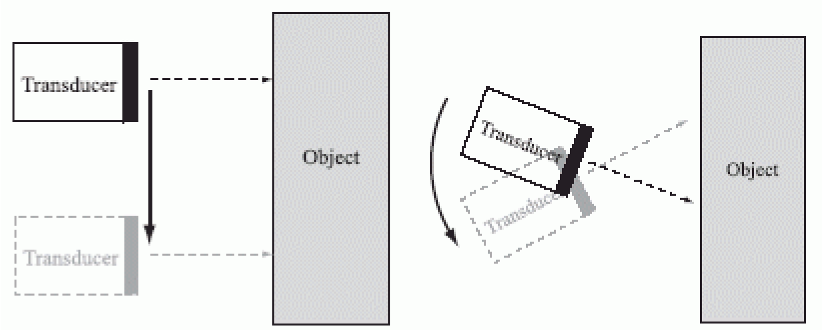
- useful in assessing rates and motion
- still used extensively in cardiac and fetal cardiac imaging

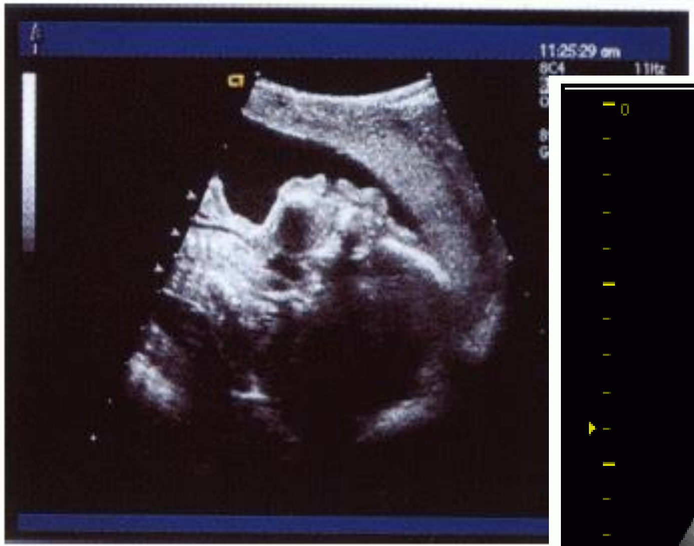


Data Acquisition: B-Mode

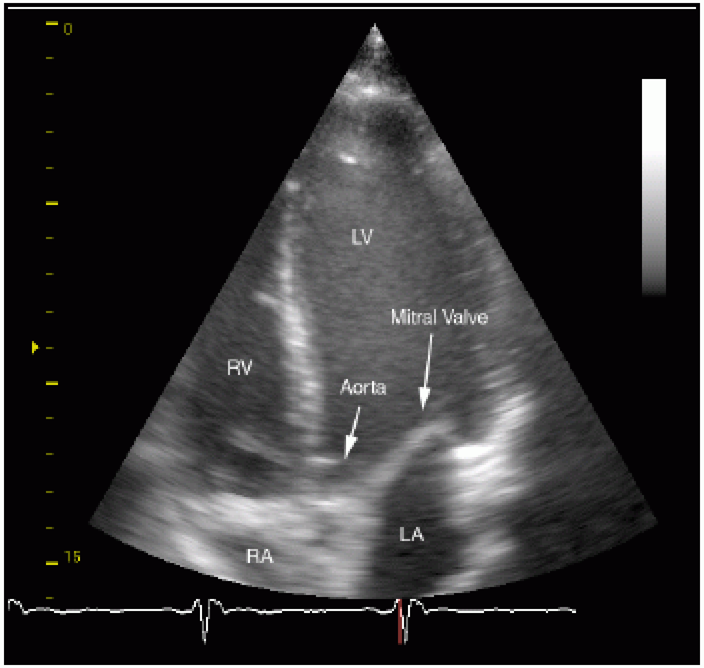
'B' for Brightness

An image is obtained by translating or tilting the transducer

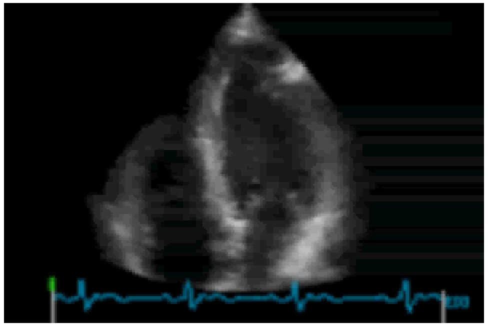




fetus



normal heart



continuous

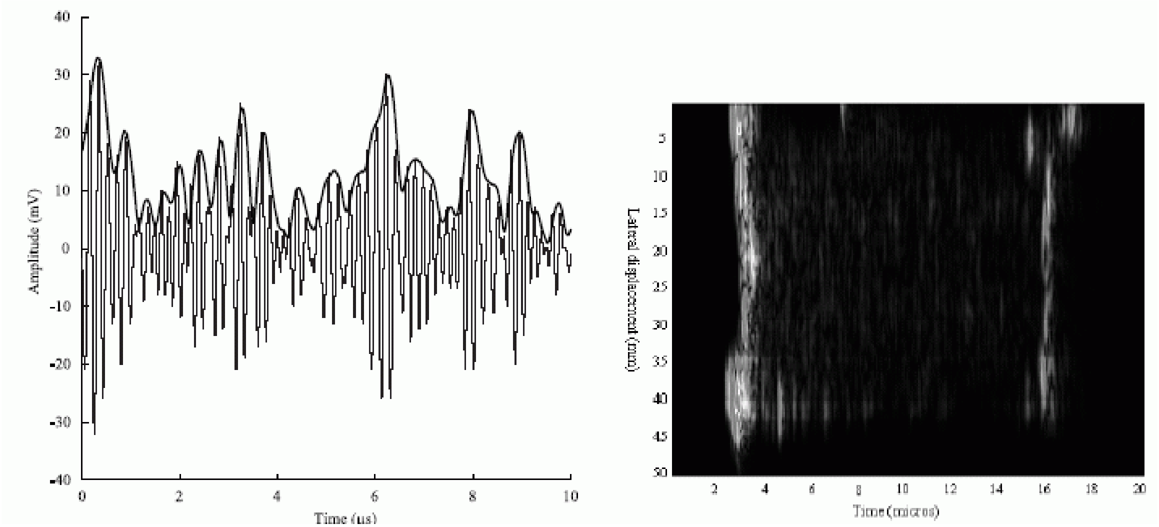
Image Reconstruction (1)

Filtering

- remove high-frequency noise

Envelope correction

- removes the high frequencies of the RF signal



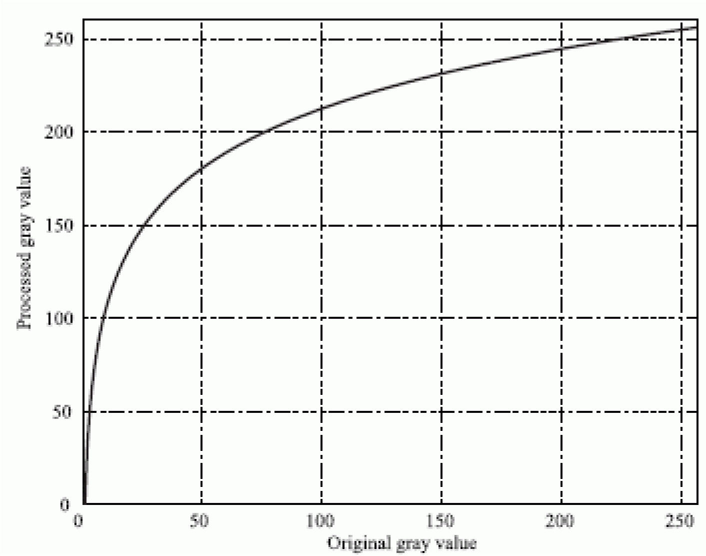
Attenuation correction

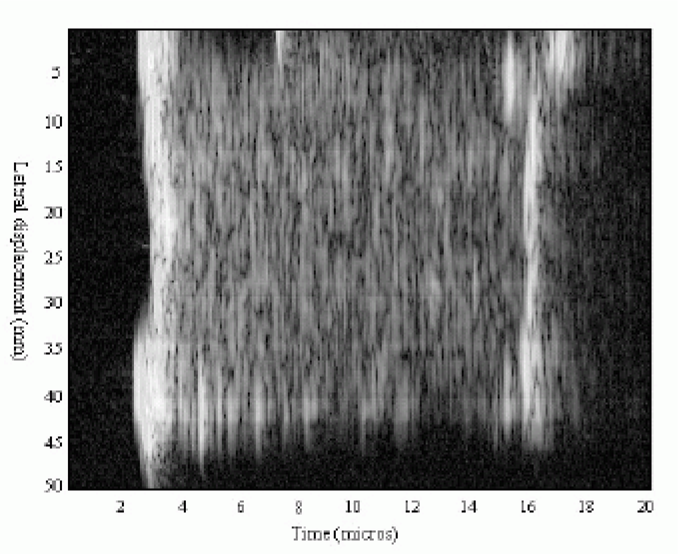
- correct for pulse attenuation at increasing depth
- use exponential decay model

Image Reconstruction (2)

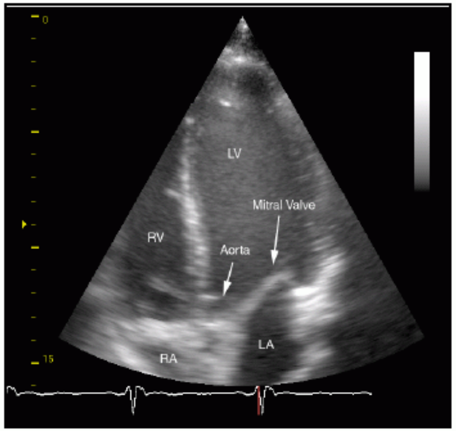
Log compression

- brings out the low-amplitude speckle noise





- speckle pattern can be used to distinguish different tissue



Acquisition and Reconstruction Time

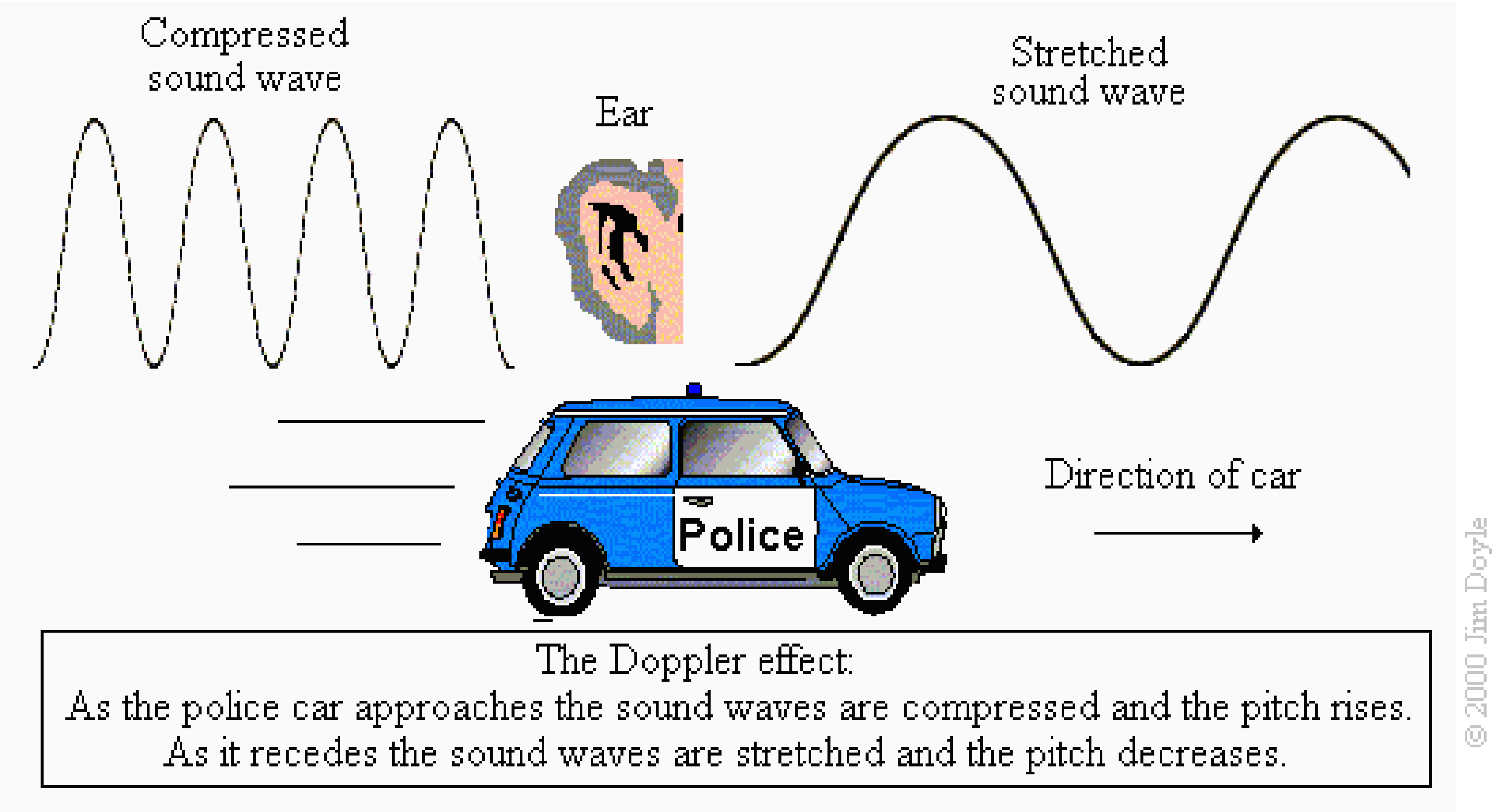
Typically each line in an image corresponds to 20 cm

- velocity of sound is 1540 m/s

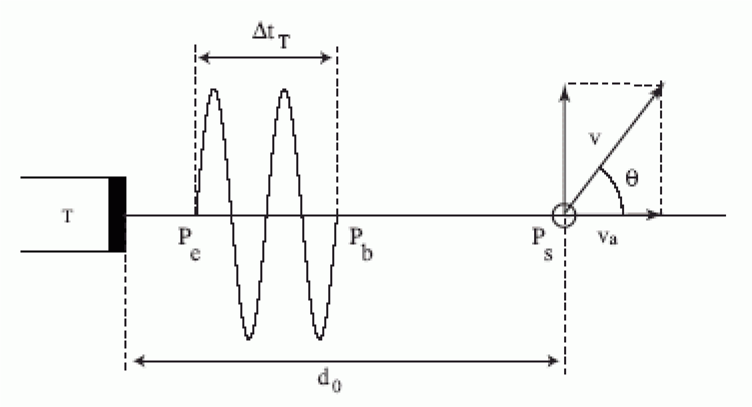
  → time for line acquisition is 267 μs
- an image with 120 lines requires then about 32 ms

  → can acquire images at about 30 Hz (frames/s)
- clinical scanners acquire multiple lines simultaneously and achieve 70-80 Hz

Doppler Effect: Introduction



Doppler Effect: Fundamentals (1)



Assume an acoustic source emits a pulse of *N* oscillation within time Δt_T

- a point scatterer P_s travels at axial velocity v_a:
- the locations of the wave and the scatterer are: $P_b(t) = ct \quad P_s(t) = d_0 + v_a t$
- the start of the wave meets P_s at:

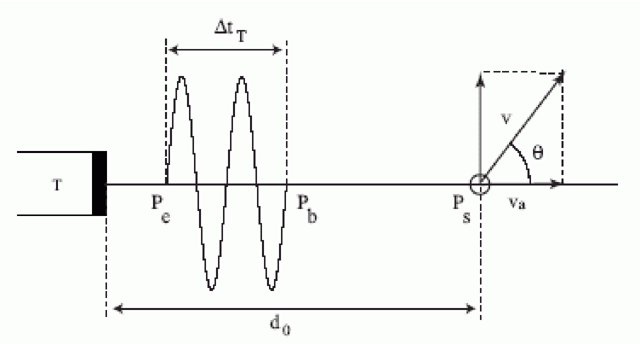
$$P_b(t_{ib}) = P_s(t_{ib}) \;\rightarrow\; t_{ib} = \frac{d_0}{c - v_a}$$

- the end of the wave meets P_s at:

$$P_b(t_{ie}) = P_s(t_{ie}) \;\rightarrow\; t_{ie} = \frac{d_0 + c\Delta t_T}{c - v_a} = t_{ib} + \frac{c}{c - v_a}\Delta t_T$$

- the start of the wave meets the transducer at $t_{rb} = 2t_{ib}$
- the end of the wave meets the transducer at $t_{re} = 2t_{ie} - \Delta t_T$

Doppler Effect: Fundamentals (2)



Received pulse (*N* oscillations)

- the duration of the received pulse is $\Delta t_R = t_{re} - t_{rb} = (\frac{2c}{c - v_a} - 1)\Delta t_T$
- writing it as frequencies $f_T = \frac{N}{\Delta t_T}$ $f_R = \frac{N}{\Delta t_R}$
- the Doppler frequency is then $f_D = f_R - f_T = \frac{-2v_a}{c + v_a} f_T \approx \frac{-2v_a \cos\theta}{c} f_T$
- to hear this frequency, add it to some base frequency f_b
- finally, to make the range smaller, f_d may have to be scaled

Example:

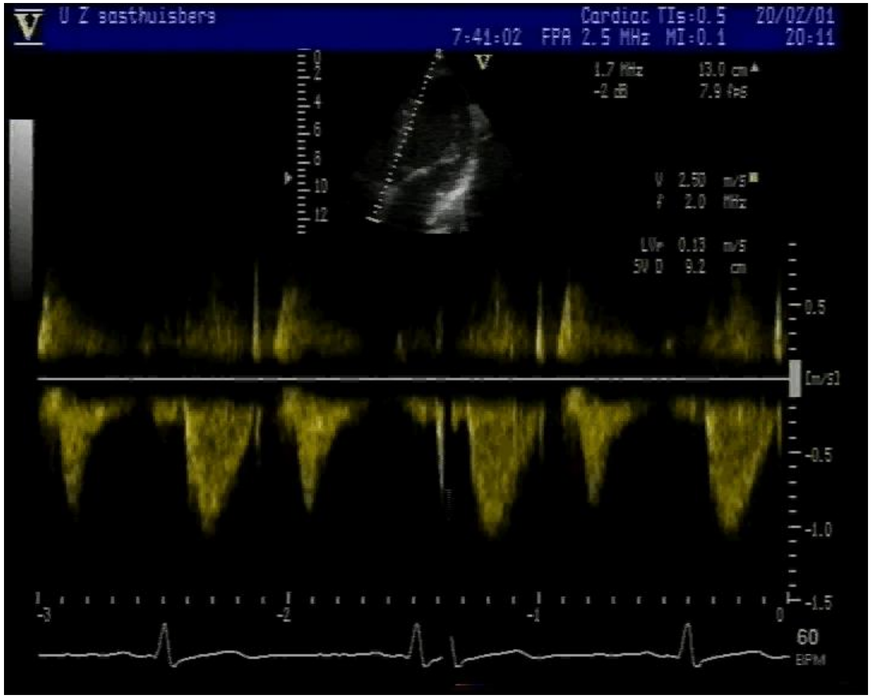
- assume a scatterer moves away at 0.5 m/s, the pulse frequency is 2.5 MHz, and a base frequency of 5 kHz, then the shift is an audible 5 kHz - 1.6 kHz = 3.4 kHz

Doppler: CW

'CW' for Continuous Wave

Compare frequency of transmitted wave f_T with frequency of received wave f_R

- the Doppler frequency is then: $f_D = f_R - f_T = \frac{-2v_a}{c + v_a} f_T \approx \frac{-2|v_a|\cos\theta}{c} f_T$
- Doppler can be made audible, where pitch is analog to velocity



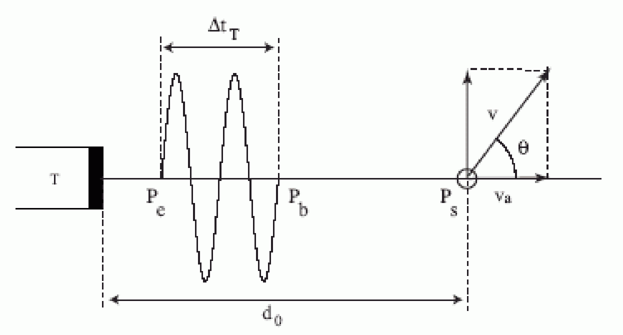
Doppler: PW (1)

'PW' for Pulsed Wave

Does not make use of the Doppler principle

- instead, received signal is assumed to be a scaled, delayed replica of the transmitted one

$$s(t) = A\sin(2\pi f_T(t - \Delta t))$$

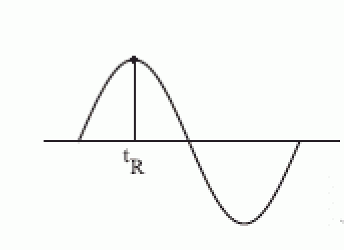


  Δt is the time between transmission and reception of the pulse

  it depends on the distance between transducer and scatterer

- in fact, we only acquire one sample of each of the received pulses, at t_R:

$$s(t_R) = A\sin(2\pi f_T(t_R - \Delta t))$$



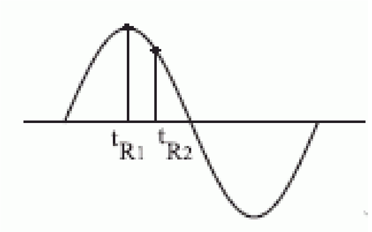
- now, if the scatterer moves away at velocity v_a, then the distance increases with $v_a T_{PRF}$ (T_{PRF}: pulse repetition period)

- this increases the time Δt (or decreases if the scatterer comes closer)

Doppler: PW (2)

Thus, the sampled sequence s_j is:

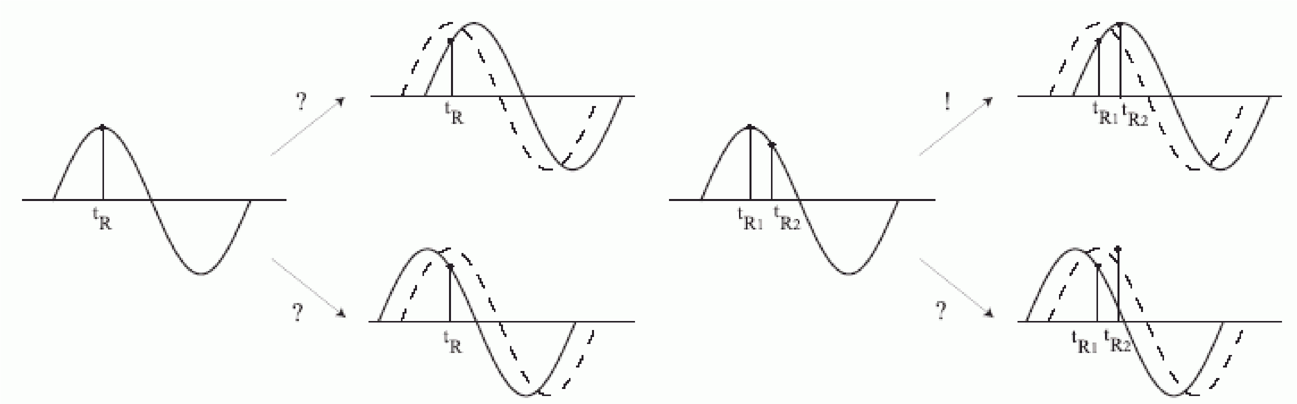
$$s_j = A\sin(-2\pi f_T (j \cdot \frac{2 \cdot v_a T_{PRF}}{c}) + B)$$



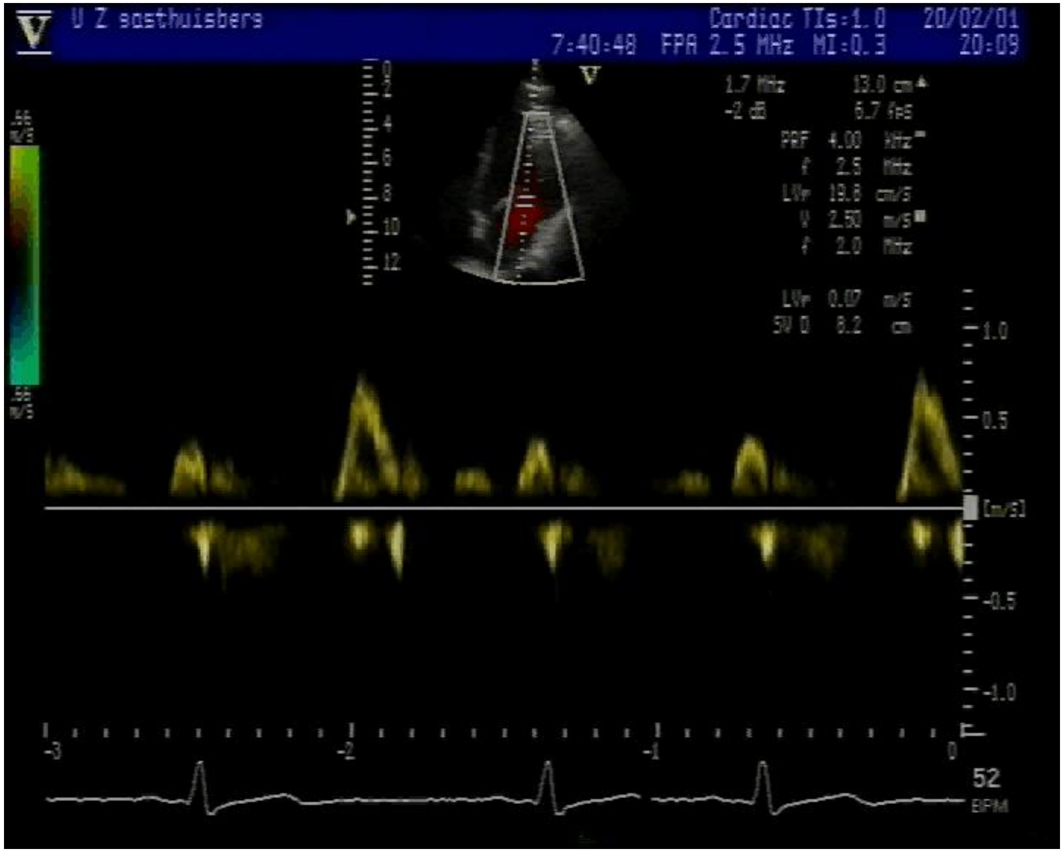
- therefore, the greater v_a, the higher the frequency of the sampled sinusoid:

$$f_D = -\frac{2v_a}{c} f_T$$

- to get direction information, one must sample more than once per pulse (twice per half oscillation) :



Doppler: PW (3)



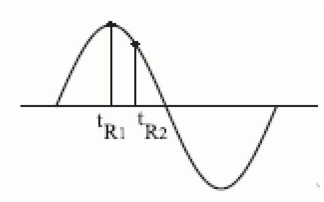
U Z gasthuisberg
Cardiac TIs:1.0
20/02/01
7:40:48 FPA 2.5 MHz MI:0.3
20:09
1.7 MHz
13.0 cm
-2 dB
6.7 fps
PRF 4.00 kHz
f 2.5 MHz
LVr 19.8 cm/s
V 2.50 m/s
f 2.0 MHz
LVr 0.07 m/s
SV D 8.2 cm
1.0
0.5
[m/s]
-0.5
-1.0
-3
-2
-1
0
52
BPM

Color Flow Imaging: Technique

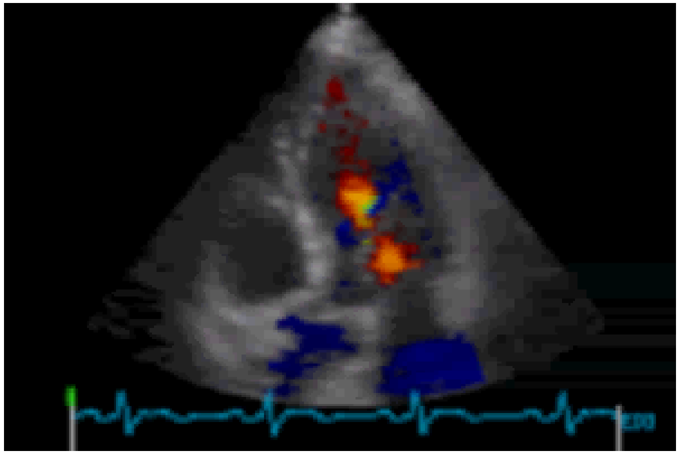
Calculates the phase shift between two subsequently received pulses

$$\Delta\varphi = 2\pi f_T \left(\frac{2 \cdot v_a T_{PRF}}{c}\right)$$

- measure the phase shift by sampling two subsequent pulses at two specific time instances t_{R1} and t_{R2}



- since this can become noisy, usually the results of 3-7 such samplings (pulses) are averaged
- divide the acquired RF line into segments (range gates) allows velocities to be obtained at a number of depths
- acquiring along a single line gives a M-mode type display
- acquiring along multiple lines enables a B-mode type display



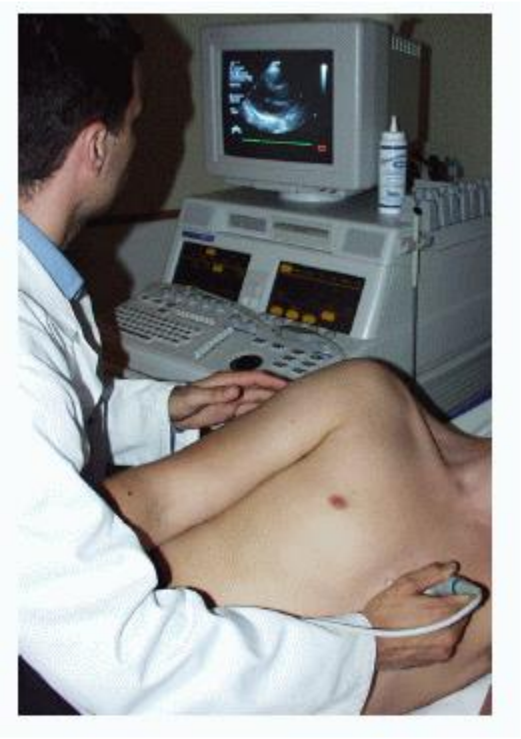
red: moving toward transducer

blue: moving away from trasnducer

Ultrasound Equipment

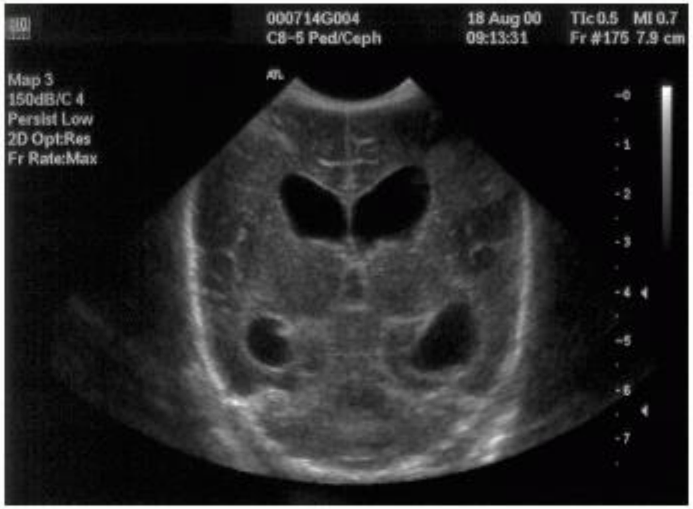
Left: Linear array transducer.

Right: Phased array transducer



commercial echocardiographic scanner

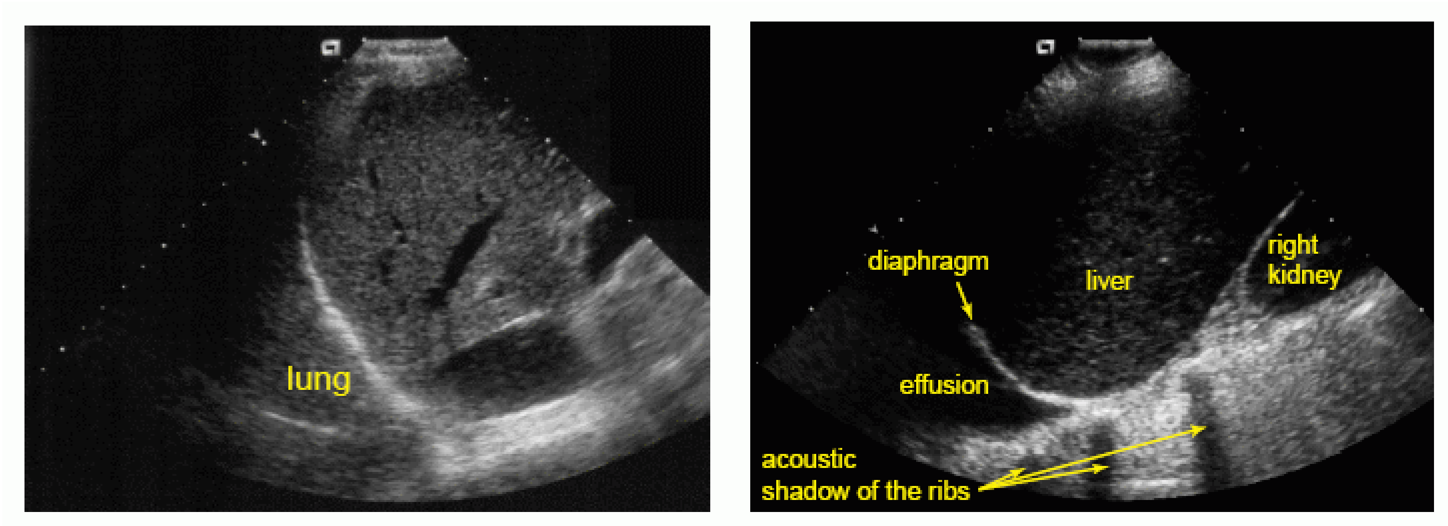
Ultrasound Applications (1)



Left: Normal cranial ultrasound.

Right: Fluid filled cerebral cavities on both sides as a result of an intraventricular haemorrhage

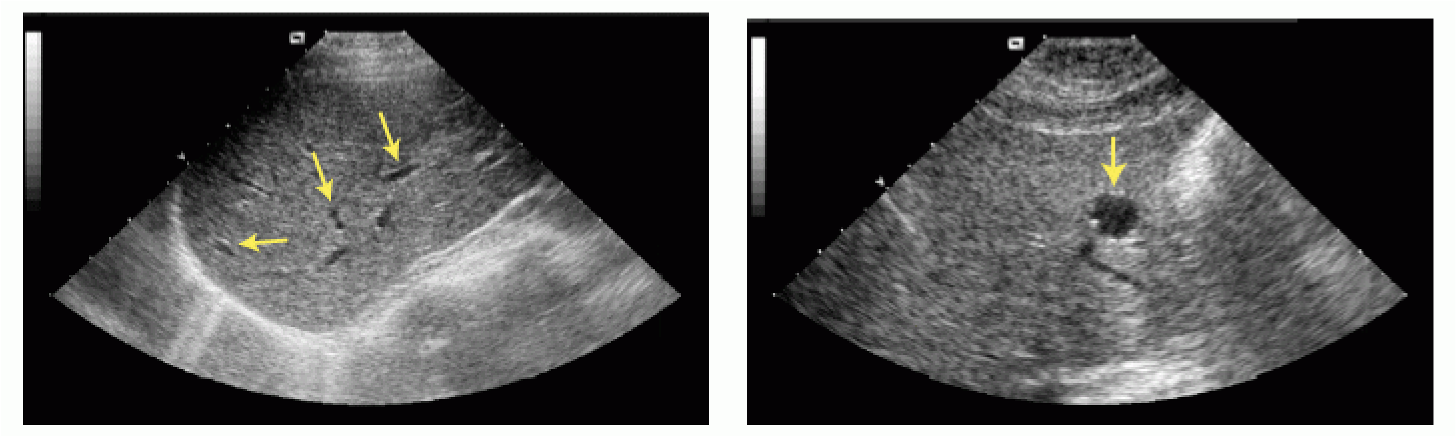
Ultrasound Applications (2)



Left: normal lung,

Right: pleural effusion

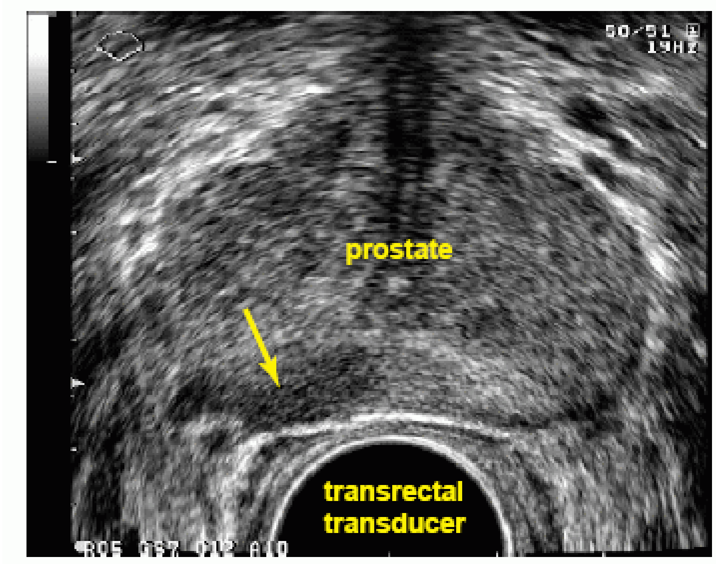
Ultrasound Applications (3)

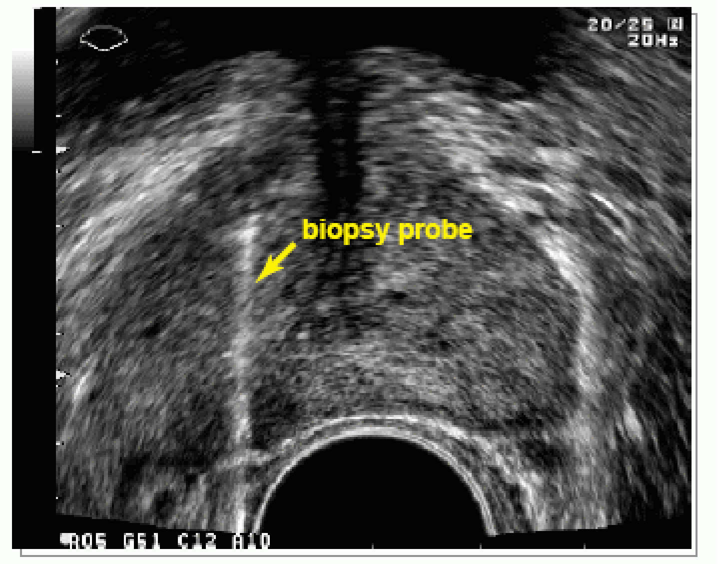


Left: normal liver

Right: liver with cyst

Ultrasound Applications (4)

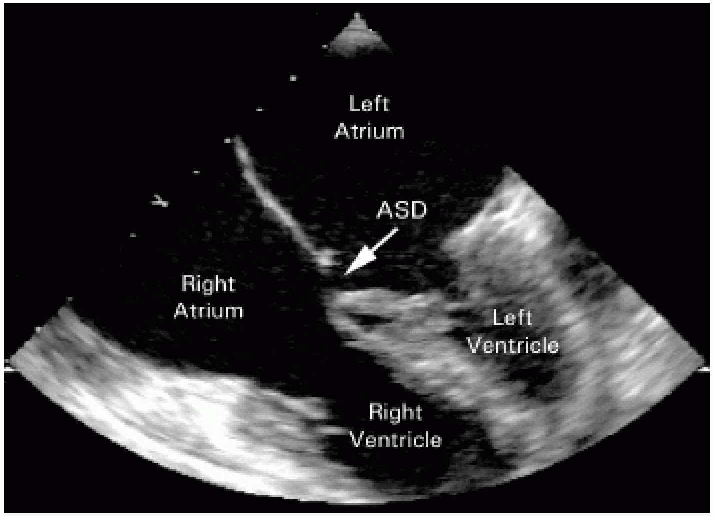




Left: prostate showing a hypoechoic lesion suspicious for cancer

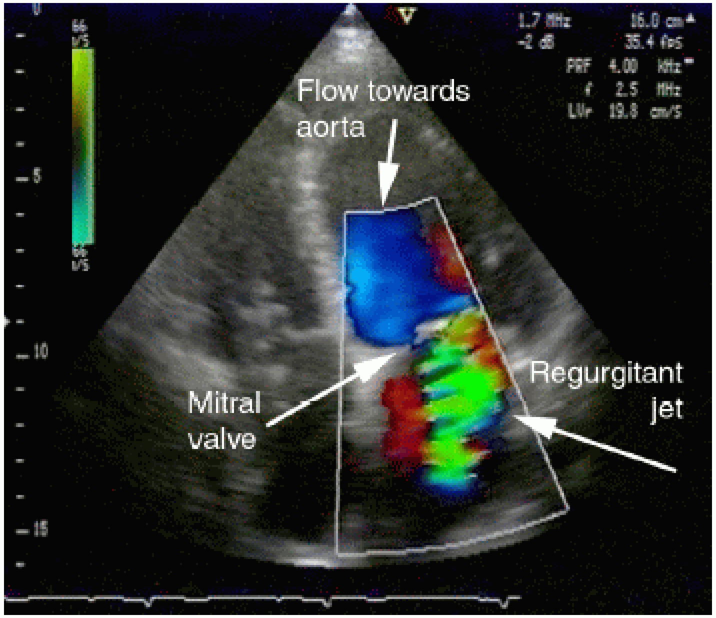
Right: with biopsy needle

Ultrasound Applications (5)



Atrial septal defect (ASD)

Ultrasound Applications (6)



Doppler color flow image of a patient with mitral regurgitation in the left atrium. The bright green color corresponds to high velocities in mixed directions, due to very turbulent flow leaking through a small hole in the mitral valve.