

# Medical ultrasound imaging

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# Part I

## Introduction to medical ultrasound

## Introduction

### Ultrasound acoustics

- Waves

- Wave equation

- Reflection and refraction

- Interface reflection

- Attenuation

### Medical ultrasound

- Devices

- Cardiologic US

- Intravascular US

- Artefacts

### Generation/detection

- Generation

- Steering/Beamforming

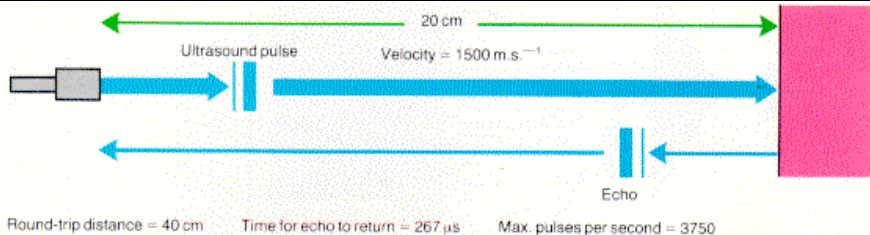
- Focusing

- Processing and control

# Medical ultrasound basics

- ▶ Acoustic waves, frequency  $2 \sim 50$  MHz
- ▶ Measure the time and intensity of the echo
- ▶ Harmless
- ▶ Stopped by air and dense tissues (bone)

# Ultrasound Principle



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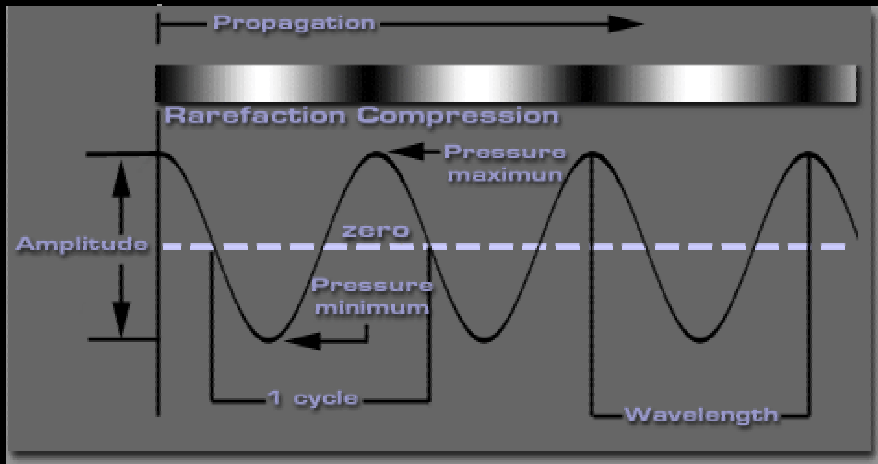
### Generation/detection

- Generation

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Sinusoidal pressure source

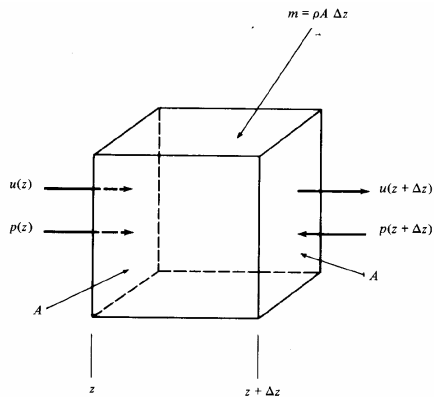
# Fyzikální základy – rozsahy veličin

Měřená veličina	Symbol	Jednotka (rozměr)	Rozsah obvyklých hodnot měřené veličiny v klinické praxi
Rychlost	$c$	$\text{m}\cdot\text{s}^{-1}$	1540 $\text{m}\cdot\text{s}^{-1}$ (měkká tkáň)
Vlnová délka	$\lambda$	mm	0,6 až 0,15 mm (měkká tkáň)
Kmitočet	$f$	hertz	2,5 až 10 MHz
Modul pružnosti	$E$	pascal	25 GPa (kost)
Akustická impedance	$Z$	$\text{kg}\cdot\text{m}^{-2}\cdot\text{s}^{-1}$	$1,63\cdot 10^5 \text{ kg}\cdot\text{m}^{-2}\cdot\text{s}^{-1}$
Hustota	$\rho$	$\text{kg}\cdot\text{m}^{-3}$	1000 $\text{kg}\cdot\text{m}^{-3}$ (voda)
Intenzita	$I$	$\text{W}\cdot\text{cm}^{-2}$	typicky 1 až 10 $\text{mW}\cdot\text{cm}^{-2}$
Tlak	$p$	pascal nebo bar	0,6 baru nebo 0,06 MPa





## Elementary volume



Speed  $u$ , pressure  $p$ , density  $\rho$ , area  $A$ , mass  $m$ .

## Newton's law

Motion along  $z$ :

$$F = ma = m \frac{du}{dt} = m \left( \frac{\partial u}{\partial t} + \frac{\partial u}{\partial z} \frac{\partial z}{\partial t} \right) \approx m \frac{\partial u}{\partial t}$$

force  $F = pA$ :

$$(p(z) - p(z + \Delta z)) A = m \frac{\partial u}{\partial t}$$

for  $\Delta z \ll z$ :

$$-\frac{\partial p}{\partial z} \Delta z A = m \frac{\partial u}{\partial t}$$

as  $m = \rho A \Delta z$

$$-\frac{\partial p}{\partial z} = \rho \frac{\partial u}{\partial t}$$

## Conservation of mass law

Difference of entering and exiting mass, density change:

$$A\left(u(z + \Delta z)\rho(z + \Delta z) - u(z)\rho(z)\right) = -A \Delta z \frac{\partial \rho}{\partial t}$$

for  $\Delta z \ll z$ :

$$\frac{\partial \rho u}{\partial z} = -\frac{\partial \rho}{\partial t}$$

density  $\rho = \rho_0 + \rho_1$ ,  $\rho_0 = \text{const}$ ,  $\rho_1 \ll \rho_0$ :

$$\rho_0 \frac{\partial u}{\partial z} + \frac{\partial \rho_1}{\partial t} = 0$$

Compressibility (stlačitelnost)  $\frac{\rho_1}{\rho_0} = Kp$ ,  $K = 1/E$ :

$$\frac{\partial u}{\partial z} + K \frac{\partial p}{\partial t} = 0$$

# 1D wave equation

$$\rho \frac{\partial u}{\partial t} + \frac{\partial p}{\partial z} = 0 \quad \text{derive by } z$$

$$\frac{\partial u}{\partial z} + K \frac{\partial p}{\partial t} = 0 \quad \text{derive by } t$$

$$\rho \frac{\partial^2 u}{\partial t \partial z} + \frac{\partial^2 p}{\partial z^2} = 0$$

$$\frac{\partial^2 u}{\partial z \partial t} + K \frac{\partial^2 p}{\partial t^2} = 0$$

subtract

$$\frac{\partial^2 p}{\partial z^2} - K \rho \frac{\partial^2 p}{\partial t^2} = 0$$

similarly

$$\frac{\partial^2 u}{\partial z^2} - K \rho \frac{\partial^2 u}{\partial t^2} = 0$$

# Wave equation solution

Harmonic wave:

$$p = p_+ \cos(\underbrace{\omega t - kz}_{\phi})$$

where  $k$  is the wave number (vlnové číslo) [rad/m].

Wave speed (phase velocity):

$$\phi_0 = \omega t - kz \quad \rightarrow \quad z = \frac{\omega}{k}t - \frac{\phi_0}{k}$$

$$c = \omega/k$$

$$c = \lambda f \quad \text{because} \quad \omega = 2\pi f, \quad k = \frac{2\pi}{\lambda}$$

## Wave speed

$$p = p_+ \cos(\underbrace{\omega t - kz}_{\phi})$$

$$\frac{\partial^2 p}{\partial z^2} = -p_+ k^2 \cos(\omega t - kz)$$

$$\frac{\partial^2 p}{\partial t^2} = -p_+ \omega^2 \cos(\omega t - kz)$$

The wave equation

$$\frac{\partial^2 p}{\partial z^2} = K\rho \frac{\partial^2 p}{\partial t^2}$$

holds if

$$k^2 = \rho K \omega^2 \quad \rightarrow \quad c = \frac{1}{\sqrt{\rho K}} = \sqrt{\frac{E}{\rho}} \quad \text{because} \quad c = \frac{\omega}{k}$$

# Speed of Sound in Tissue



- The speed of sound in a human tissue depends on the average density  $\rho$  ( $\text{kg}\cdot\text{m}^3$ ) and the compressibility  $K$  ( $\text{m}^2\cdot\text{N}^{-1}$ ) of the tissue.

$$c = \frac{1}{\sqrt{\rho_0 K}}$$

## Other wave equation solution

$$p = p_- \cos(\omega t + kz)$$

Any forward or backward wave (by linearity and harmonic decomposition).

$$p = f_+(z + ct) + f_-(z - ct)$$

Forward and backward wave combination:

$$p = p' \left( \cos(\omega t - kz) + \cos(\omega t + kz) \right)$$

Standing wave:

$$p = 2p' \cos(\omega t) \cos(kz)$$



# Tissue Characteristics



- Engineers and scientists working in ultrasound have found that a convenient way of expressing relevant tissue properties is to use characteristic (or acoustic) impedance  $Z$  ( $\text{kg}\cdot\text{m}^{-2}\cdot\text{s}^{-1}$ )

$$Z = \rho_0 c$$

## Acoustic impedance

$$Z_a = \frac{p \text{ (pressure)}}{I \text{ (flow)}} \text{ [Pa} \cdot \text{s/m}^3\text{]}$$

“acoustic Ohm”.

For an infinite tube:

$$Z_a = \frac{\rho_0 c}{S}$$

$Z = \rho_0 c$  is a specific acoustic impedance.

Unit  $[\text{kg/s} \cdot \text{m}^2] = 1 \text{ Rayl}$ .

# Fyzikální základy - veličiny

Materiál	Rychlost zvuku $c$ $\text{m}\cdot\text{s}^{-1}$	Hustota $\rho$ $\text{kg}\cdot\text{m}^{-3}$	Akustická impedance $Z$ $\text{kg}\cdot\text{m}^{-2}\cdot\text{s}^{-1}$ ( $\times 10^6$ )	Koeficient absorpce $\alpha$ $\text{dB}\cdot\text{cm}^{-1}\cdot\text{MHz}^{-1}$
Vzduch	330	1,3	0,00043	
Tuk	1470	970	1,42	0,6
Ricínový olej	1500	933	1,40	
Voda	1492	1000	1,48	
Měkká tkáň	1500	<1000	~1,45	1,0
Mozek	1530	1020	1,56	0,85
Krev	1570	1020	1,60	0,18
Ledviny	1561	1030	1,61	
Játra	1549	1060	1,64	0,9
Sval	1568	1040	1,63	
Sval (podélná vlákna)			1,2	1,65
Sval (příčná vlákna)			3,3	1,65
Oční čočky	1620	1130	1,83	2,0
Kost	4080	1700	6,12	6,1
Plast			3,2	2,0

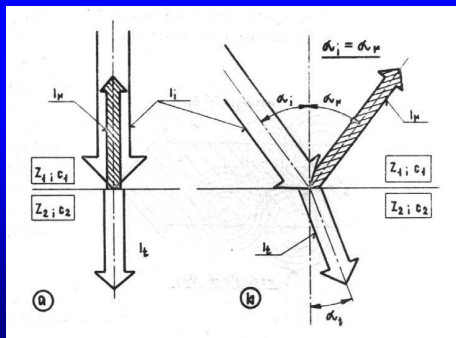
# Specular Reflection



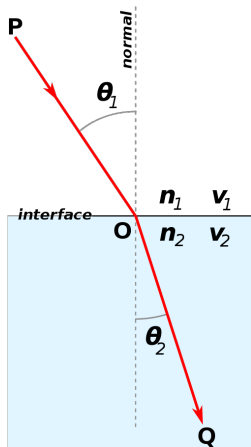
- The first, specular echoes, originate from relatively large, strongly reflective, regularly shaped objects with smooth surfaces. These reflections are angle dependent, and are described by reflectivity equation . This type of reflection is called specular reflection.

# Primární parametrické pole a modulace ultrazvukového signálu

- útlum UZV energie,
- odraz a lom UZV vln,



# Snell's law



$$\frac{\sin \theta_2}{\sin \theta_1} = \frac{v_2}{v_1} = \frac{c_2}{c_1}$$

# Reflectivity



$$R = \frac{p_r}{p_i} = \frac{\frac{Z_2}{\cos\theta_t} - \frac{Z_1}{\cos\theta_i}}{\frac{Z_2}{\cos\theta_t} + \frac{Z_1}{\cos\theta_i}}$$

At normal incidence,  $\theta_i = \theta_t = 0$  and

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

# Reflectivity for Various Tissues



<i>Materials at Interface</i>	<i>Reflectivity</i>
Brain-skull bone	0.66
Fat-muscle	0.10
Fat-kidney	0.08
Muscle-blood	0.03
Soft tissue-water	0.05
Soft tissue-air	0.9995



# Scattered Reflection



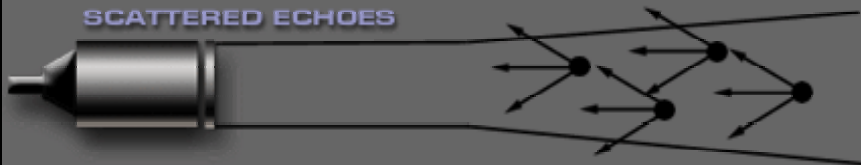
- The second type of echoes are scattered that originate from small, weakly reflective, irregularly shaped objects, and are less angle-dependent and less intense. The mathematical treatment of non-specular reflection (sometimes called “speckle”) involves the Rayleigh probability density function. This type of reflection, however, sometimes dominates medical images, as you will see in the laboratory demonstrations.



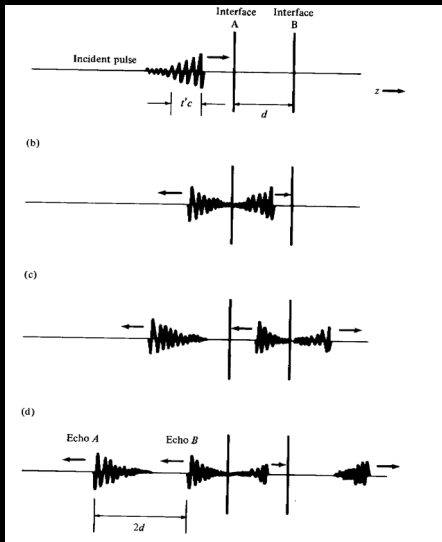
### SPECULAR ECHOES



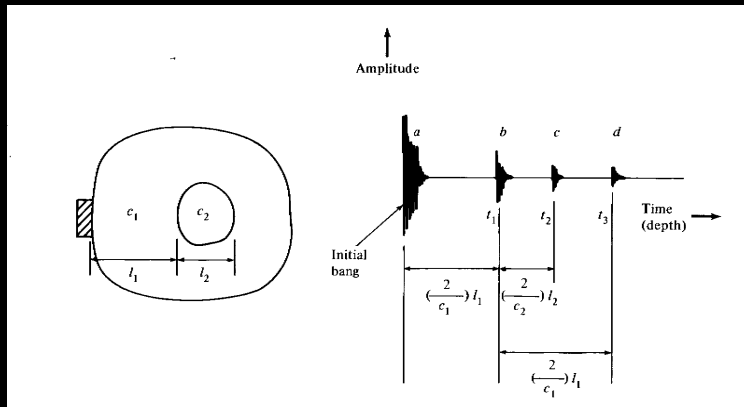
### SCATTERED ECHOES



# Echoes from Two Interfaces



# Echoes from Internal Organ



# Attenuation



- Most engineers and scientists working in the ultrasound characterize attenuation as the “half-value layer,” or the “half-power distance.” These terms refer to the distance that ultrasound will travel in a particular tissue before its amplitude or energy is attenuated to half its original value.

# Attenuation



- Divergence of the wavefront
- Elastic reflection of wave energy
- Elastic scattering of wave energy
- Absorption of wave energy

# Ultrasound Attenuation



<i>Material</i>	<i>Half-power distance (cm)</i>
Water	380
Blood	15
Soft tissue	5 to 1
except muscle	1 to 0.6
Bone	0.7 to 0.2
Air	0.08
Lung	0.05

# Attenuation



- As a general rule, the attenuation coefficient is doubled when the frequency is doubled.

$$I_{avg} = I_0 \exp\{-2\alpha z\}$$



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# Medical ultrasound devices



# Medical ultrasound devices



## Medical ultrasound devices



# Medical applications of ultrasound imaging

- ▶ Cardiology (heart)
- ▶ Gynecology: breast, fetus (pregnancy)
- ▶ Internal organs: liver, kidney, thyroid gland
- ▶ Intravascular ultrasound
- ▶ Therapeutic ultrasound: shock wave (kidney stone), thermal effects (rehabilitation)

# Imaging modes

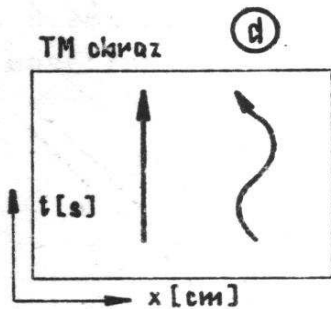
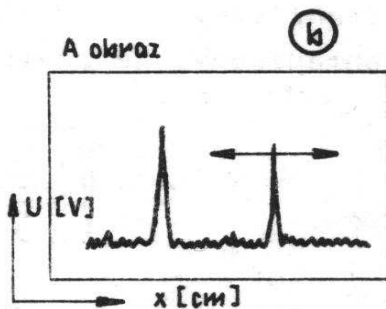
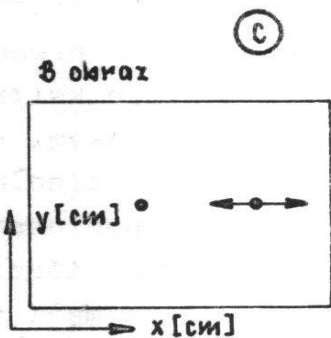
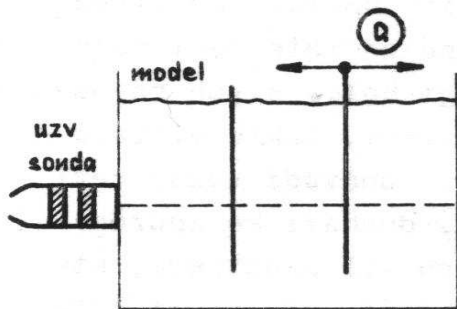
A osciloscopic, intensity/time

B **2D in the probe plane**

C 2D perpendicular

TM 1D+time

Q Doppler (speed)





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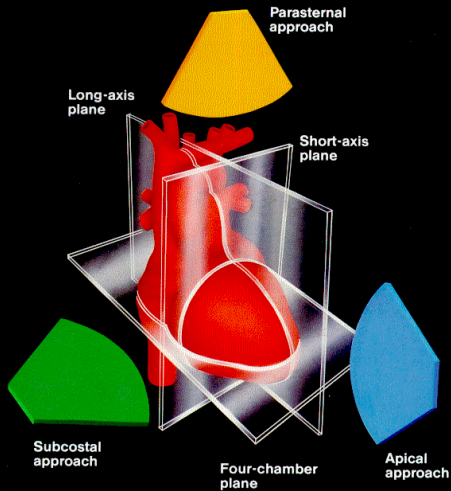
Generation

Steering/Beamforming

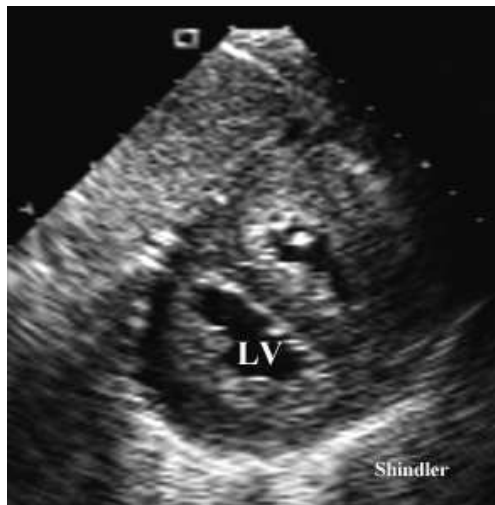
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Processing and control

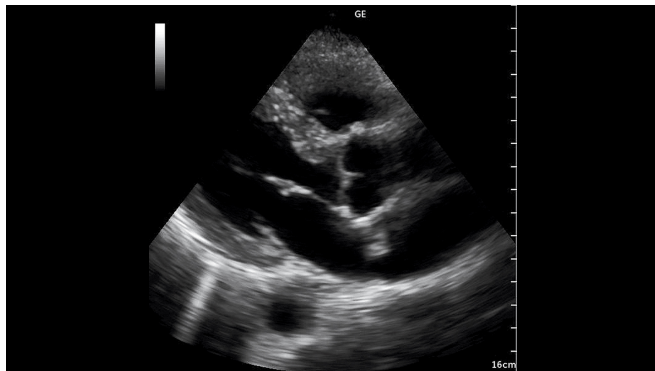
# Conventional Cardiac 2D Ultrasound



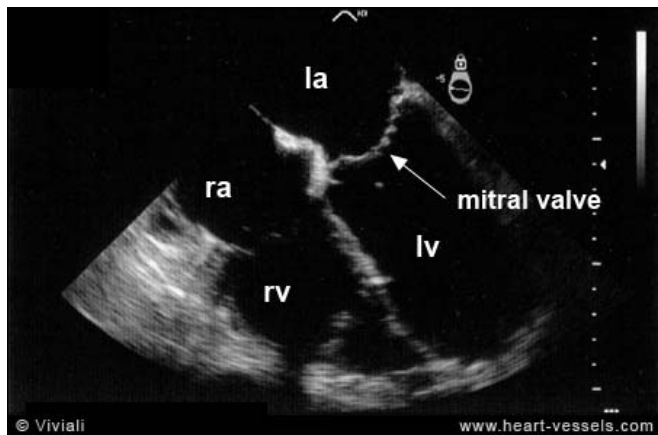
# Heart



# Heart



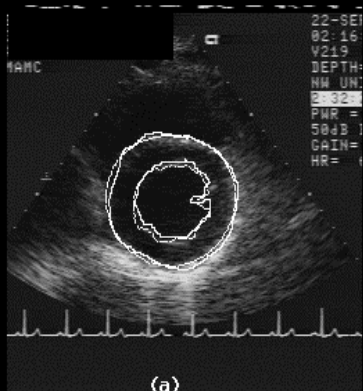
# Heart



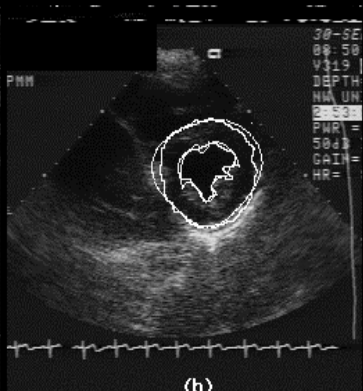
# B-mode Image of Heart



# Traditional Ultrasound Images



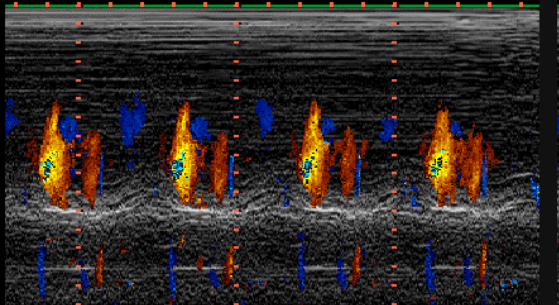
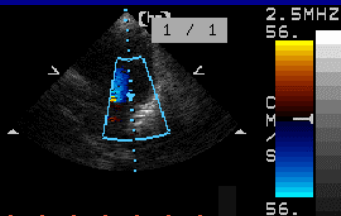
End-diastole



End-systole

TIS:0.7  
2.5/2.0-S/M GAIN 74 COMP 70  
RUSH PRES. ST. 16CM  
LUKES HOSP. PROC 2/0/A/B/A  
HP Adult  
ID:JS

20 SEP 96  
14:19:32





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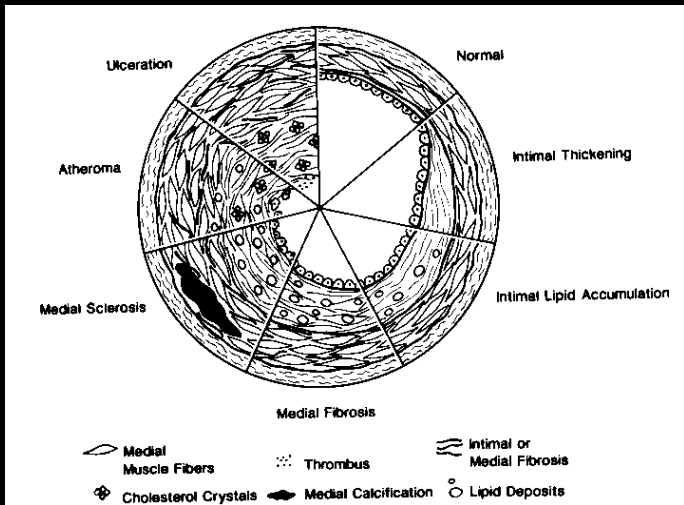
Generation

Steering/Beamforming

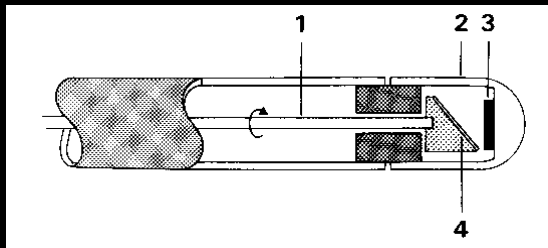
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# Progression of Vascular Disease

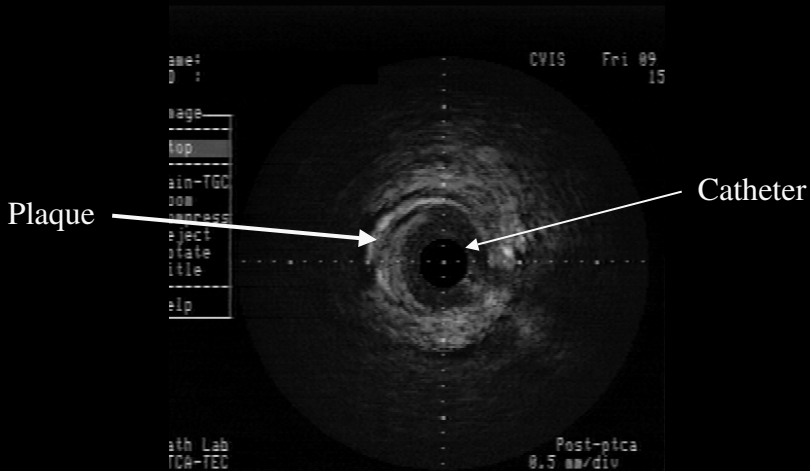


# IVUS Catheter

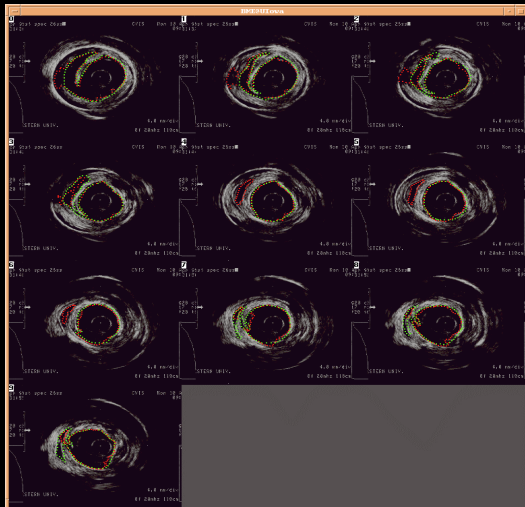


- 1 - Rotating shaft
- 2 - Acoustic window
- 3 - Ultrasound crystal
- 4 - Rotating beveled acoustic mirror

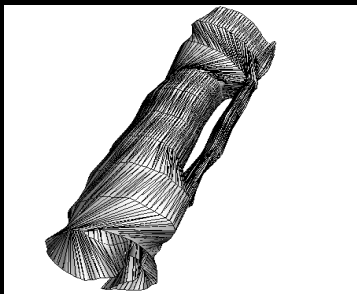
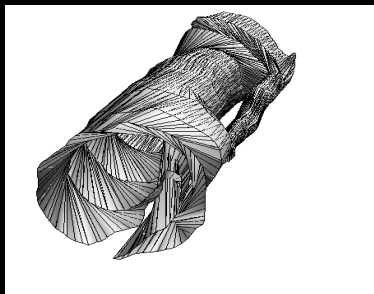
# Slightly Diseased Artery in Cross-section



# An array of Images



# 3D IVUS



## Other ultrasound examples



## Other ultrasound examples





# Other ultrasound examples



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**Artefacts**

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Generation

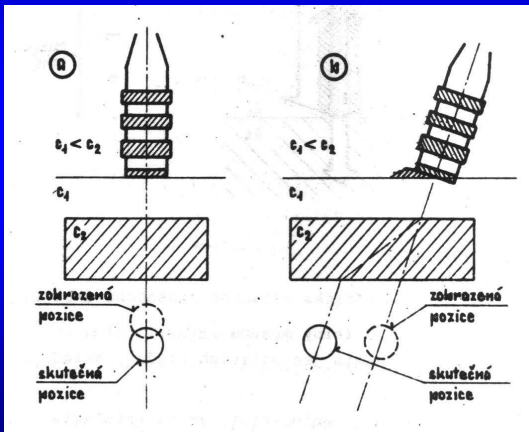
Steering/Beamforming

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Processing and control

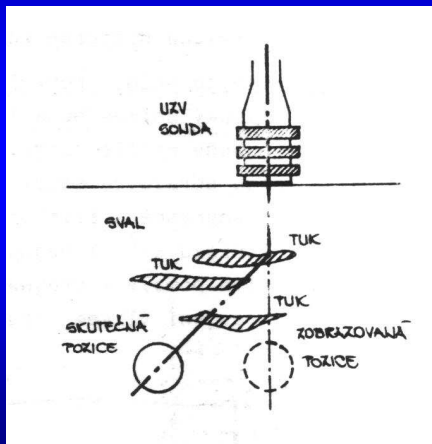
# Geometrická distorze UZV zobrazení

- změnou rychlosti šíření UZV vlny,



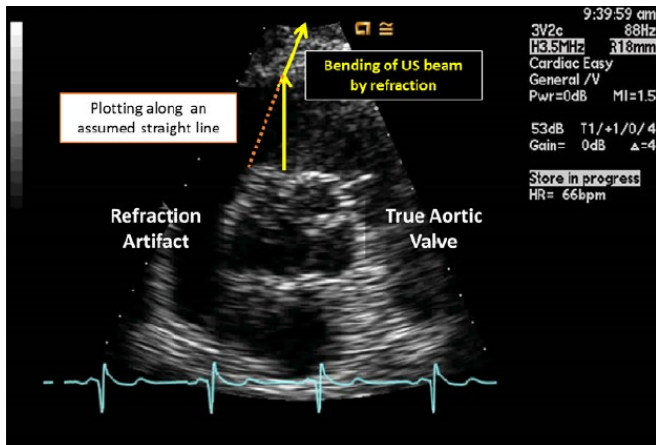
# Geometrická distorze UZV zobrazení

- skladbou tkání,



# Refraction artefacts

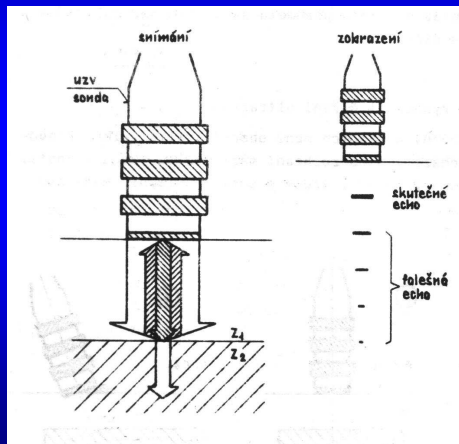
## Beam bending



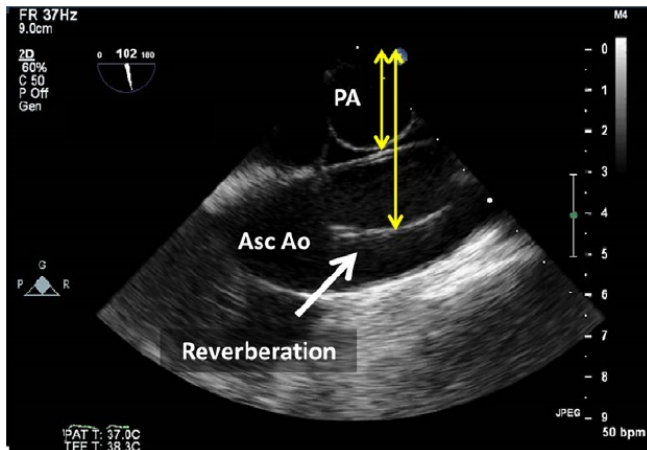
Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018

# Geometrická distorze UZV zobrazení

- násobnou reflexí,

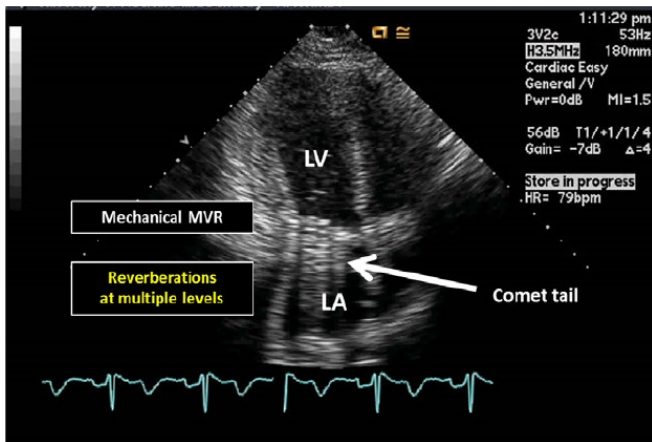


# Reverberation



Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018

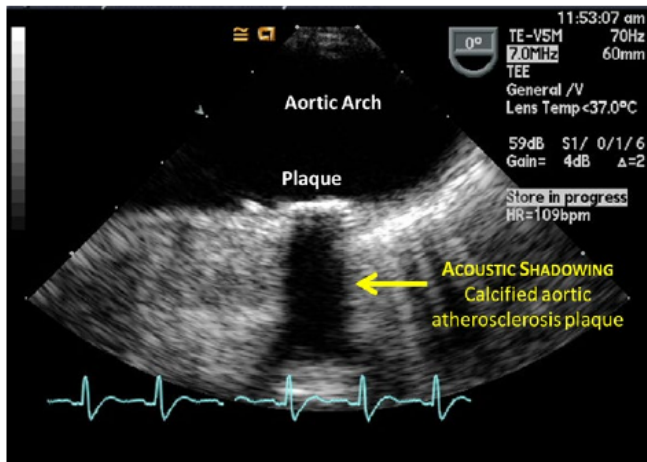
# Mitral prosthesis reverberation



Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018



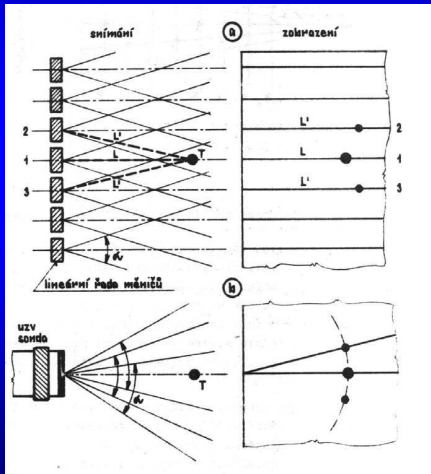
## Shadow (aortic plaque)



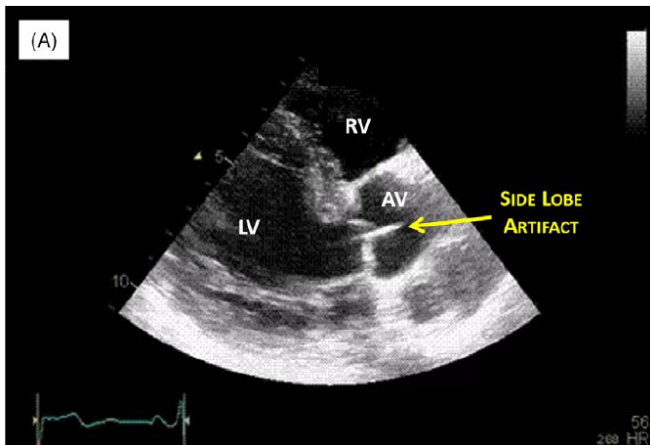
Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018

# Geometrická distorze UZV zobrazení

- vlivem konečné šířky UZV svazku,



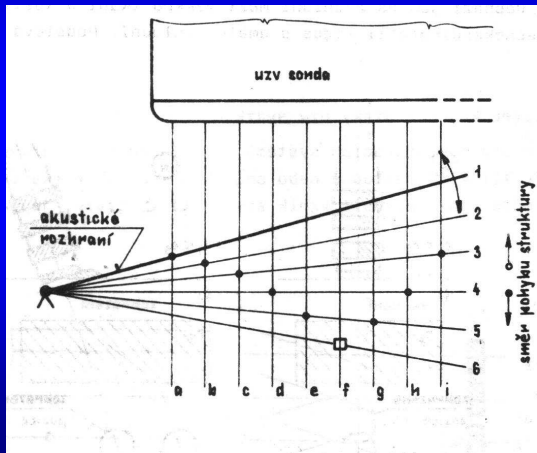
## Side lobe artifacts



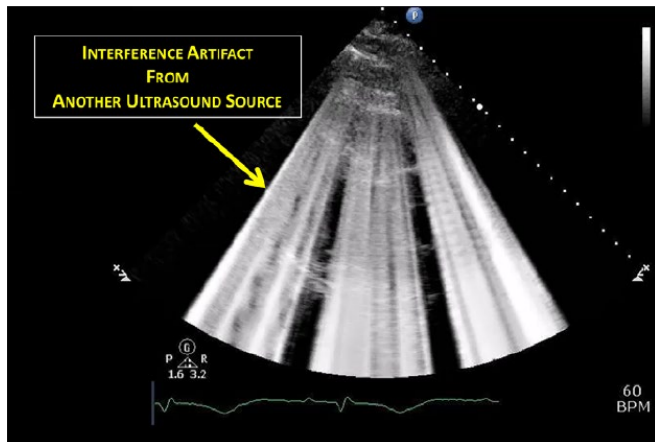
Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018

# Geometrická distorze UZV zobrazení

- pohybem tkáňových struktur,



# Interference artefacts



Quien, Saric: Ultrasound imaging artifacts: How to recognize them and how to avoid them. Echocardiography. 2018

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# Generace a detekce UZV signálu

---

- požadavky na konstrukci systému,
  - generace UZV impulsu 10 až 100 mW/cm<sup>2</sup>,
  - vysoký dosažený odstup S/Š,
  - malá akustická vazba mezi jednotlivými měniči,
  - krátký generovaný impuls ~ 2μs,
  - vysoká účinnost přenosu energie mezi měniči,
  - tlumení zpětné akustické vlny,
  - dosažení širokého úhlového krytí snímaného pole,
  - potlačení vibrací,
  - lehká, snadno manipulovatelná konstrukce,
- princip vstupní jednotky digitálního sonografu,

# Zdroje ultrazvukového vlnění

---

- zdrojem UZV vlnění UZV měnič v sondě,
- přímý a nepřímý piezoelektrický jev,
- charakteristickým parametrem sondy je rezonanční frekvence, určená tl. měniče,
- co nejkratší impuls při vysílání x velká citlivost,

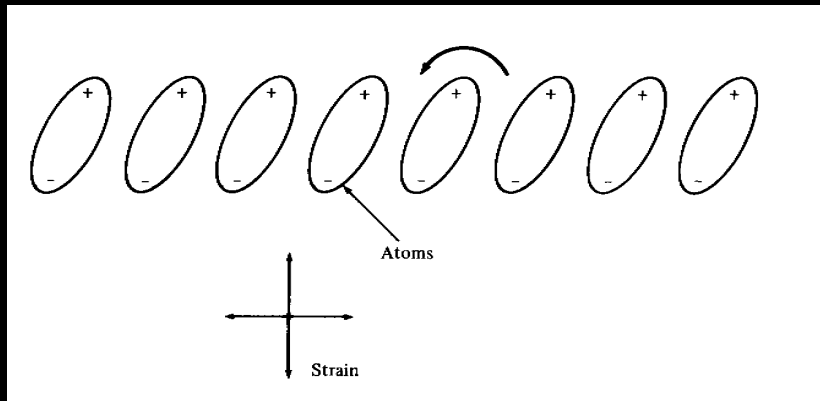


# Pressure Generation



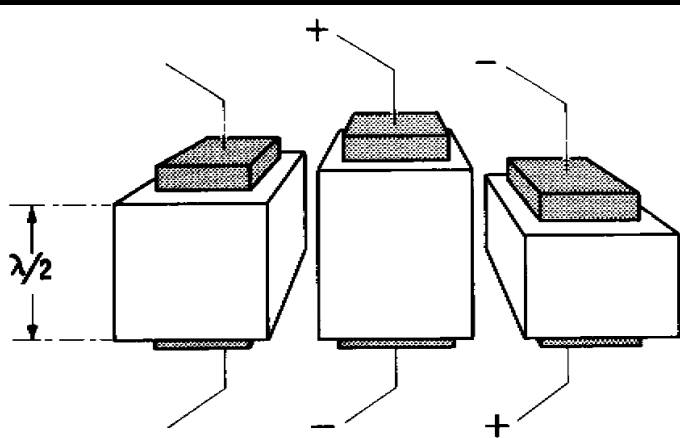
- Piezoelectric crystal
- 'piezo' means pressure, so piezoelectric means
  - pressure generated when electric field is applied
  - electric energy generated when pressure is applied

# Charged Piezoelectric Molecules



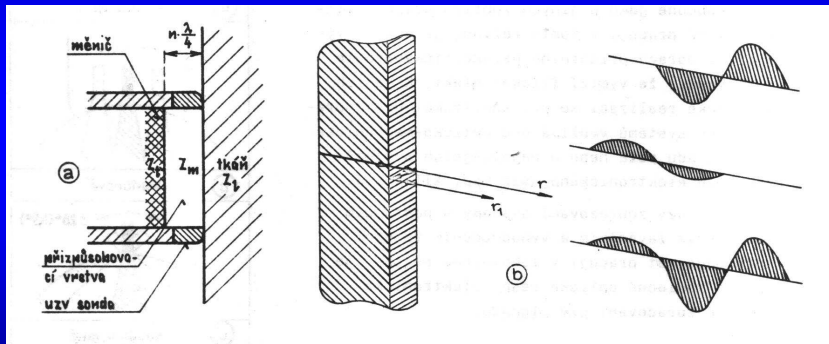
Highly simplified effect of  $E$  field

# Piezoelectric Effect

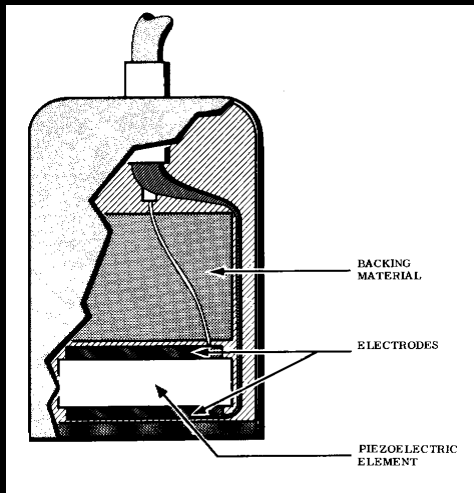


# Zpracování UZV signálu

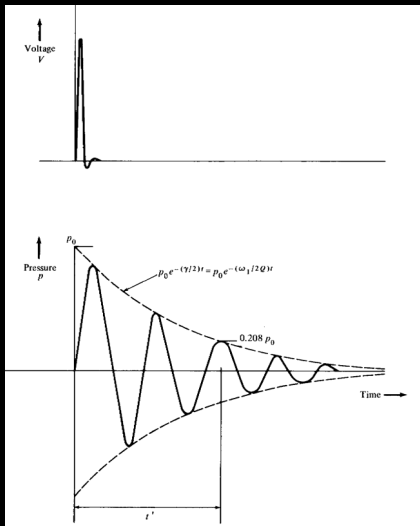
- přizpůsobení akustických impedancí



# Transducer

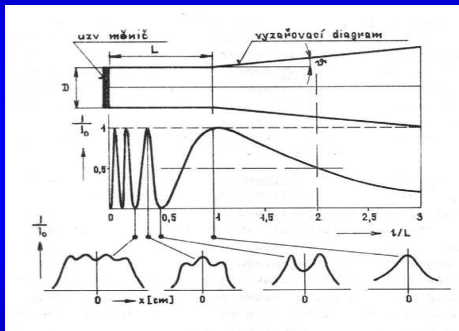


# Pressure Radiated by Sharp Pulse



# Zdroje ultrazvukového vlnění

- ultrazvukové pole,



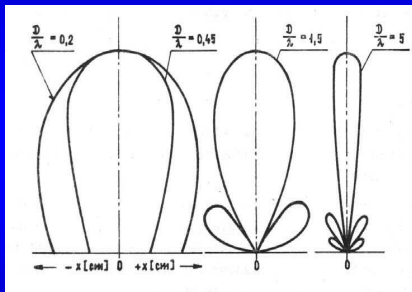
$$L = \frac{D^2 - \lambda^2}{4\lambda}$$

- blízké pole (blízká Fresnelova oblast),

- vzdálené pole (vzdálená Fraunhoferova oblast),

# Zdroje ultrazvukového vlnění

- vyzařovací diagram sondy,



- Fraunhoferova formule,

$$\sin \vartheta = 1,22 \frac{\lambda}{D}$$

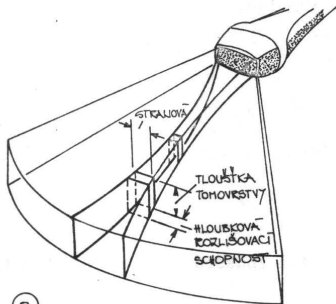
$$\sin \vartheta = \frac{\lambda}{b}$$

- významnou úlohu sehrává poměr  $\frac{D}{\lambda}$  ,

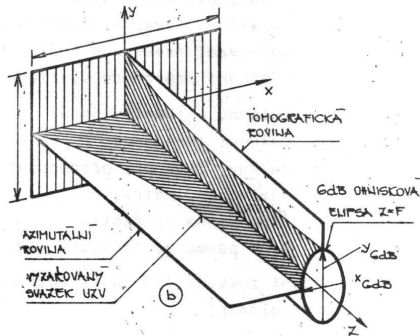
- postranní laloky - tlumení x akustická vazba,



# Rozlišovací schopnost



(a)



(b)



## Introduction

## Ultrasound acoustics

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Artefacts

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Generation

**Steering/Beamforming**

Focusing

Processing and control

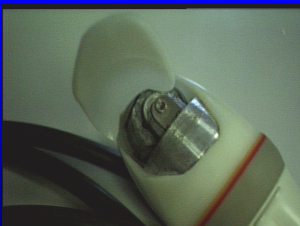
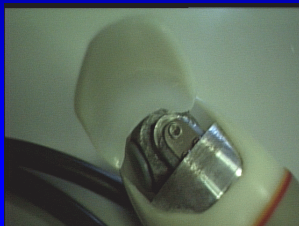
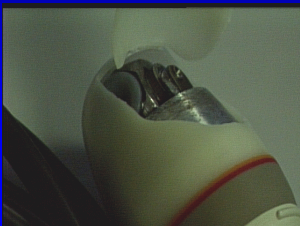
# Zpracování UZV signálu

---

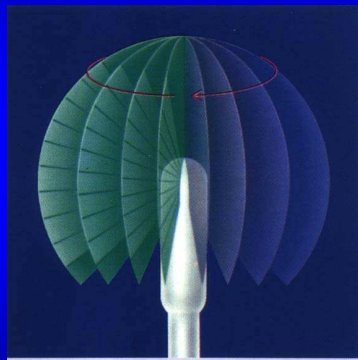
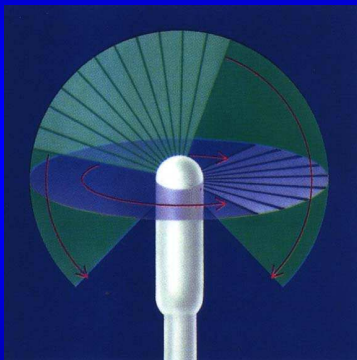
- vychylování UZV svazku - poziční jednotka
  - mech. systémy s lineárním snímáním,
  - mech. systémy se sektorovým snímáním,
    - rotační systém,
    - systém s kývající sondou,
  - elektronické systémy s lineárním snímáním,
  - elektronické systémy se sektorovým snímáním,
- fokuzace UZV svazku

# UZV sonda s mech. rozkladem - Siemens

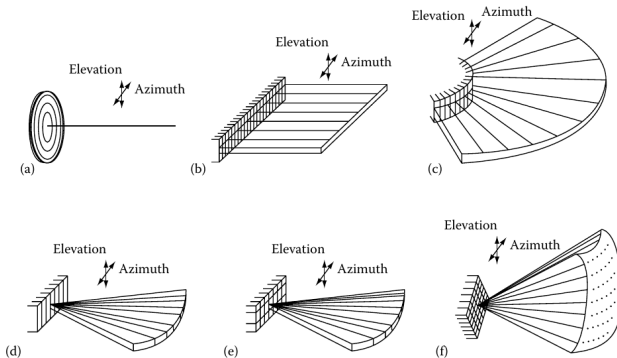
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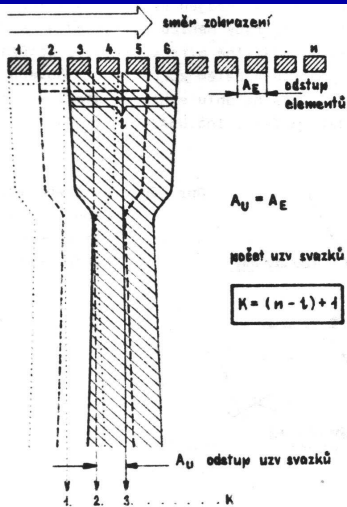
# UZV sonda s mech. rozkladem - Siemens



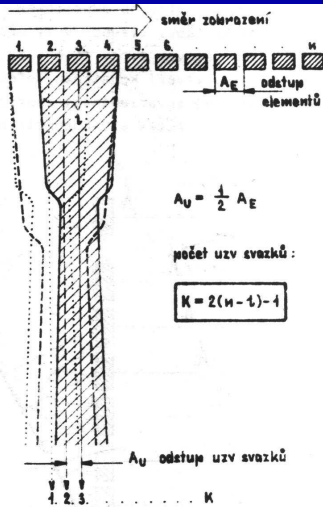
# Transducer array types



# El. systémy s lineárním snímáním



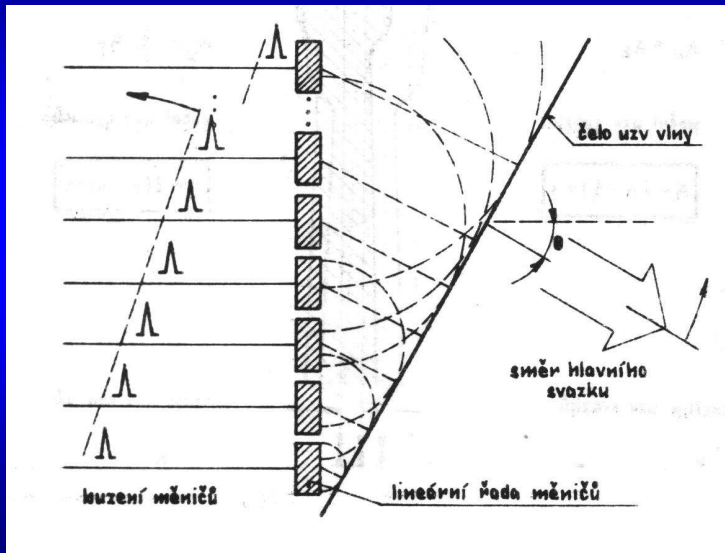
ⓐ



ⓑ



# El. systémy se sektorovým snímáním





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**Focusing**

Processing and control

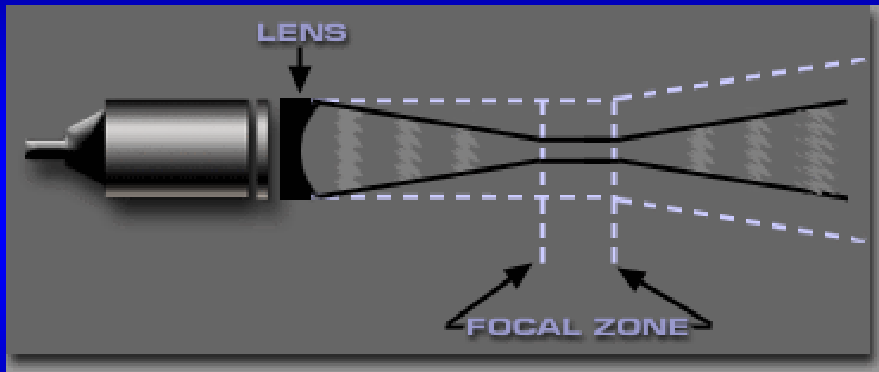
# Fokuzace svazku UZV signálu - typy

---

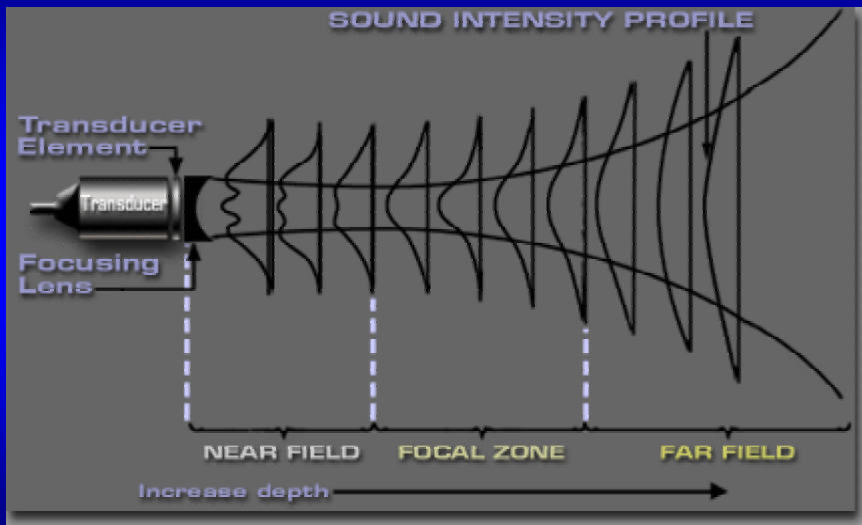
- fokuzace UZV čočkou,
- fokuzace zrcadly,
- elektronická fokuzace,
  - statická,
    - s lineární řadou měničů,
      - v režimu vysílání,
      - v režimu příjmu,
      - velikostí apertury,
      - elektronicko-optická,
    - s anulární sondou,
  - dynamická,
    - v režimu vysílání,



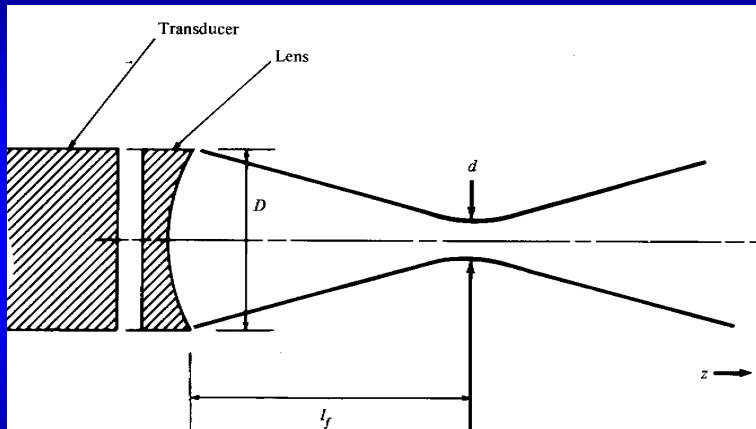
# Fokuzace čočkou



# Fokuzace čočkou



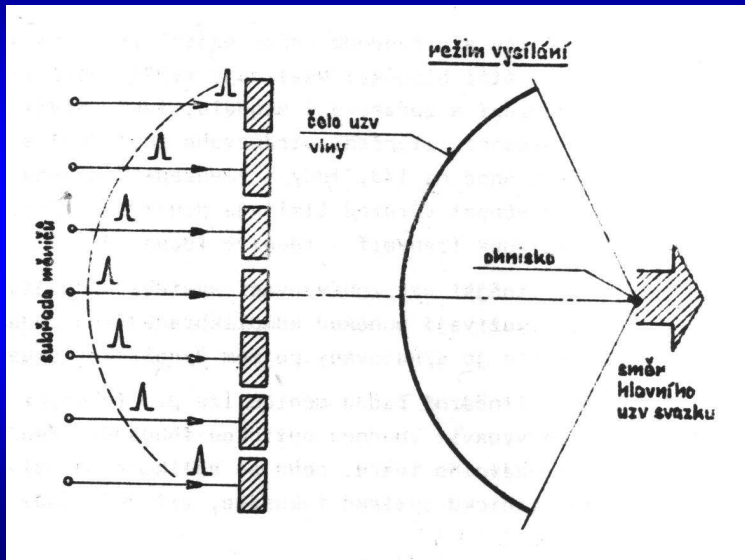
# Fokuzace čočkou



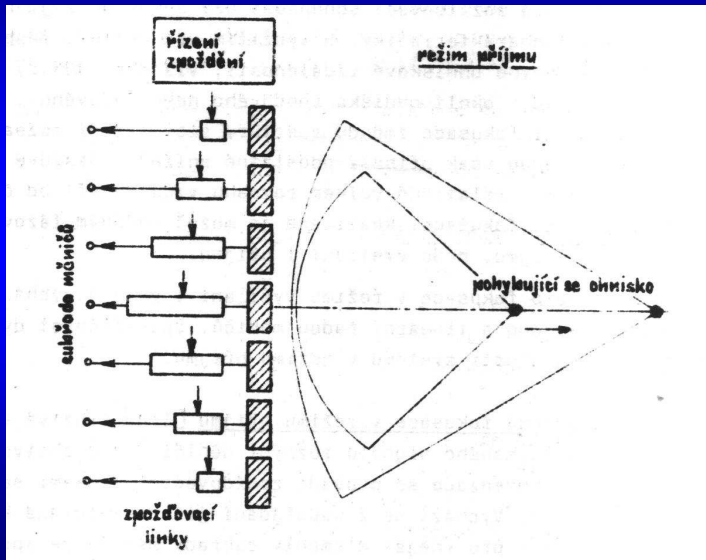
$$d = 2,44 \cdot \left( \frac{l_f}{D} \right) \cdot \lambda$$



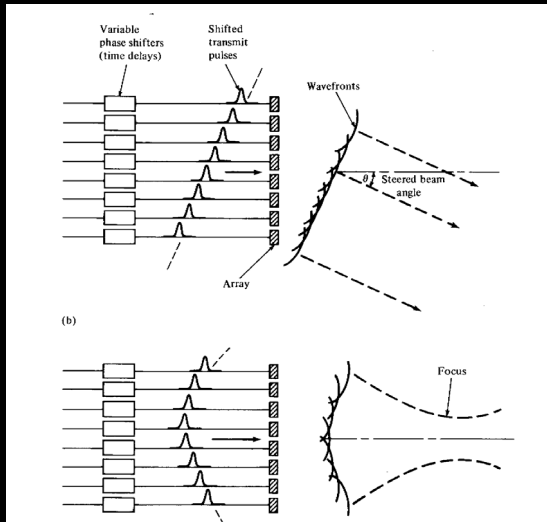
# El. fok. stat. s lin. řad. měn. v r. vysílání



# El. fok. stat. s lin. řad. měn. v r. příjmu

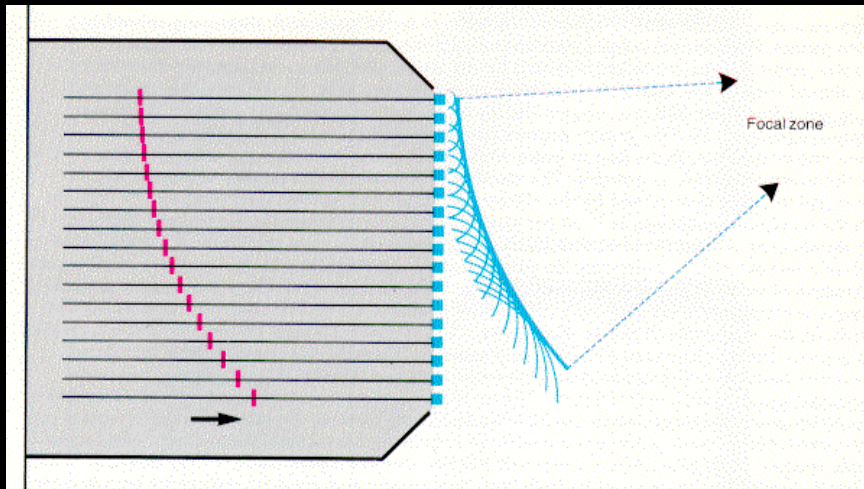


# Phased Linear Array

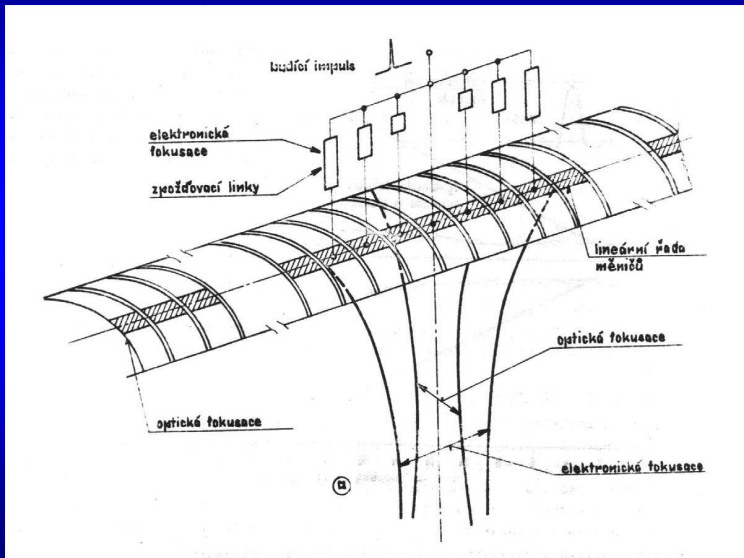




# Beam Direction



# El.-optická fok. stat. s lin. řad. měn.



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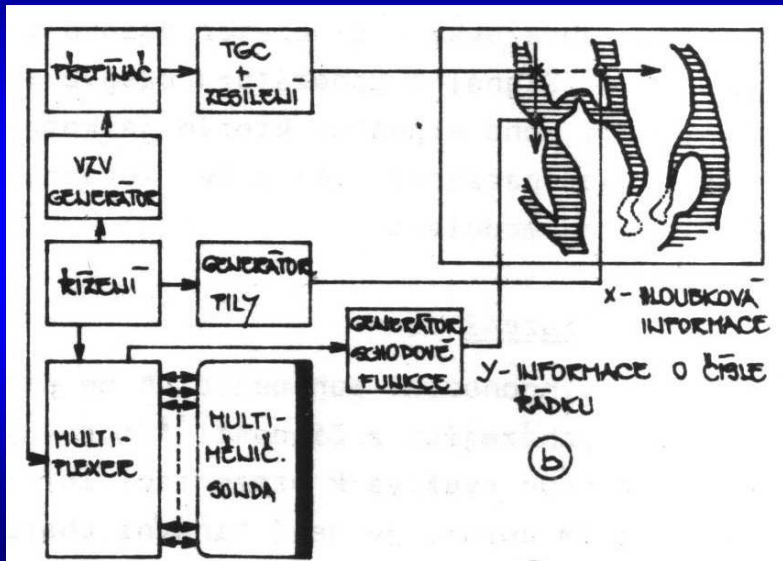
Generation

Steering/Beamforming

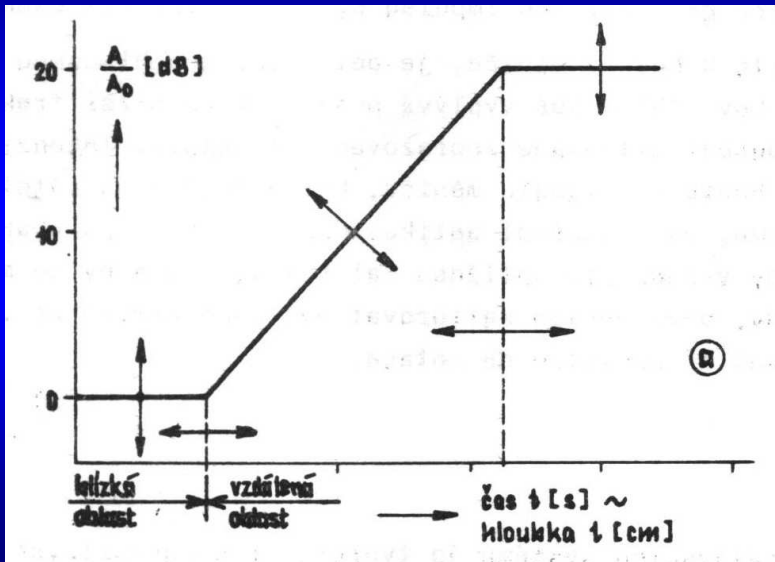
Focusing

Processing and control

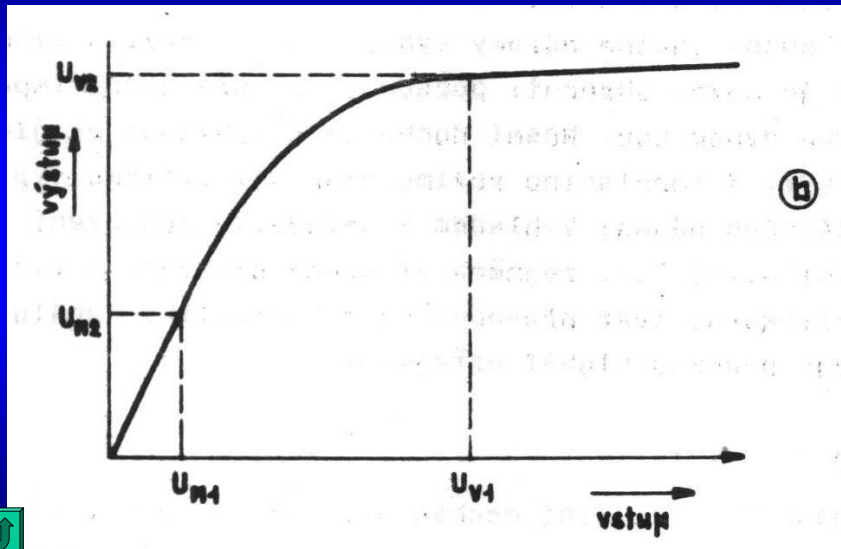
# Zpracování elektrického sign. - B mód



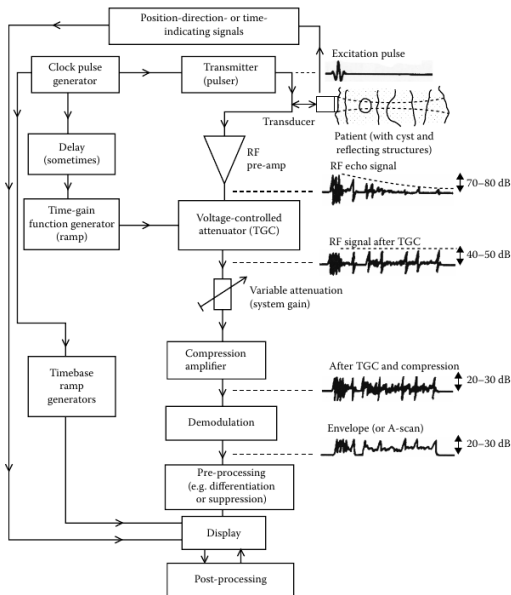
# Zesilovače s časově řízeným zesílením



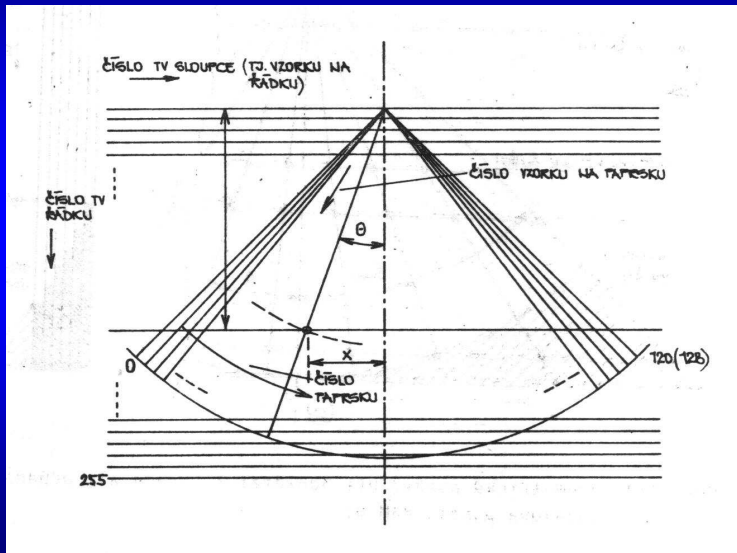
# Amplitudově řízené zesilovače



# Scanner block diagram



# Geom. vztah sekt. sním. a TV zobr. rastru





## Part II

### Modern ultrasound imaging

Doppler ultrasound

US contrast agents

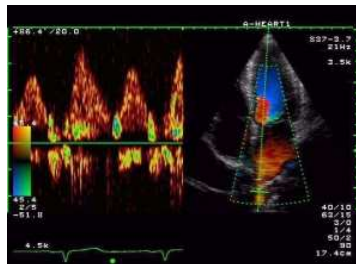
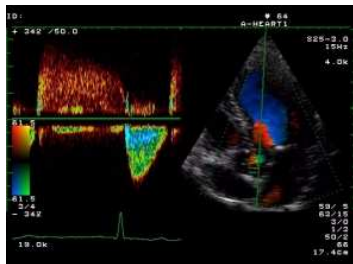
Harmonic imaging

3D US imaging

# DOPPLEROVSKÁ ULTRASONOGRAFIE

(principy přístrojů CW, PW, CDI)

Ing. Jiří Hozman



37ZSL2 Zobrazovací systémy v lékařství 2

1.3.2001

# Christian Andreas Doppler

(rakouský fyzik a matematik)

---

\* 29.11.1803 Salzburg, Austria

† 17.3.1853 Venice, Italy



1835 - počátek pobytu v Praze

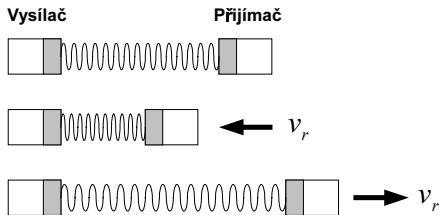
1842 - formulace Dopplerova principu

1845 - experimentální ověření

1847 - konec pobytu v Praze

## 1. Stacionární zdroj a pohybující se přijímač

$$\lambda_s = \frac{c}{f_s}$$

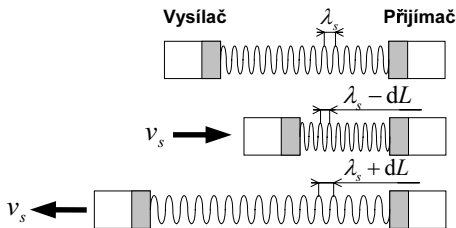


### Dopplerova frekvence

$$f_r = f_s + \frac{v_r}{\lambda_s} = f_s + \left( \frac{v_r}{c} \right) f_s \qquad f_D = \left( \frac{v_r}{c} \right) f_s$$

## 2. Stacionární přijímač a pohybující se zdroj

$$\lambda_s = \frac{c}{f_s}$$



$$dL = v_s \left( \frac{1}{f_s} \right)$$

$$\lambda_r = \lambda_s - dL = \frac{c}{f_s} - \frac{v_s}{f_s} = \frac{c}{f_r}$$

$$f_r = \frac{c}{c - v_s} f_s = \frac{1}{1 - \frac{v_s}{c}} f_s$$

**Použitím rozvoje do Taylorovy řady**

$$\frac{1}{1 - x} = 1 + x + \frac{x^2}{2} + \dots$$

**Všechny členy s  $x^2$  a vyšší mocninou zanedbáme a protože  $v/c \ll 1$  můžeme psát**

$$f_r = \left(1 + \frac{v_s}{c}\right) f_s \quad \text{Dopplerova frekvence} \quad f_D = \left(\frac{v_s}{c}\right) f_s$$

# Blood flow speed measurement

- ▶ Doppler effect: Frequency changes if the source moves with respect to the receiver.
- ▶ Reflection from red blood cells
- ▶ Red blood cells
  - ▶ Moving receiver
  - ▶ Moving source
- ▶ Doppler shift

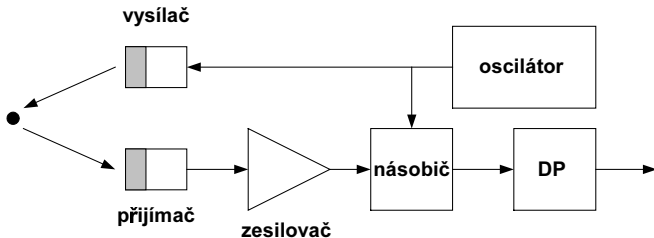
$$f_r = f_t + f_d \quad f_d \approx 2\frac{v}{c}f_c$$

- ▶ We only measure the projection along the ray.



## Demodulace Dopplerova signálu

---



**vysílaný signál**

$$E \cos(\omega_c t)$$

**přijatý signál**

$$V_i = A \cos(\omega_c t + \Phi) + B \cos(\omega_c + \omega_D)t$$

## Výsledkem násobení v demodulátoru je

$$V_A = AE \cos(\omega_c t + \Phi) \cos(\omega_c t) + BE \cos(\omega_c + \omega_D) t \cos(\omega_c t)$$
$$= \left(\frac{AE}{2}\right) \boxed{\cos(2\omega_c t + \Phi)} + \boxed{\cos(\Phi)} + \left(\frac{BE}{2}\right) \boxed{\cos(2\omega_c t + \omega_D t)} + \boxed{\cos(\omega_D t)}$$

Jednotlivé složky ve výše uvedeném vztahu znamenají:

$\boxed{\cos(2\omega_c t + \Phi)}$  <sup>potlačení</sup> dvojnásobek vysílané frekvence (DP)

$\boxed{\cos(2\omega_c t + \omega_D t)}$  <sup>potlačení</sup> dvojnásobek vysílané frekvence (DP)

$\boxed{\cos(\Phi)}$  <sup>potlačení</sup> stejnosměrná složka (HP)

$\boxed{\cos(\omega_D t) = \cos(-\omega_D t)}$  Dopplerův signál, nelze určit směr



## Směrové demodulační systémy

---

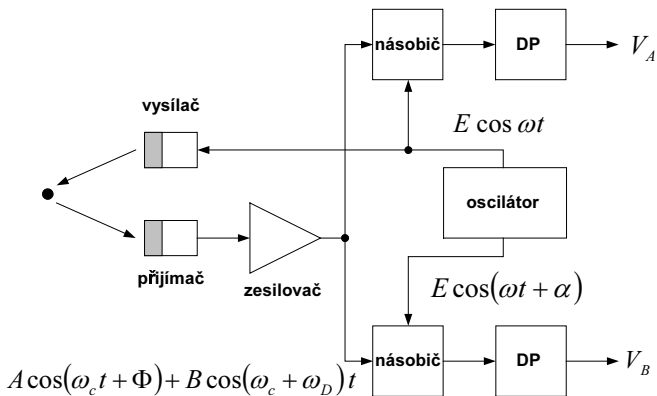
### Základní myšlenka

**Převést směrovou informaci, danou znaménkem frekvenčního posuvu na jiný indikátor směru, který by po demodulaci zůstal zachován**

### Příklad

**Pro vysílanou frekvenci 5 MHz a frekvenci Dopplerova signálu 5,8 kHz musíme odlišit kladný směr toku krve, tj. 5,0058 MHz a záporný směr toku krve, tj. 4,9942 MHz.**

## Základní uspořádání směrového demodulátoru



## Fázové zpracování kvadrurních signálů $V_A$ a $V_B$

$$V_A = \frac{1}{2} BE \cos(\omega_D t) \quad V_B = \frac{1}{2} BE \cos(\omega_D t - \alpha)$$

**fázový posuv  $\alpha = \pi/2 \rightarrow \sin$  a  $\cos \rightarrow$  kvadrurní signály**

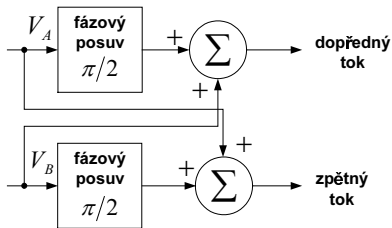
**Pokud zanedbáme amplitudy, pak dostaneme**

$$V_A = \cos \omega_D t \quad V_B = -\sin \omega_D t$$

**forward flow (dopředný tok) - od sondy, tj.  $\omega_f = -\omega_D$**

**reverse flow (zpětný tok) - k sondě, tj.  $\omega_r = \omega_D$**

$$V_A = \cos \omega_f t + \cos \omega_r t \quad V_B = \sin \omega_f t - \sin \omega_r t$$



$$\cos \omega_f t + \cos \omega_r t \quad 0 \quad \rightarrow \quad \cos \omega_f t + \cos \omega_r t \quad (1)$$

$$\pi/2 \quad \rightarrow \quad \sin \omega_f t + \sin \omega_r t \quad (2)$$

$$\sin \omega_f t - \sin \omega_r t \quad 0 \quad \rightarrow \quad \sin \omega_f t - \sin \omega_r t \quad (3)$$

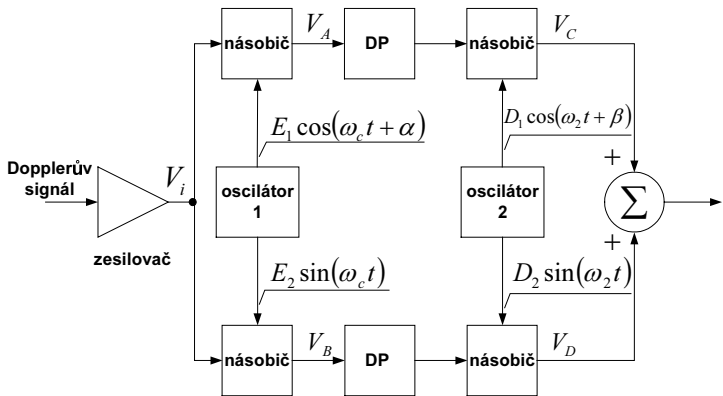
$$\pi/2 \quad \rightarrow \quad -\cos \omega_f t + \cos \omega_r t \quad (4)$$

$$(2) + (3) = \sin \omega_f t$$

$$(1) + (4) = \cos \omega_r t$$



# Úplný systém k separaci signálové složky odpovídající dopřednému a zpětnému toku ve frekvenční oblasti



**Hlavní myšlenka - posun nulové frekvence, tj. nulové rychlosti na jinou frekvenci  $\omega_2$**

$\alpha$  a  $\beta$  představují chybu ve fázovém rozdílu

$$V_i = \boxed{A \cos(\omega_c t + \Phi)} + \boxed{B \cos(\omega_c + \omega_D)t}$$

signál nosné                      Dopplerův signál

$$V_A = \frac{1}{2} B E_1 \cos(\omega_D t - \alpha) \quad V_B = -\frac{1}{2} B E_2 \sin(\omega_D t)$$

$$V_C = (B E_1 D_1 / 2) \cos(\omega_D t - \alpha) \cos(\omega_2 t + \beta)$$

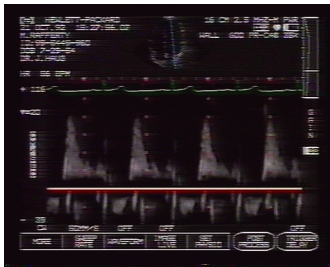
$$V_C = \frac{1}{4} B E_1 D_1 [\cos(\omega_D t + \omega_2 t - \alpha + \beta) + \cos(\omega_D t - \omega_2 t - \alpha - \beta)]$$



## Kontinuální Dopplerovské systémy CW

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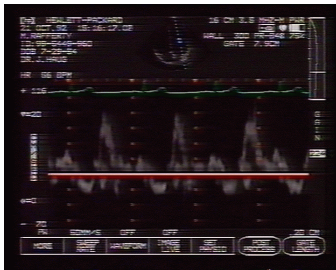
- kontinuální vysílání i příjem,
- v sondě dva piezo-elementy,
- vzorkovací objem dán šířkou UZ svazku,
- možnost detekovat velké rychlosti průtoku,
- nerozliší hloubku - množství spektrálních složek.



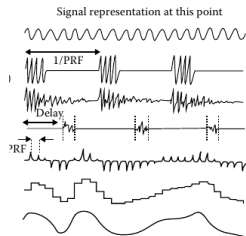
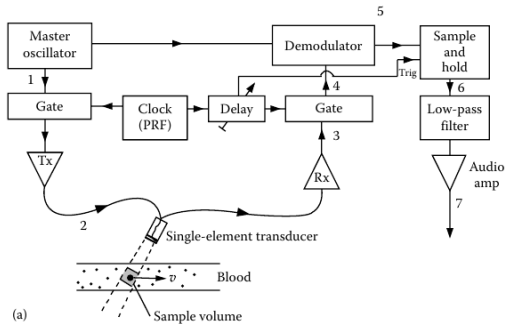
## Pulzní Dopplerovské systémy PW (Pulsed Wave)

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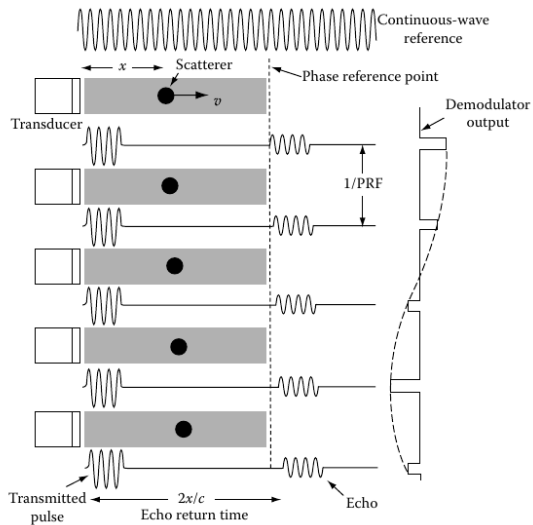
- pulzní vysílání i příjem,
- v sondě jeden piezo-element,
- vzorkovací objem dán délkou pulzu,
- omezení rozsahu detekovaných rychlostí průtoku,
- rozliší hloubku - méně spektrálních složek.



# PW Doppler block diagram



# PW Doppler aliasing



## Omezení rozsahu detekovaných rychlostí průtoku u PW

$$\Phi = f_c T_D \quad \frac{d\Phi}{dt} = f_c \frac{T_D}{dt} \quad T_D = \frac{2z}{c}$$

$$\frac{dT_D}{dt} = \frac{2}{c} \frac{dz}{dt} = \frac{2v}{c} \quad f_D = \frac{d\Phi}{dt} = f_c \frac{T_D}{dt} = f_c \frac{2v}{c}$$

$$T_p = \frac{1}{f_p}$$

$$f_{D_{\max}} = \frac{1}{2T_p} = \frac{2v_{\max}}{c} f_c \quad \text{dáno } f_{\text{Nyquist}}$$

$$z_{\max} = \frac{c}{2} T_p$$

$$T_p = \frac{c}{4v_{\max} f_c} = \frac{2z_{\max}}{c}$$

$$v_{\max} z_{\max} = \frac{c^2}{8f_c}$$

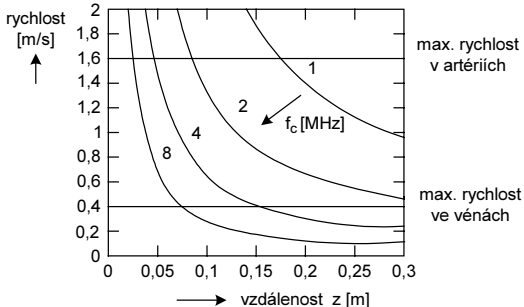
## Typický příklad

$$f_c = 2\text{MHz} \quad c = 1500\text{ms}^{-1} \quad v_{\max} z_{\max} = 0,14\text{m}^2\text{s}^{-1}$$

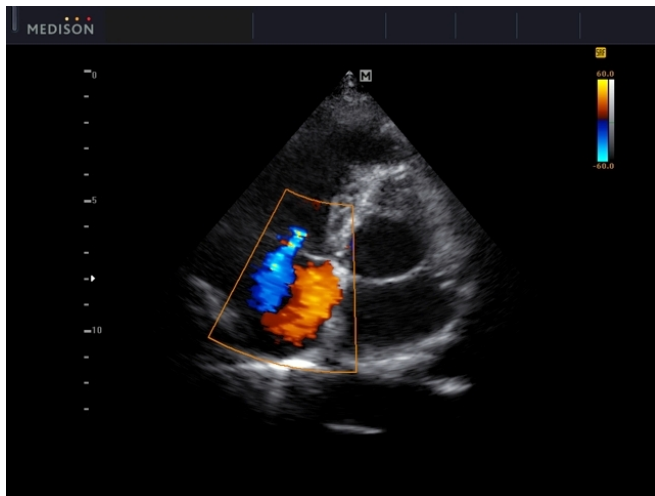
Ve vzdálenosti 10cm můžeme naměřit maximální rychlost

$$1,4\text{ms}^{-1}$$

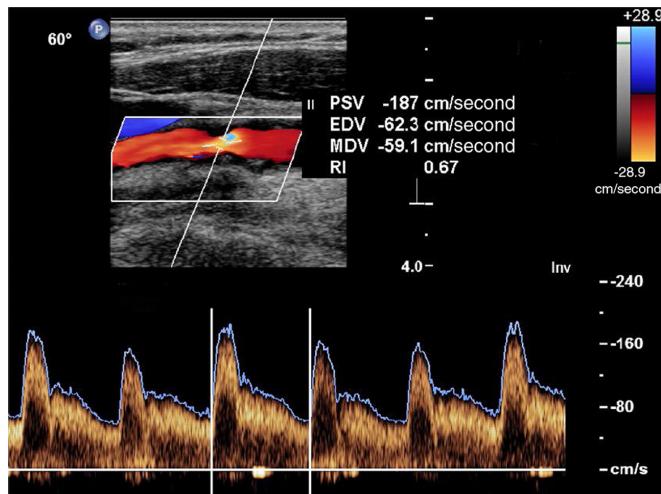
$$v_{\max} = \frac{c^2}{8f_c z_{\max}}$$



# Doppler US — examples

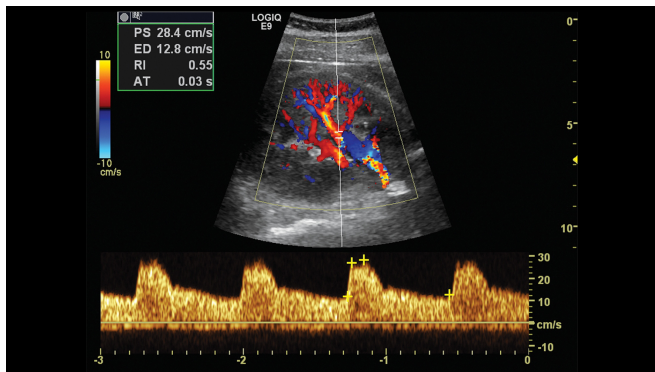


# Doppler US — examples





# Doppler US — examples



Doppler ultrasound

US contrast agents

Harmonic imaging

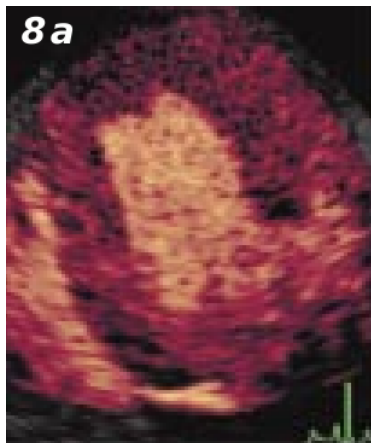
3D US imaging

## Contrast agents

- ▶ 1968, Gramiak, saline injection
- ▶ Mikrobubbles ( $2 \sim 5 \mu\text{m}$ )
- ▶ Asymmetric compression/expansion
- ▶ Stabilization (synthetic polymers), up to 5 – 10 min.
- ▶ Injection.
- ▶ Albutex, Optison, Echovist, Levovist. . .

# Flash Contrast Imaging

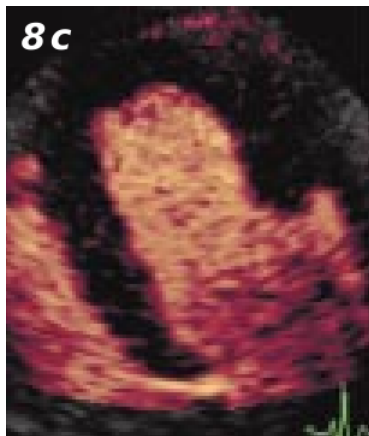
US bubble destabilization.



normal

# Flash Contrast Imaging

US bubble destabilization.



flash, bubbles broken

## Flash Contrast Imaging

US bubble destabilization.



filling up

Myocardial perfusion evaluation.

Doppler ultrasound

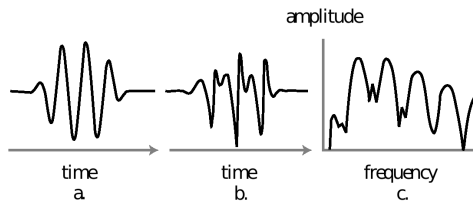
US contrast agents

Harmonic imaging

3D US imaging

# Nonlinear response

## Assymmetric bubble compression



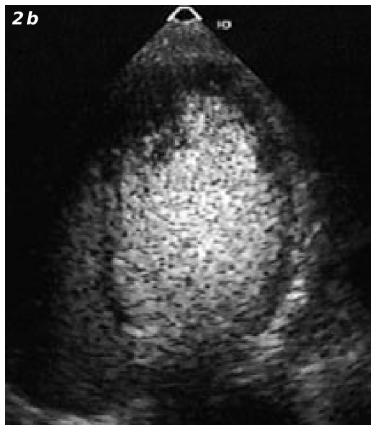


## Harmonic imaging

- ▶ Transmit  $f_0$ , receive  $2f_0$



standard US



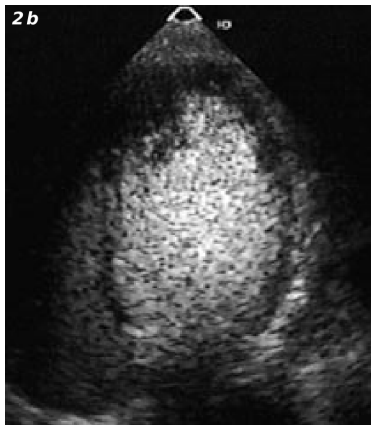
2<sup>nd</sup> harmonic

## Harmonic imaging

- ▶ Transmit  $f_0$ , receive  $2f_0$
- ▶ Bandwidth limitation



standard US



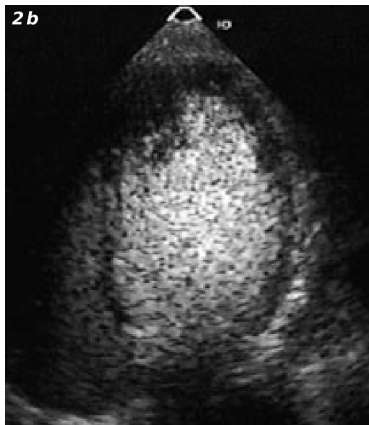
2<sup>nd</sup> harmonic

## Harmonic imaging

- ▶ Transmit  $f_0$ , receive  $2f_0$
- ▶ Bandwidth limitation
- ▶ Bubbles not needed, tissue nonlinearity



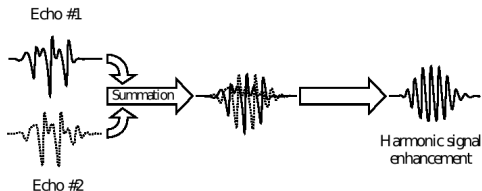
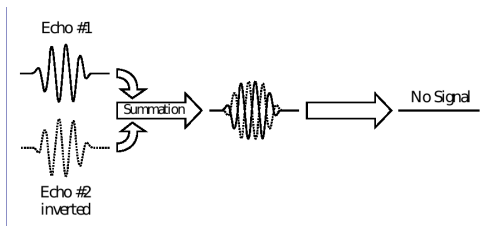
standard US



2<sup>nd</sup> harmonic

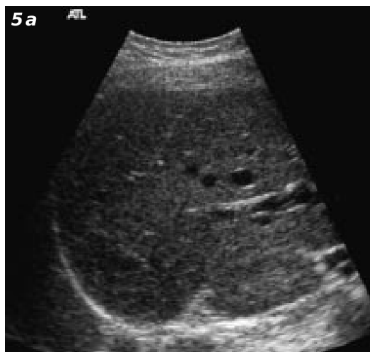
# Pulse Inversion Harmonic Imaging

- ▶ Two pulses, second inverted
- ▶ Responses summed
- ▶ Filtration not needed

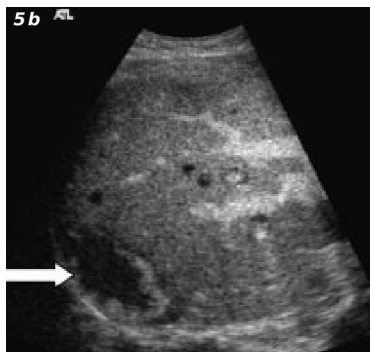


## Pulse Inversion Harmonic Imaging

- ▶ Two pulses, second inverted
- ▶ Responses summed
- ▶ Filtration not needed



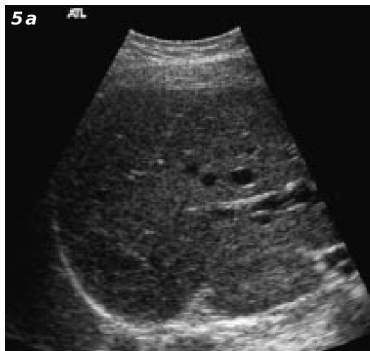
standard image (liver)



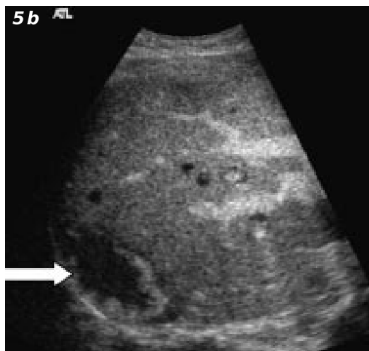
pulse inversion

## Pulse Inversion Harmonic Imaging

- ▶ Two pulses, second inverted
- ▶ Responses summed
- ▶ Filtration not needed
- ▶ Several pulses (Power Pulse Inversion)



standard image (liver)



pulse inversion

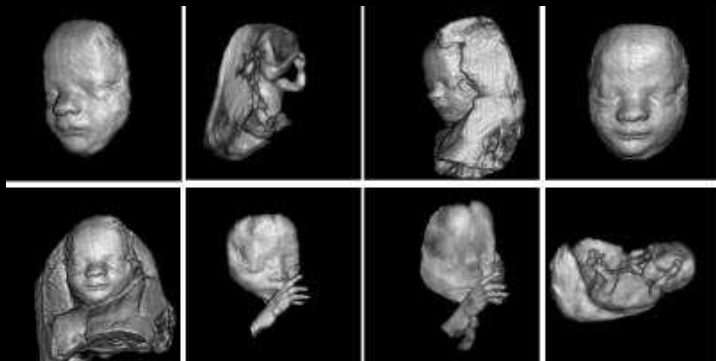
Doppler ultrasound

US contrast agents

Harmonic imaging

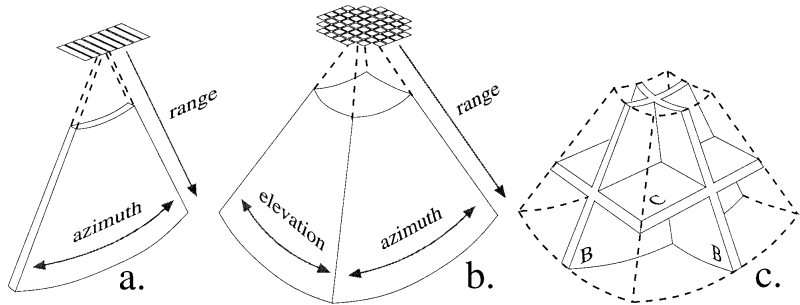
3D US imaging

# 3D Reconstruction





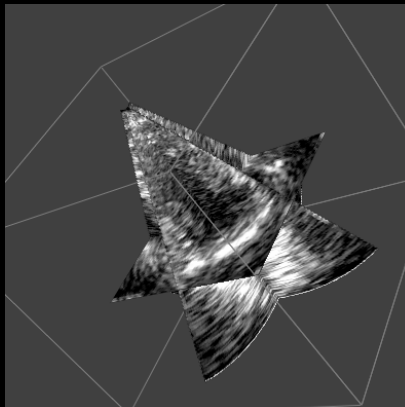
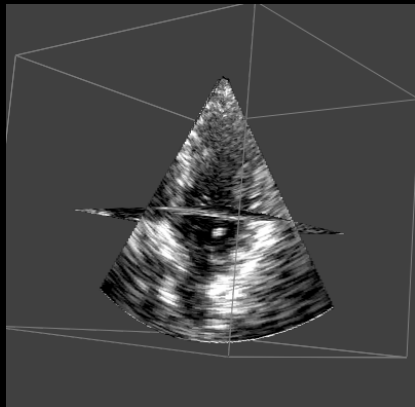
# 3D Ultrasound



Traditional 2D

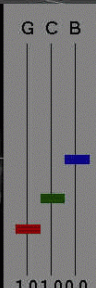
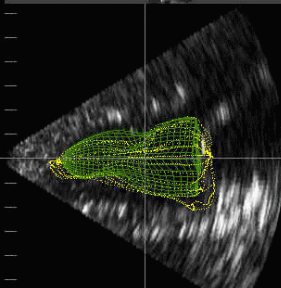
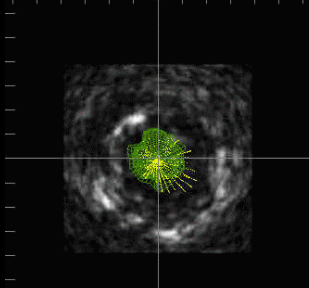
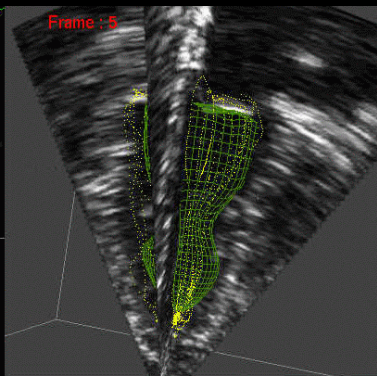
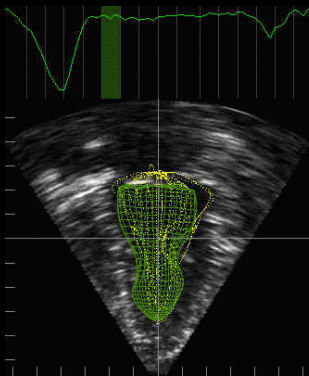
New 3D

# Real-time 3D Ultrasound

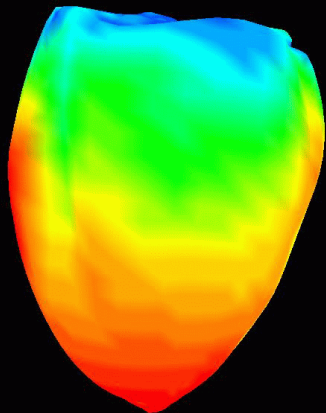




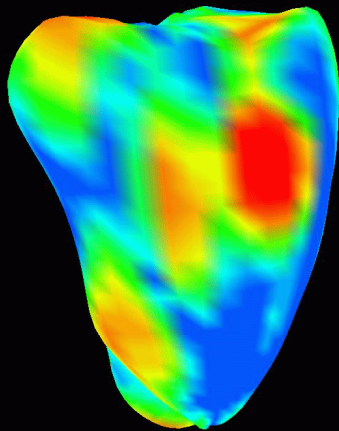
Frame : 5



# Velocity of Contraction



Normal



Abnormal

# Conclusions

- ▶ Non-invasive, affordable and portable imaging technique
- ▶ Excellent soft tissue imaging
- ▶ Lower image quality (wrt CT or MRI) due to speckle but improving
- ▶ Low penetration depth versus resolution
- ▶ Does not pass through air or gas
- ▶ Does not pass through bones, shadows
- ▶ Modern techniques — 3D, contract agents, Doppler