Diagnostic Ultrasound: Principles and Applications

Daniel F. Leotta, PhD

University of Washington
Applied Physics Laboratory
Center for Industrial and Medical Ultrasound

CSE/EE577
Fall 2011

Ultrasound Imaging

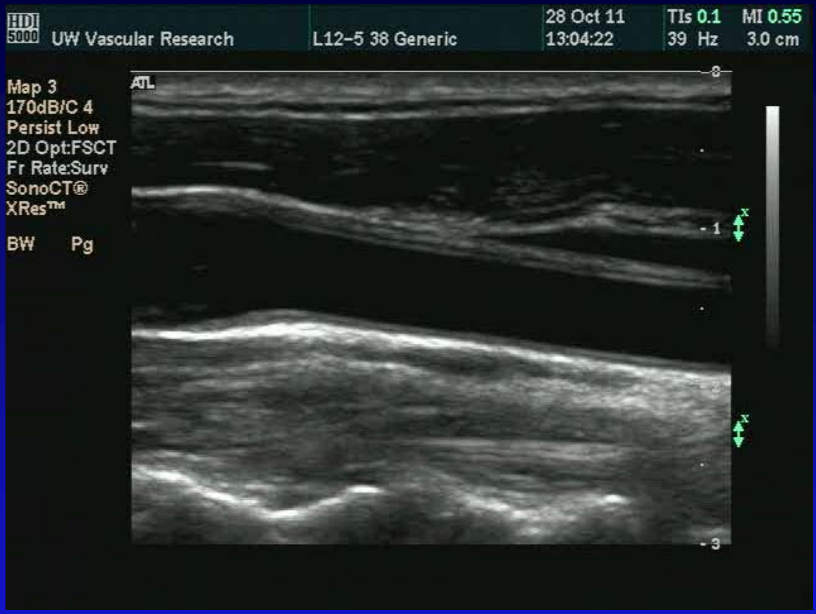
Medical imaging modality based on high-frequency sound waves

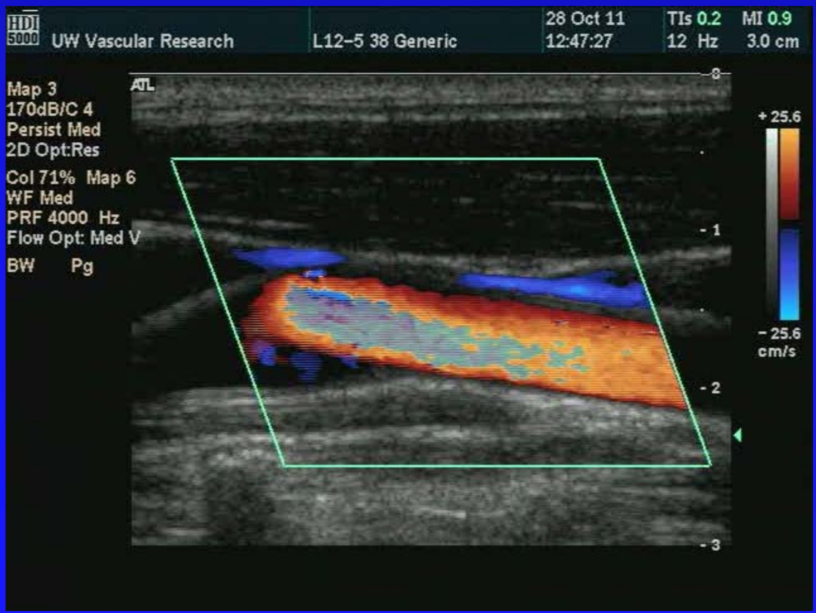
Ultrasound Imaging

- Ultrasound imaging is based on echo-ranging principles
  - Short-duration sound pulses are transmitted into the body
  - Received echoes are used to construct 2D images of tissue

Ultrasound Instruments

- Key advantages
  - No ionizing radiation
  - Real-time display
  - Anatomy and physiology (Doppler)
  - Relatively inexpensive
  - Portable





Ultrasound Instruments

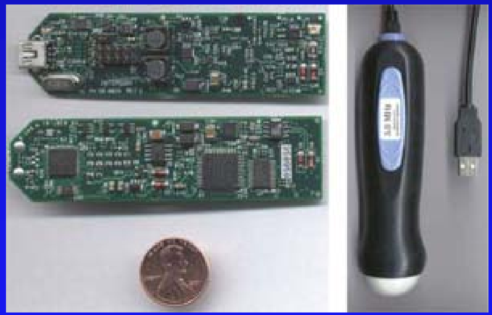
iU22 (Philips Ultrasound, Bothell)

180PLUS
(SonoSite, Bothell)
- Late 1990s

Ultrasound Instruments

- GE LogiqBook
- GE Vscan, 2009

Ultrasound Instruments



- Cell phone imager (2009)
  - Richard and Zar, Washington U, St. Louis

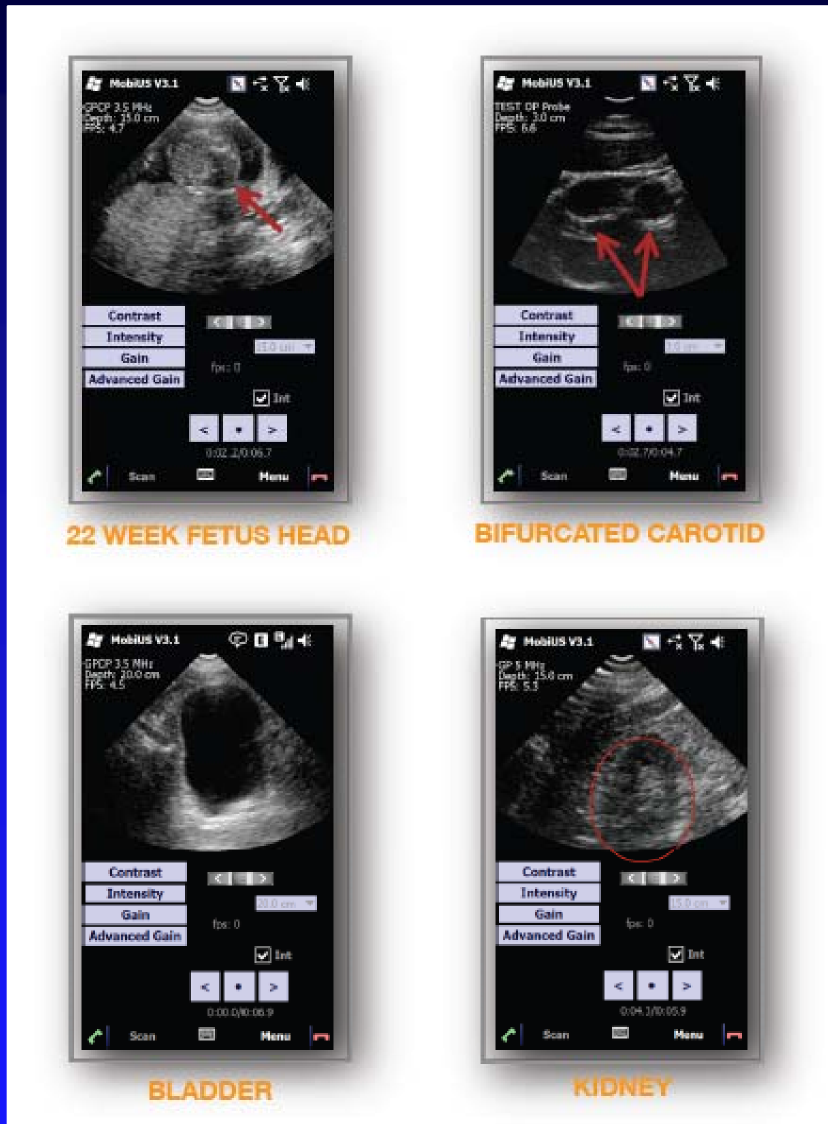
Ultrasound Instruments

MobiSante

AFFORDABLE DIAGNOSTICS
AT THE POINT OF CARE

Mobisante, Inc. is a privately-held mHealth company and we are building ***the world's first*** SmartPhone-based, Low-cost, Easy-to-use, Portable Ultrasound Imaging Systems. These systems are based on technology developed by a top university research lab and partially funded by Microsoft Research. The basic systems, which are expected to be extremely affordable, will make Ultrasound imaging accessible to primary care physicians, physician extenders, field workers in telemedicine settings, veterinarians and other cost-sensitive medical institutions in rural and emerging markets locally, regionally and around the globe. If you are interested in learning more, please contact us - we are interested in hearing from you.

http://www.mobisante.com



- MobiSante, Redmond, WA (2010)

Ultrasound Imaging

- Limitations
  - Operator dependent
    - Strong angle dependence
  - Complex spatial resolution parameters
  - No record of image locations (in general)
  - Artifacts

Ultrasound Imaging

Overview: Sound

- Sound is transmitted and received by a piezoelectric transducer
- Returned echoes vary depending on tissue characteristics
- Timing is used to determine echo depth

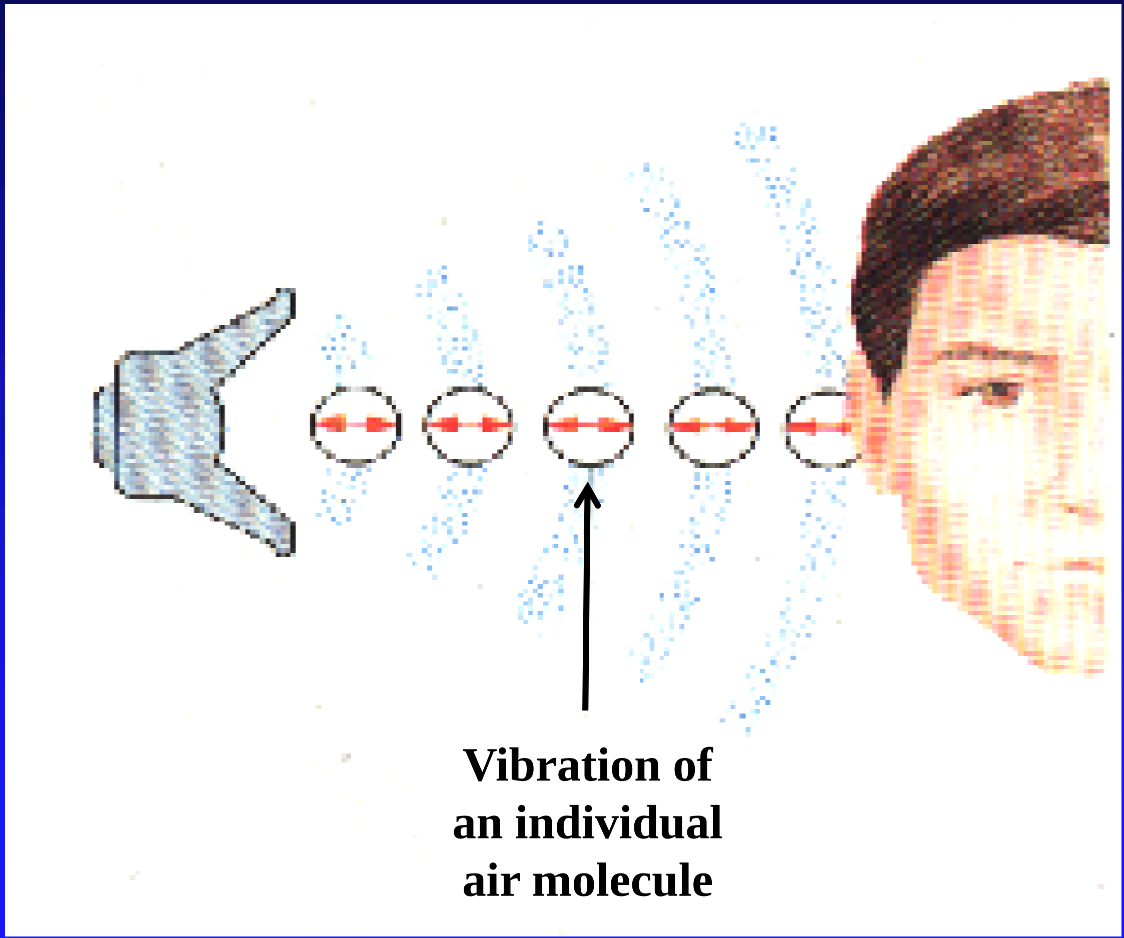
Sound

Vibration of particles in a medium

– OR –

A mechanical pressure wave

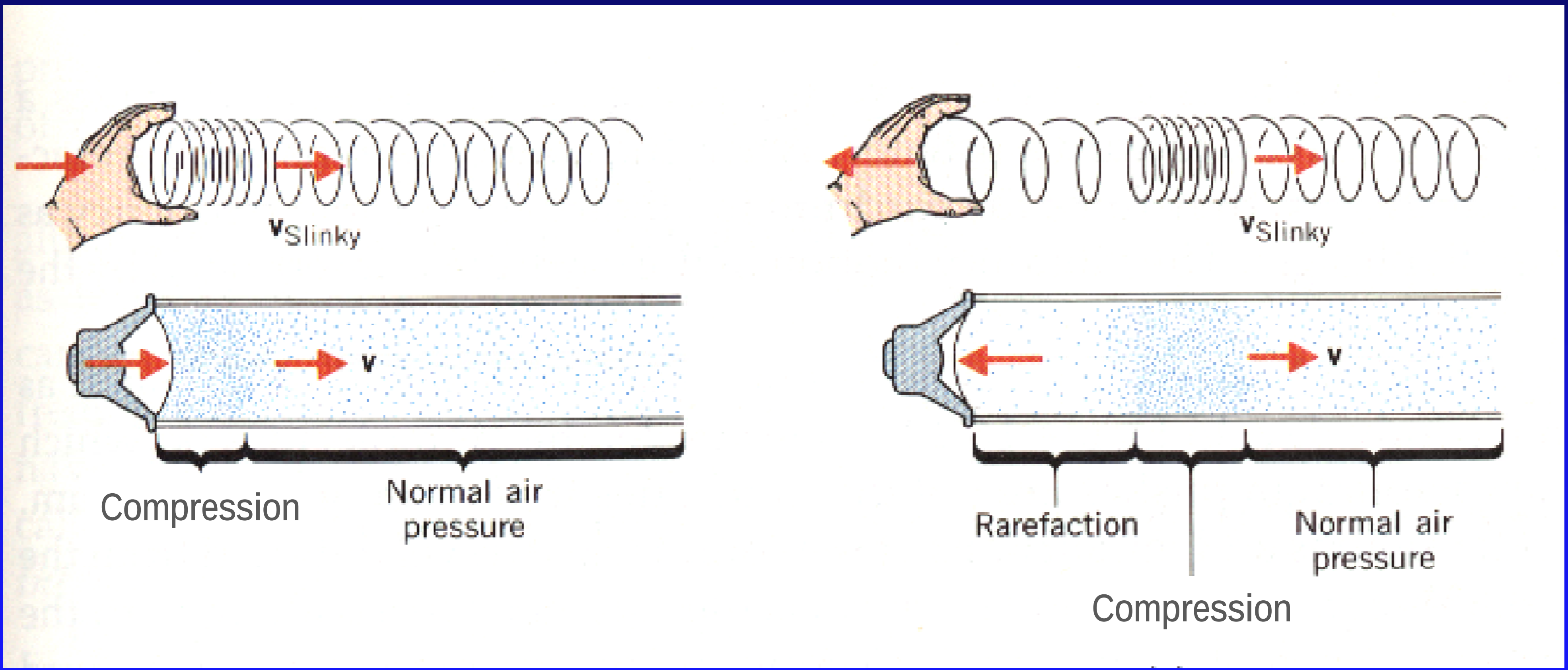
Sound is a longitudinal wave: particle motion is parallel to the direction of propagation



Physics, Cutnell & Johnson, 1992

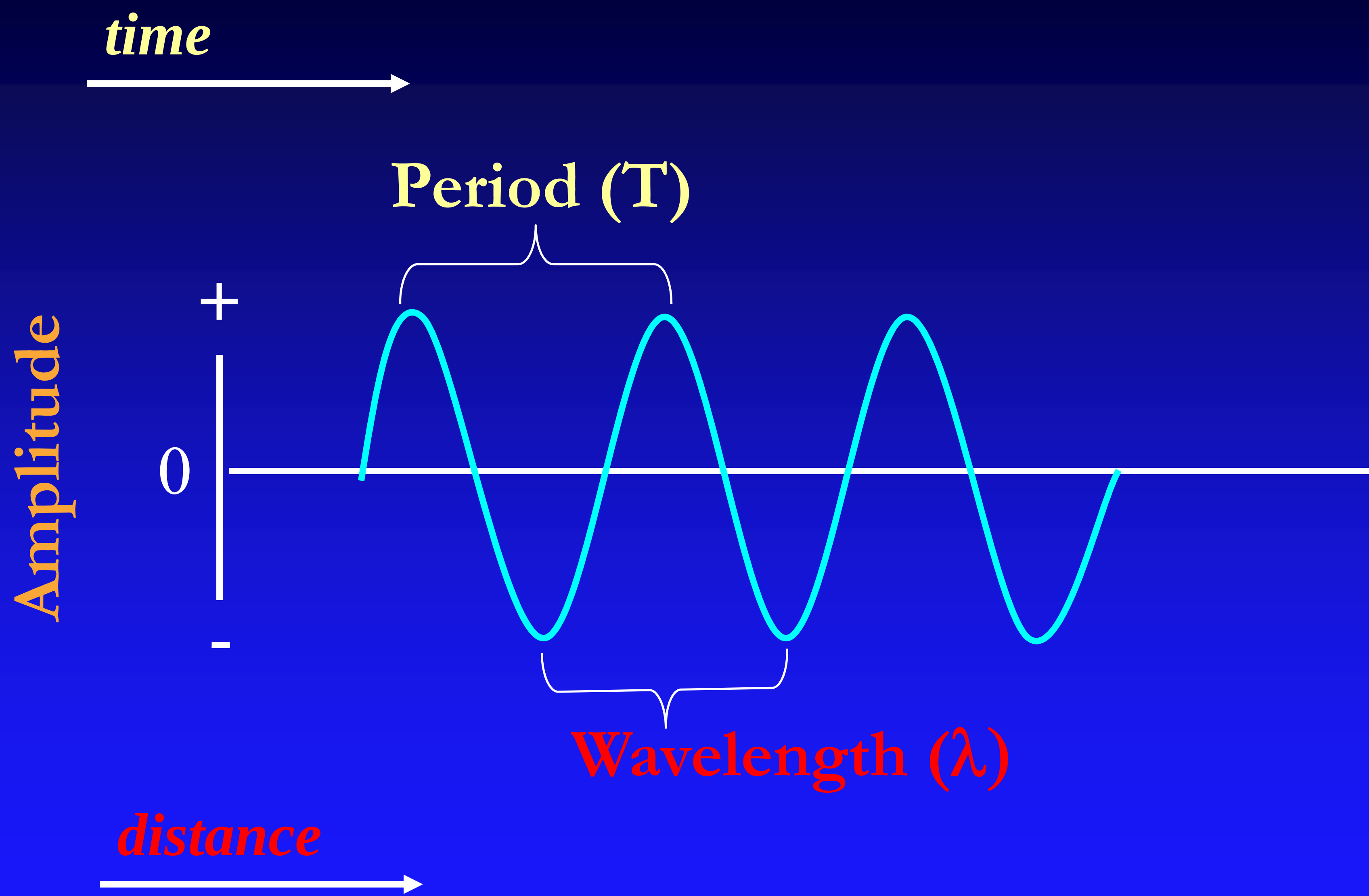
Sound

Compression and Rarefaction



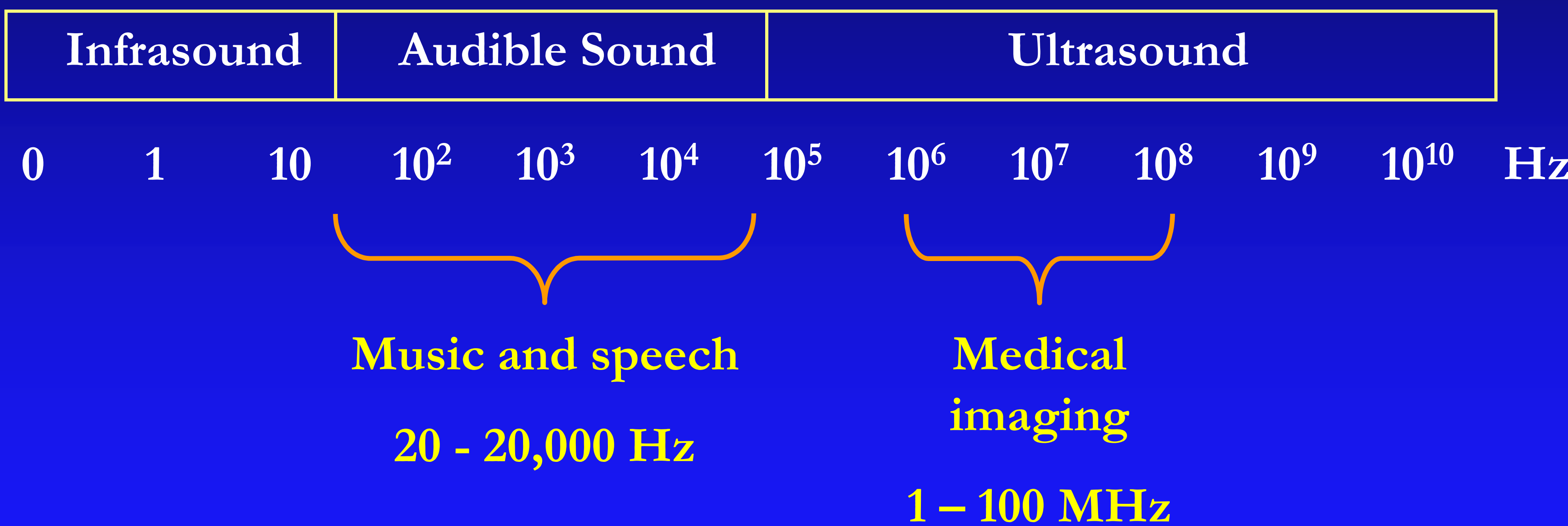
Physics, Cutnell & Johnson, 1992

Sound



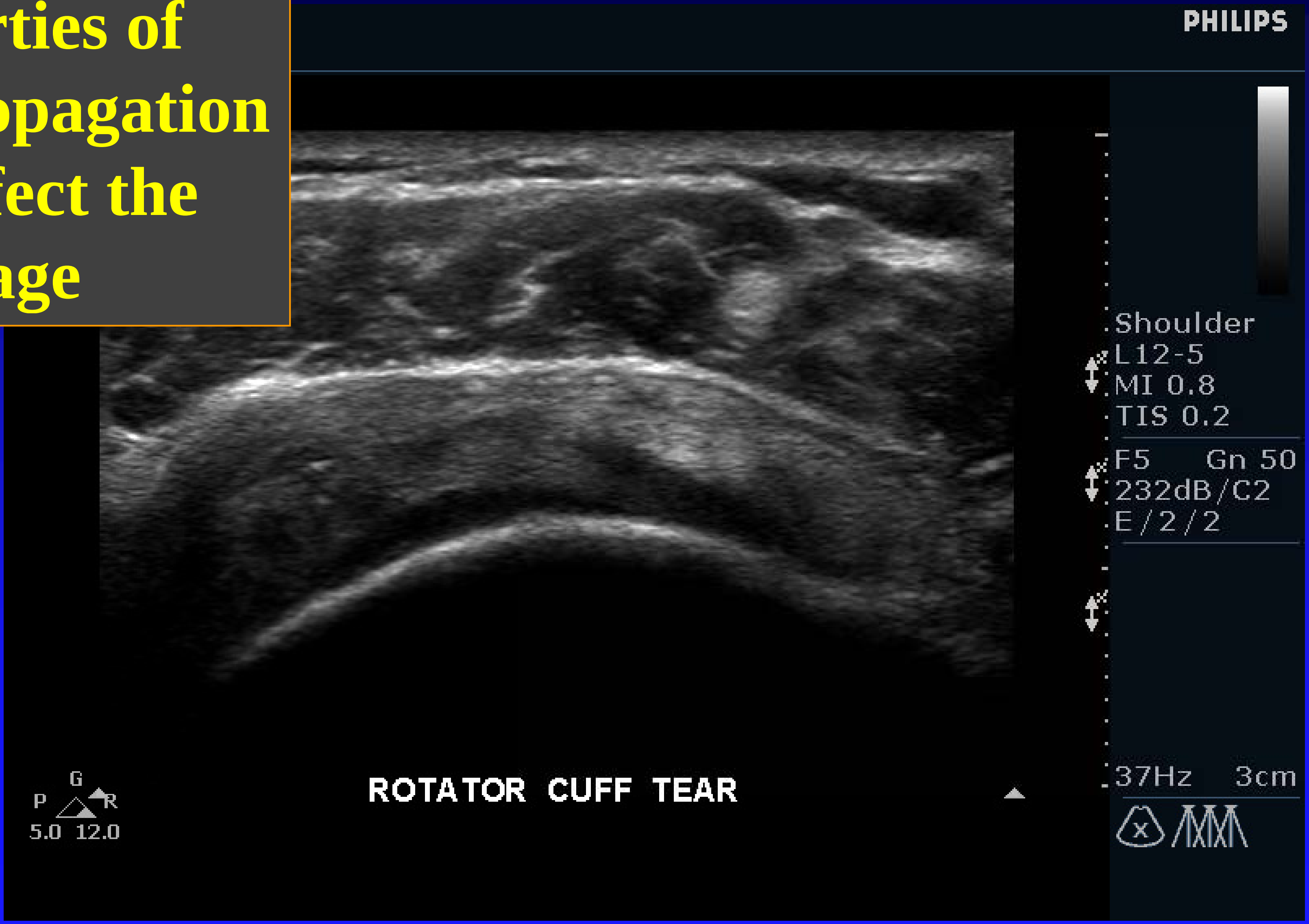
Frequency = cycles/sec

Spectrum of Sound



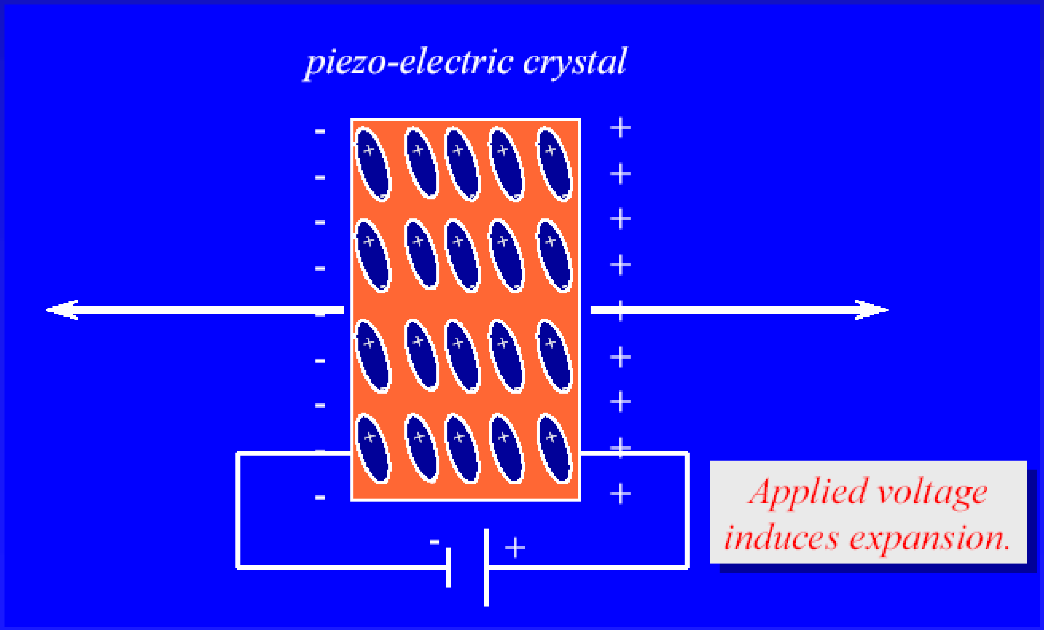
Ultrasound Imaging

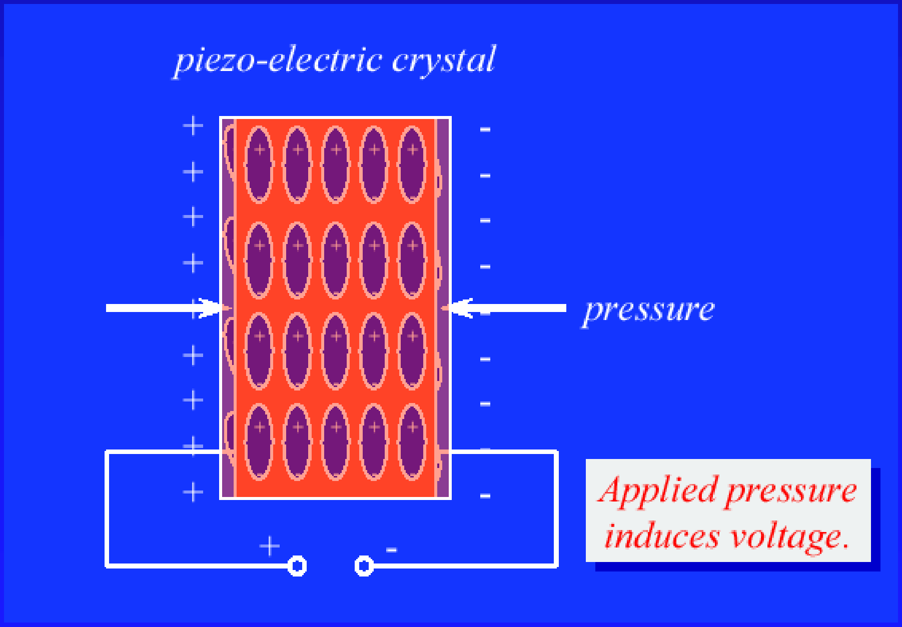
Properties of sound propagation that affect the image



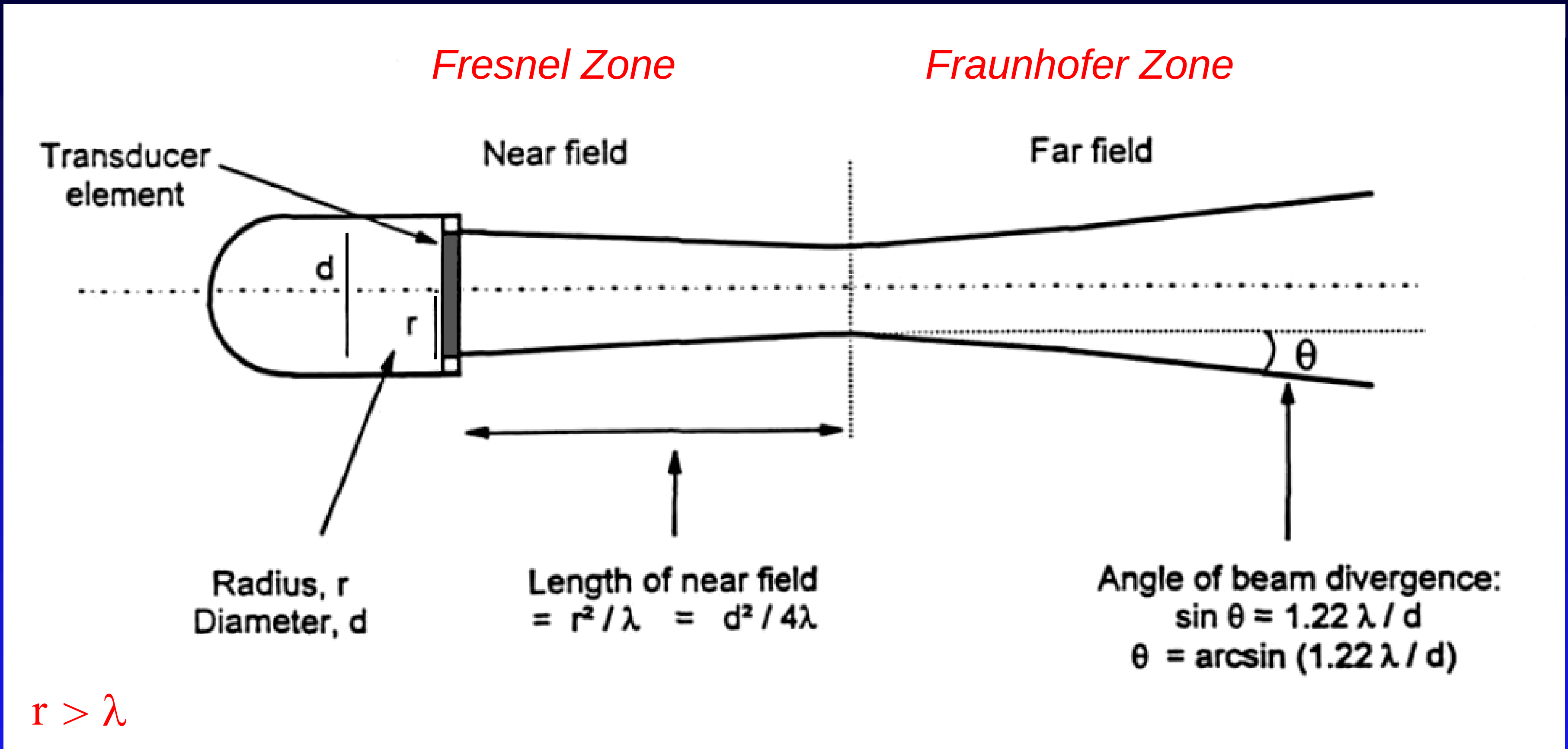
Piezoelectric Effect

- **Transducers constructed from piezoelectric materials**
  - **converts electrical energy to mechanical energy**
  - **converts mechanical energy to electrical energy**
  - **the material, shape and size of the transducer influence the frequency it can generate**



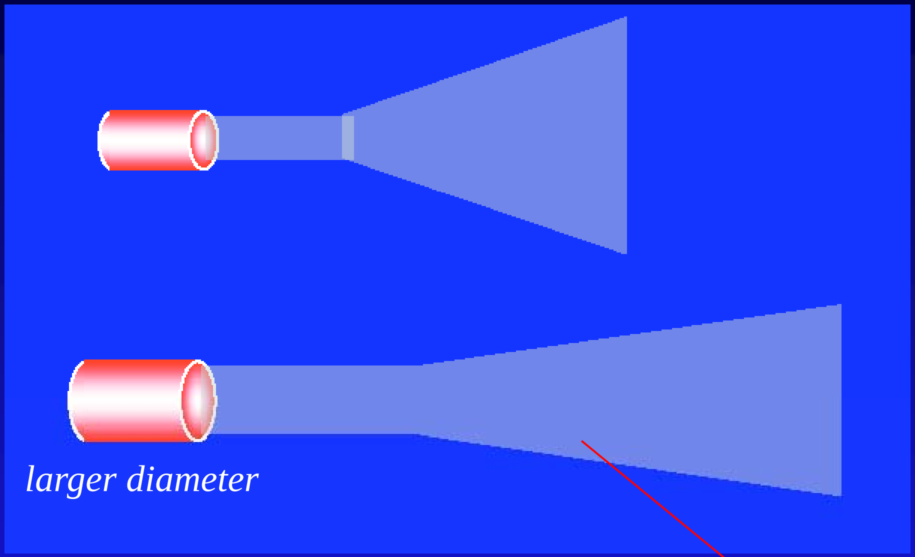


Beam Pattern

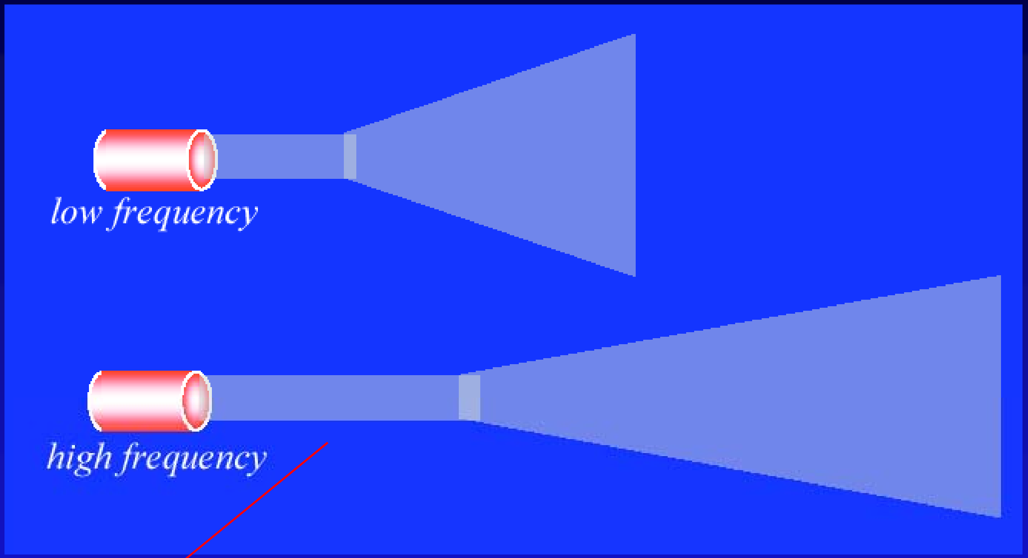


- Characteristic beam pattern for single-element circular transducer
  - Near field (Fresnel Zone): non-uniform intensity, non-divergent
  - Far field (Fraunhofer Zone): uniform intensity, divergent
  - Natural focus at transition between zones

Beam Pattern



Diameter



Frequency

- Deeper near field
- Less divergent far field

- Beam diameter determined by:
  - distance from transducer
  - transducer diameter
  - frequency

Properties of Sound

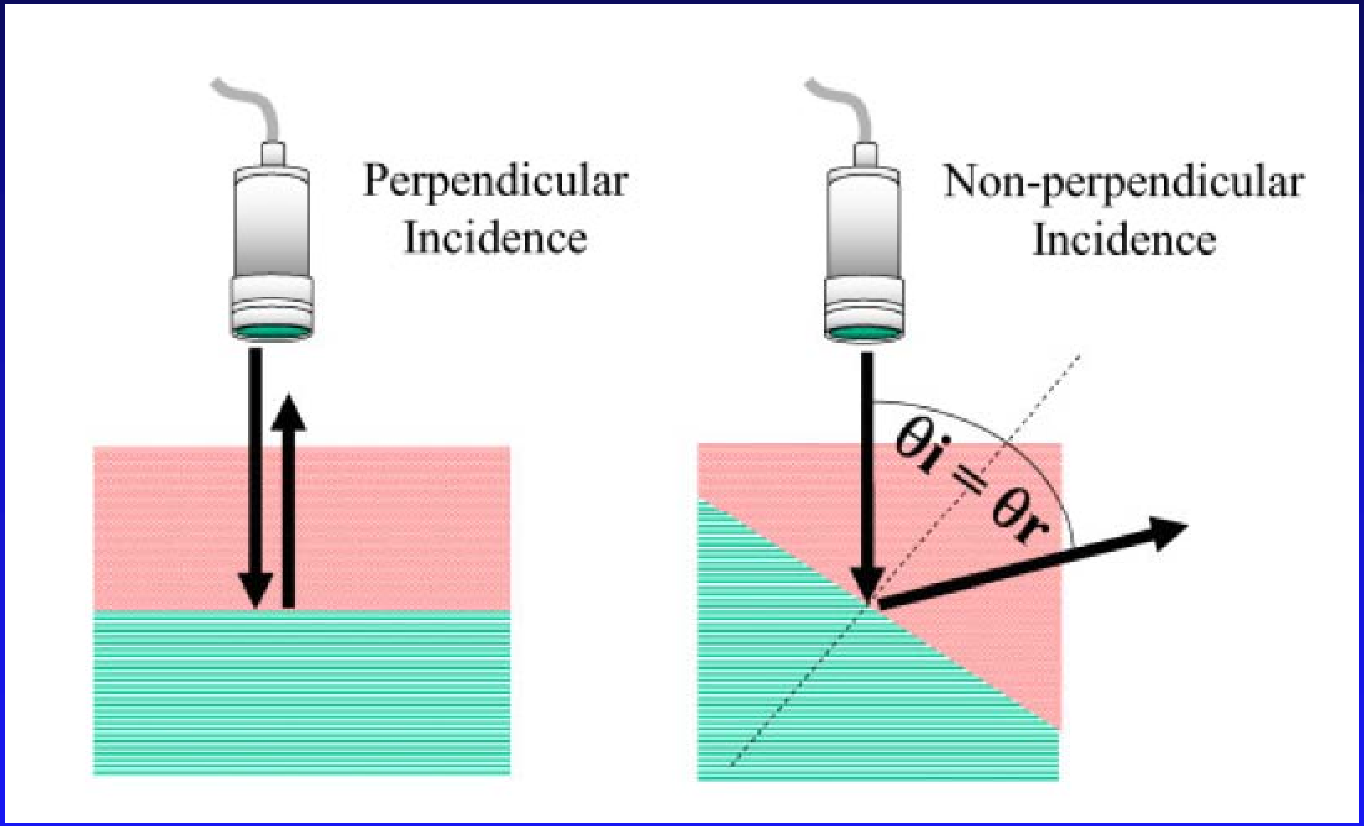
Reflection

- **Production of echoes at interfaces of tissues with different physical properties**
  - **sound waves that do not transmit across an interface are redirected back into the medium from which they originated**

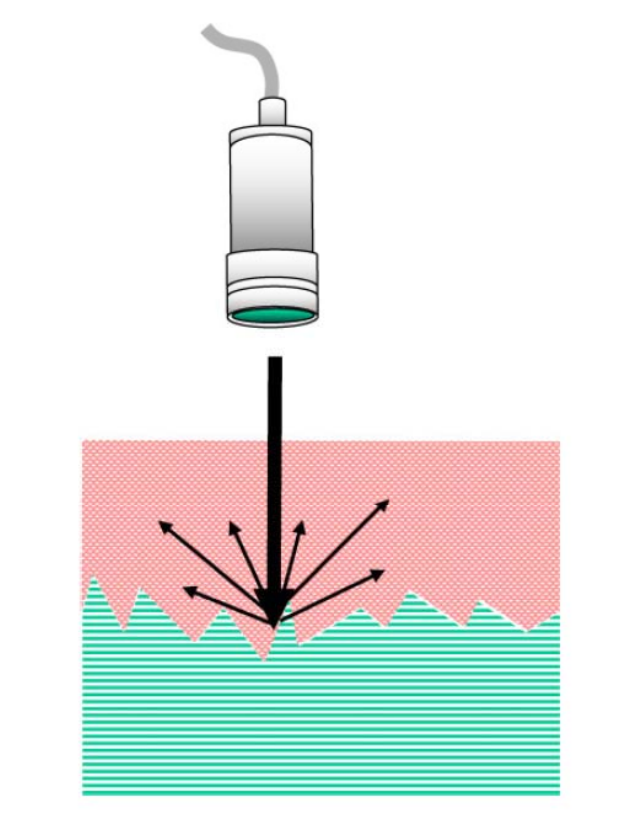
Types of Reflection

- **Specular**
  - **Structures larger than the wavelength**
    - Angle of reflection = Angle of incidence
- **Scattering**
  - **Structures smaller than the wavelength**
    - Multi-directional

Types of Reflection



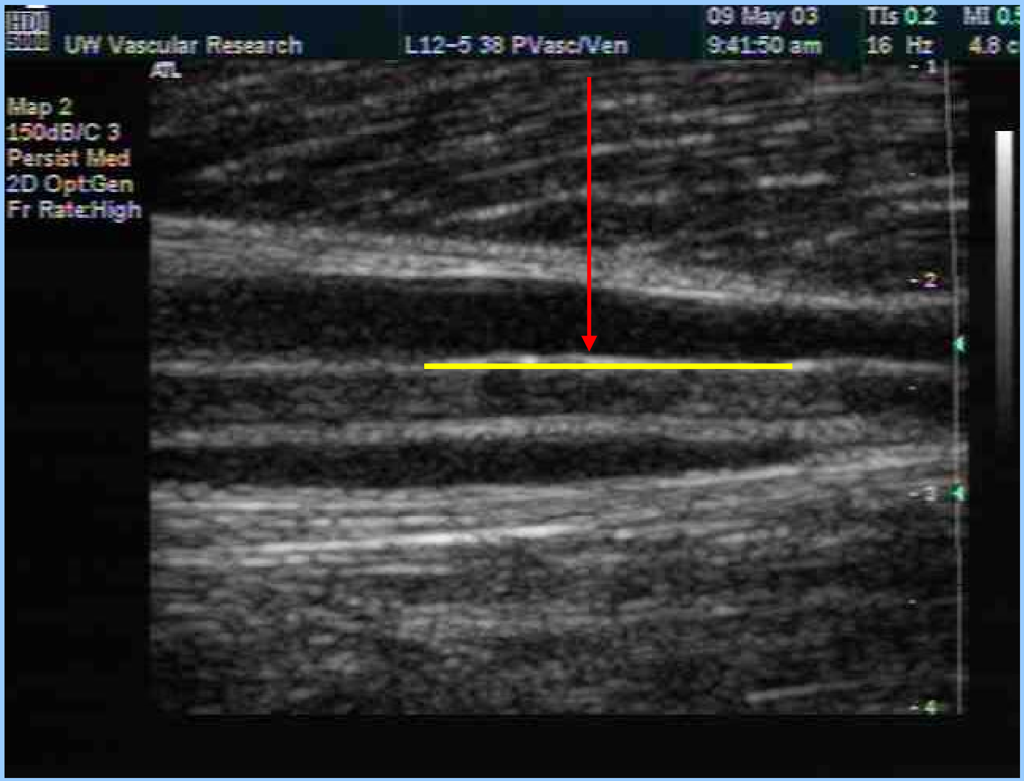
Specular reflection

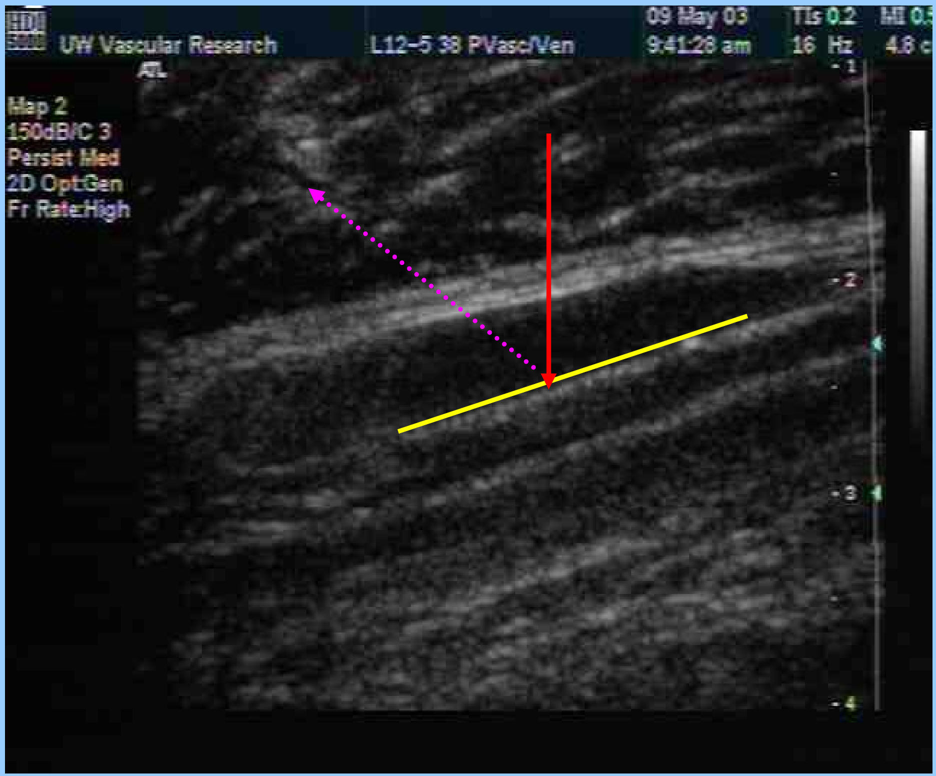


Scattering

Halliwell

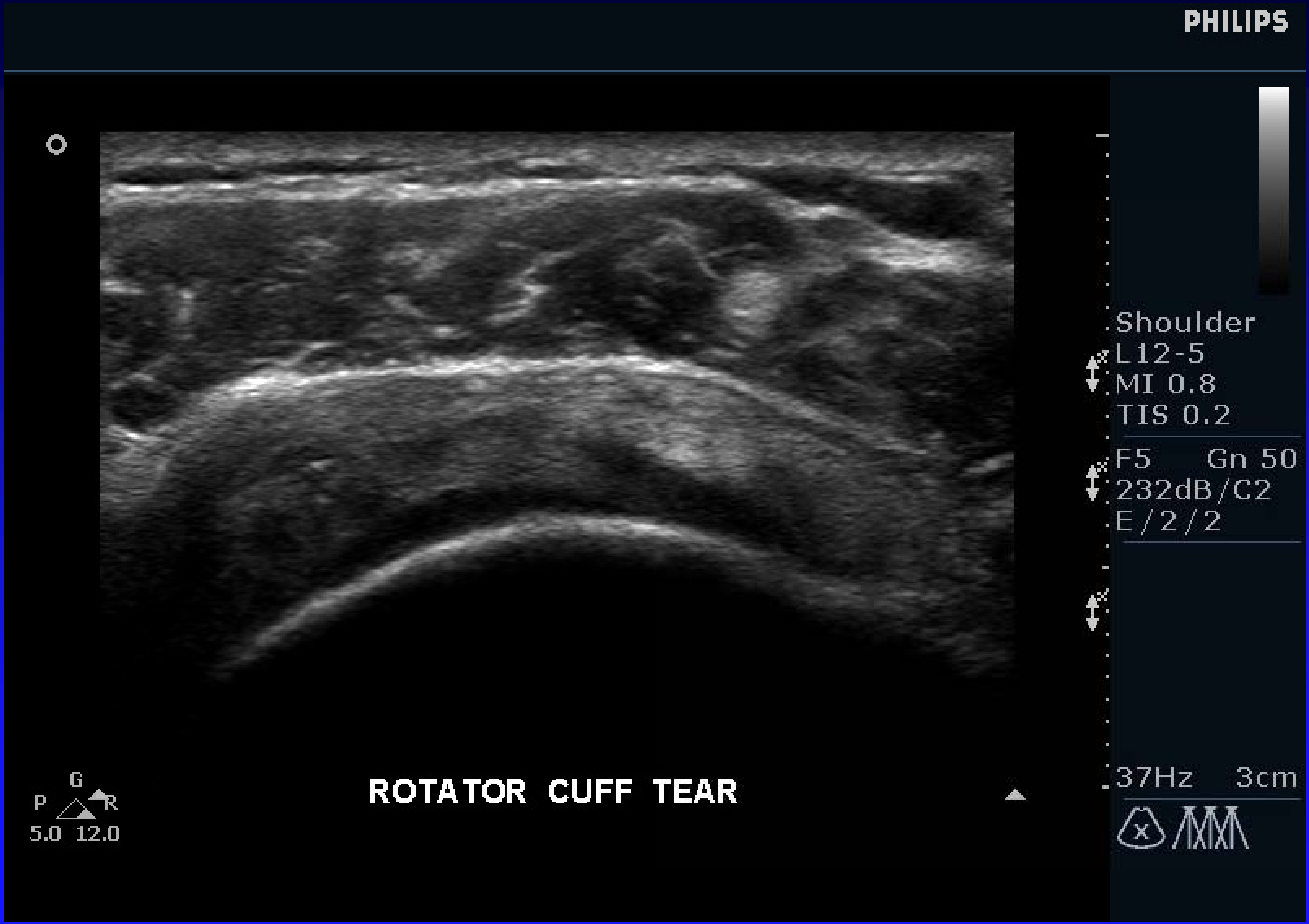
Reflection





Angle dependence

Reflection



Angle dependence

Amount of Reflection

- **Depends on the difference in Acoustic Impedance (Z) of the media**

  - how much sound is reflected is directly proportional to the impedance mismatch

$$Z = \rho\ c$$

ρ = density = mass/volume
c = propagation speed

Acoustic Impedance (Z)

Medium	Z (Rayls)
Air	0.0004×10^6
Fat	1.38×10^6
Water	1.48×10^6
Soft tissue	1.63×10^6
Muscle	1.70×10^6
Bone	7.80×10^6

Rayl: $kg/m^2/s$

Properties of Sound

Speed of Sound

How fast sound travels in a given medium depends on the structure of the medium

- **Density of particles**
- **Stiffness**
  - **Bulk Modulus: resistance to compression**

Speed of Sound (c)

Medium	m/sec
Air	331
Fat	1450
Water	1482
Soft tissue	1540
Blood	1570
Muscle	1585
Bone	4080
Steel	5960

Distance Equation

Convert time to distance to create accurate anatomic images

$$D = c\ t$$

c = sound propagation speed (m/s)

t = time (sec)

Distance Equation

Example 1

$$D = c\,t$$

c = 1540m/sec

How long does it take for sound to travel 1cm into the body?

0.01m = 1540m/sec x t

t = 0.0000065sec or 6.5μsec

Distance Equation

Example 2

$D = c\ t$

c = 1540m/sec

How long does it take for the transducer to receive a signal 4cm deep?

It takes 6.5µsec for 1cm of travel

6.5µsec x 4cm x 2 = 52µsec

Round-trip travel

Properties of Sound

<u>Attenuation</u>

The decrease in the intensity of sound as it travels through a medium (loss of energy)

- Absorption
- Reflection

- **Proportional to <u>distance</u> and <u>frequency</u>**
  - Longer path: increased attenuation
  - Higher frequency: increased attenuation

Attenuation Coefficients

Tissue	Attenuation Coefficient (dB/cm/MHz)
Liver	0.5
Fat	0.6
Brain	0.6
Kidney	0.9
Muscle	1.0
Heart	1.1

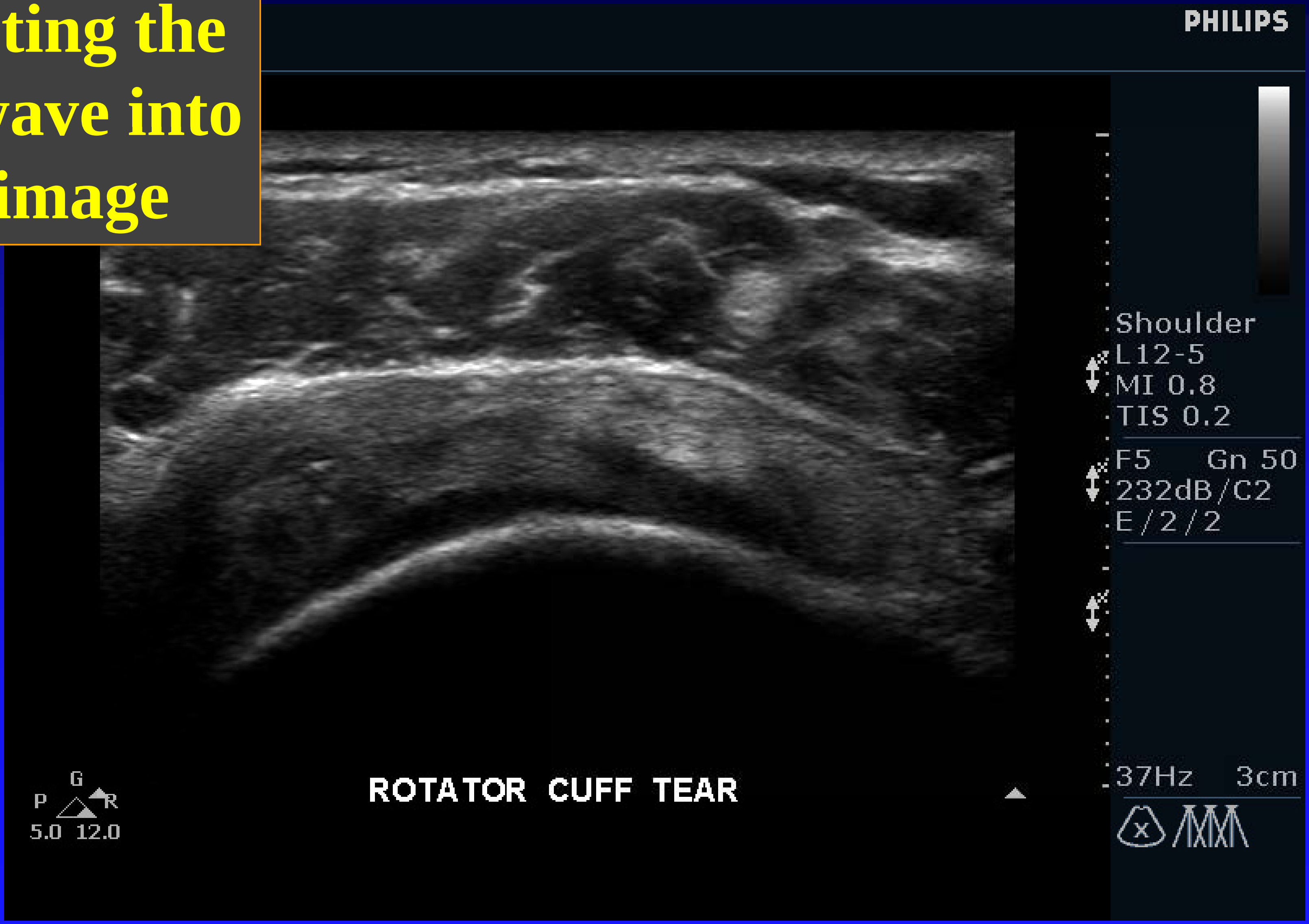
0.8 dB/cm/MHz average value for soft tissue

Attenuation

- **To calculate attenuation, multiply the Attenuation Coefficient by round trip distance and frequency:**
  - **3.5 MHz sound, 4 cm sound travel**
    - **attenuation = 0.5 dB/cm/MHz x 4 cm x 3.5 MHz = 7 dB**
  - **5 MHz sound, 10 cm sound travel**
    - **attenuation = 0.5 dB/cm/MHz x 10 cm x 5 MHz = 25 dB**

Ultrasound Imaging

Converting the sound wave into a 2D image



Ultrasound Imaging

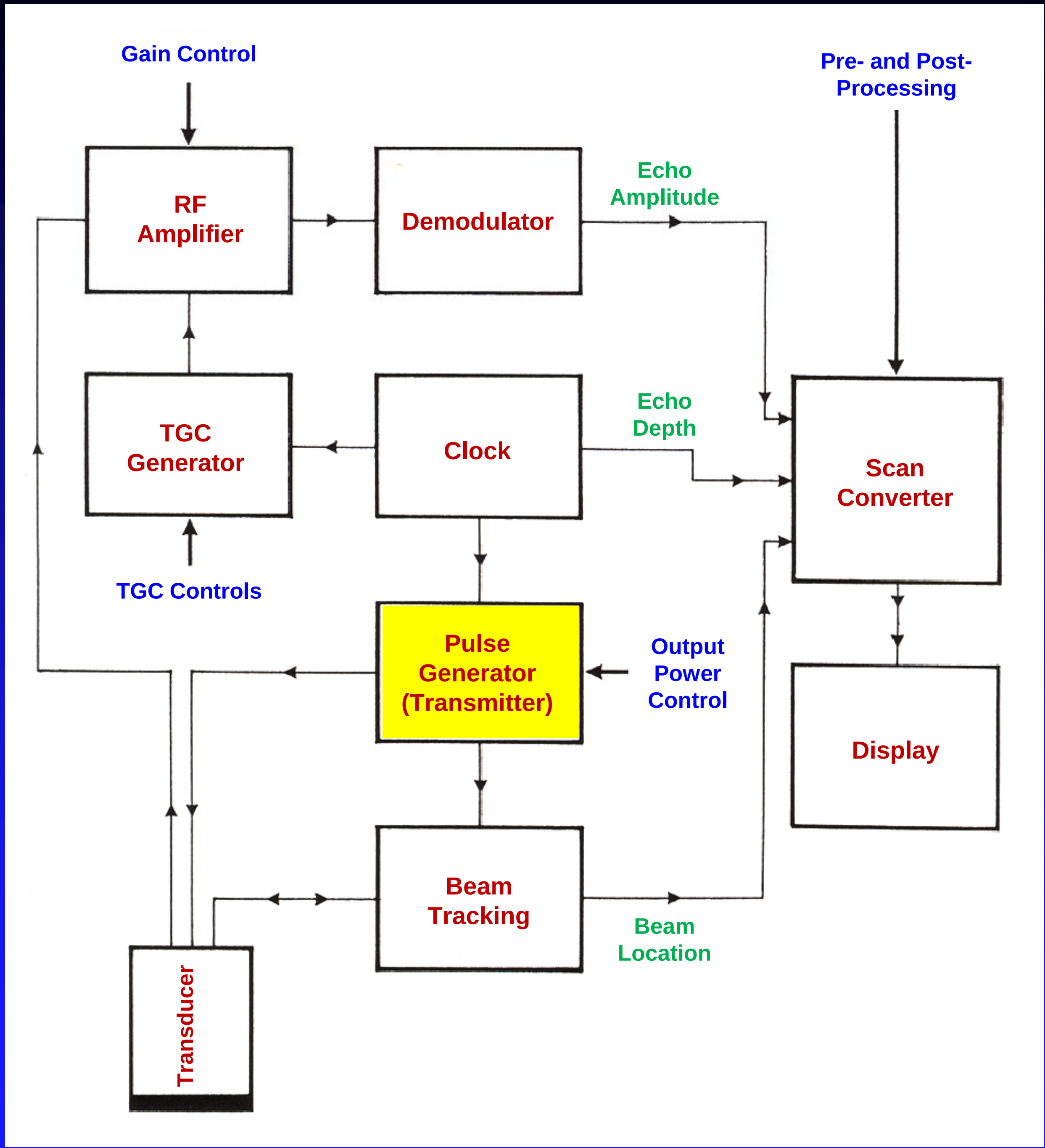
Overview: Instrument

- Transmit short pulse into tissue
- Receive echoes
- Perform amplitude demodulation
- Maintain time record time for depth calculation
- Repeat to create 2D map of echo amplitude

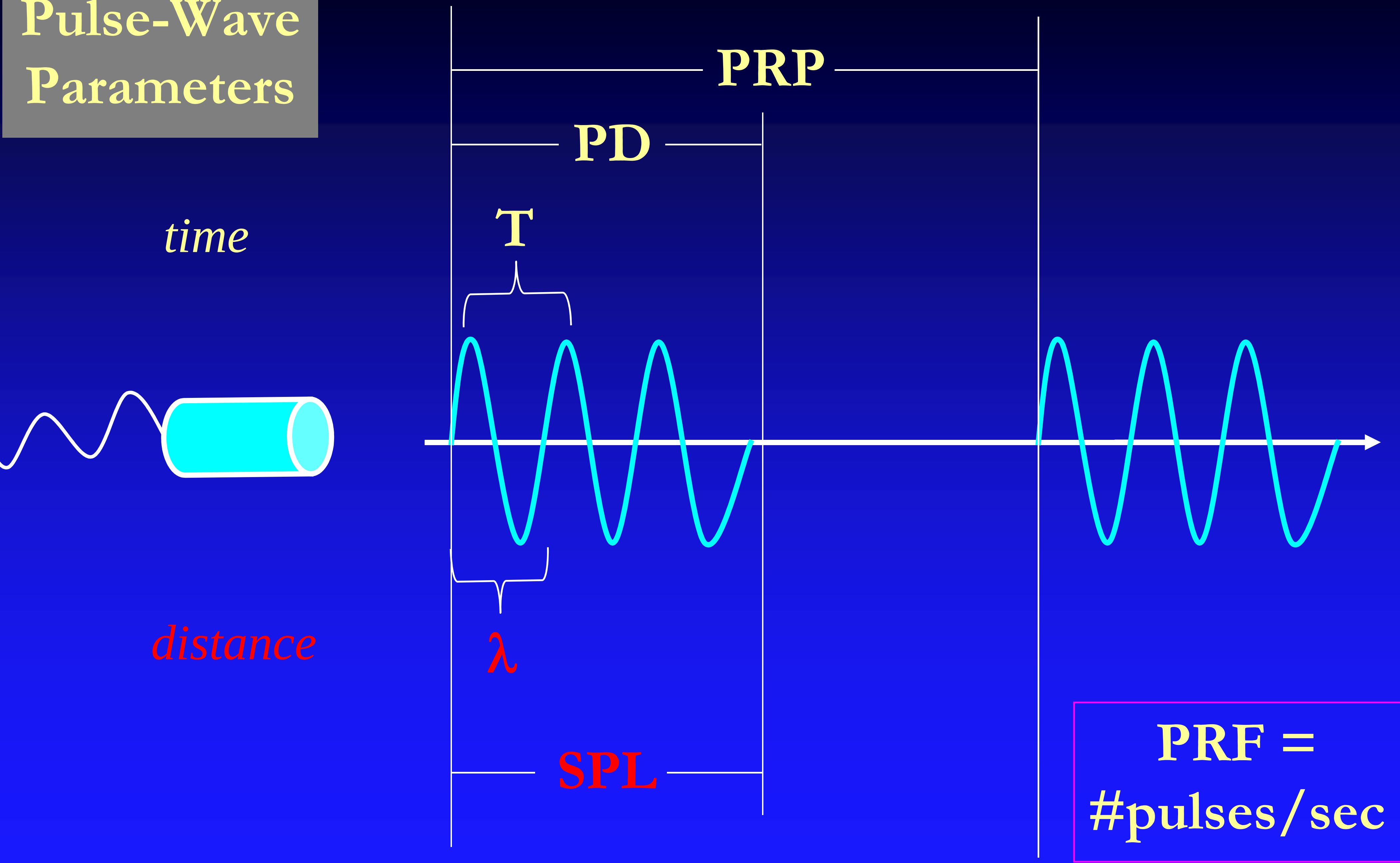
Components of an Ultrasound Scanner

2D gray scale imaging

• Transmitter



Pulse-Wave Parameters



Pulse-Wave Parameters

- PRF = *Pulse Repetition Frequency*
- PRP = *Pulse Repetition Period*
  = 1/PRF
  = time until pulse repeats itself
- PD = *Pulse Duration* = time pulse lasts
- DF = *Duty Factor* $= \frac{\text{PD}}{\text{PRP}} = \frac{\text{active time}}{\text{total time}}$
- SPL = *Spatial Pulse Length*
  = length of pulse in space
  = (#cycles/pulse) x (λ)

Related to temporal resolution

Related to spatial resolution

Typical pulse times

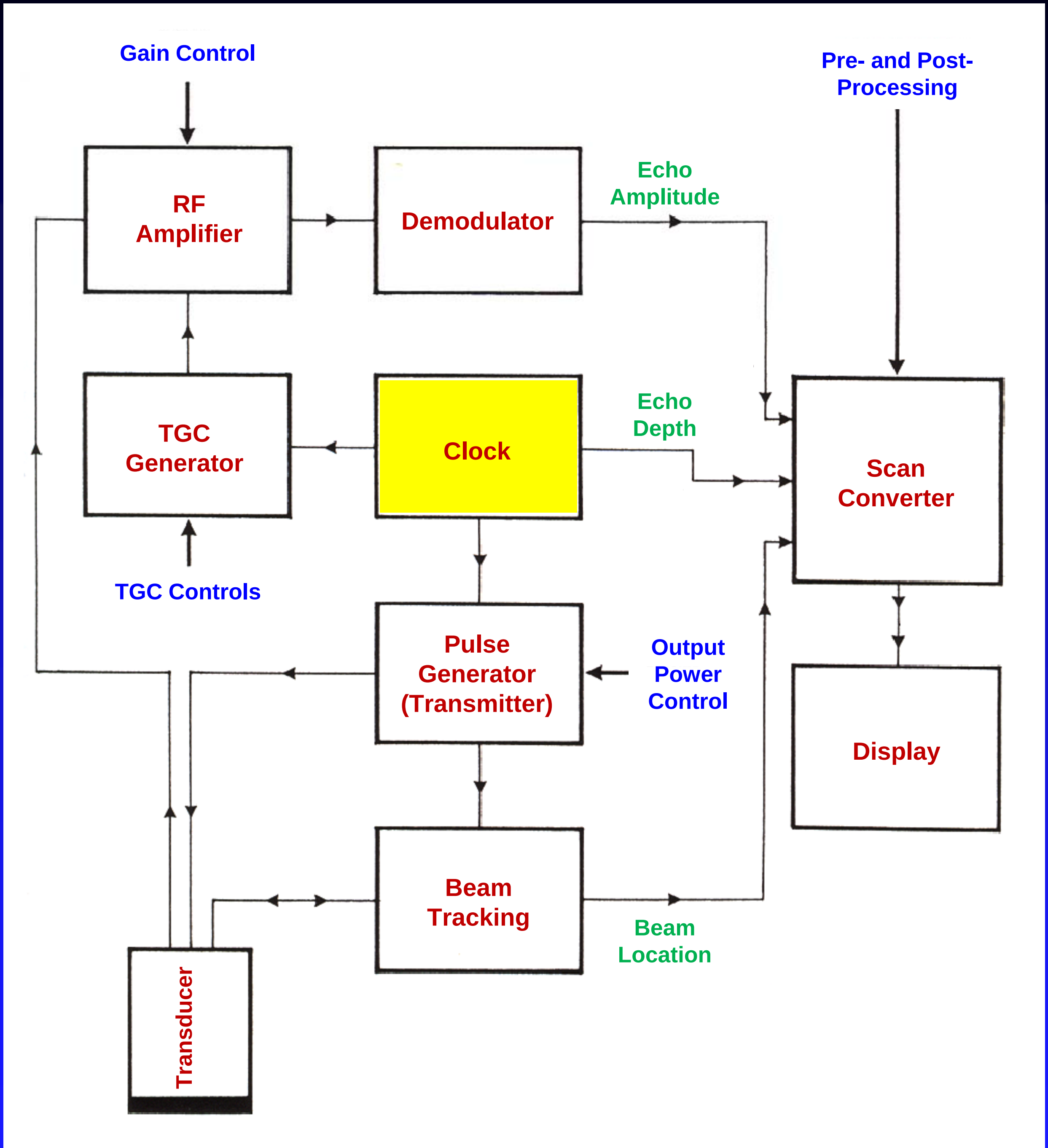
For 3-cycle 3 MHz pulse: PD = 1 microsecond

For 10 cm depth: PRP = 130 microsecond

Components of an Ultrasound Scanner

2D gray scale imaging

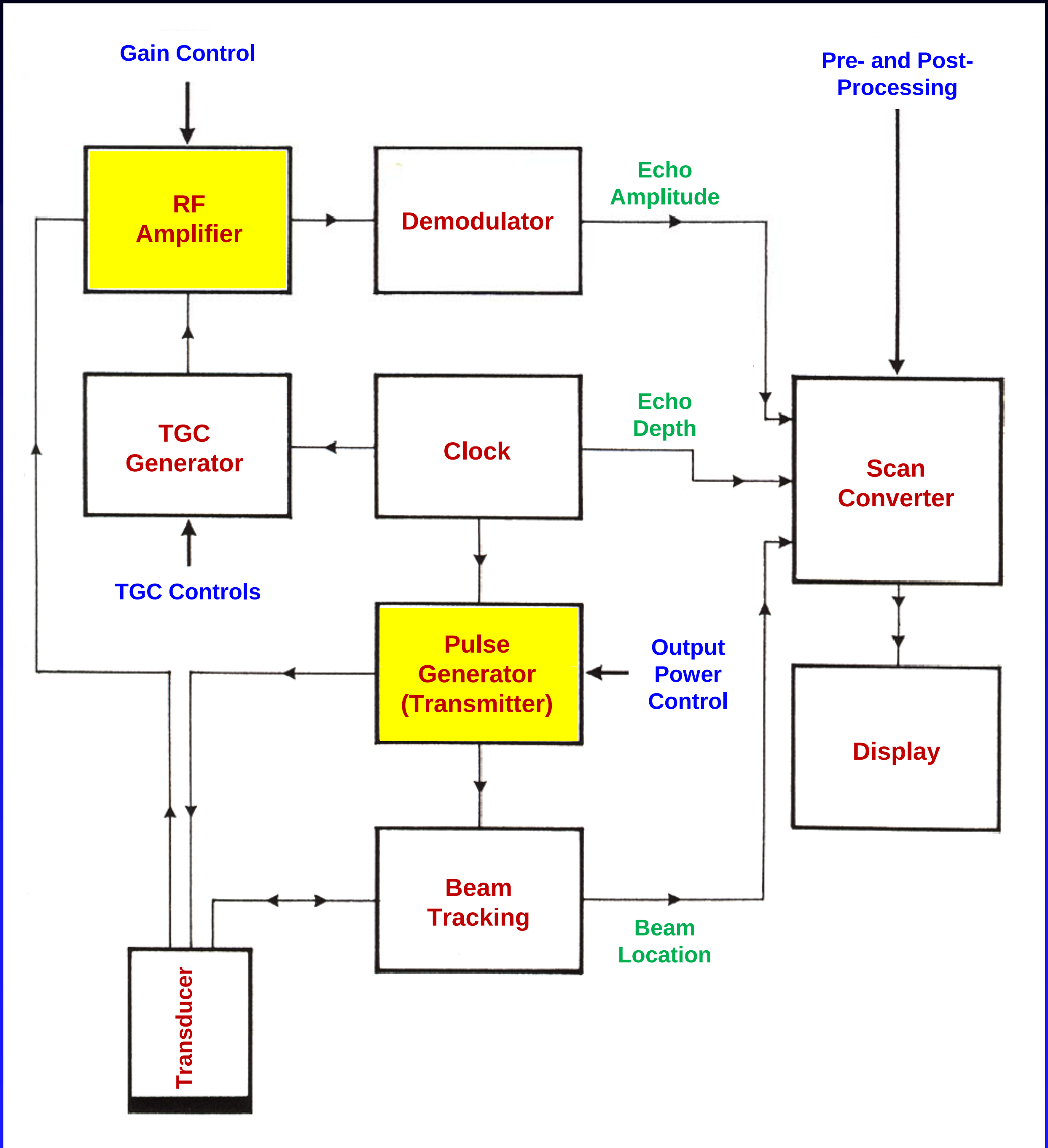
• Master clock for pulse and echo timing



Components of an Ultrasound Scanner

2D gray scale imaging

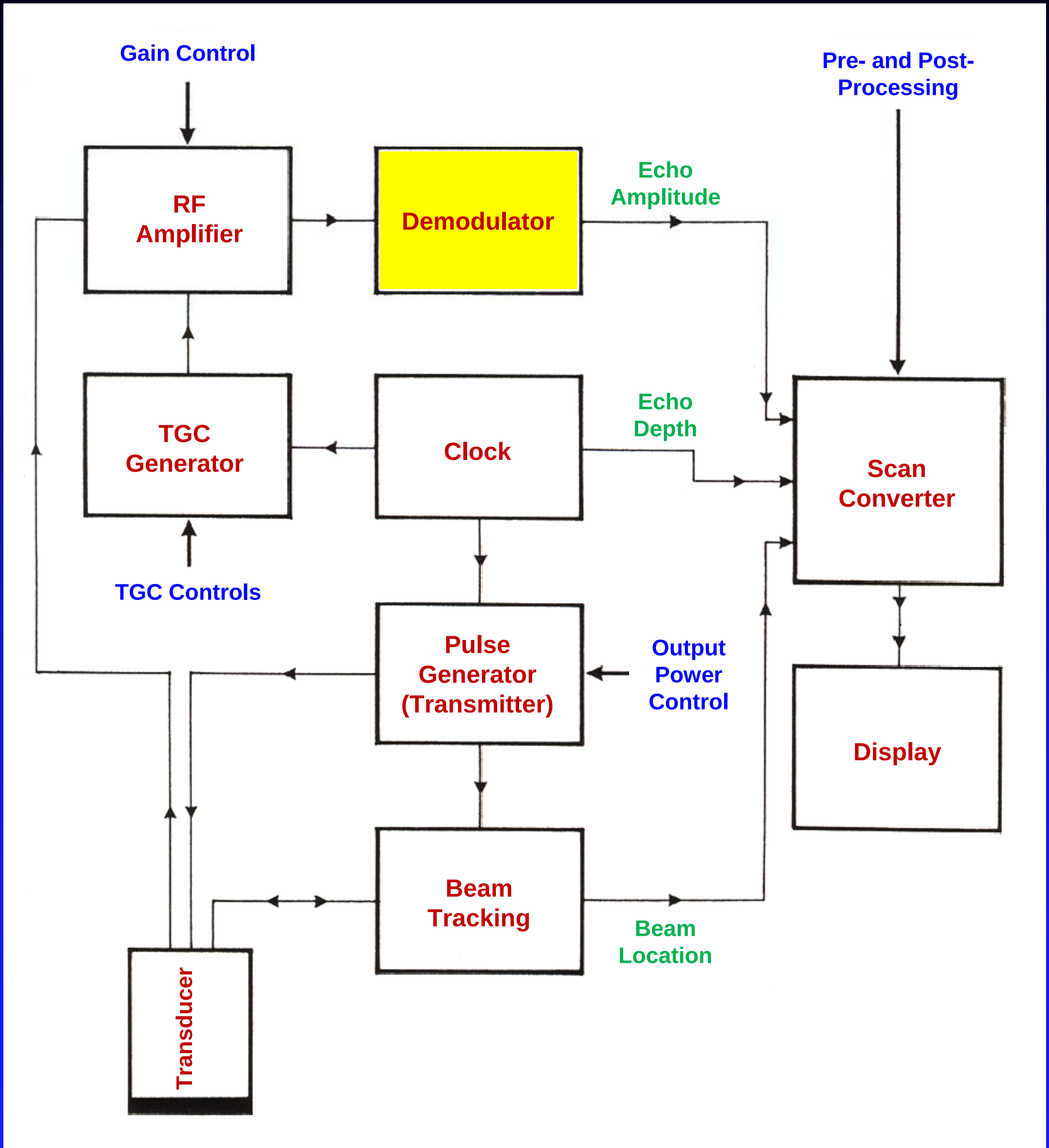
- Short pulse transmitted
- Echoes received and amplified



Components of an Ultrasound Scanner

2D gray scale imaging

• Echo 'detection'

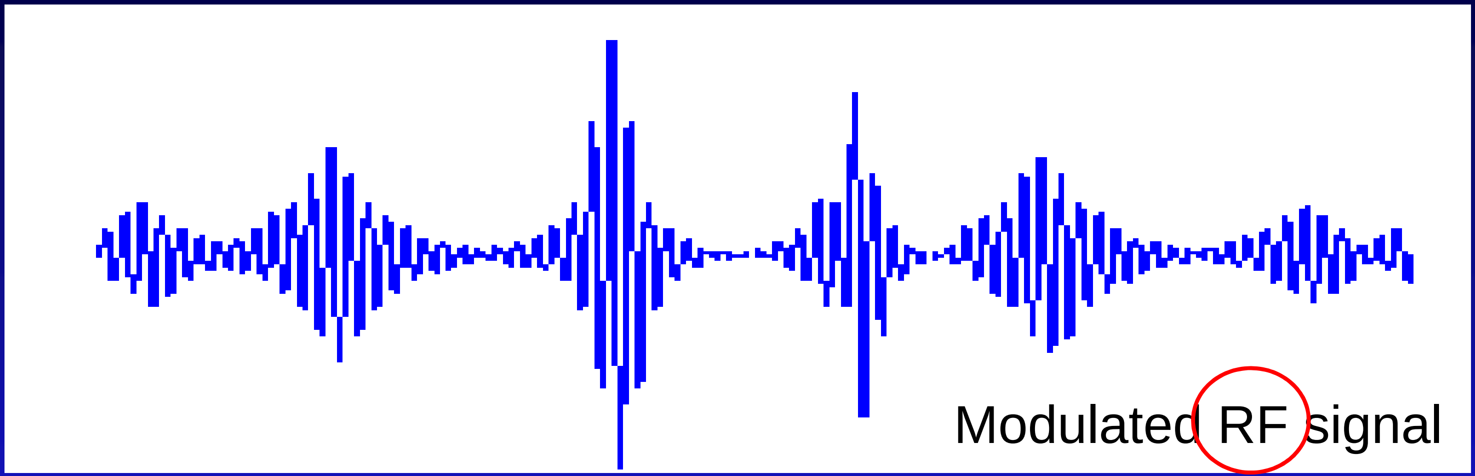


Signal Processing

- **Amplitude modulation**
  - Changes in amplitude of received energy provide information about tissue characteristics
    - Low-frequency amplitude variations superimposed on high-frequency signal
  - ‘Detection’ or ‘Demodulation’ extracts the information from the returned echoes
- **Frequency modulation**
  - Changes in frequency of received energy provide information about tissue motion
  - Doppler ultrasound

Signal Processing

A short pulse is transmitted and a series of echoes is received



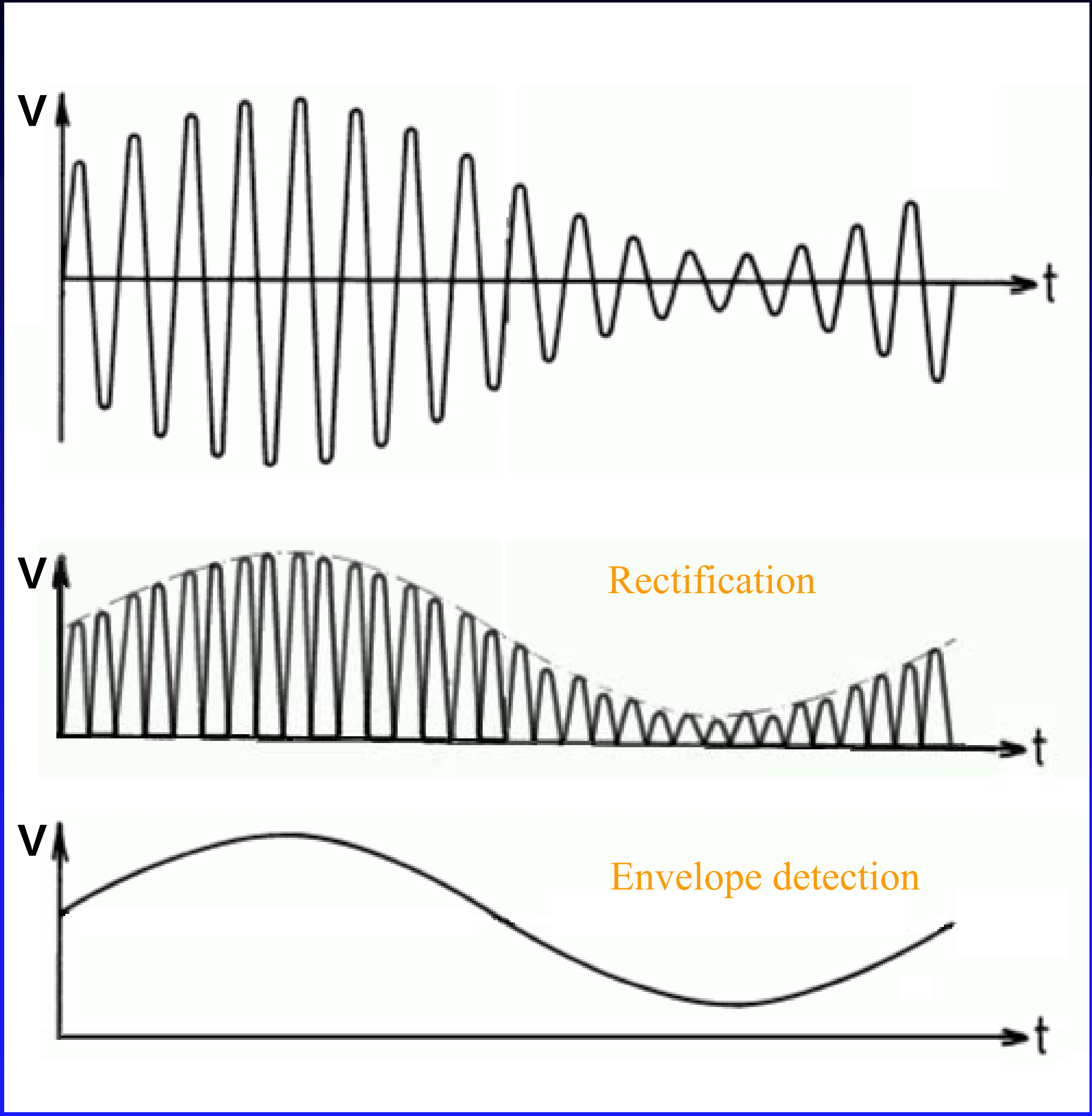
The ultrasound echo is referred to as a 'Radio Frequency' signal because it is in the frequency range of radio waves in the electromagnetic spectrum.

Electromagnetic Spectrum

Region	Frequency
Radio	10^6 - 10^9
Microwave	3×10^{12}
Infrared	4×10^{14}
Visible	7×10^{14}
Ultraviolet	3×10^{17}
X-Ray	3×10^{19}
Gamma Ray	$> 3 \times 10^{19}$

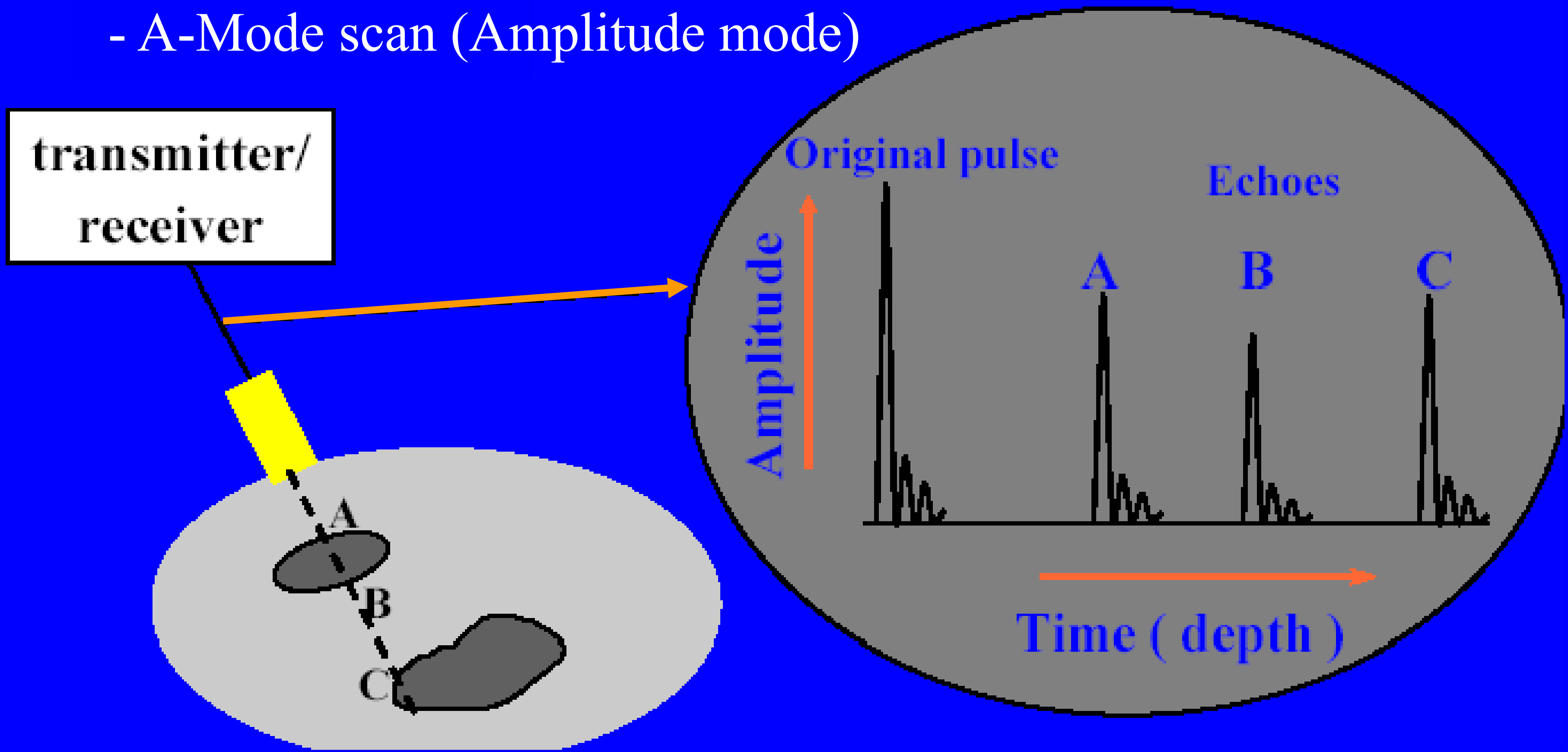
Medical ultrasound frequencies ≈ 2-20 MHz

- **Amplitude Demodulation**
  - **Separation of carrier and modulating waveform**
    - Rectification
      - Absolute value
    - Envelope detection
      - Low-pass filter



A-Mode

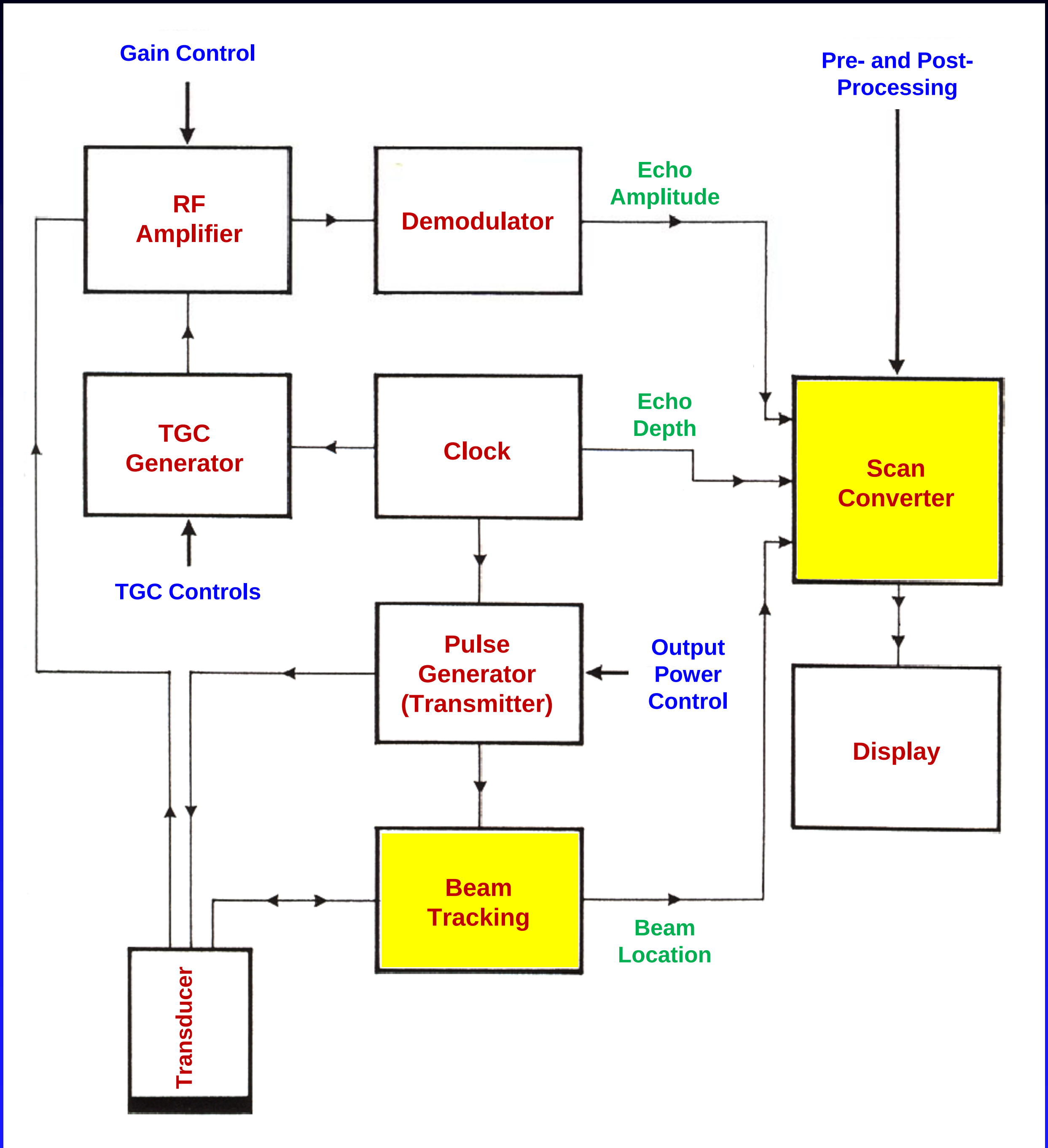
- Short pulses are transmitted and the echo times are measured
  - A-Mode scan (Amplitude mode)



Components of an Ultrasound Scanner

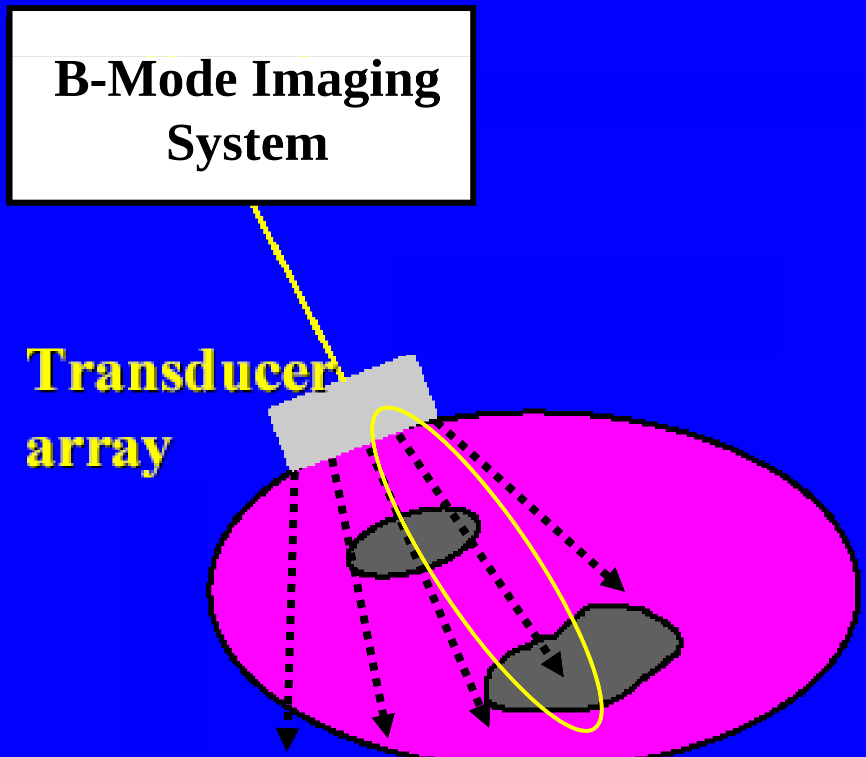
2D gray scale imaging

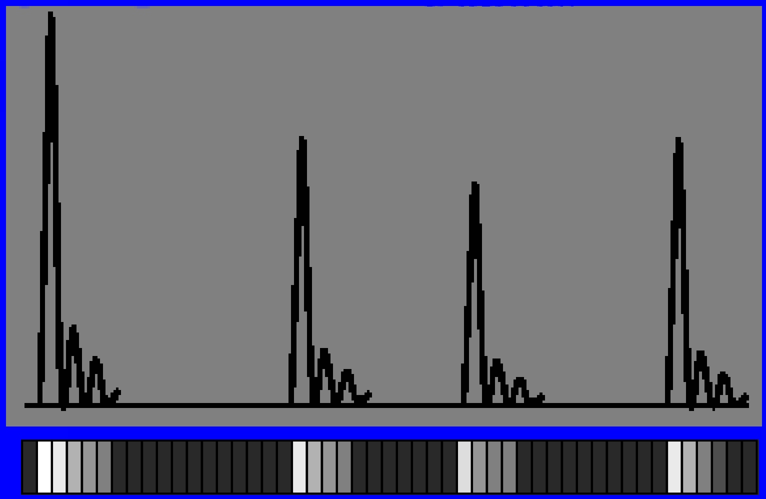
- Beam location saved in scan converter memory
- Echo amplitude mapped to gray scale
- 2D image displayed on screen



B-Mode

- Pulses are transmitted in multiple directions and the echoes are mapped to brightness on a 2D display
  - B-Mode scan (Brightness mode)





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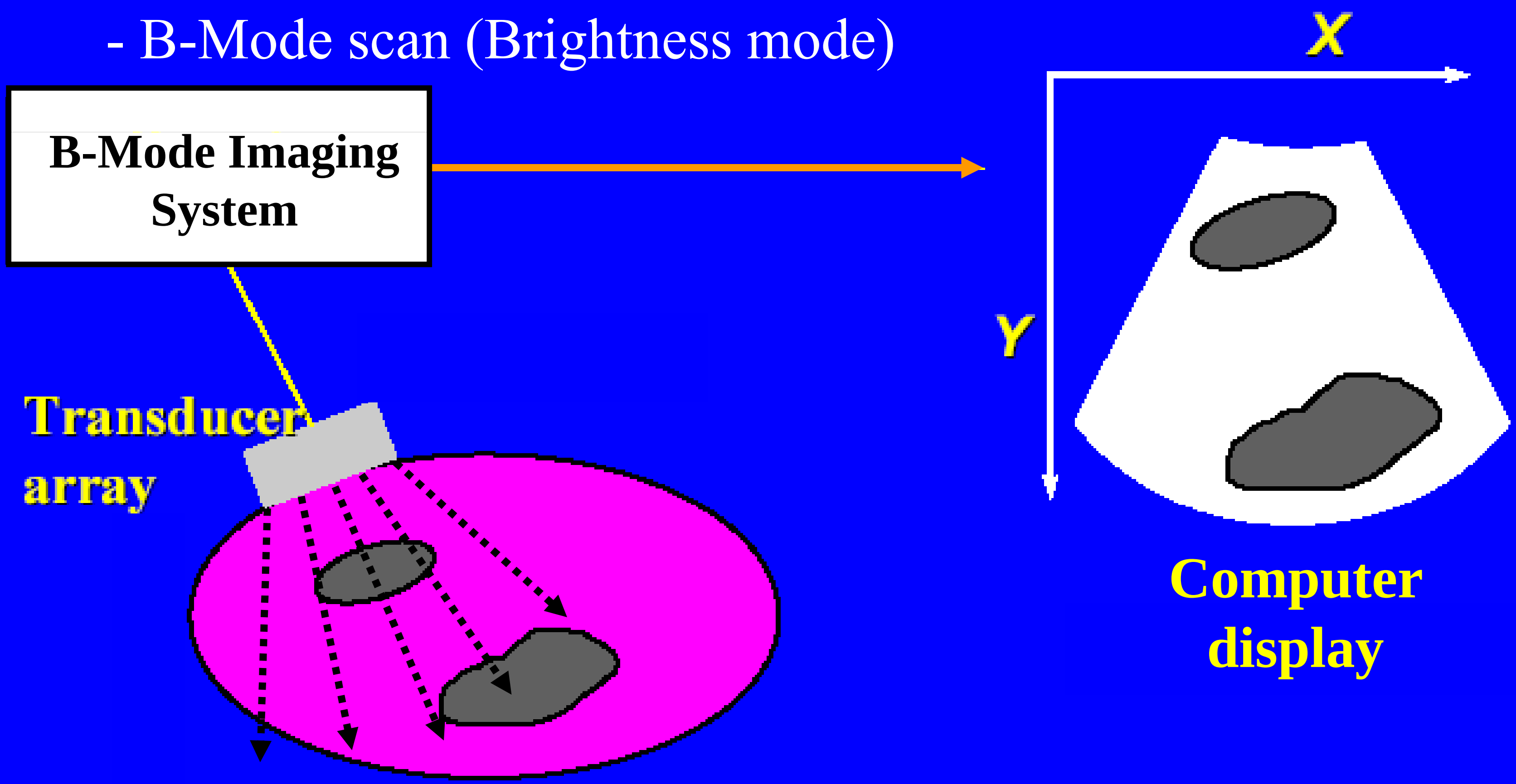
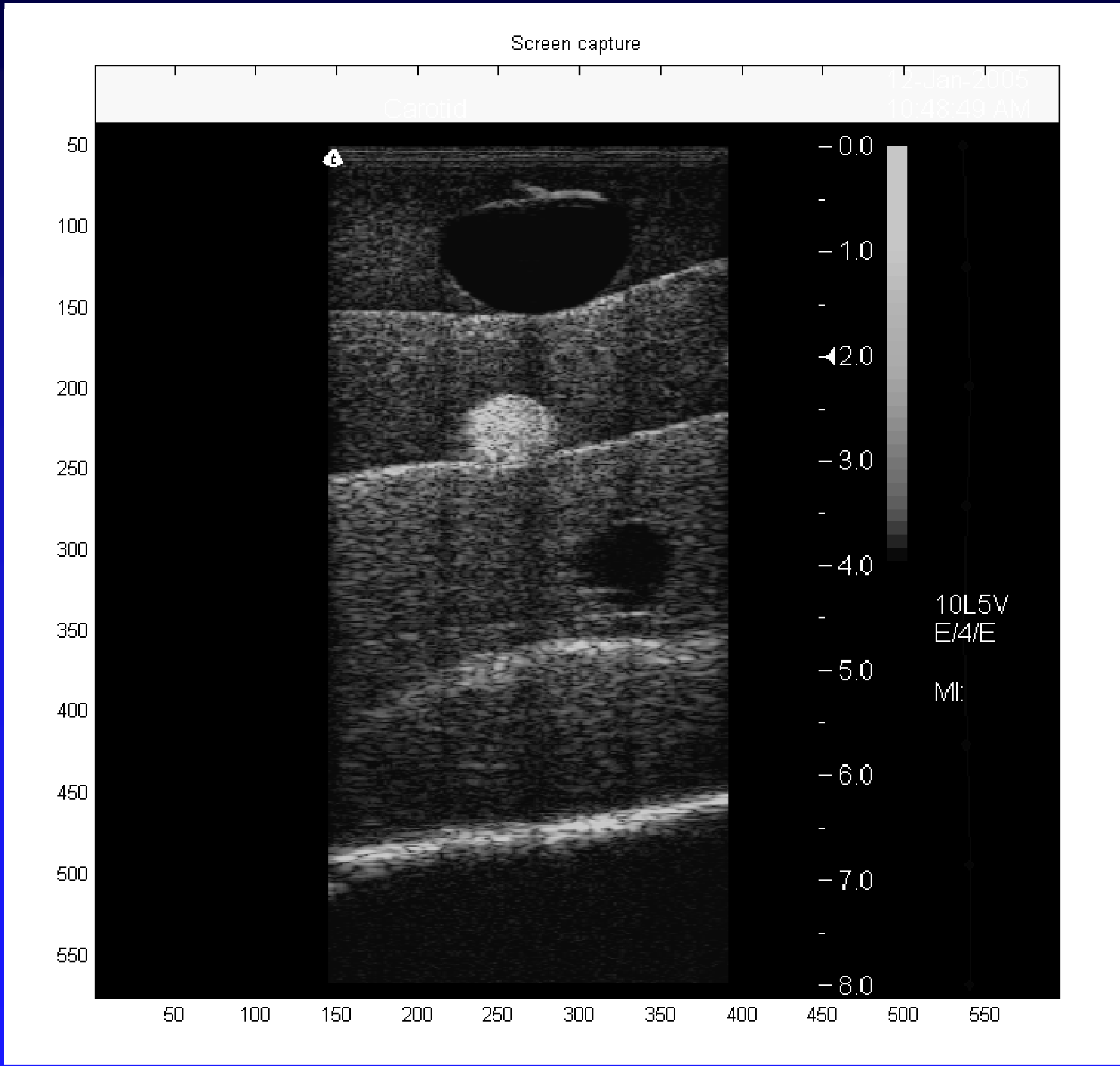


Processing Example: RF to B-Mode

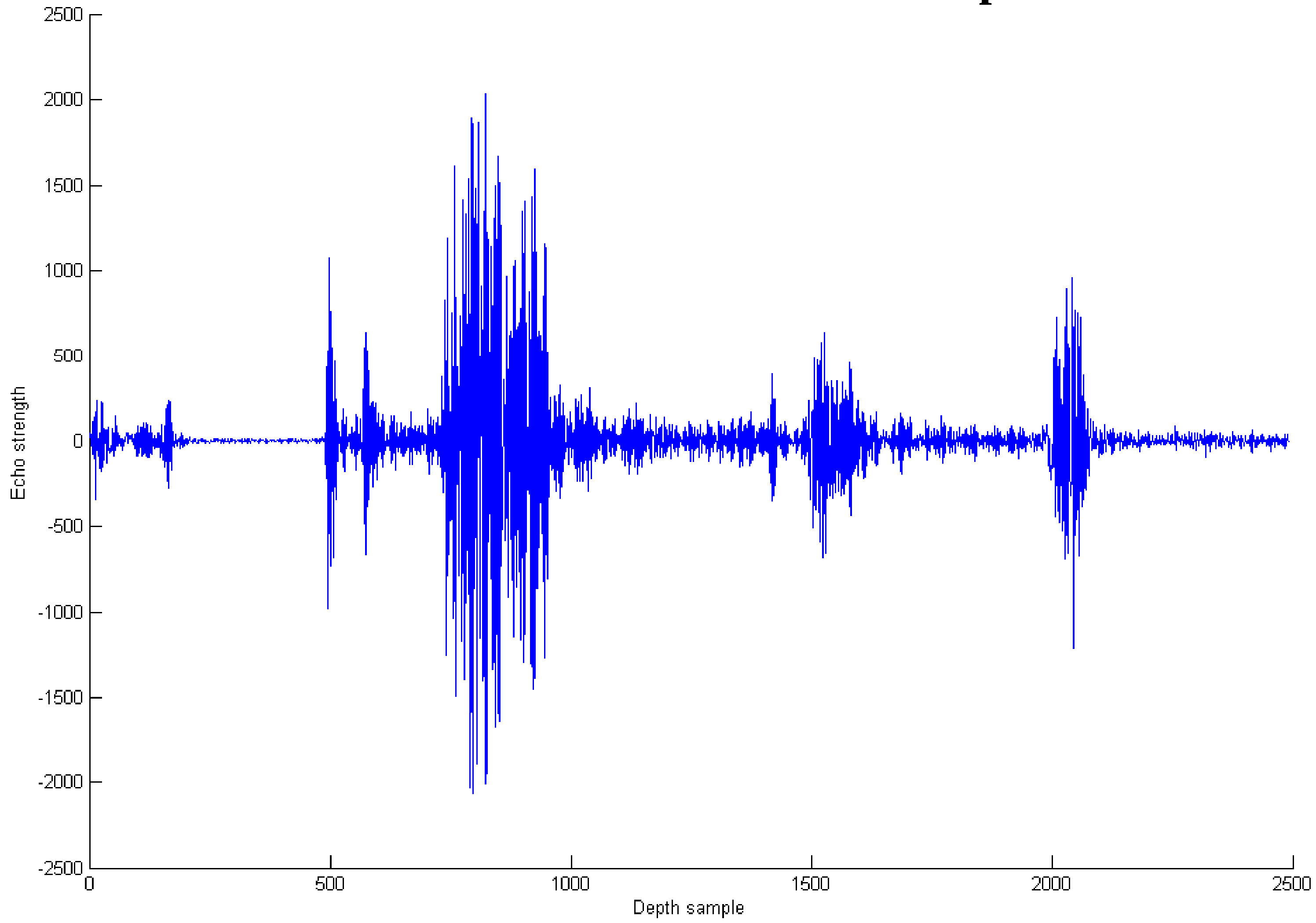
Images of test phantom with Terason portable ultrasound scanner



Bitmap image

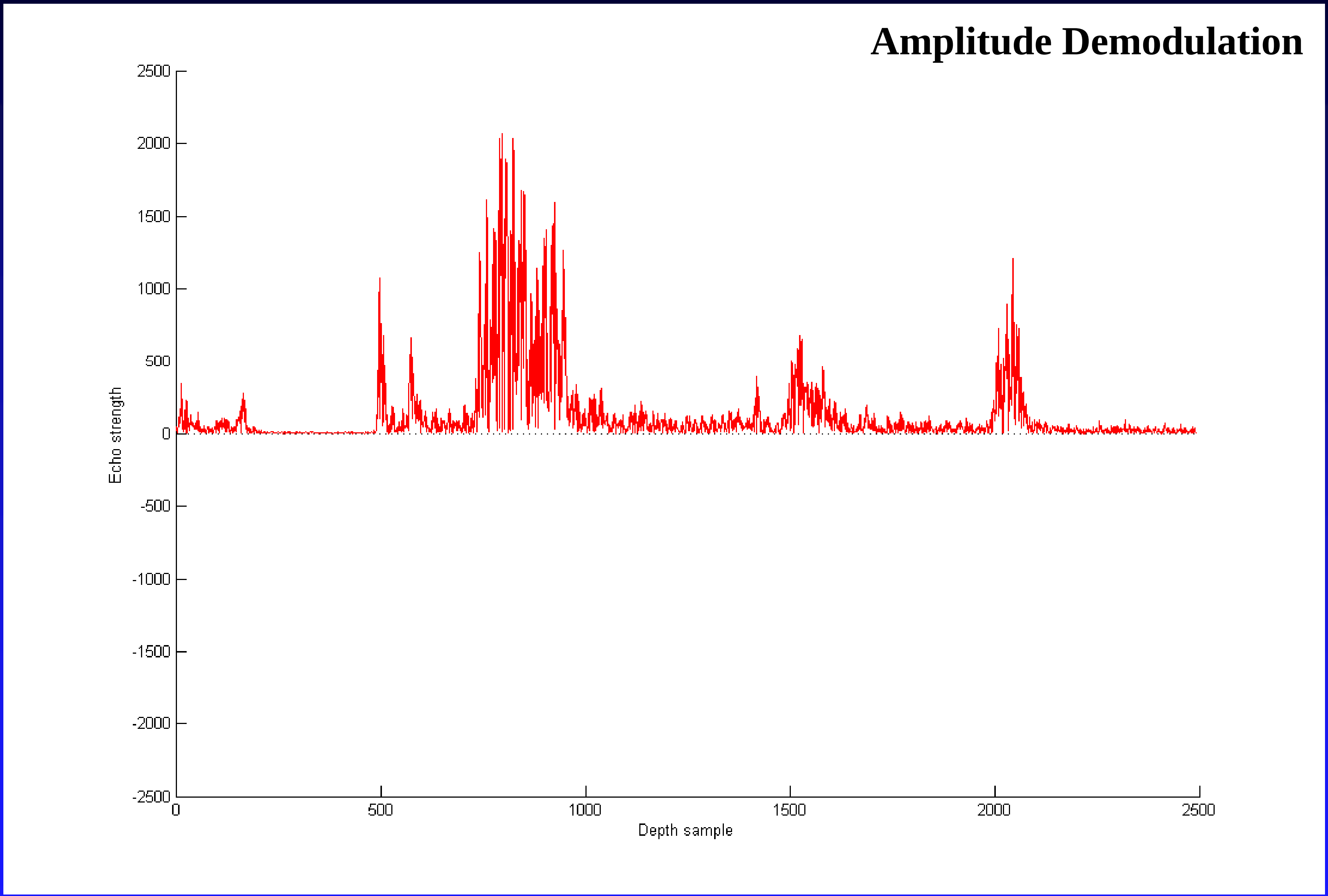
RF to B-Mode

Amplitude Demodulation



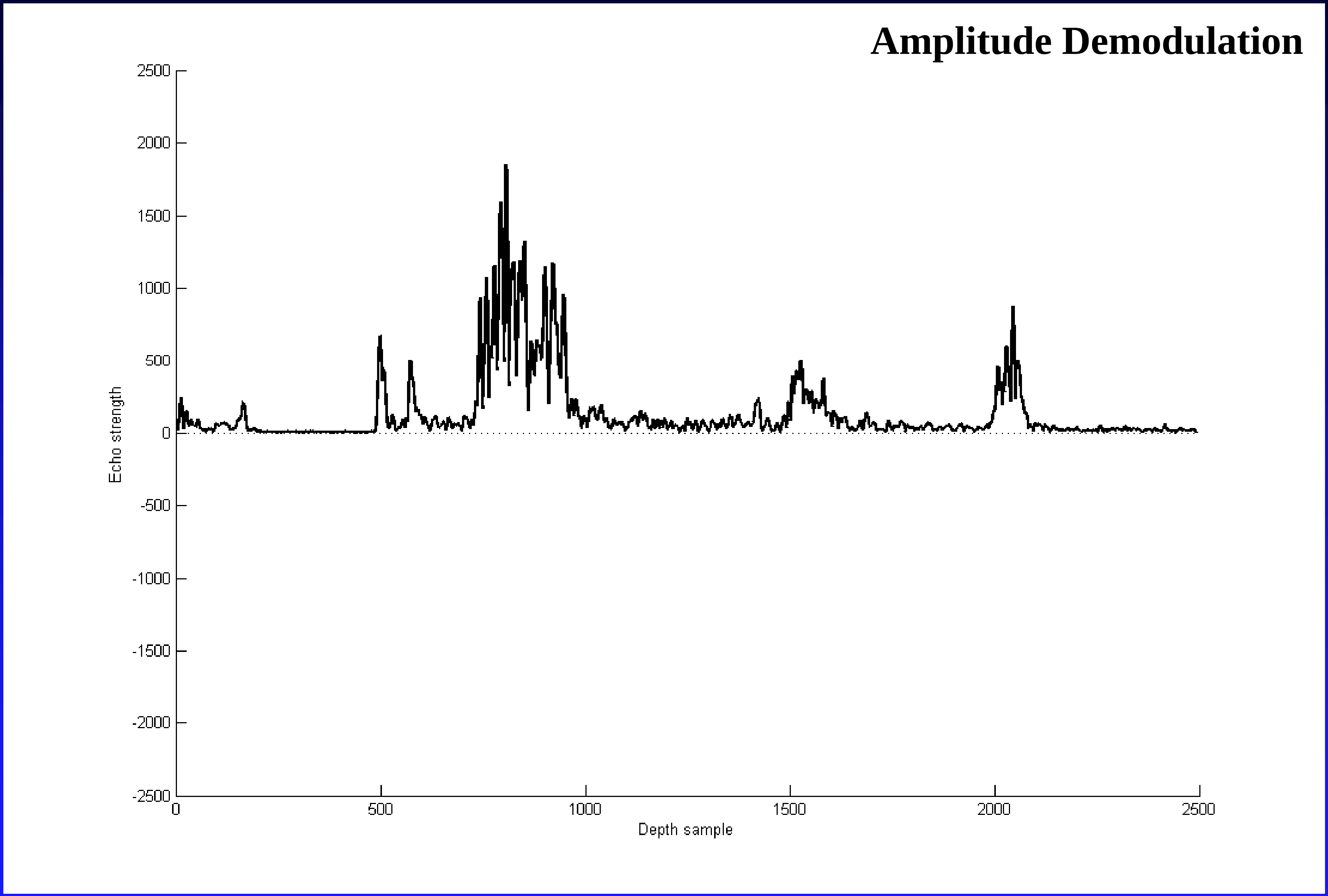
Single RF line

RF to B-Mode



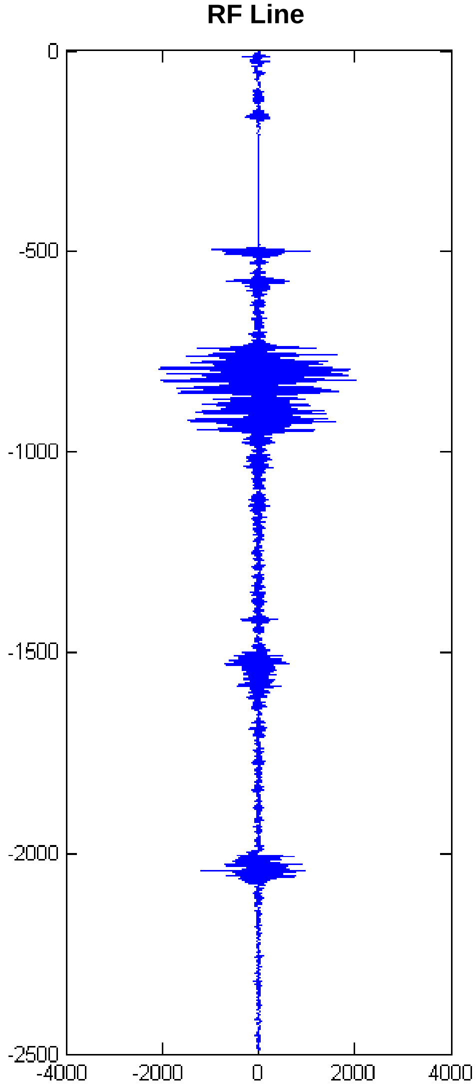
Rectified RF

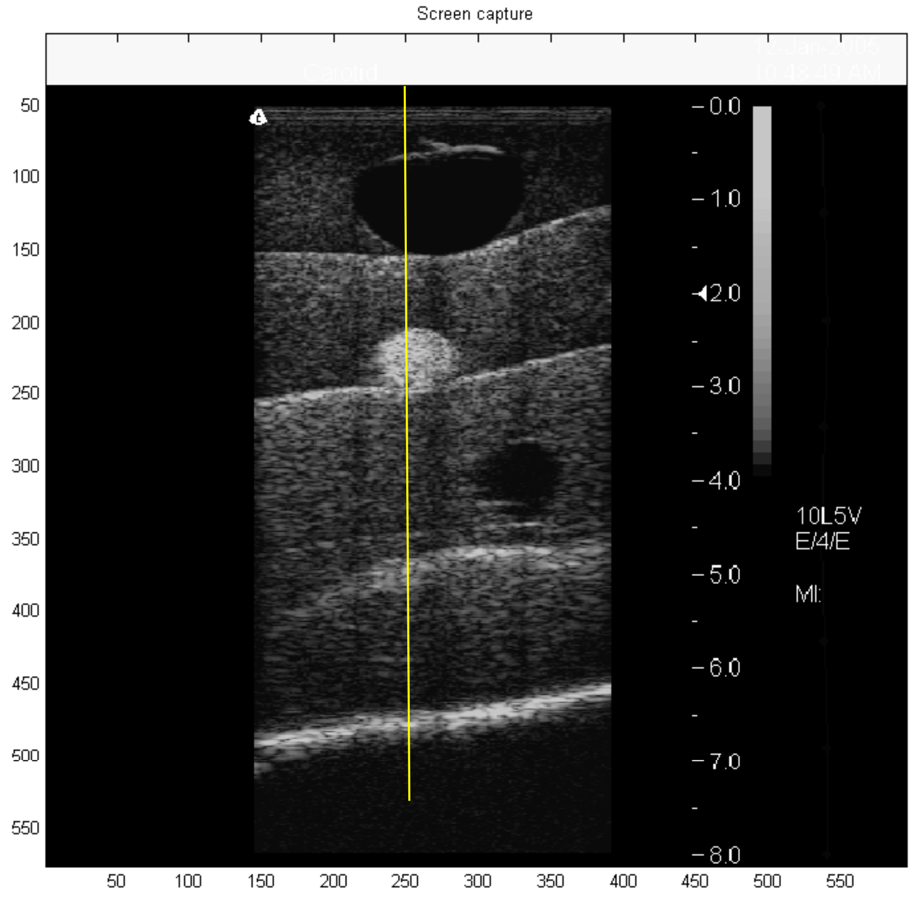
RF to B-Mode



Envelope (A-Mode)

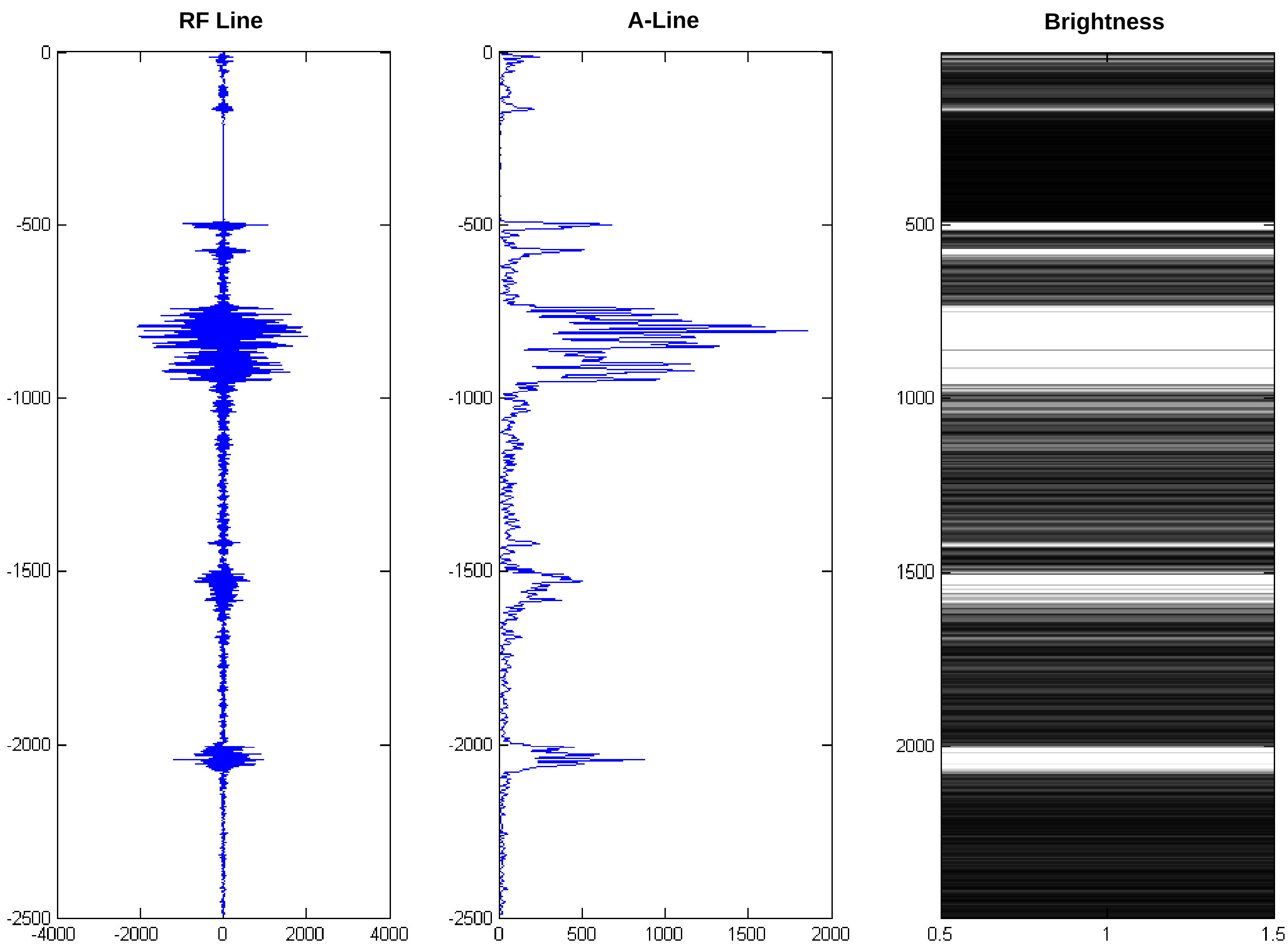
RF to B-Mode





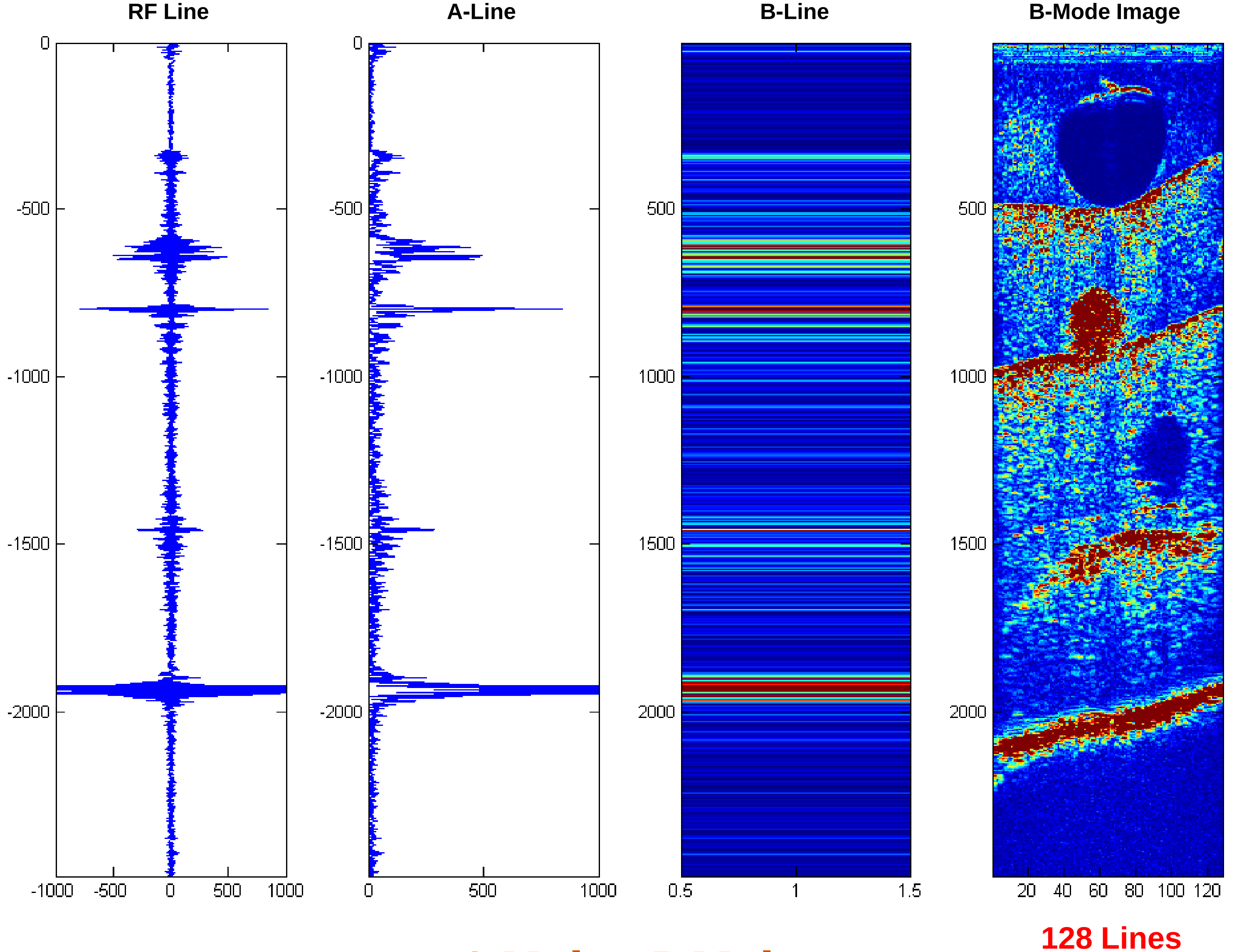
A-Mode to B-Mode

RF to B-Mode



A-Mode to B-Mode

RF to B-Mode

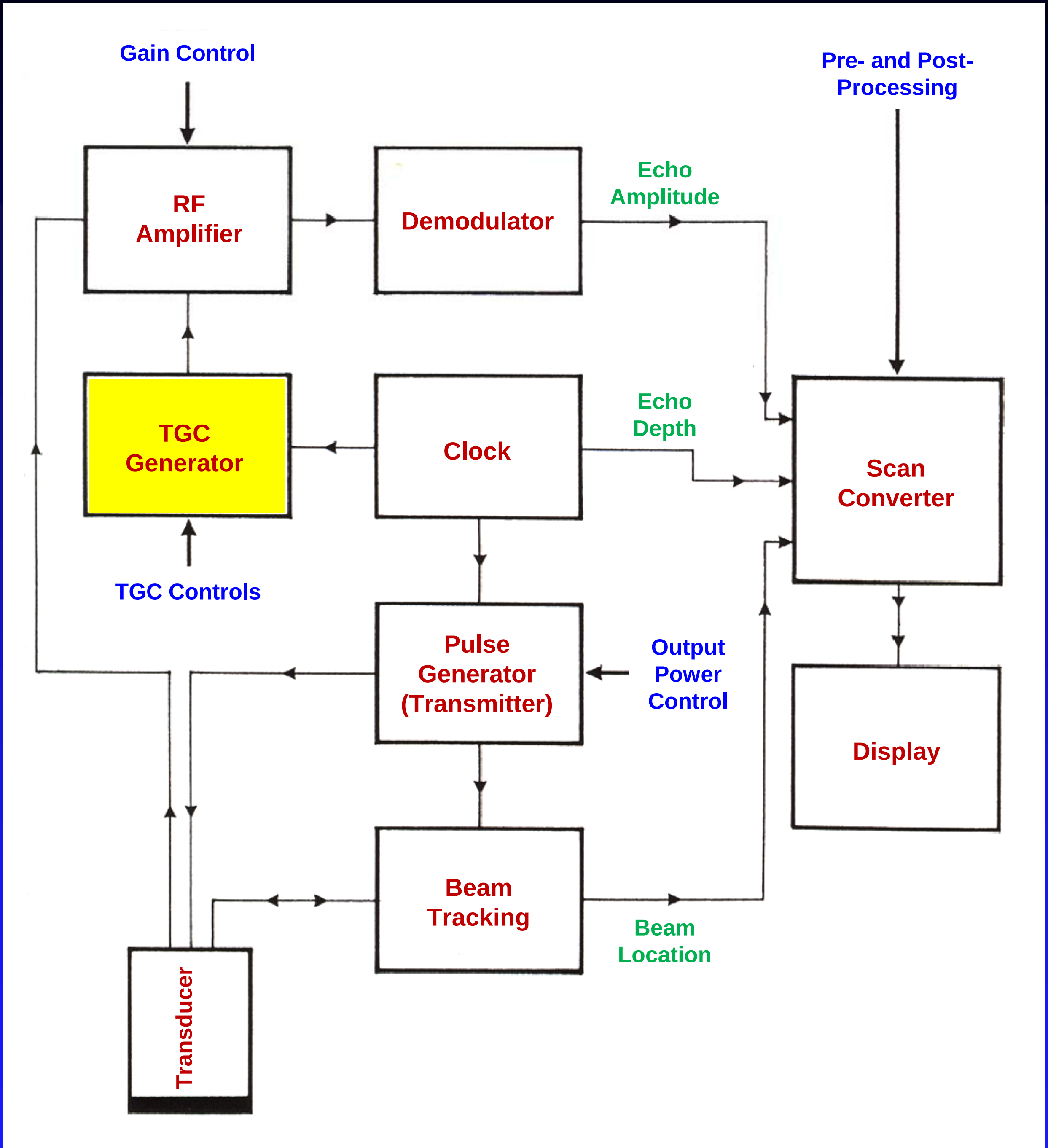


A-Mode to B-Mode

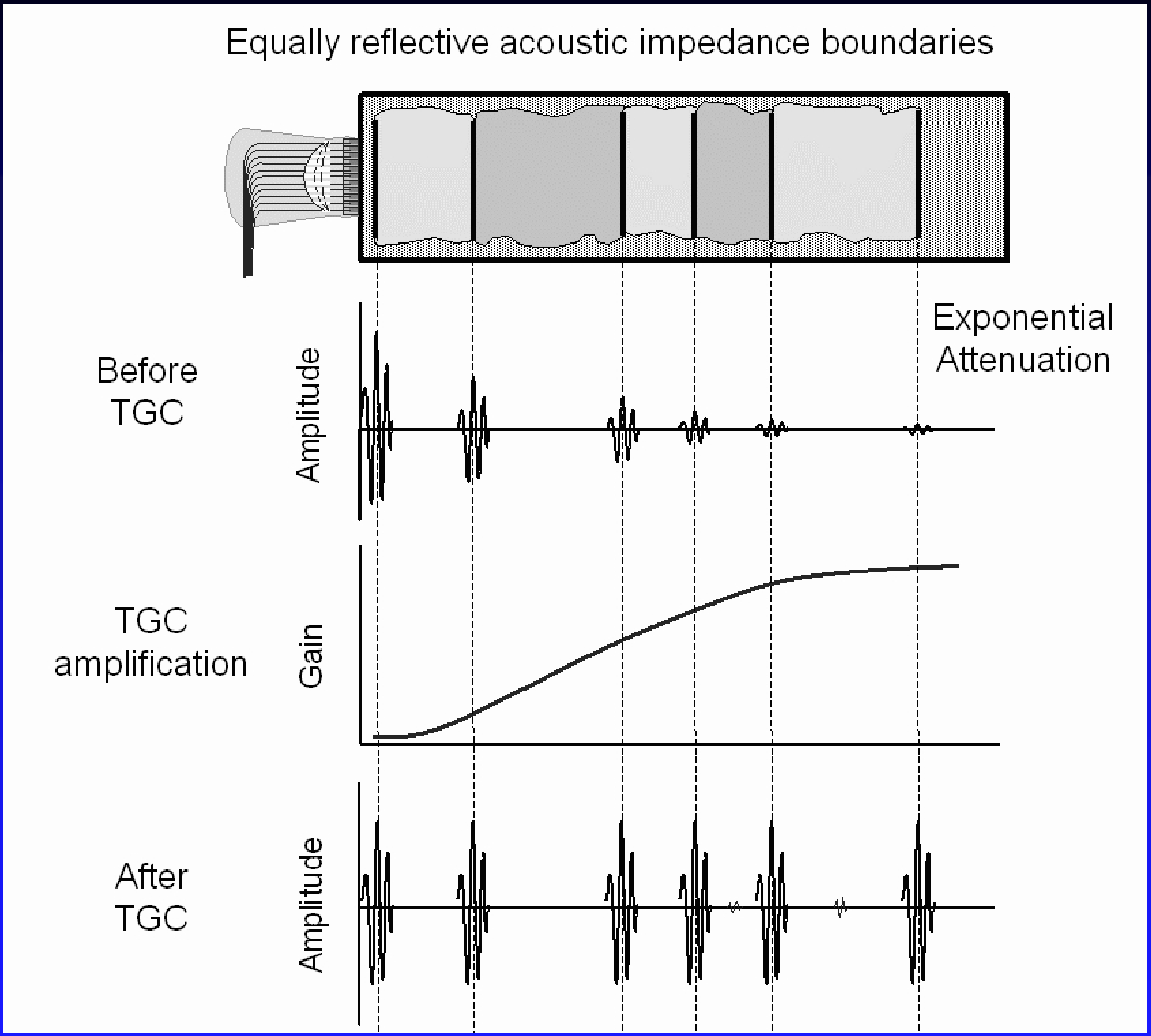
Components of an Ultrasound Scanner

2D gray scale imaging

- Time-Gain Compensation: adjust gain as a function of time (depth) to compensate for attenuation



Time-Gain Compensation (TGC)



Compensate for signal attenuation as a function of depth

Time-Gain Compensation

- Average attenuation rate: 0.8 dB/cm/MHz
- Variable TGC allows gain adjustment at different depths
- Multiple slider controls

HDI 5000
UW Vascular Research
L12-5 38 PVasc/Ven
09 Jan 04
9:10:51 am
TIs 0.2
MI 0.7
18 Hz
3.9 cm
Map 2
150dB/C 3
Persist Med
2D Opt:Gen
Fr Rate:High
ATL
0
1
2
3

TGC: correct

TGC: incorrect

Ultrasound Scanheads

- Scanhead construction and operation determine the format and characteristics of the ultrasound scan plane
  - Scanhead design affects resolution
    - Spatial and temporal
  - Range of designs available for specific imaging applications

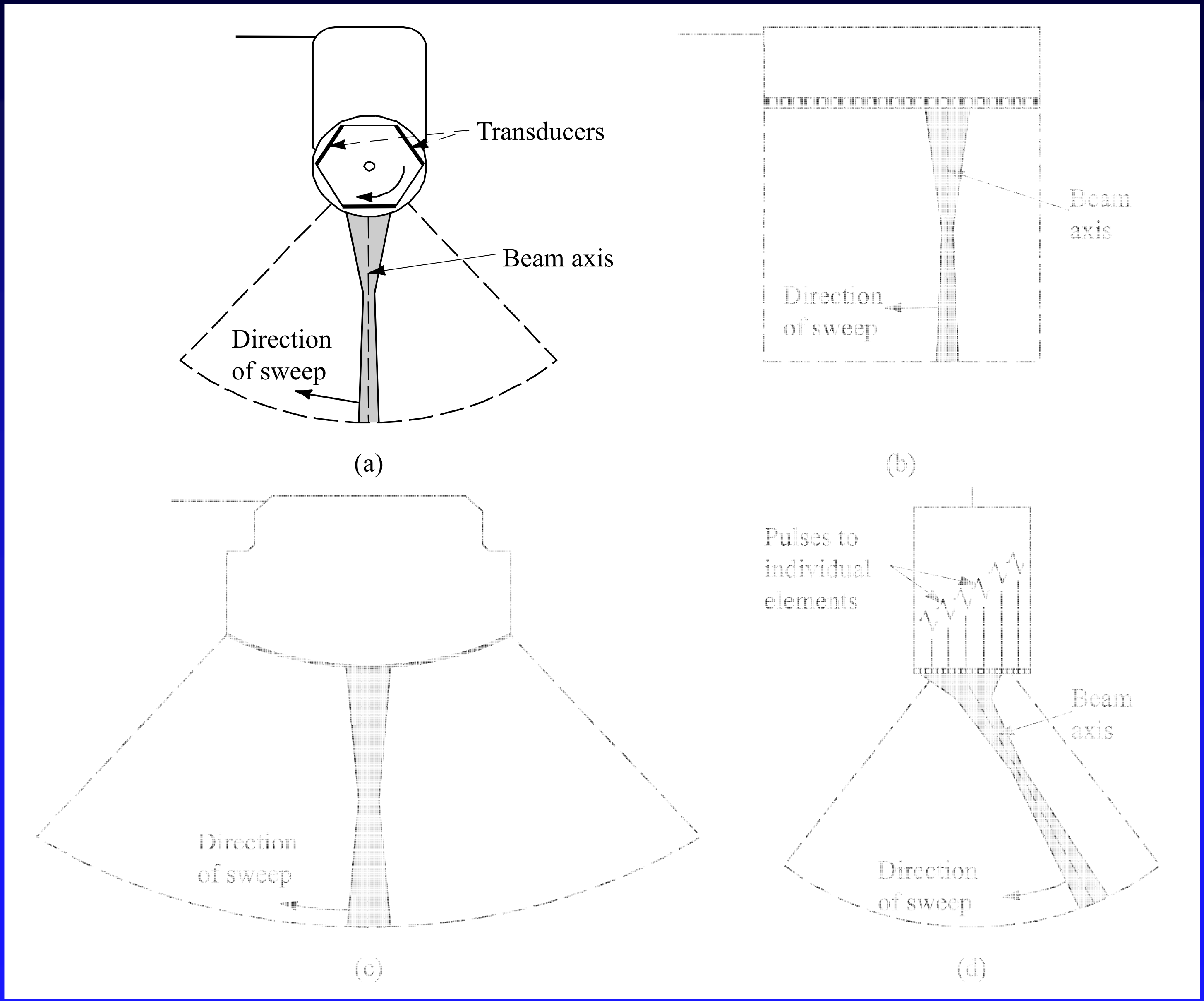
Ultrasound Scanheads

(a) Rotating mechanical device

(b) Linear array: scans an area the same width as the scanhead

(c) Curved linear array: sweeps a sector

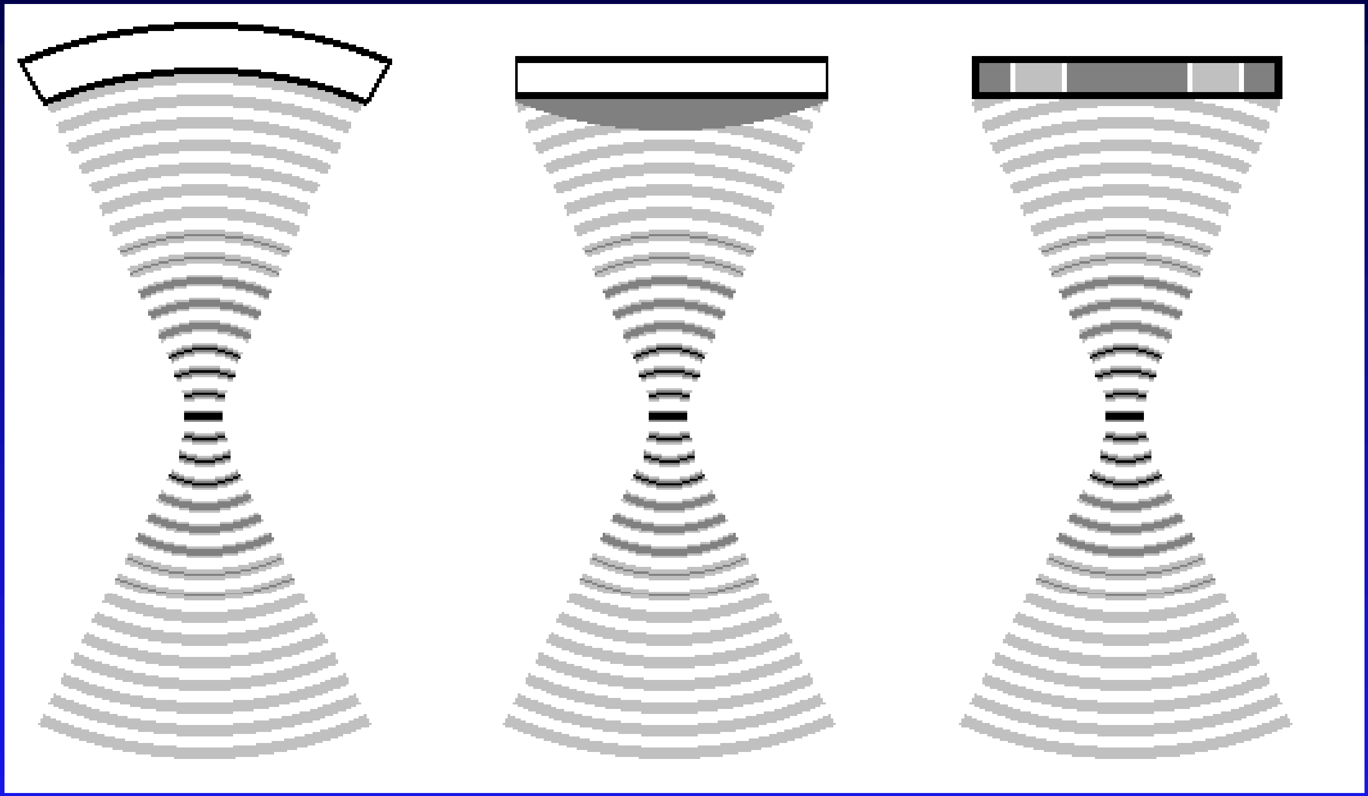
(d) Phased array: variable timing of the excitation across elements steers the beam so that a small transducer sweeps a large area



Mechanical Scanhead

- Single-element transducer is swept across the image plane by a motor
  - Prone to wear and damage over time
  - Fixed focus

Focusing Methods



Curved transducer face

Lens

Electronic focusing (phasing)

Only electronic focusing allows for variable focus; all other methods have fixed focus

Ultrasound Scanheads

(a) Rotating mechanical device

(b) Linear array: scans an area the same width as the scanhead

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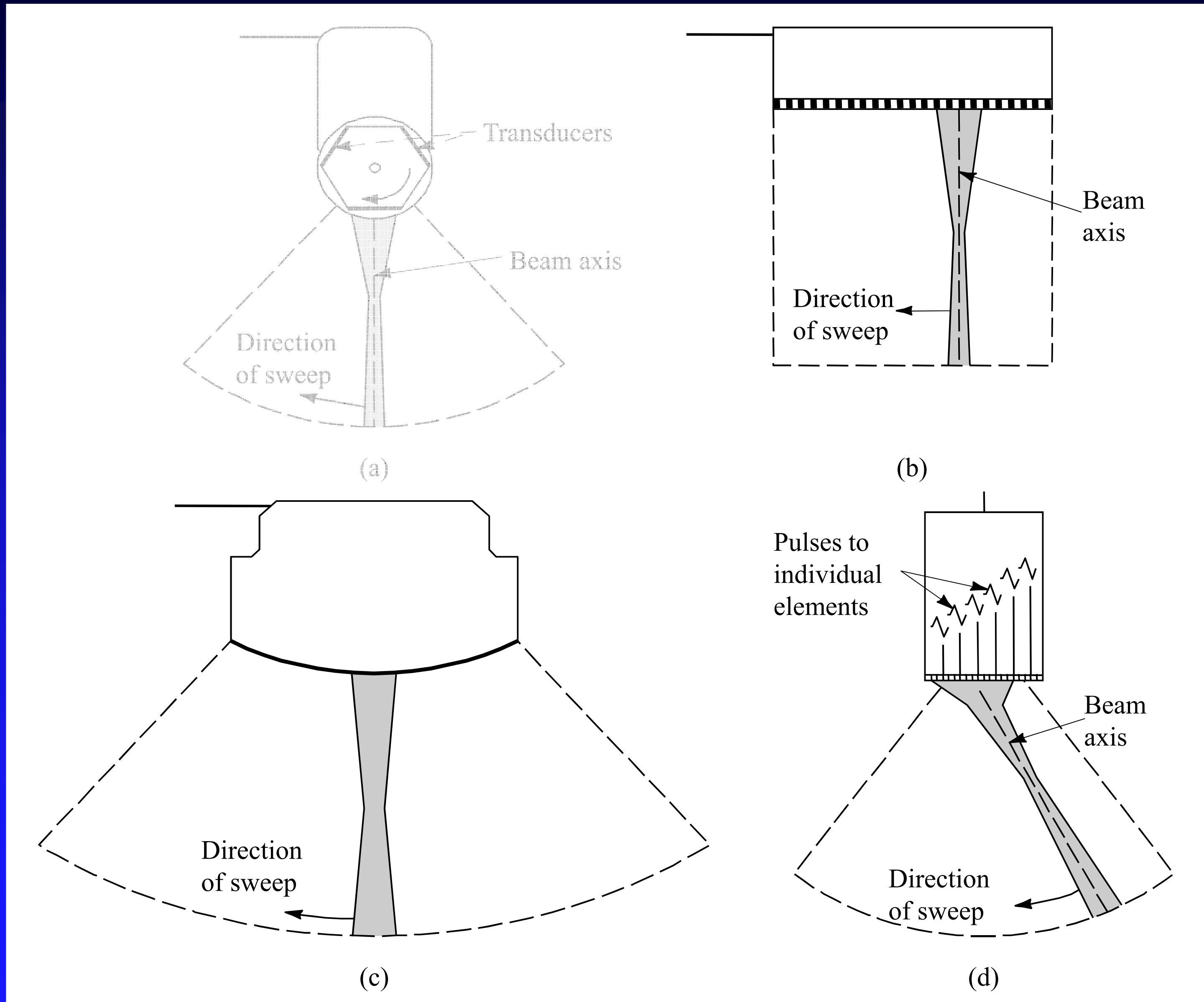
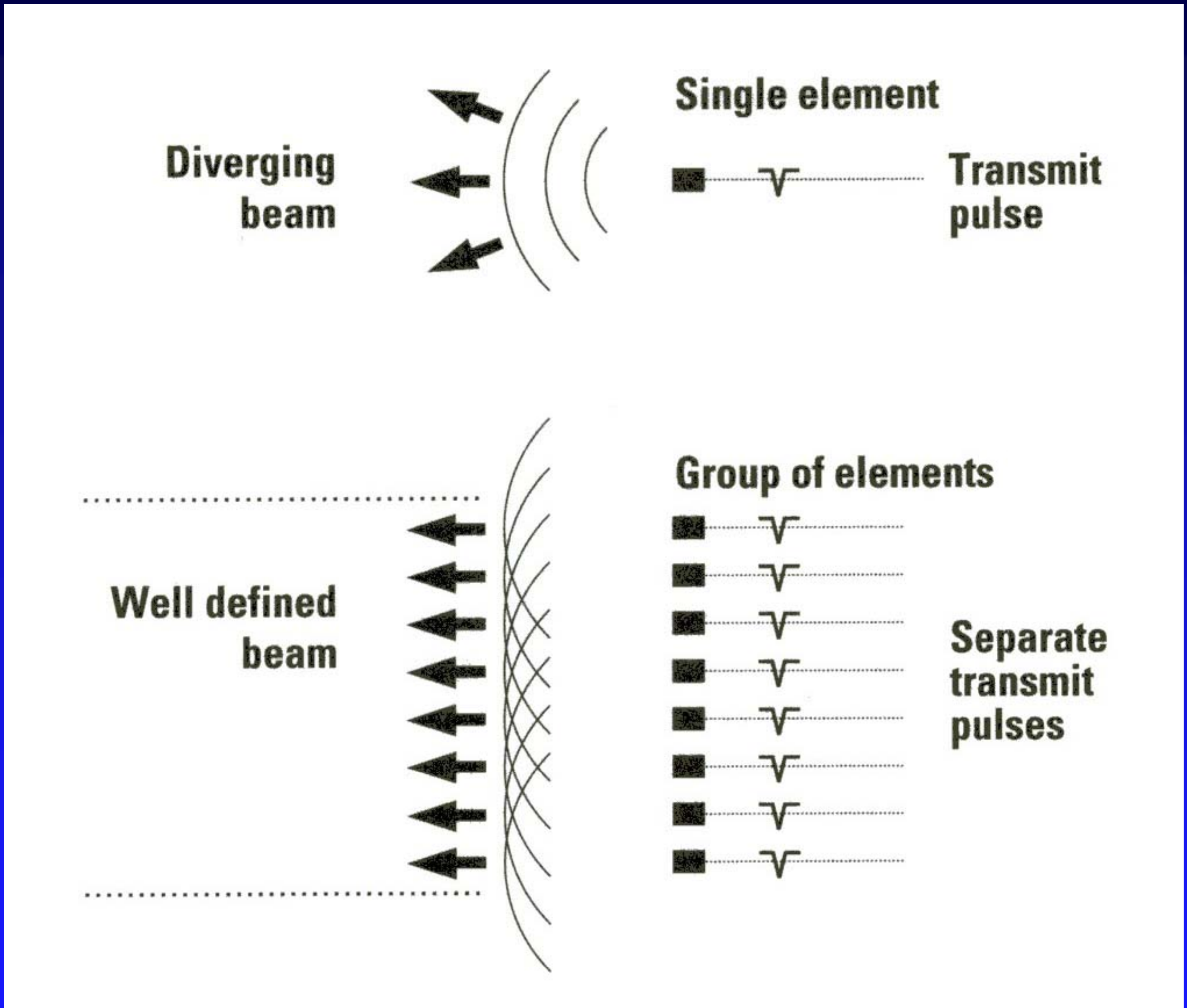
(d) Phased array: variable timing of the excitation across elements steers the beam so that a small transducer sweeps a large area



Array Scanheads

Arrays

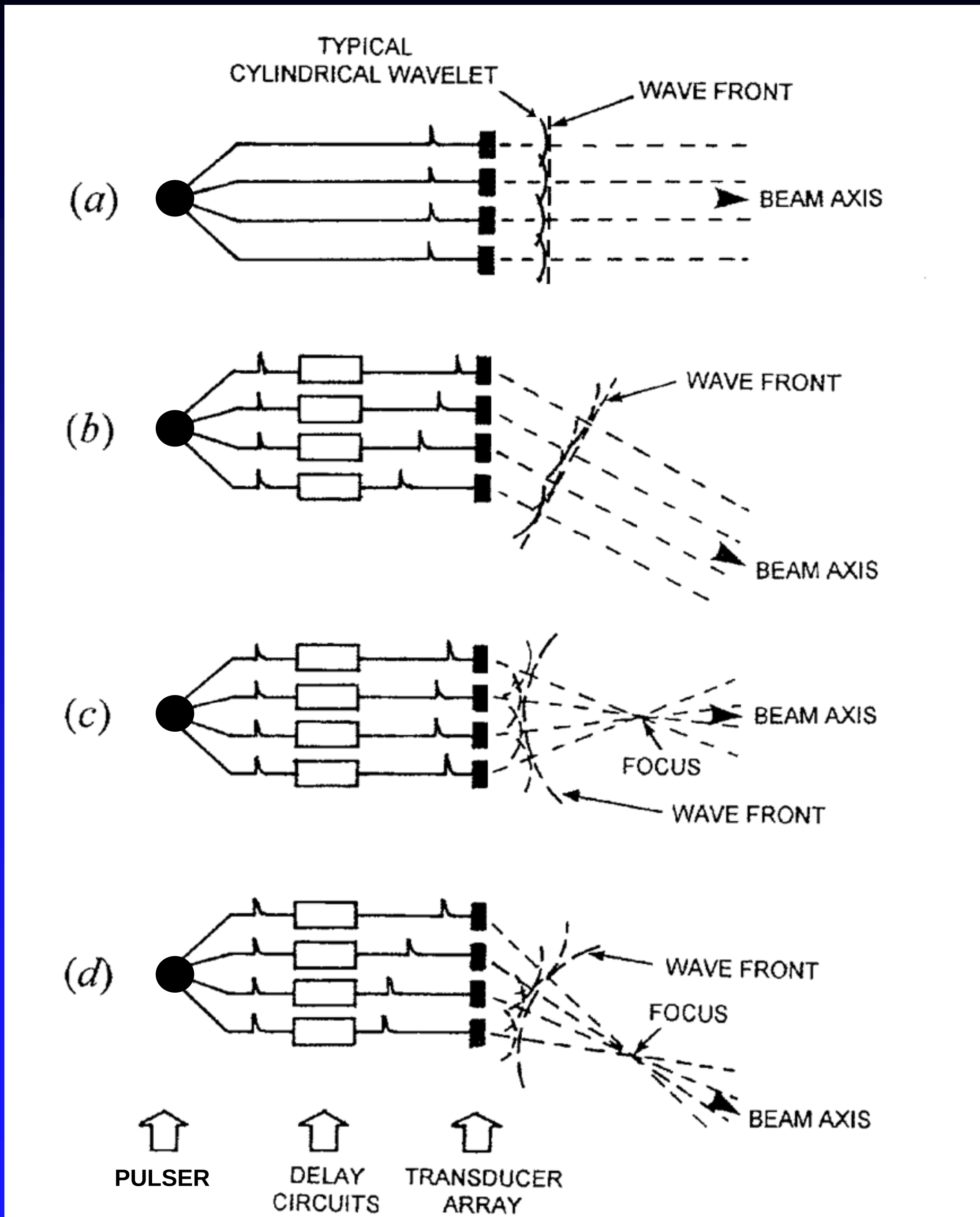


- Ultrasound waves from different elements sum
- Adjust <u>timing</u> of excitation across the elements to steer and focus the beam

Zagzebski

Arrays

Electronic arrays control the excitation time of multiple transducer elements to steer and focus the ultrasound beam



Steered

Focused

Sector scan

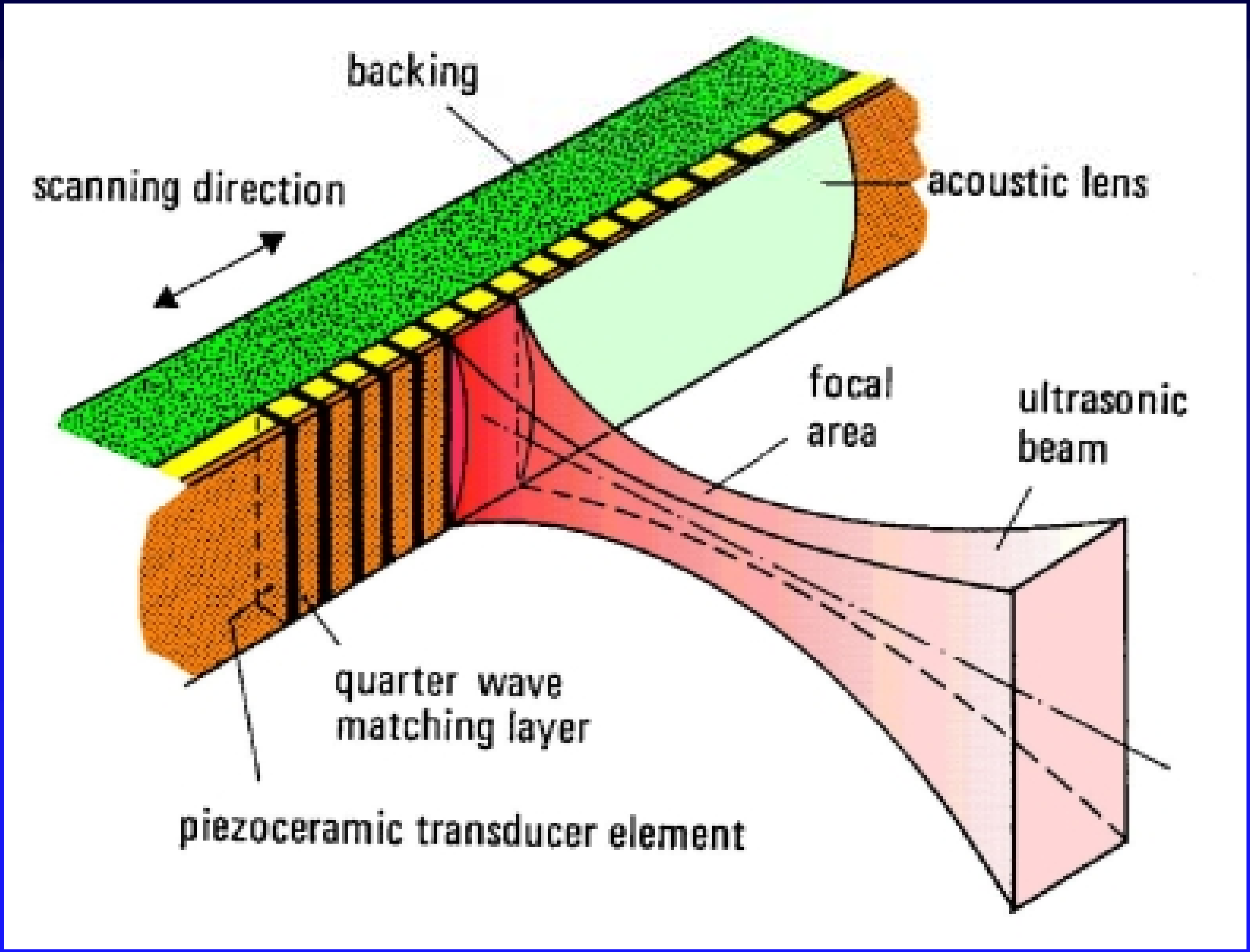
Steered and Focused

Resolution

- **Detail (geometric) resolution**
- **Temporal (frame-rate) resolution**

Arrays

- **Transducer elements in linear electronic arrays are <u>not</u> <u>symmetric</u>**
  - **Beam pattern is not symmetric**



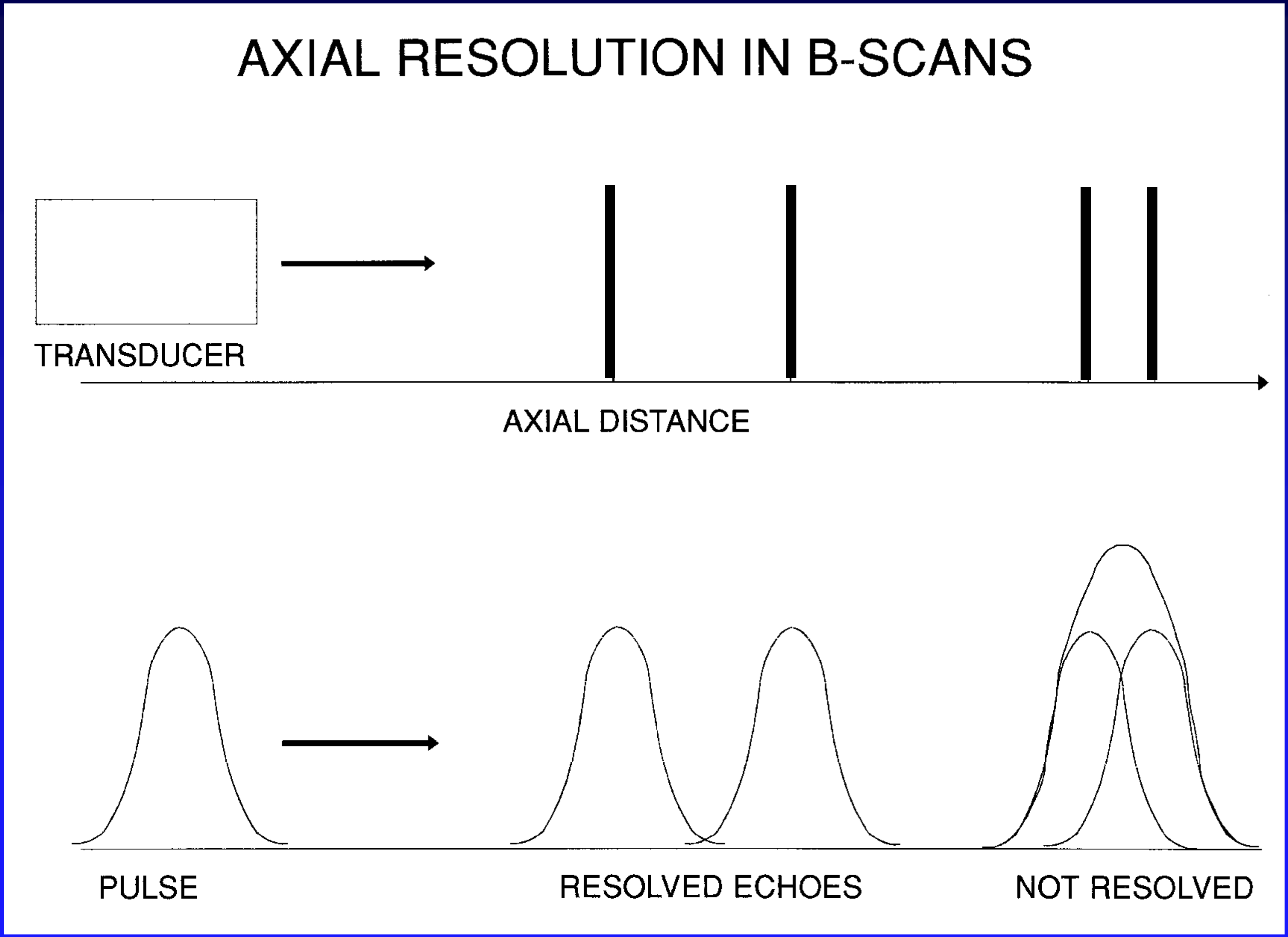
Klein

Detail Resolution

- Axial: along the scan line (depth)
  - Axial Resolution = 1/2 x (Spatial Pulse Length)
  - Constant with depth
  - Improves with increased frequency

- Lateral: perpendicular to the scan line within the image plane
  - Lateral Resolution = Beam width
  - Varies with depth
  - Improves with focusing and with increased frequency

Axial Resolution



Axial Resolution

- **Axial Resolution = (Spatial Pulse Length) / 2**
- **SPL = (# cycles/pulse) x λ**
- **Improve axial resolution by**
  - reduced number of cycles
  - increased frequency

$\lambda = c/f$

c = propagation speed
f = transmit frequency

- Wavelength is affected by frequency and the medium
- Ultrasound wavelengths in tissue are less than 1 mm

Axial Resolution

Example: 3-cycle pulse

5 MHz transducer

$\lambda = c/f$

$= (1540 \text{ m/s}) / (5 \times 10^6 \text{ Hz})$

$= 0.308 \text{ mm}$

Axial resolution $= (3 \times \lambda) / 2$

$= 0.462 \text{ mm}$

10 MHz transducer

$\lambda = c/f$

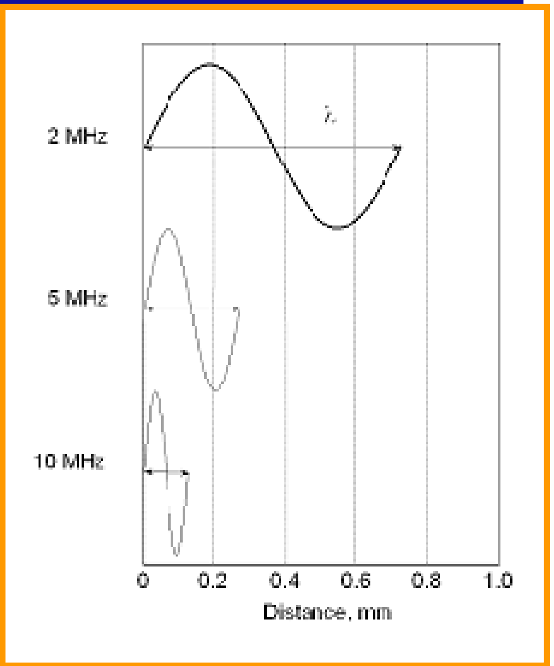
$= (1540 \text{ m/s}) / (10 \times 10^6 \text{ Hz})$

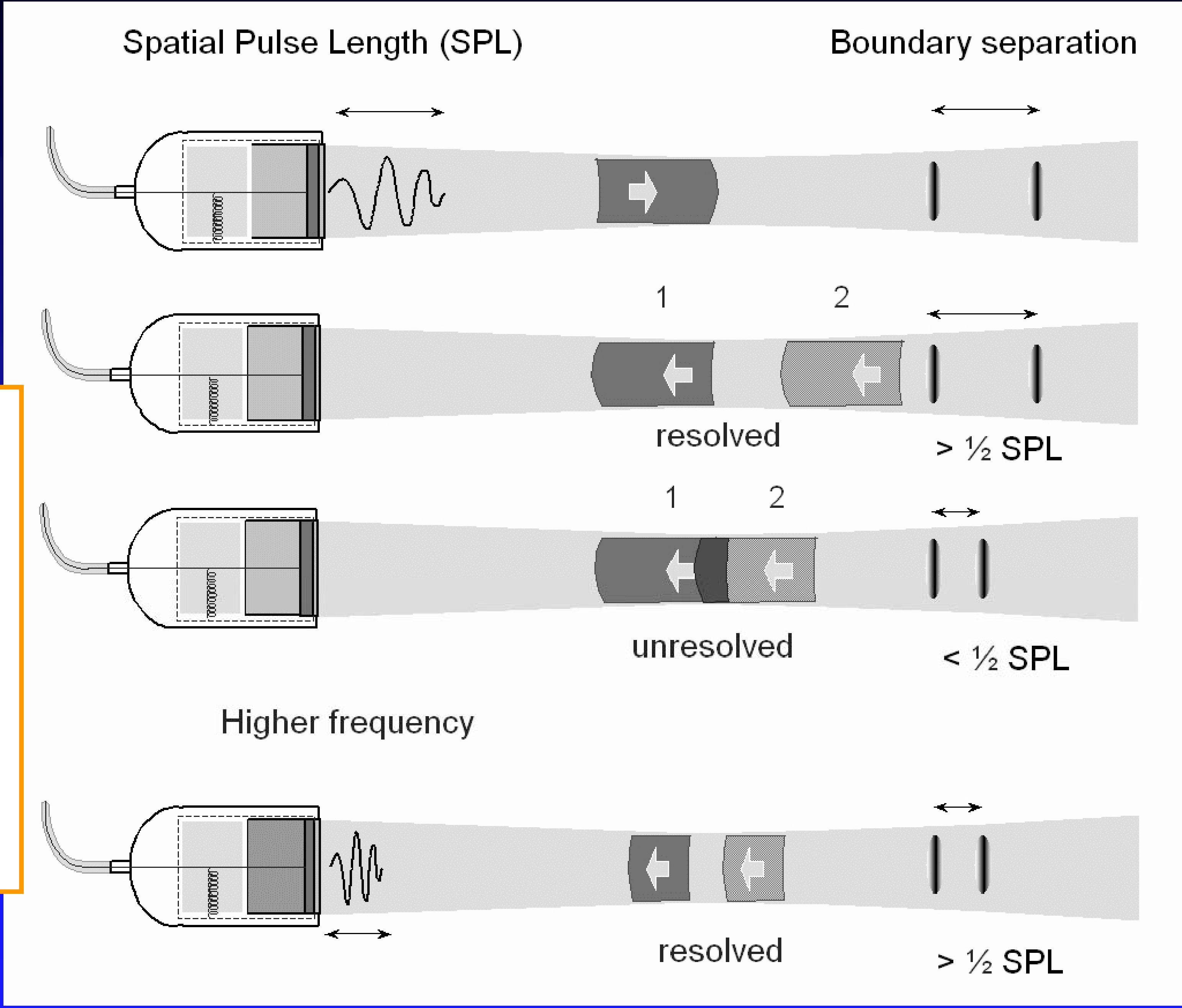
$= 0.154 \text{ mm}$

Axial resolution $= (3 \times \lambda) / 2$

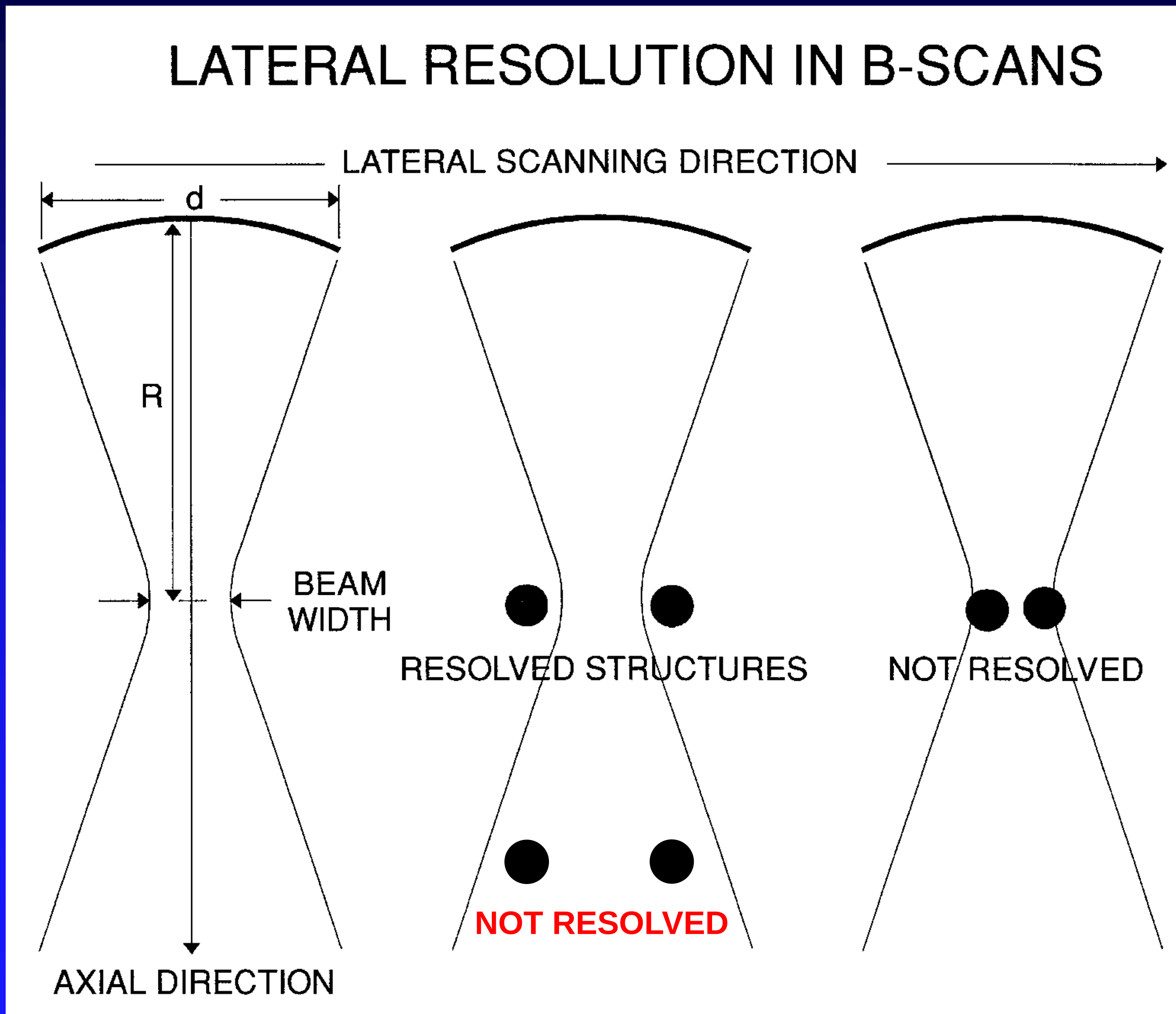
$= 0.231 \text{ mm}$

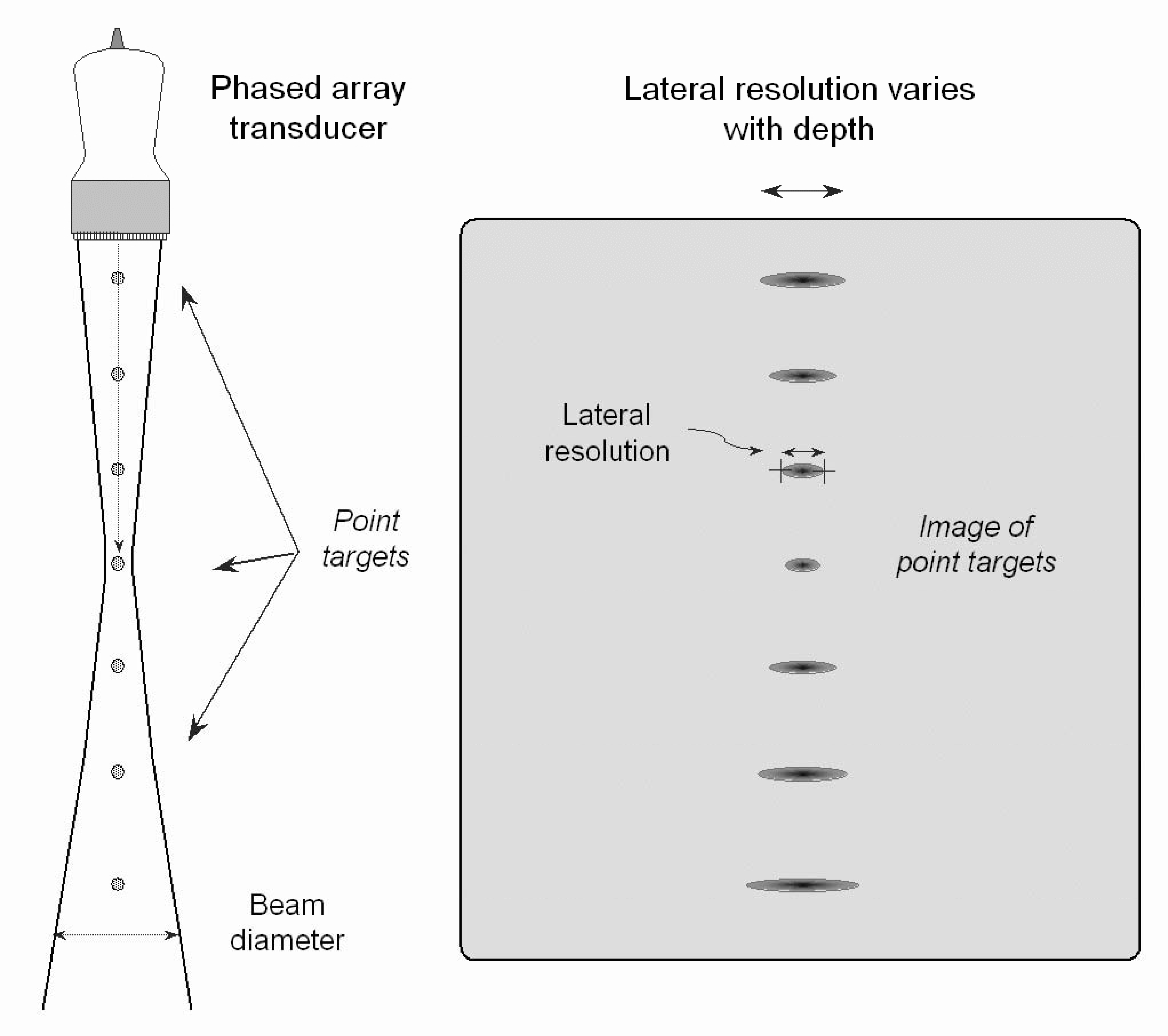
Wavelengths





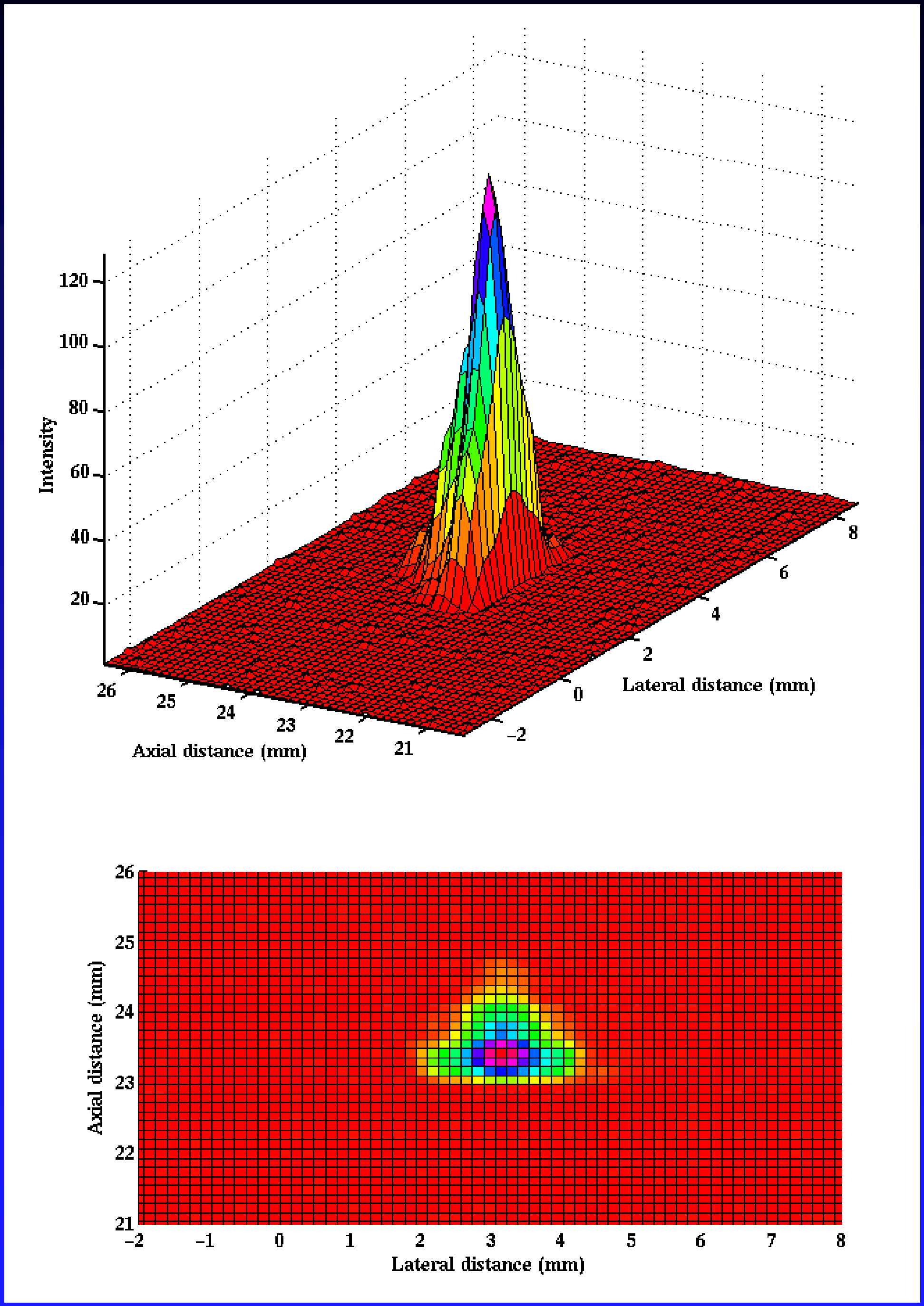
Lateral Resolution




Phased array transducer
Lateral resolution varies with depth
Lateral resolution
Point targets
Image of point targets
Beam diameter

Axial and Lateral Resolution

Axial and Lateral Resolution



Plot of image brightness from a string target

Imaging depth: 4.8 cm

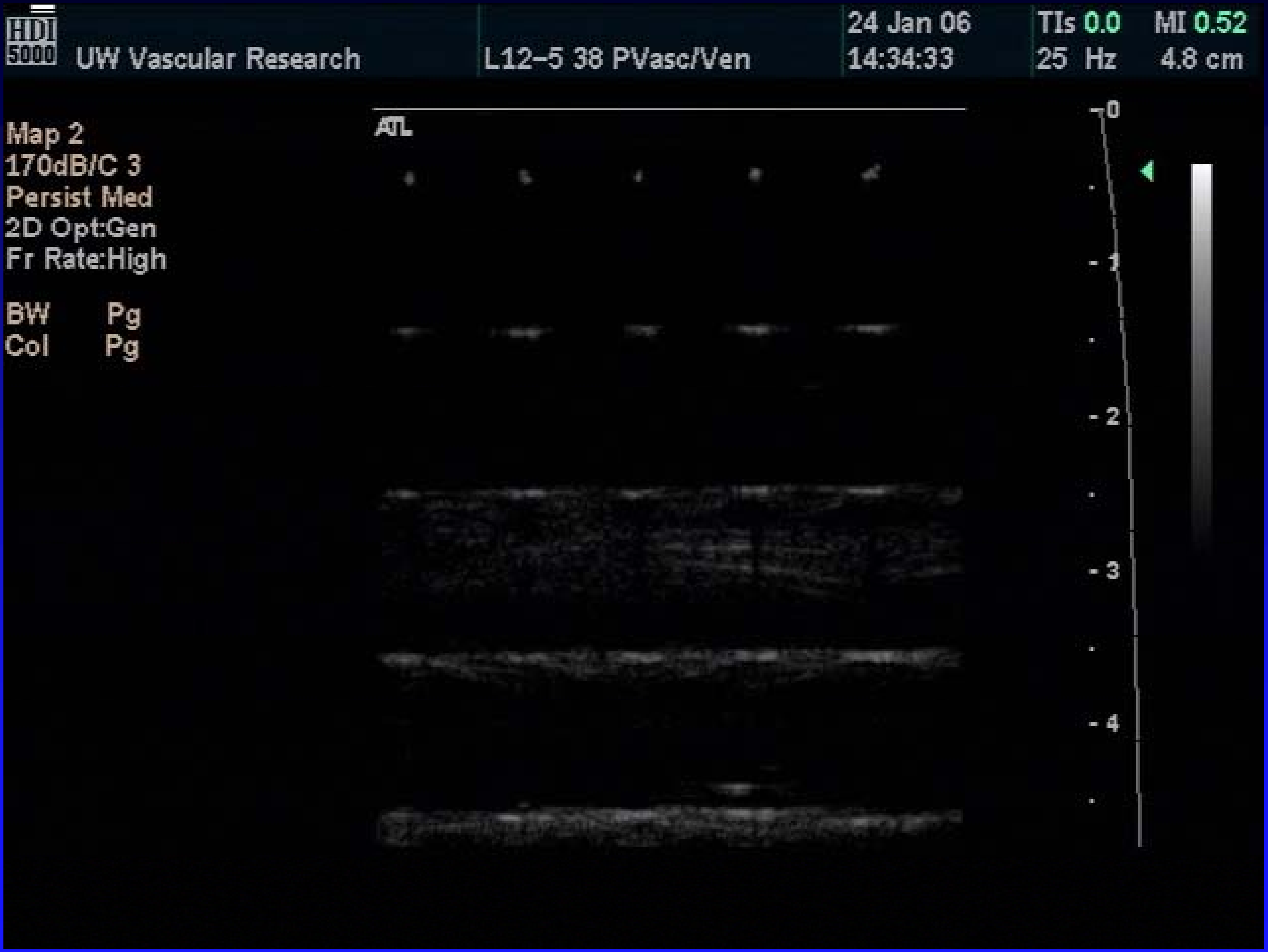
Target depth: 2.3 cm

Focal depth: 3.5 cm

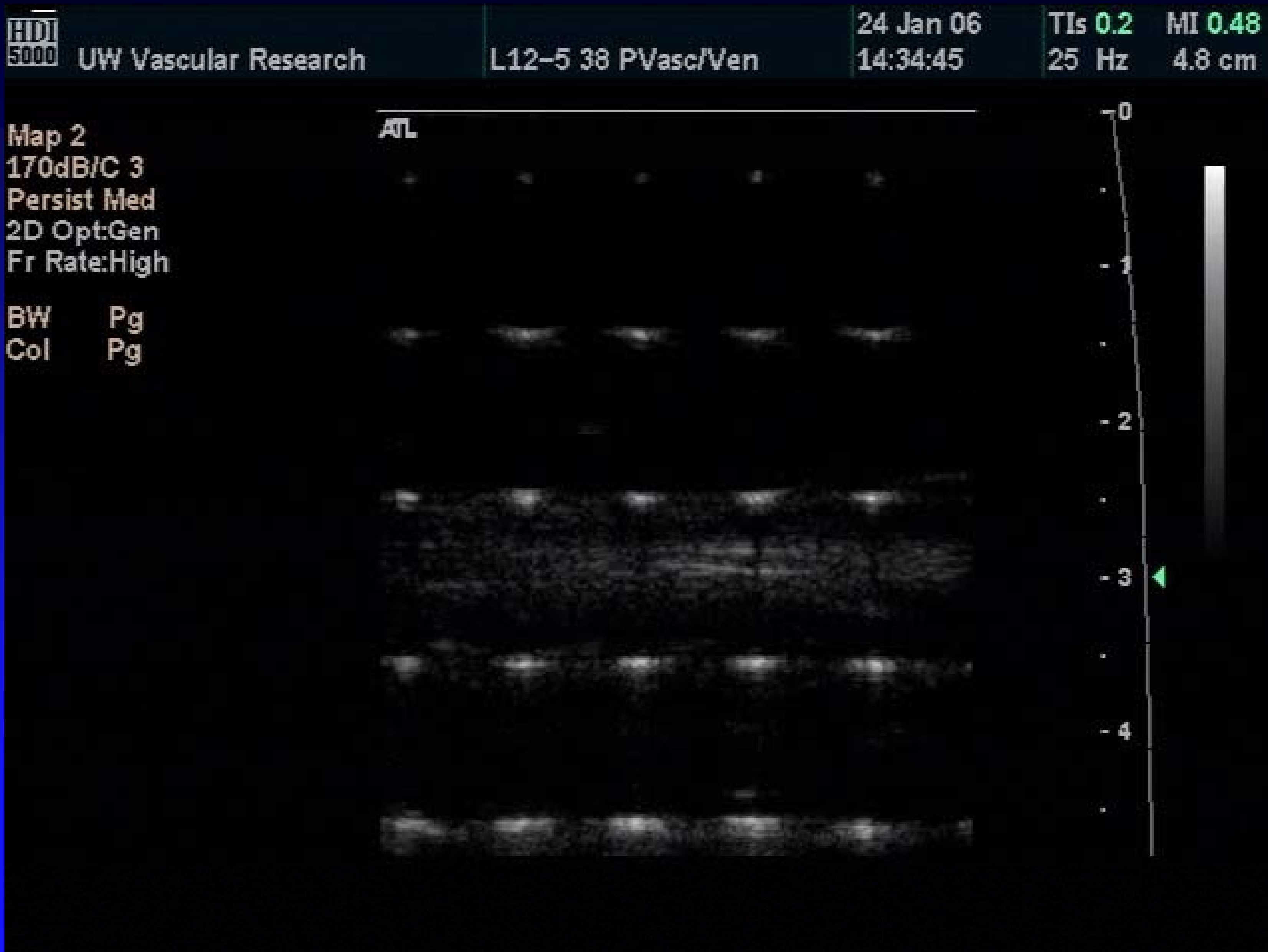
Beam direction

ATL L10-5 on HDI 3000

Leotta 1998


HDI 5000
UW Vascular Research
L12-5 38 PVasc/Ven
24 Jan 06
14:34:33
TIs 0.0
25 Hz
MI 0.52
4.8 cm
Map 2
170dB/C 3
Persist Med
2D Opt:Gen
Fr Rate:High
BW Pg
Col Pg
ATL
0
1
2
3
4

Focus Depth


HDI 5000
UW Vascular Research
L12-5 38 PVasc/Ven
24 Jan 06
14:34:45
TIs 0.2
25 Hz
MI 0.48
4.8 cm
Map 2
170dB/C 3
Persist Med
2D Opt:Gen
Fr Rate:High
BW Pg
Col Pg
ATL
0
1
2
3
4

Focus
Depth

Detail Resolution

MHz	**Axial resolution**	Lateral resolution	**Wave length (mm)**
3.0	1.1 mm	2.8 mm	0.5
4.0	0.8 mm	1.5 mm	0.375
5.0	0.6 mm	1.2 mm	0.3
7.5	0.4 mm	1.0 mm	0.2
10.0	0.3 mm	1.0 mm	0.15

Tradeoff: **Increased frequency → Improved resolution**
Increased frequency → Increased attenuation

Frequency Tradeoffs

Increased frequency → Improved resolution
Increased frequency → Increased attenuation

- Use lower frequencies for deeper structures
- Use highest frequency that can penetrate to the depth of interest

Transcranial
1.5 - 2.0 MHz

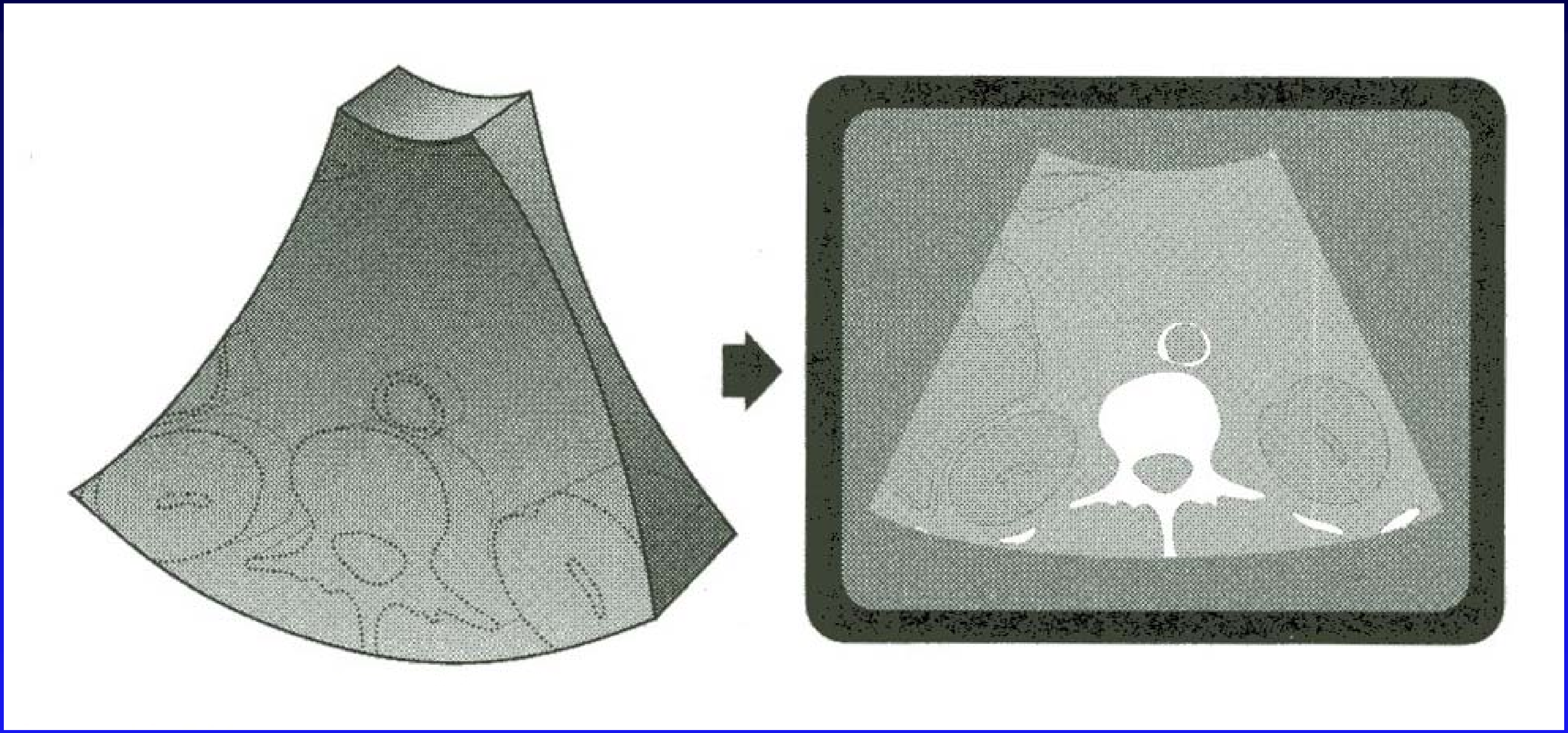
Cardiac
2.0 - 5.0 MHz

Abdominal
2.0 - 5.0 MHz

Musculoskeletal
5.0 - 12.0 MHz

Peripheral Vascular
7.0 - 15.0 MHz

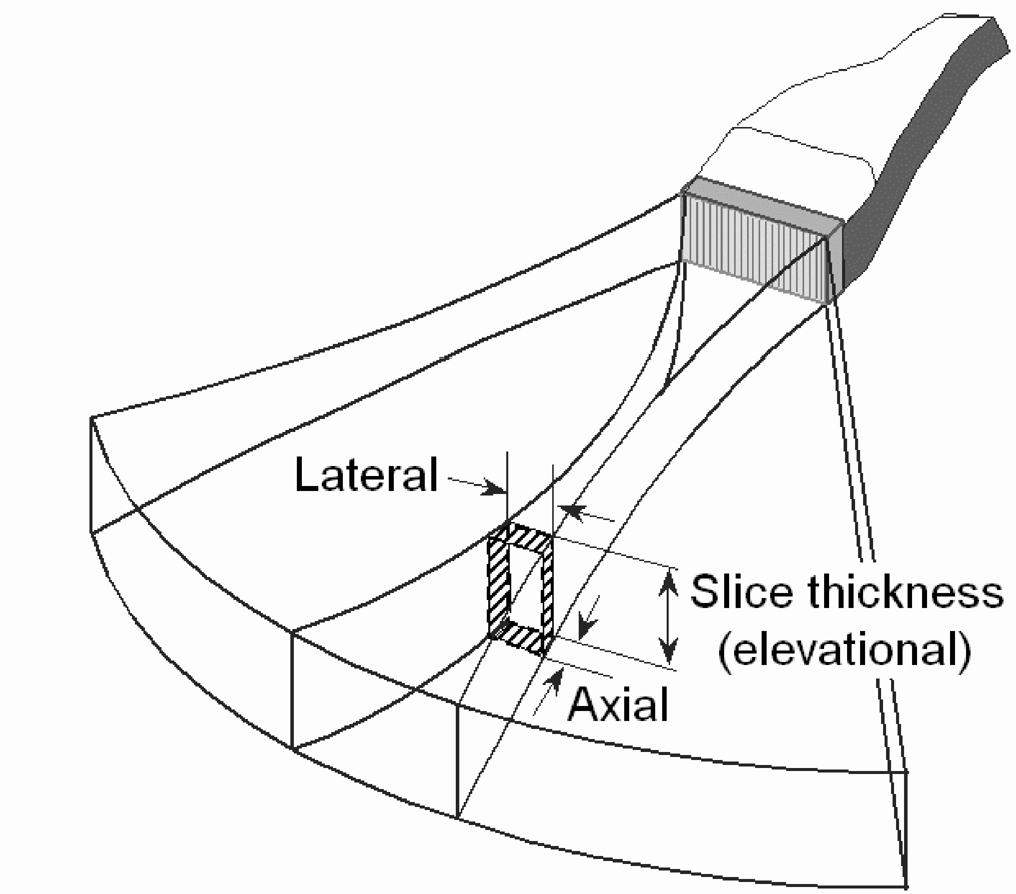
Beam Thickness



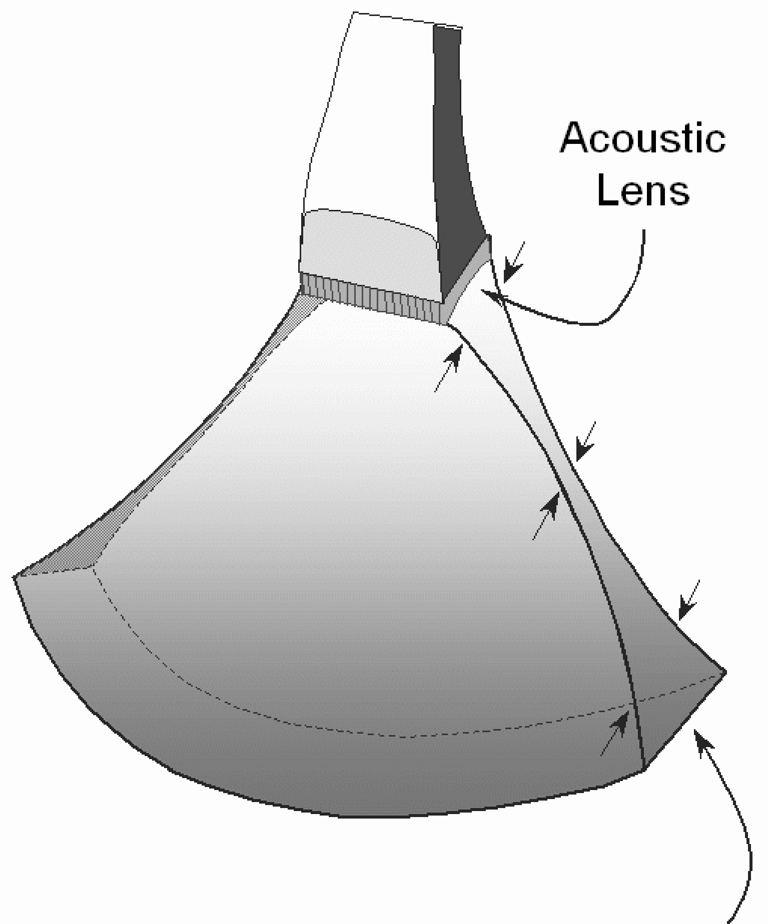
- **Beam pattern perpendicular to 2D image plane**
- **Beam thickness (elevation) generally larger than lateral beam width**
- **Fixed focus set by acoustic lens**

Zagzebski

Beam Thickness



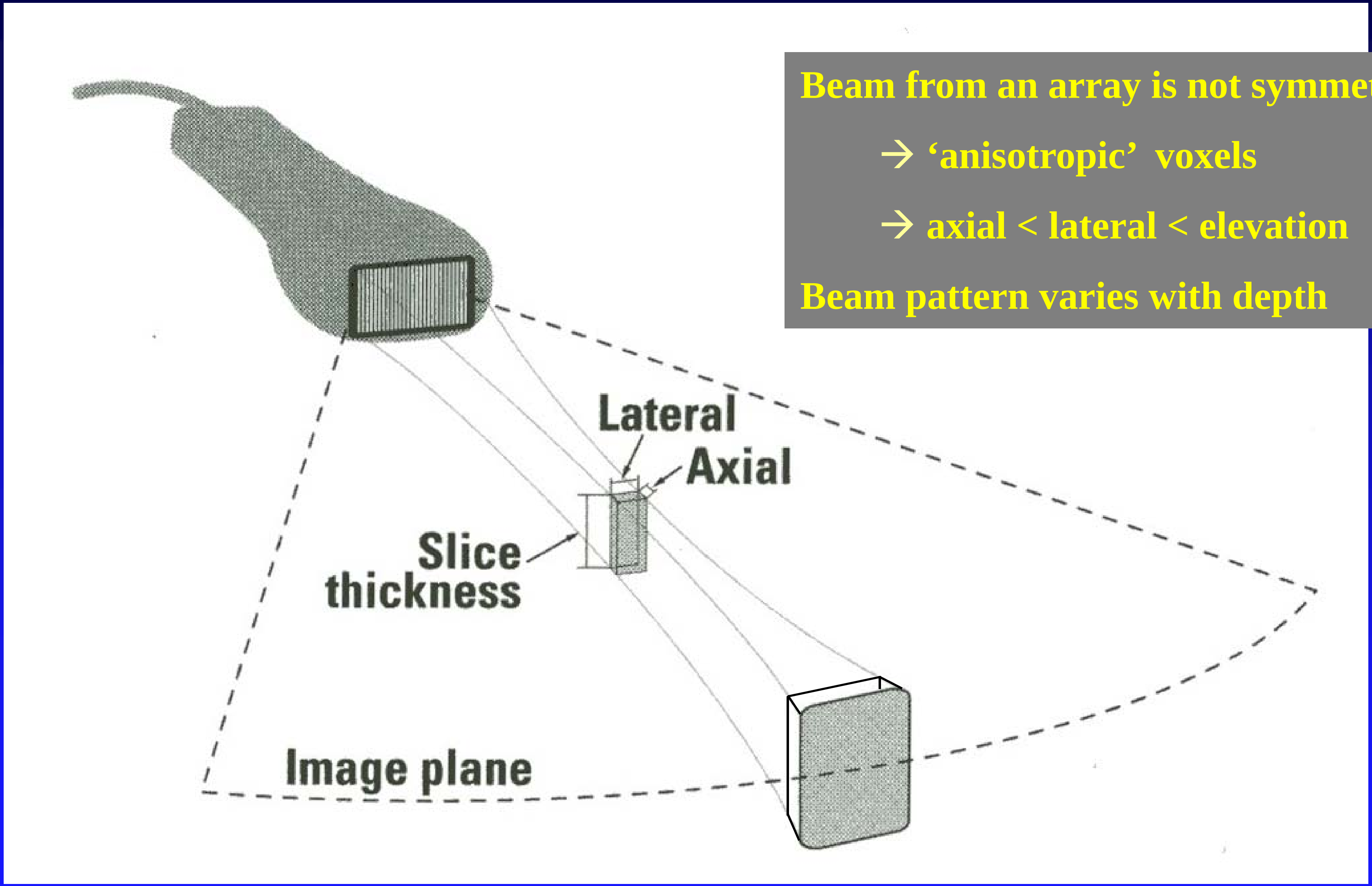
A Resolution components in 3-D space.



B Elevational Profile of Ultrasound Beam with depth

'Beam Thickness'

Array Resolution



Beam from an array is not symmetric

→ 'anisotropic' voxels

→ axial < lateral < elevation

Beam pattern varies with depth

Resolution

- **Detail (geometric) resolution**
- **Temporal (frame-rate) resolution**

Temporal Resolution

Frame Rate: **the number of 2D images that can be produced per second**

- Decreases with increasing imaging depth
- Decreases with increasing number of scan lines

Temporal Resolution

Pulse Repetition Frequency

- If echoes can arrive from as far as depth R, then we must wait at least until $t = 2R/c$ to transmit the next pulse
- Therefore the Pulse Repetition Frequency (PRF) must be $\leq c/2R$

Frame Rate

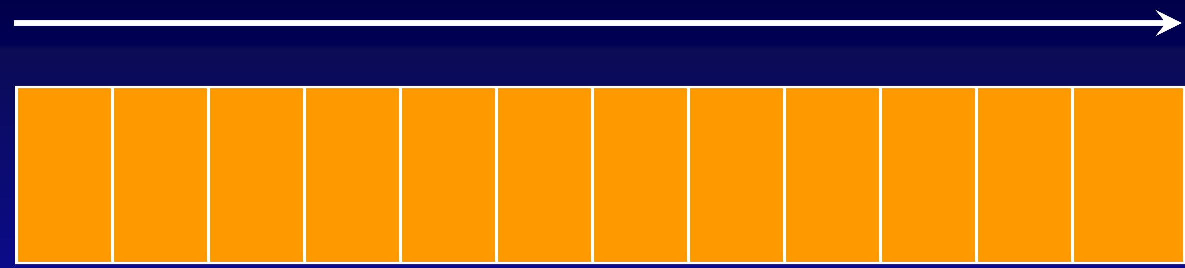
- Each pulse echo is used to construct one line of a 2D B-mode image
- If the B-mode image is made up of n lines, then the time to scan one 2D frame is $n(2R/c) = n/PRF$
- Therefore the maximum Frame Rate is

$$FR_{max} = PRF/n = c/2Rn$$

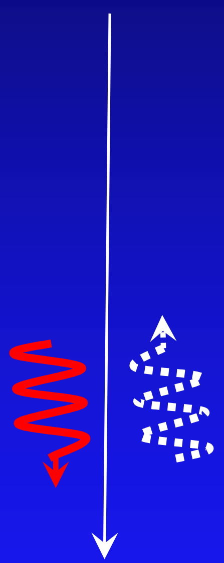
R = range = maximum depth

n = number of scan lines per frame

Frame Rate



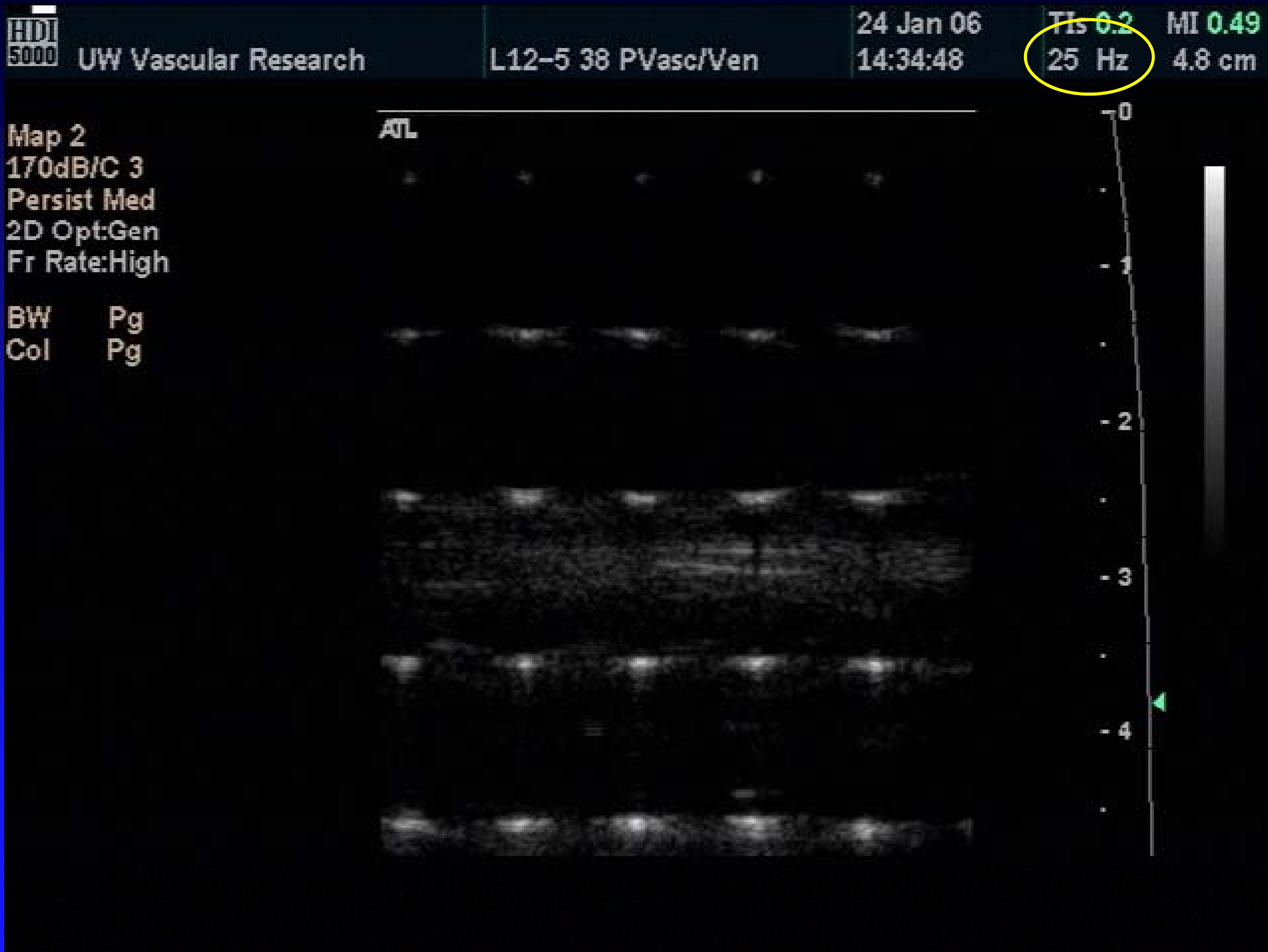
Example



- **1 pulse per scan line**
- **8-cm imaging depth (R)**
- **128 scan lines per frame (n)**

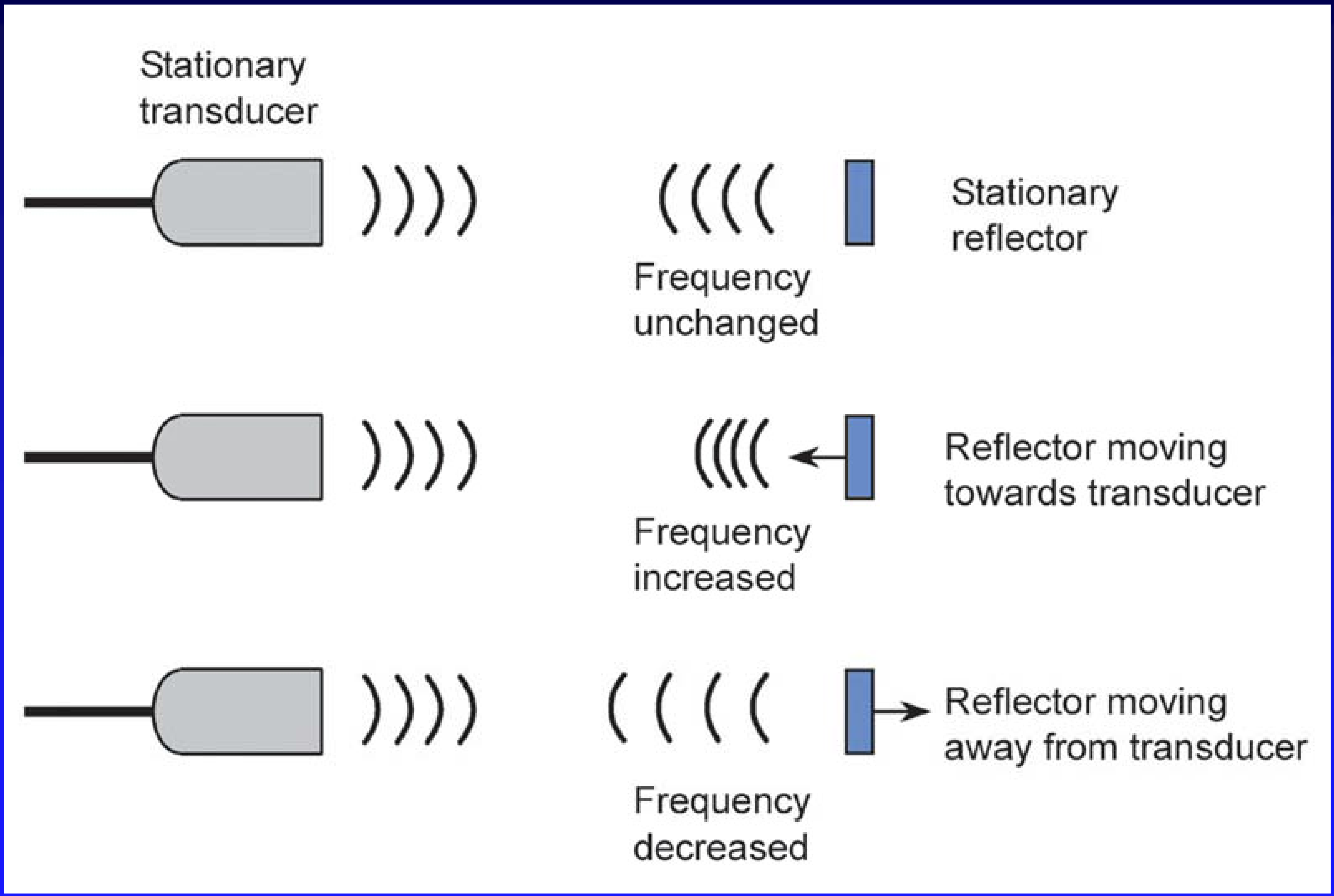
→ maximum PRF = c/2R = 9625 Hz

→ maximum Frame Rate = PRF/n = 75 frames/sec


HDI 5000
UW Vascular Research
L12-5 38 PVasc/Ven
24 Jan 06
14:34:48
TIs 0.2
25 Hz
MI 0.49
4.8 cm
Map 2
170dB/C 3
Persist Med
2D Opt:Gen
Fr Rate:High
BW Pg
Col Pg
ATL
0
1
2
3
4

Doppler Ultrasound

Doppler Imaging



Amersham Medical

The Doppler Shift

- The Doppler shift is the change in the frequency of sound due to motion of the source of the sound or the observer (or both)
- It equals 2 times the transmit frequency multiplied by the velocity and the cosine of the angle of incidence, all divided by the propagation speed of sound in human soft tissue

$$\Delta f = \frac{2\, v\, f_t \cos\theta}{c}$$

v = velocity

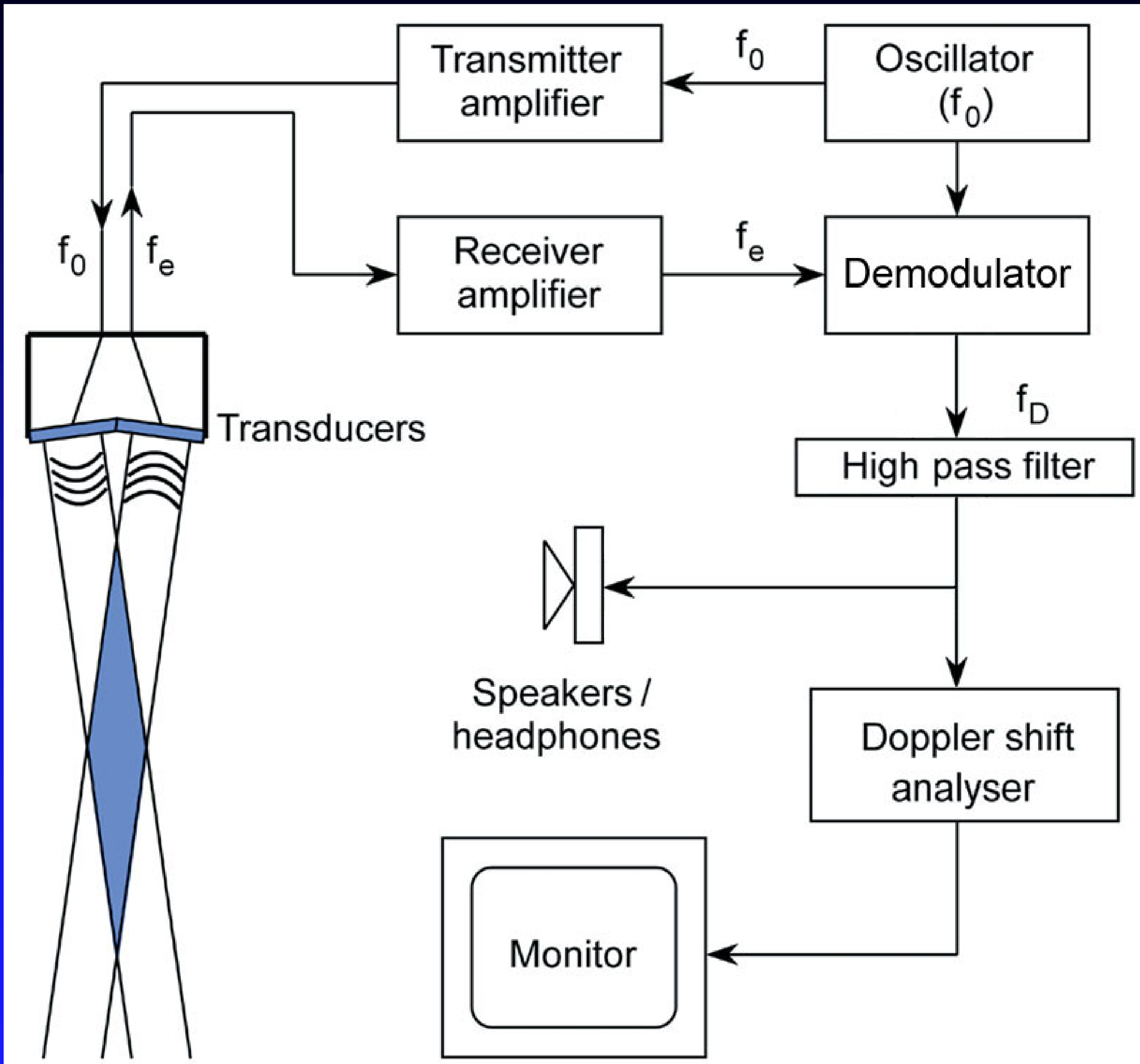
f_t = transmit frequency

θ = angle of insonation

c = speed of sound in human soft tissue

Doppler Imaging

- Visualization of anatomy and blood flow
  - Frequency shifts due to moving scatterers (red blood cells) are measured and displayed
- Color Doppler
  - 2D image showing presence, speed, direction, and character of blood flow
- Spectral Doppler
  - Detailed flow measurement at a single location



Basic Doppler Instrument

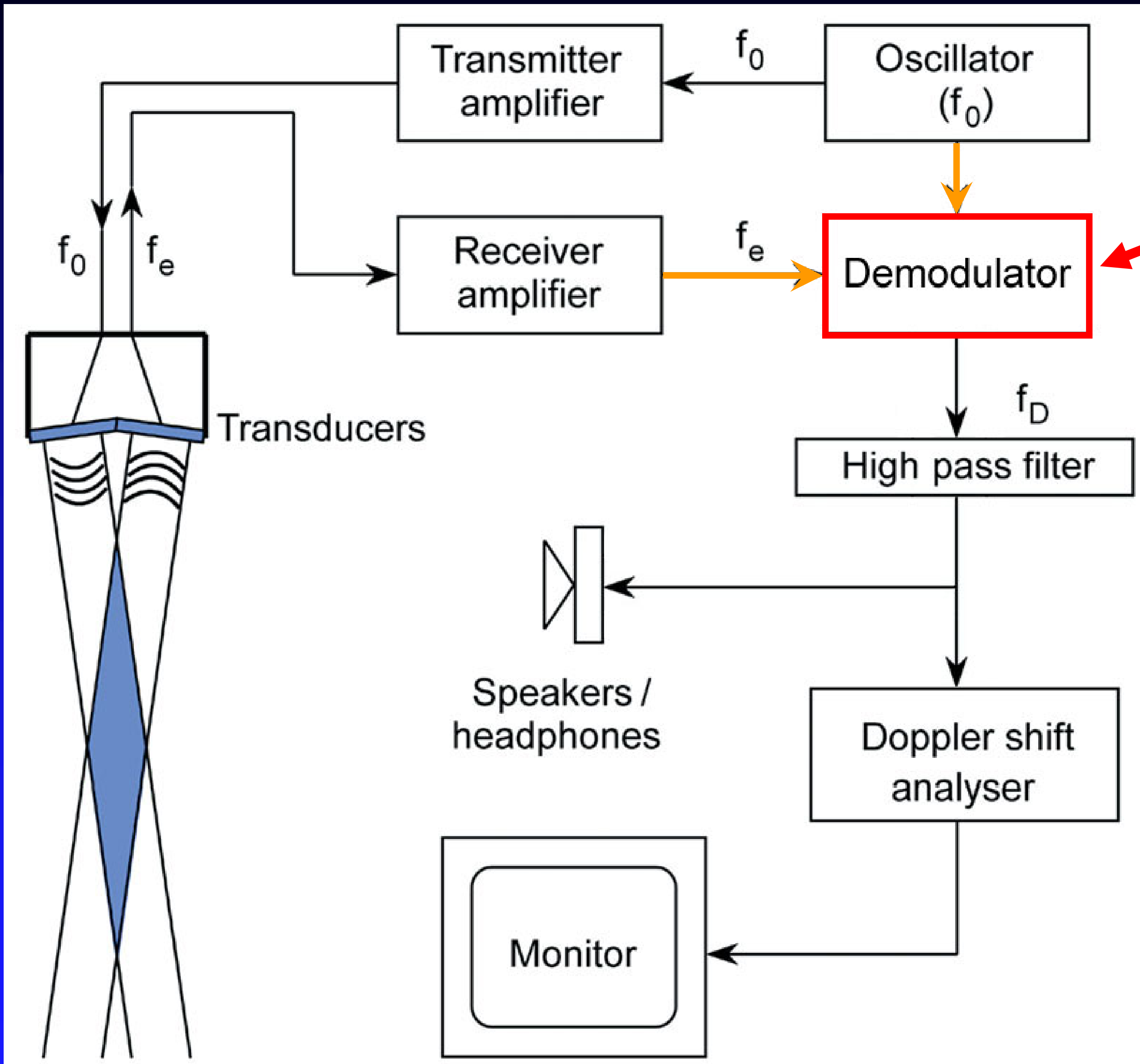
Continuous Wave

f_0 = transmit frequency

f_e = echo frequency

f_D = Doppler frequency

Adapted from Amersham Medical



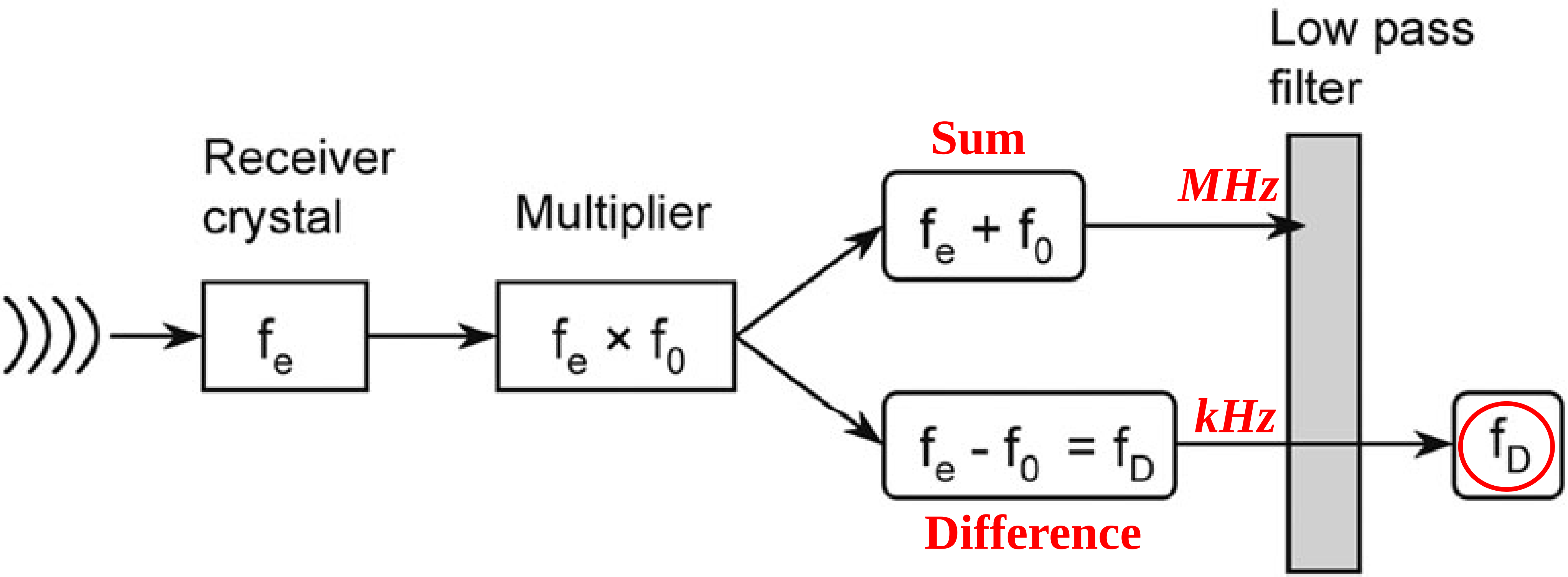
Detect changes in frequency of the reflected signal

Transmit frequency is used as a reference

f_0 = transmit frequency
f_e = echo frequency
f_D = Doppler frequency

Adapted from Amersham Medical

Demodulation



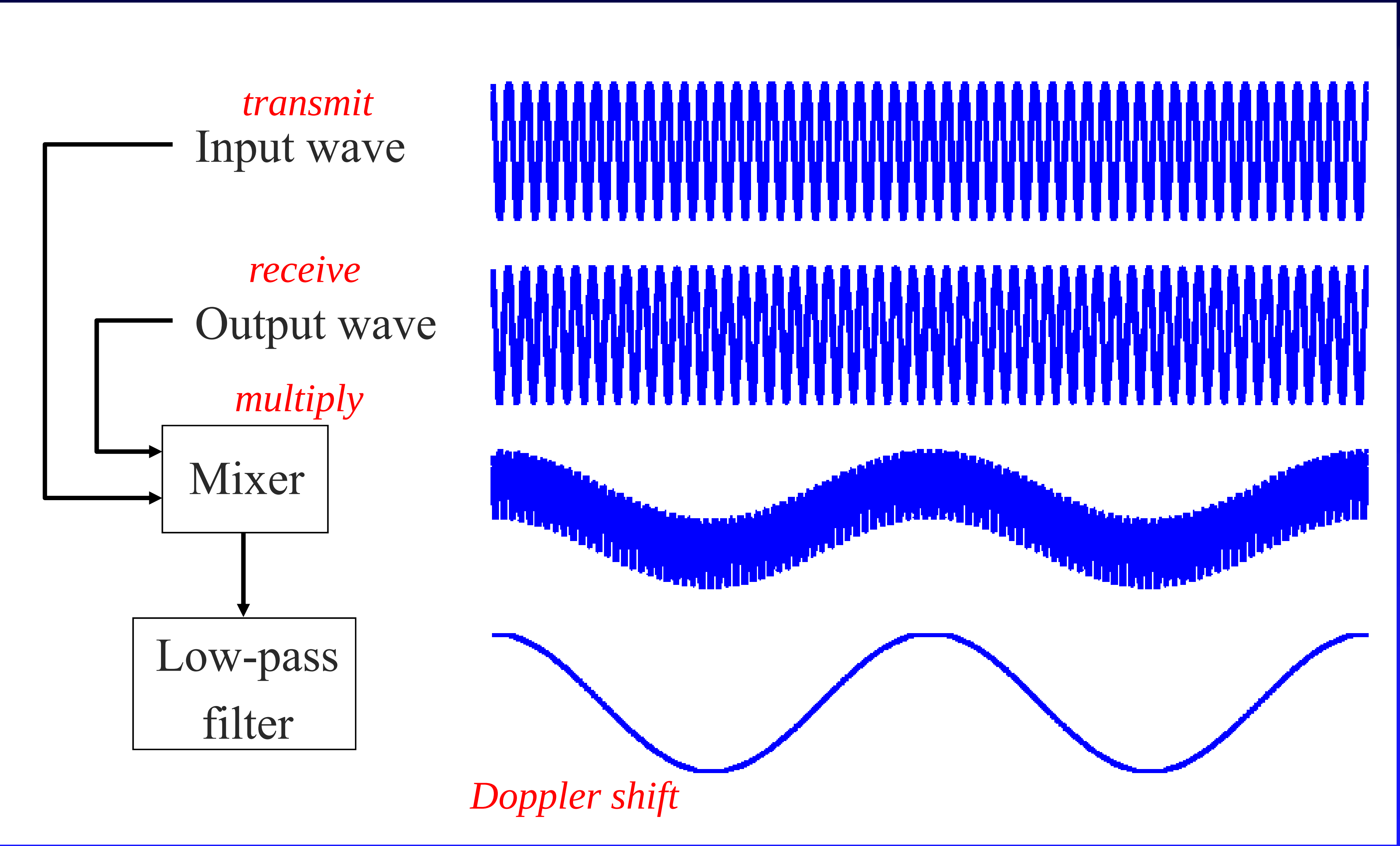
f_0 = transmit frequency

f_e = echo frequency (received)

f_D = Doppler frequency

Amersham Medical

Demodulation



Doppler Imaging

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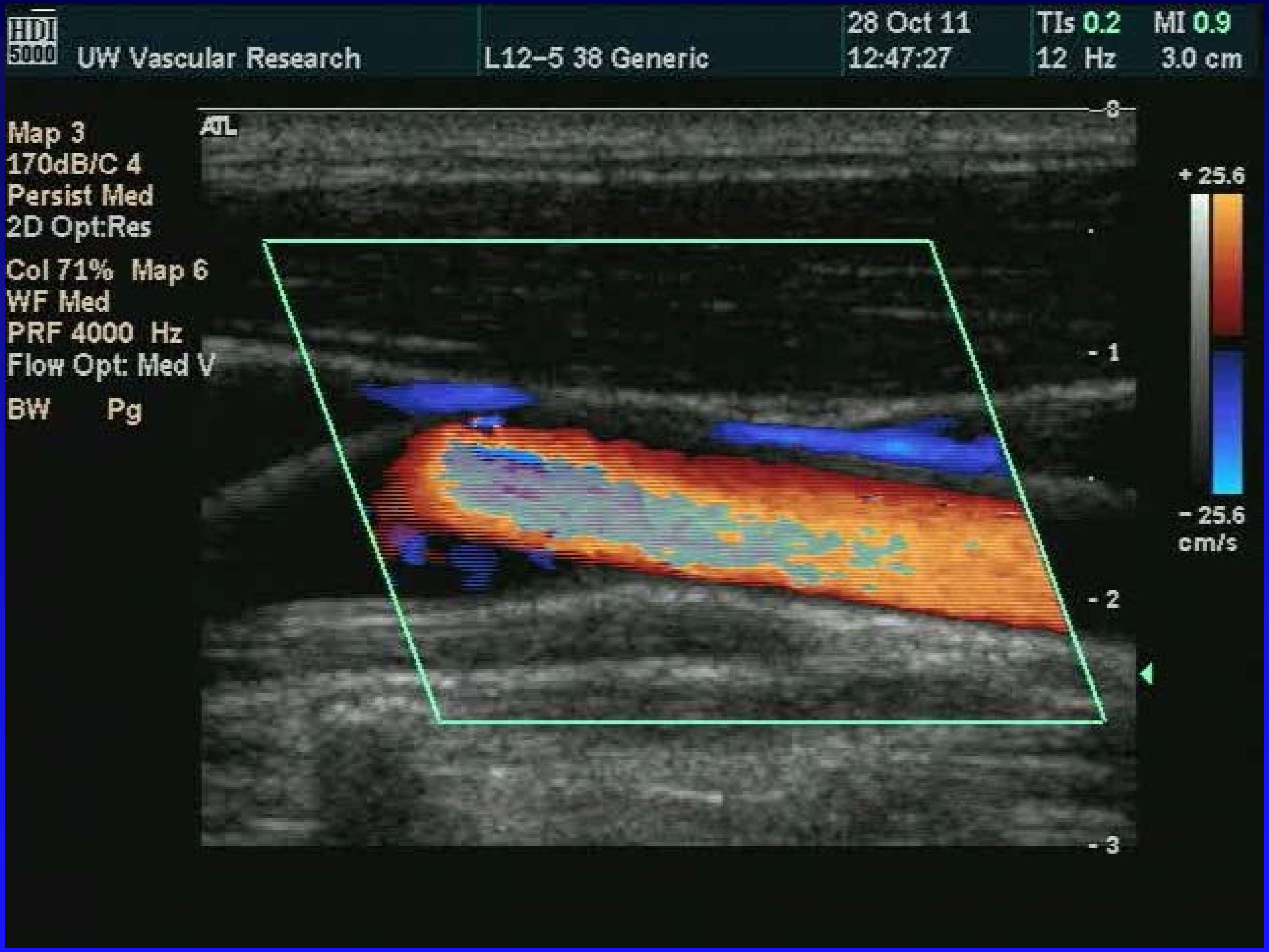
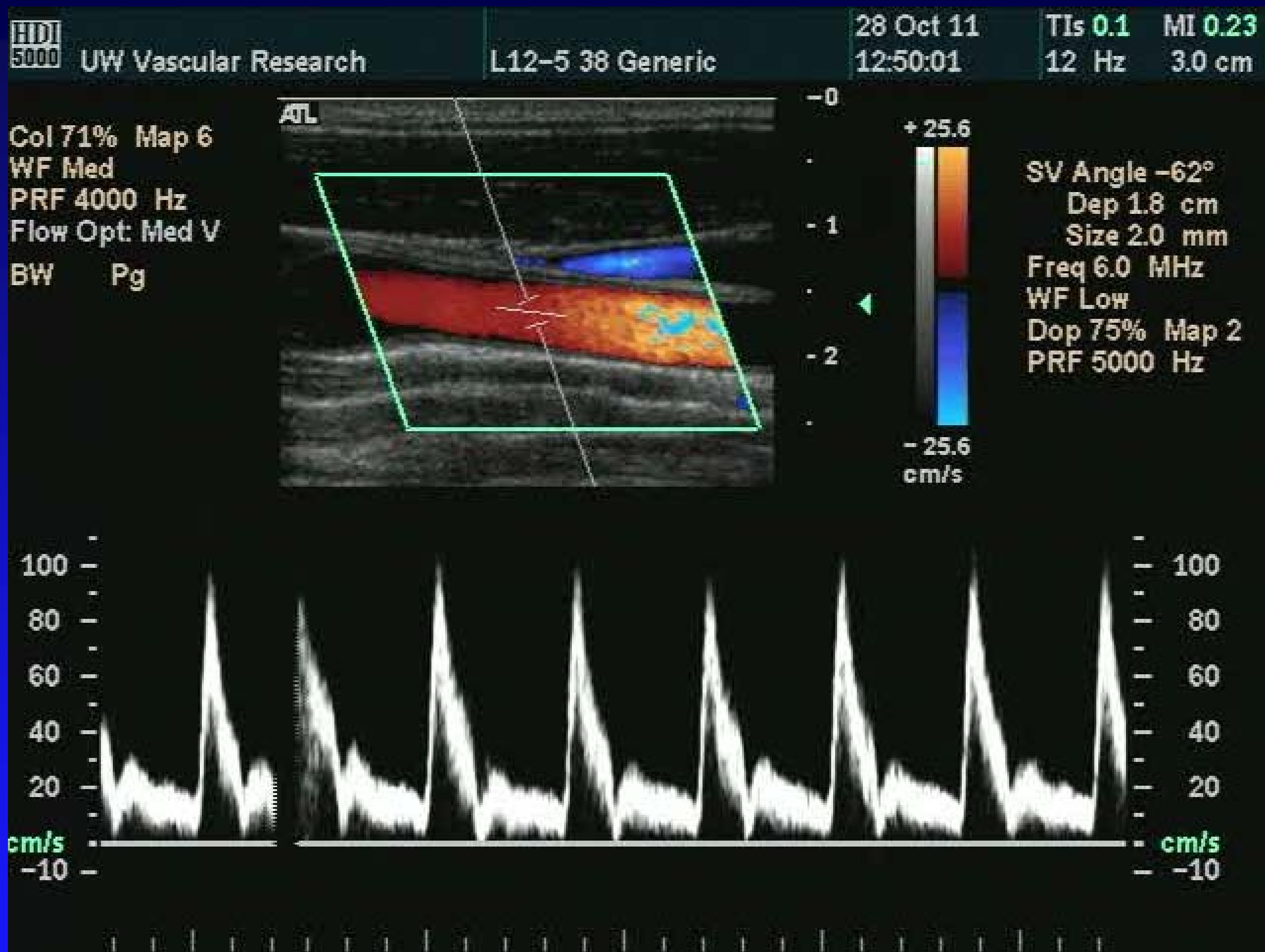


Color Doppler

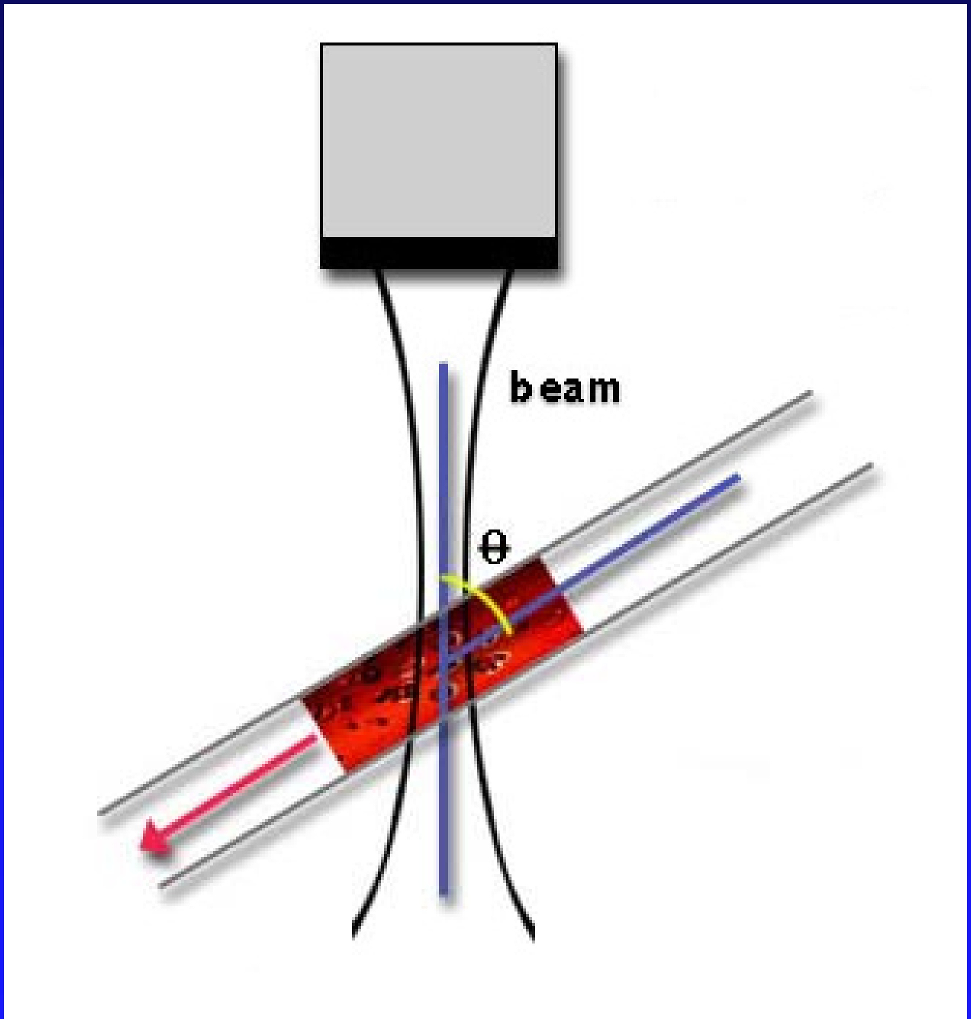


Spectral Doppler

Doppler Angle

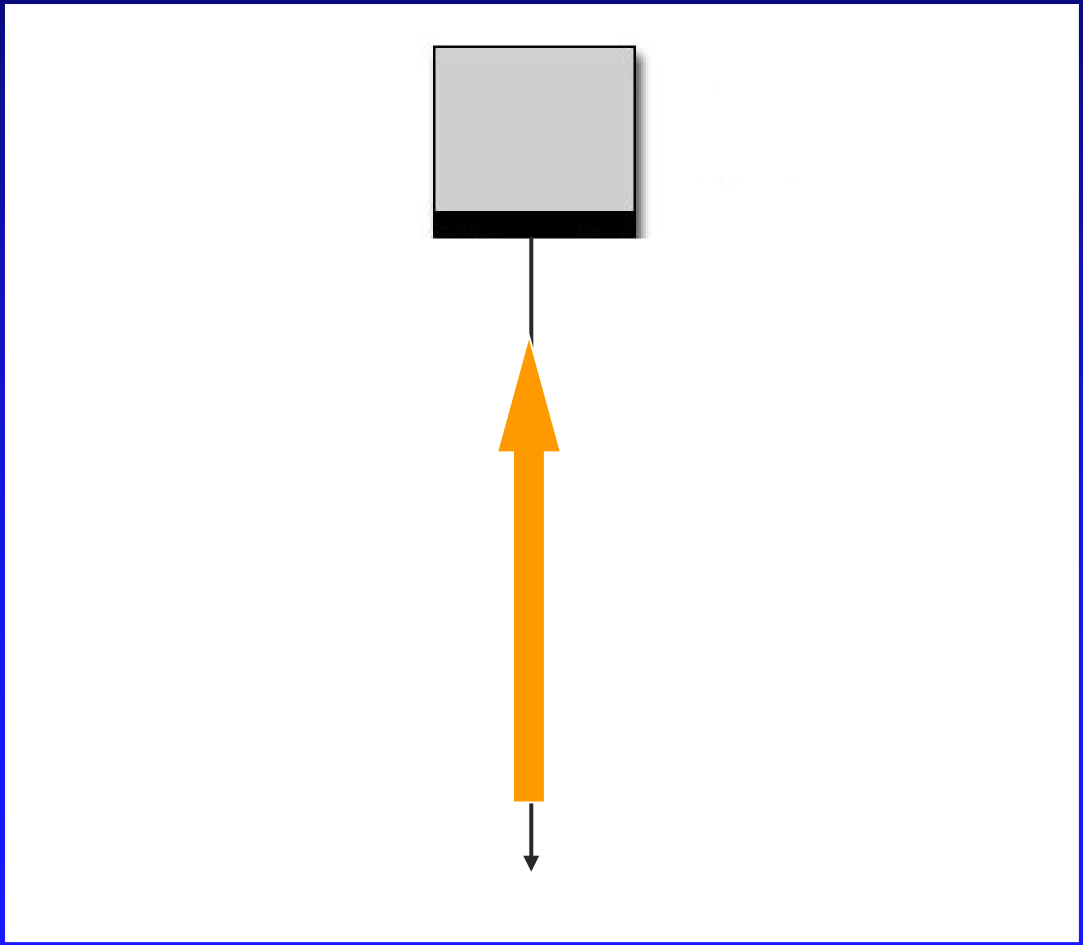
- **Angle between sound travel & flow**
- **0 degrees**
  - **flow in direction of sound travel**
- **90 degrees**
  - **flow perpendicular to sound travel**
  - **no Doppler shift**

$$\Delta f = \frac{2\, v\, f_t \cos\theta}{c}$$



Doppler Angle

$$\Delta f = \frac{2\, v\, f_t \cos\theta}{c}$$

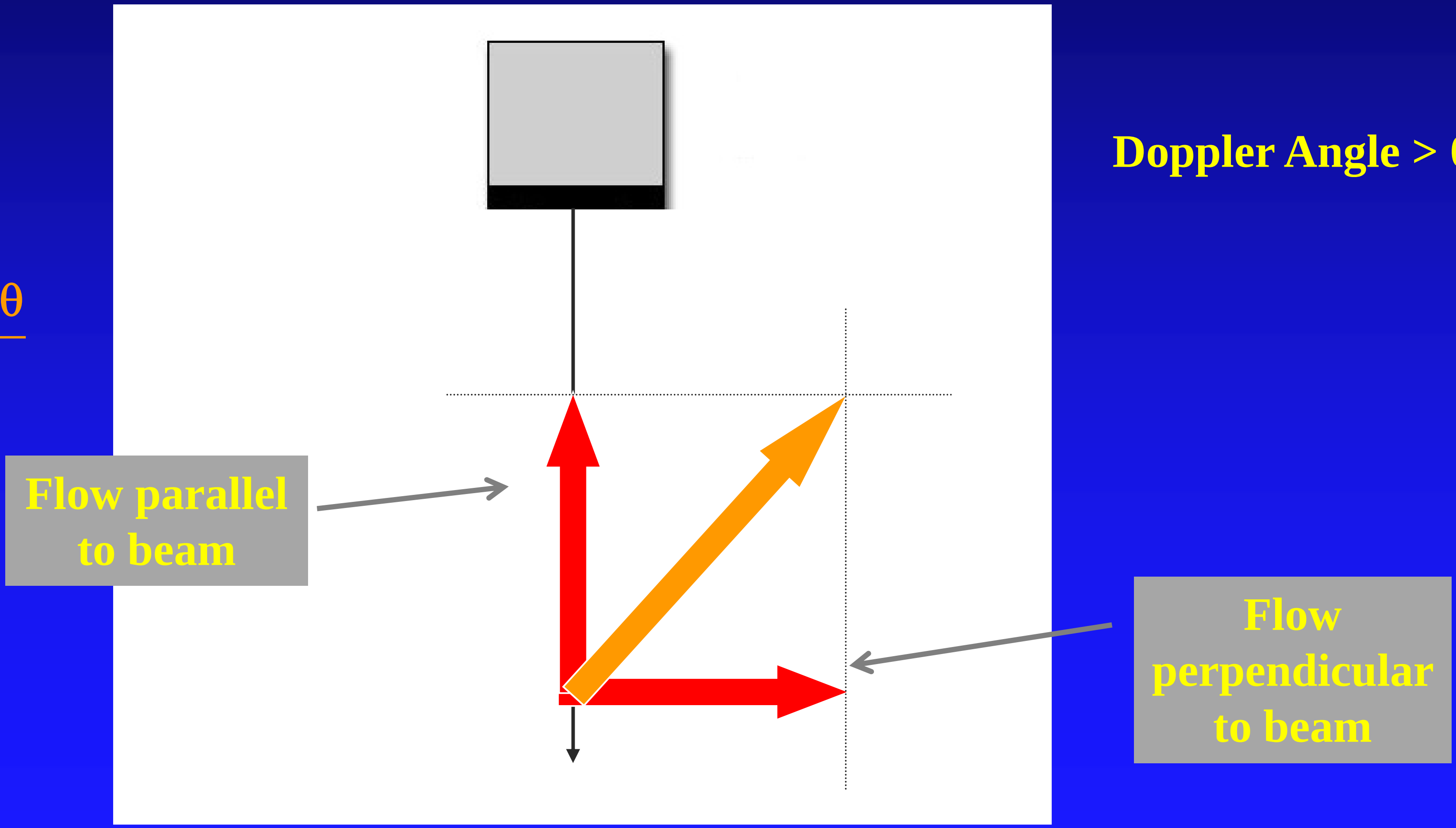


Doppler Angle = 0

Doppler Angle

- Flow vector can be separated into two vectors

$$\Delta f = \frac{2\, v\, f_t \cos\theta}{c}$$

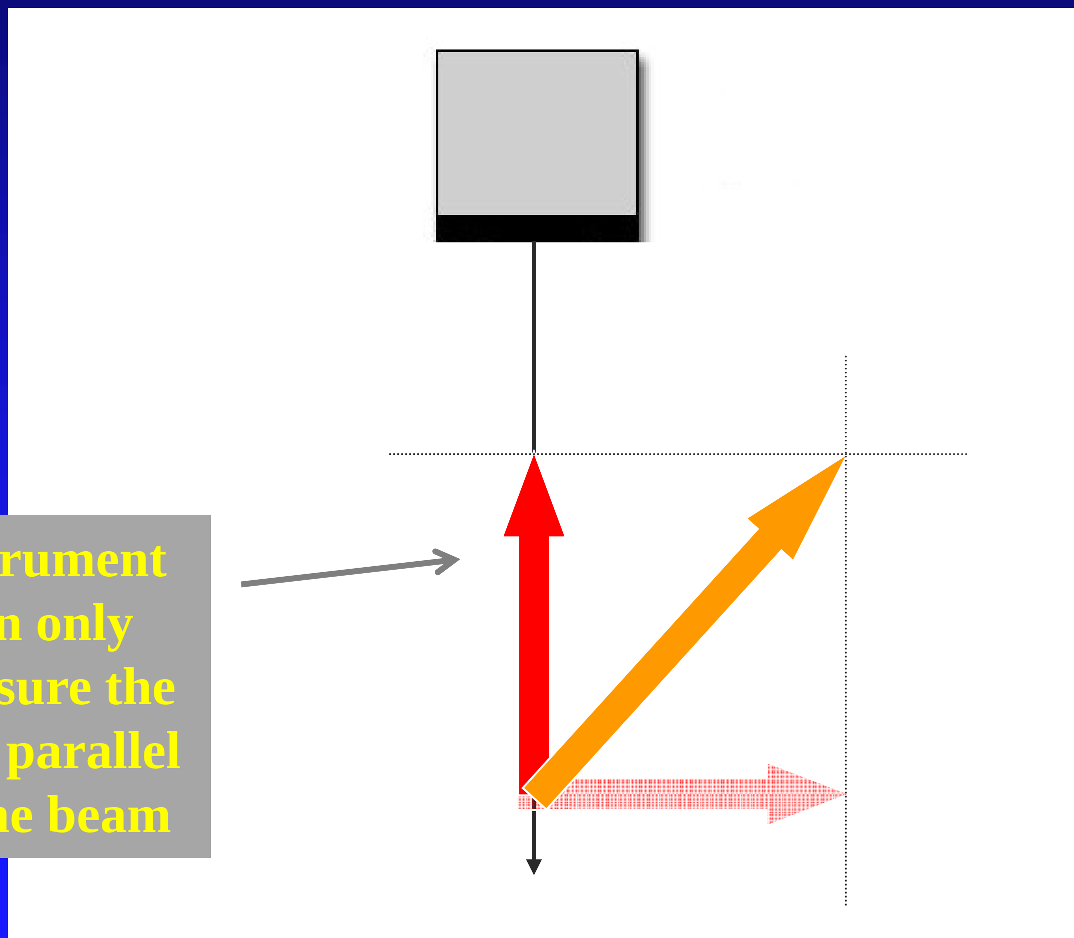


Doppler Angle > 0

Doppler Angle

- **Flow vector can be separated into two vectors**

$$\Delta f = \frac{2\, v\, f_t \cos\theta}{c}$$



Instrument can only measure the flow parallel to the beam

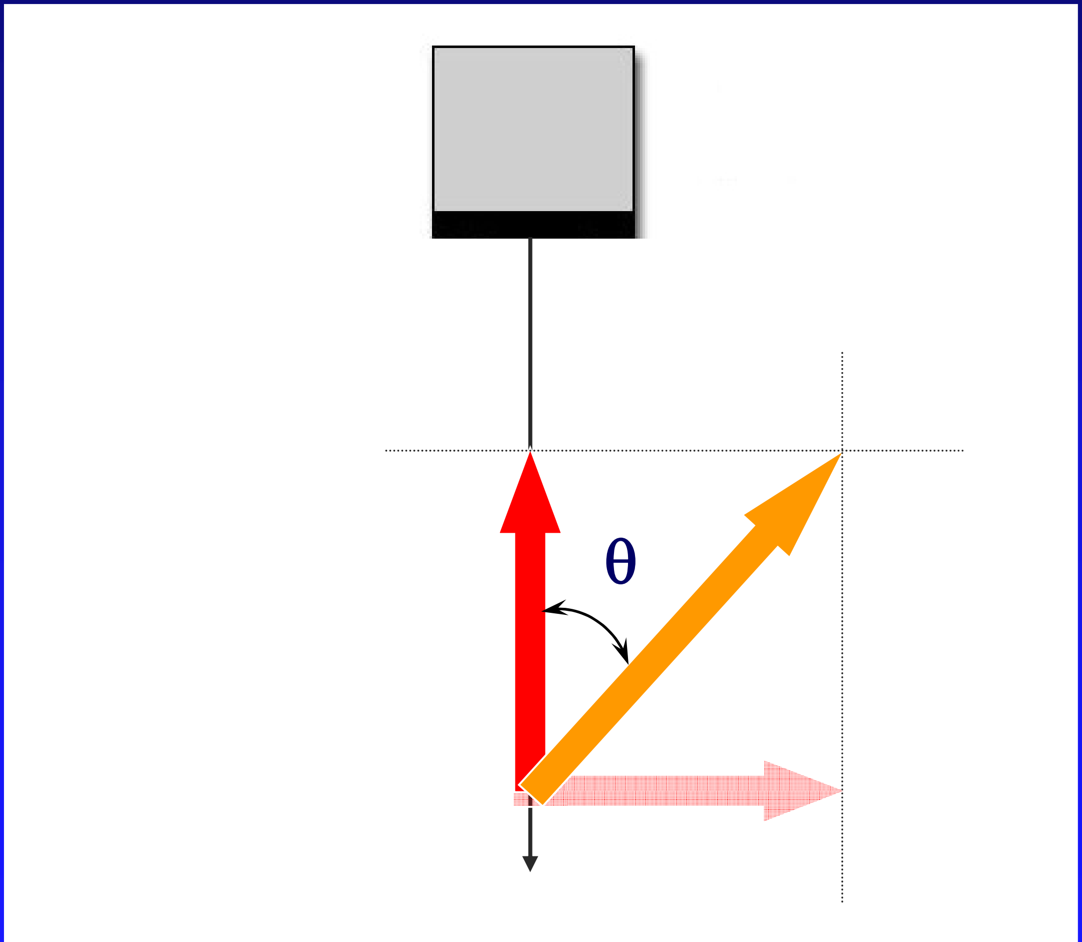
Doppler Angle > 0

Doppler Angle

- Flow vector can be separated into two vectors

$$\Delta f = \frac{2\,v\,f_t \cos\theta}{c}$$

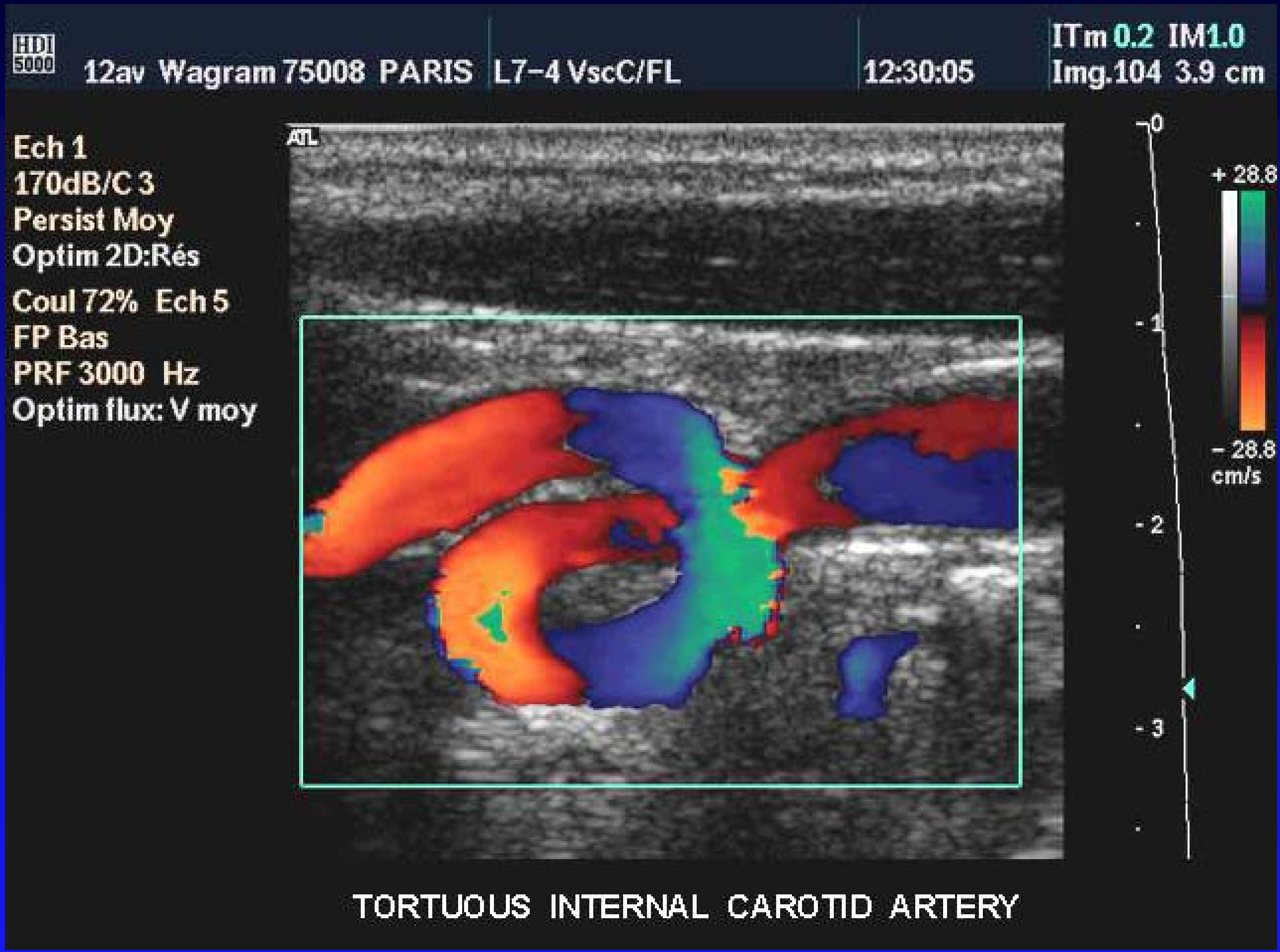
$$v = \frac{c\,\Delta f}{2\,ft\cos\theta}$$



Doppler Angle > 0

- Sensed flow always ≤ actual flow
- Cosine corrects for angle in the velocity calculation

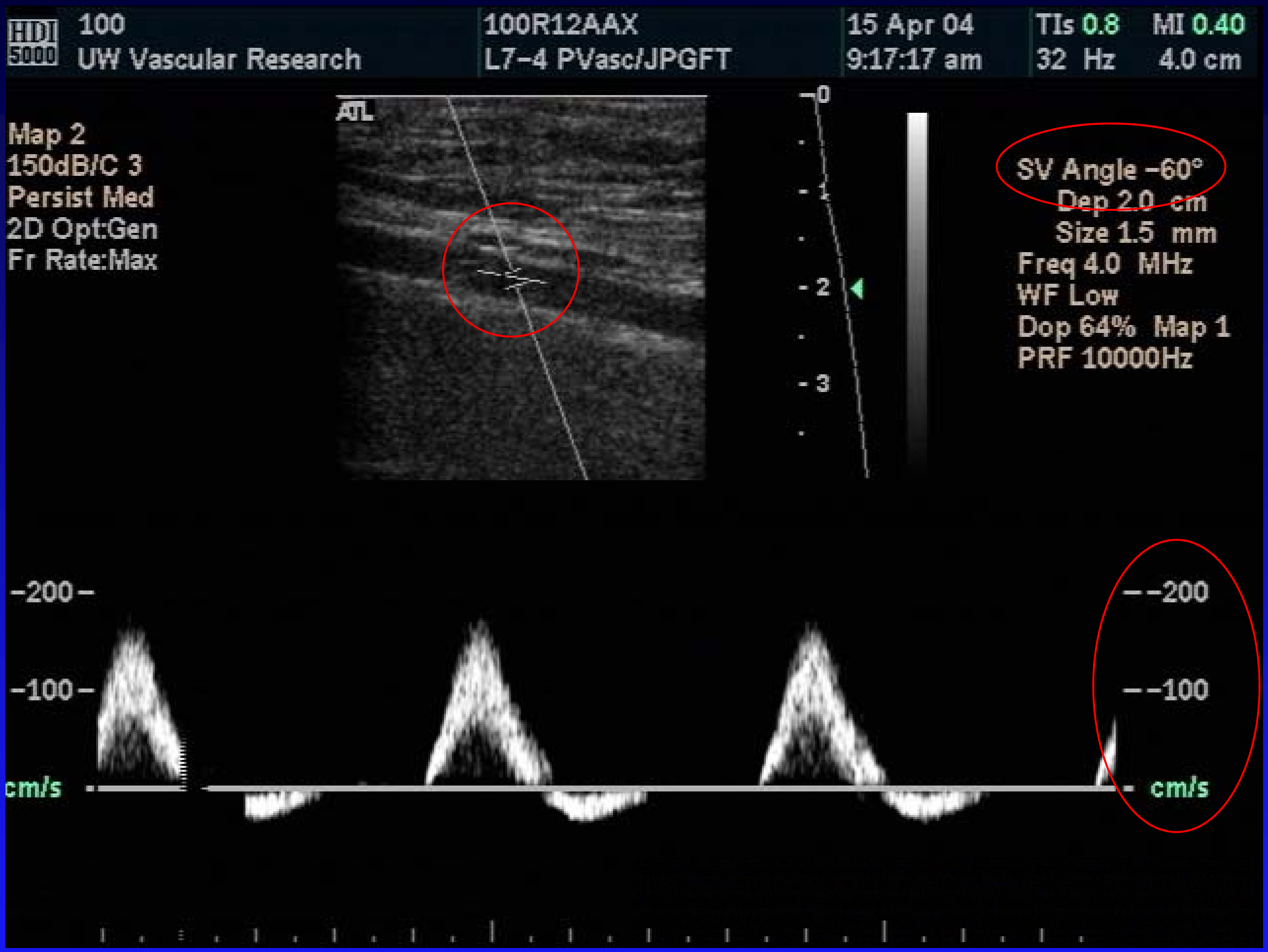
Color Doppler



Color flow systems represent the velocity in each pixel as a single value represented by a color selected from a color scale

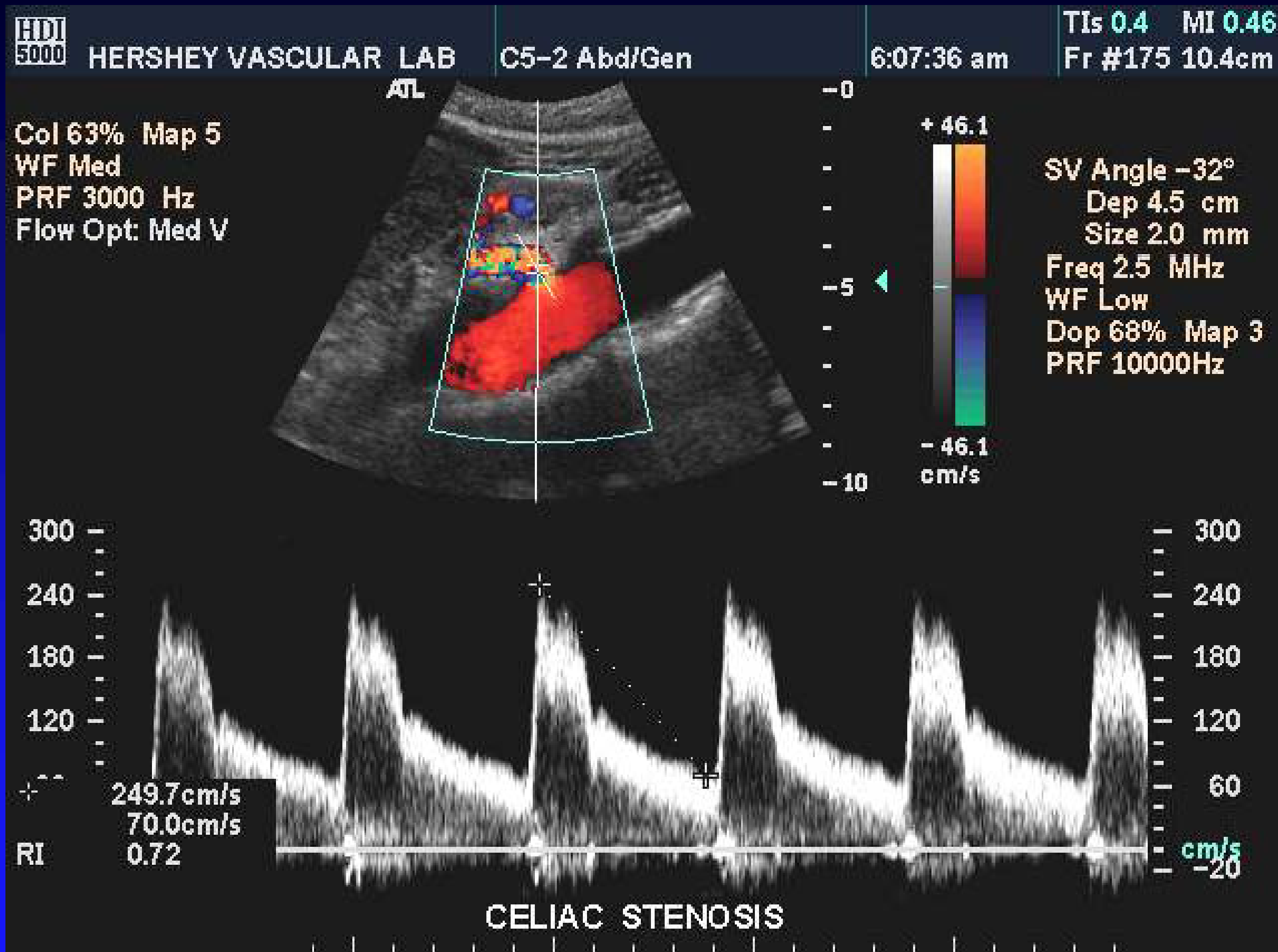
Note: velocity estimate changes with angle

Pulse Wave (Spectral) Doppler



Pulse wave systems measure the Doppler frequency spectrum at a specified depth and display the calculated velocities as a function of time

Angle estimate provided by operator

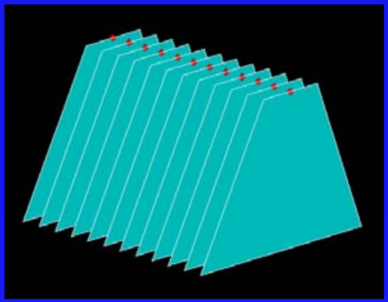


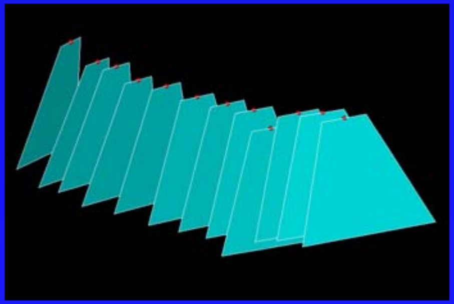
Color Doppler guidance to sites of interest for Spectral Doppler

<u>Three-Dimensional Ultrasound Imaging</u>

3D Ultrasound

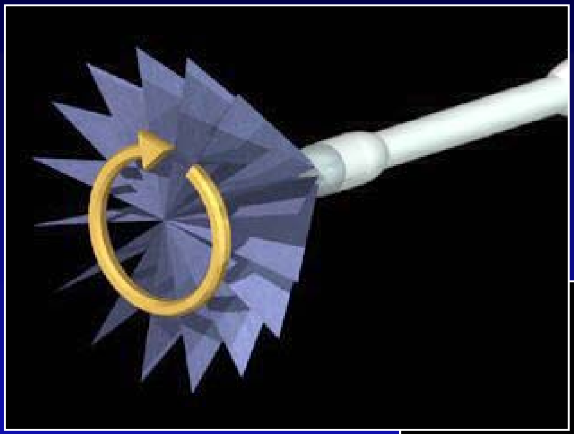
- Limitations of 2D ultrasound
  - 2D slices through a 3D structure
  - spatial relationships (between images and studies) are not preserved
- Benefits of 3D ultrasound
  - robust displays enhance interpretation
  - measurements require fewer geometric assumptions
- Acquisition methods
  - relate multiple 2D images in a 3D coordinate system
  - capture data in a 3D volume

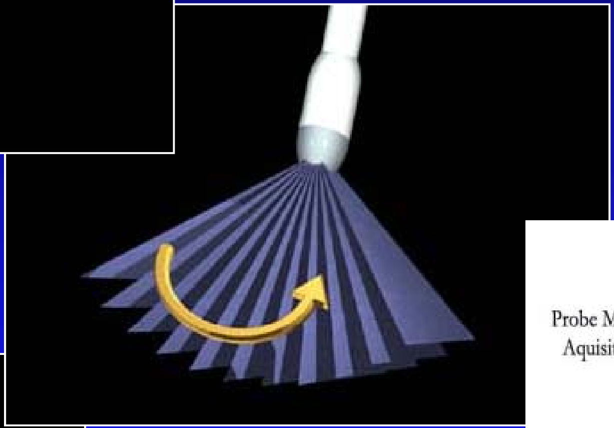




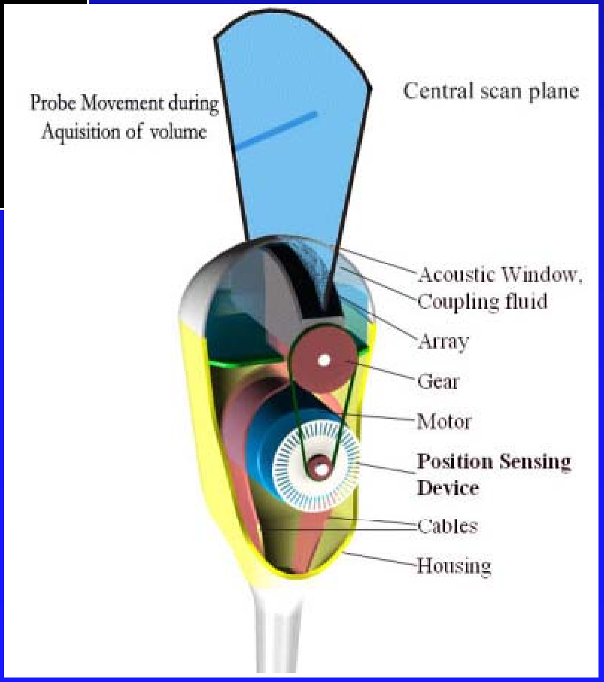
3D Ultrasound Methods: Mechanical

Medison





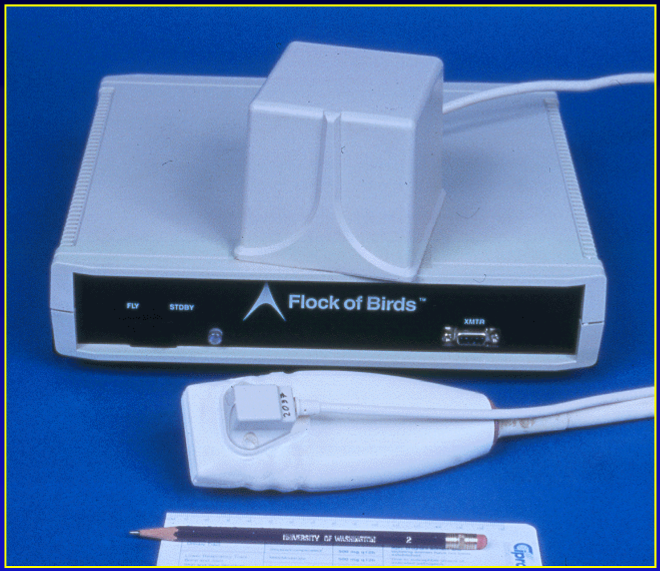
Mechanical scans



GE

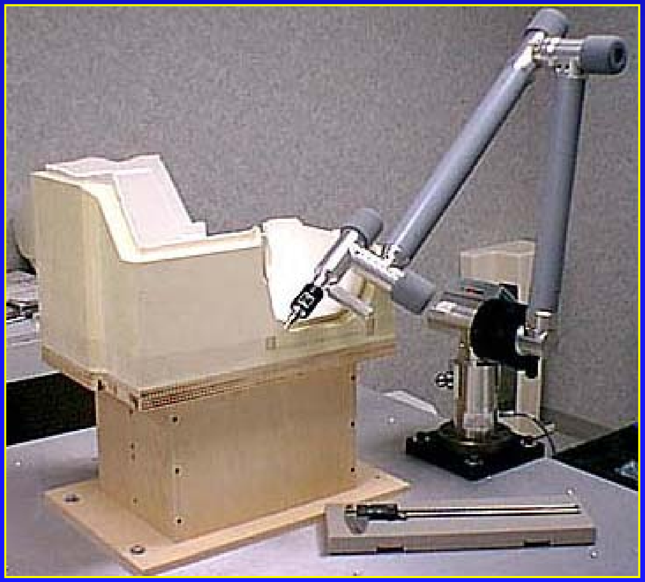
3D Ultrasound Methods: Freehand

Freehand systems



Magnetic
Ascension Technology

Optical
Image Guided Technologies

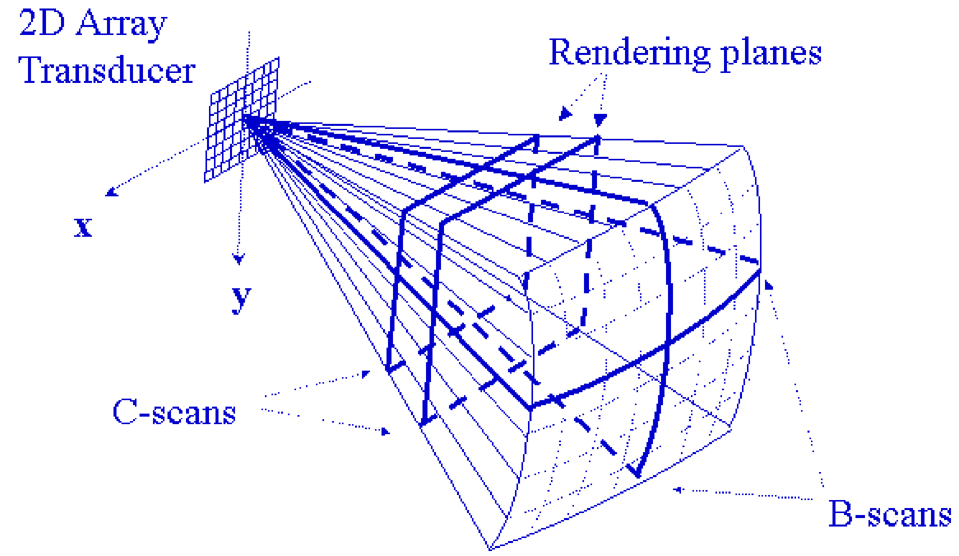


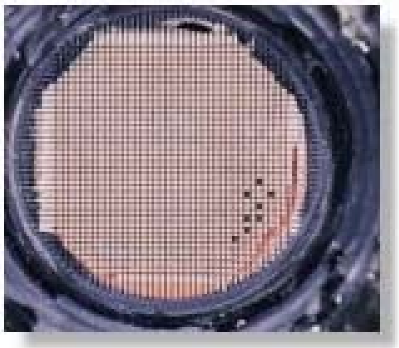
Articulated arm
FARO

3D Ultrasound Methods: Volume Scan

2D array transducer

- **N x N arrays are used to steer the ultrasound beam in both the azimuth and elevation directions**
- **Interrogate a pyramidal-shaped region and produce a volumetric image at high speeds without moving the transducer**
- **Recently-developed transducers include a 64 x 64 = 4096 element array operating at 3.5 MHz**



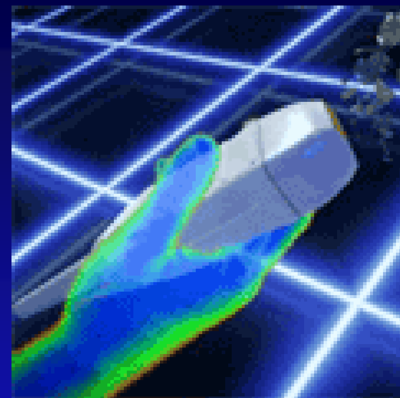


2-D matrix-array at Duke

Stennet, von Ramm
Carnegie Mellon / Duke

3D Ultrasound Methods: Volume Scan

2D array transducer



Tricuspid / Mitral

- **N x N arrays are used to steer the ultrasound beam in both the azimuth and elevation directions**

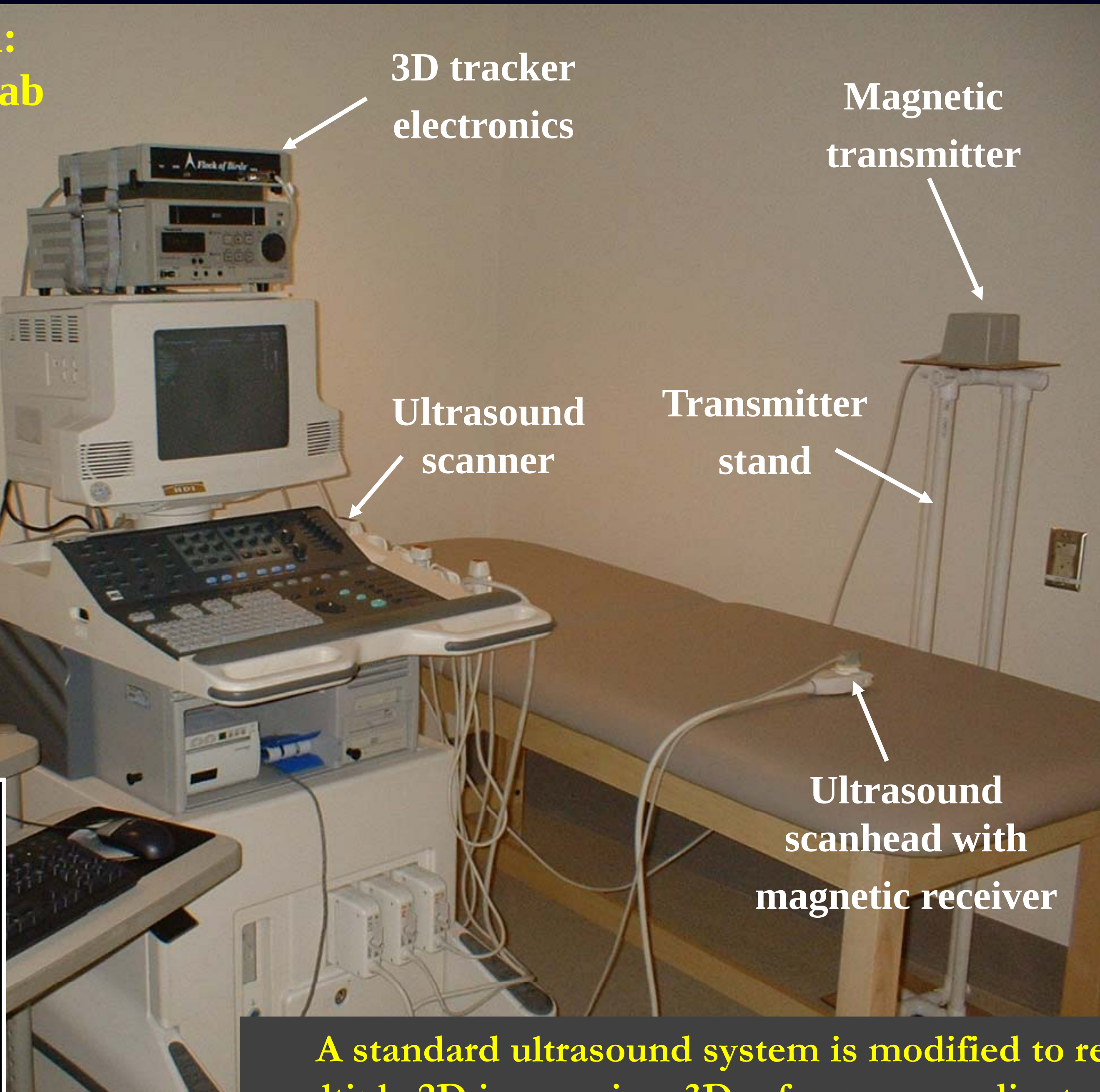
- **Interrogate a pyramidal-shaped region and produce a volumetric image at high speeds without moving the transducer**

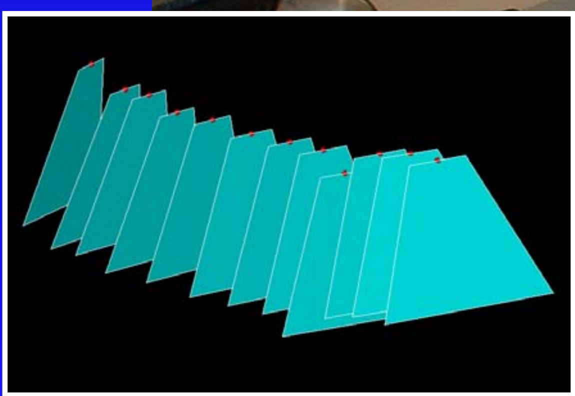
- **Recently-developed transducers include a 64 x 64 = 4096 element array operating at 3.5 MHz**

Aortic / Tricuspid / Mitral

Philips Sonos 7500

3D Ultrasound: UW Research Lab

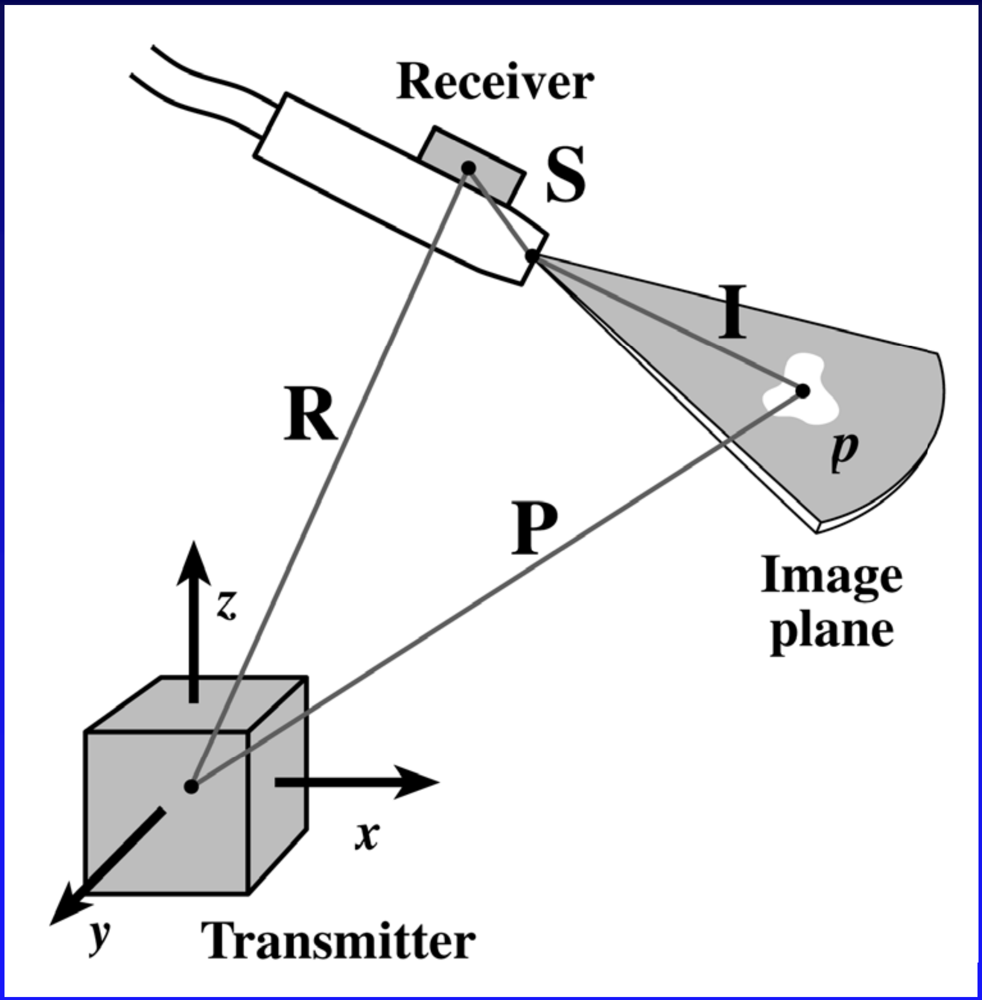




Arbitrary image position and orientation

A standard ultrasound system is modified to relate multiple 2D images in a 3D reference coordinate system

3D Ultrasound: Scanhead Tracking



- Relate pixel locations in ultrasound image with points in the 3D reference coordinate system
  - R: tracking output
  - S: calibration
  - I: in-plane pixel location
  - P: 3D pixel location

Volume Reconstruction

- Insert 2D image data into a regular 3D grid
- No manual interaction required
- Variety of display options

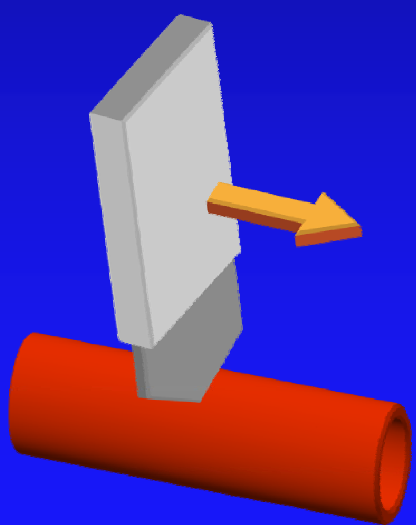


Image acquisition

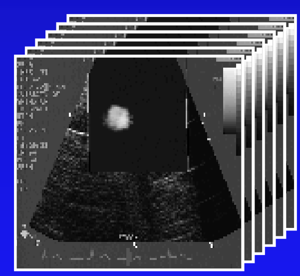
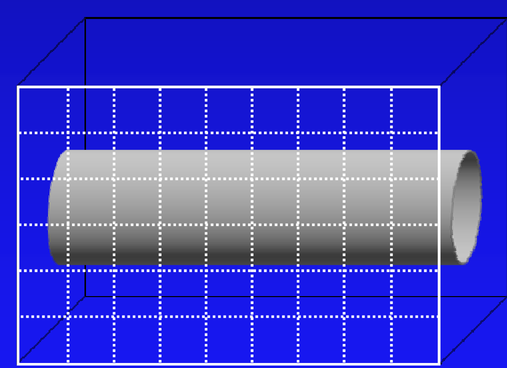
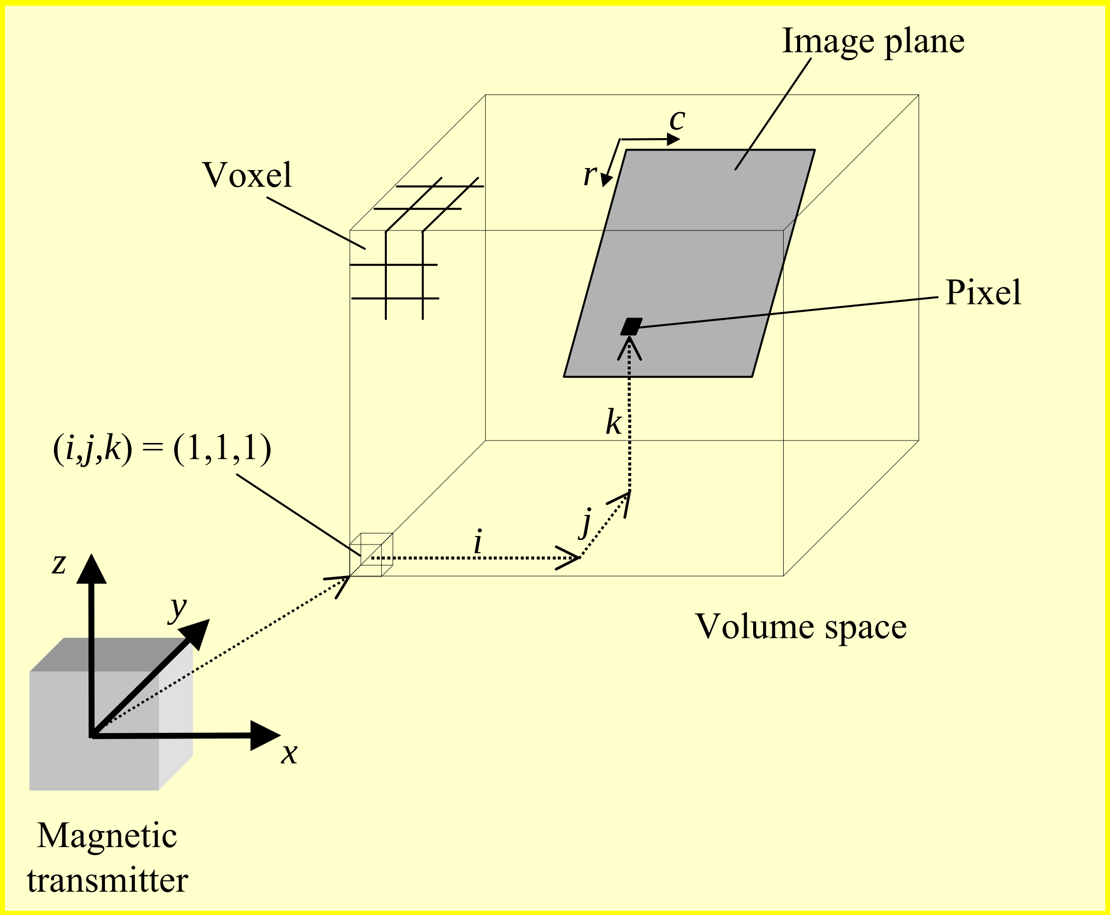


Image stack



3D volume

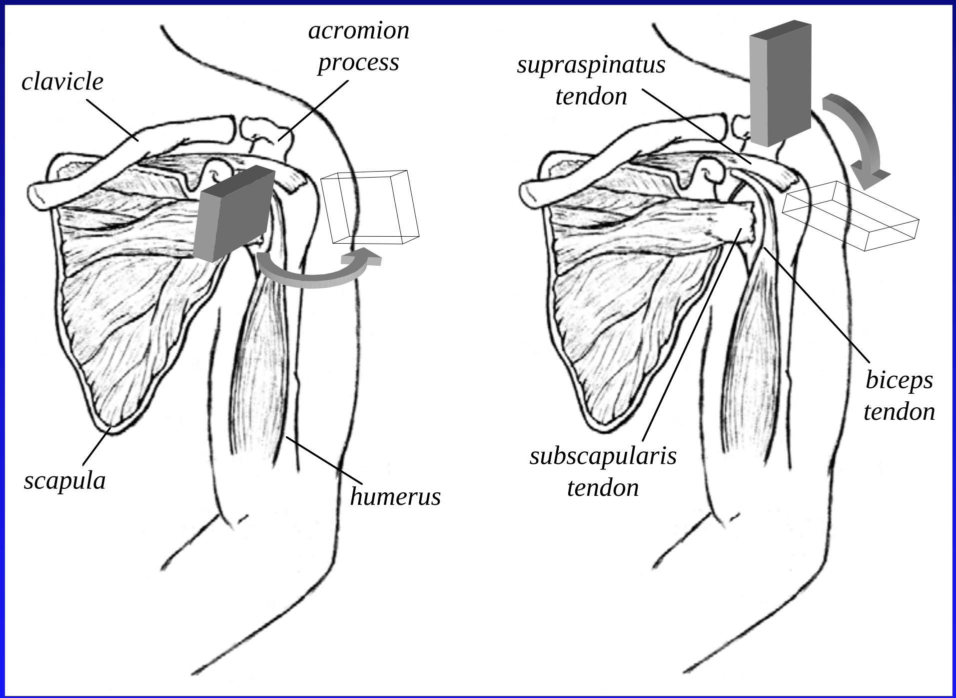
Image-to-Volume Processing



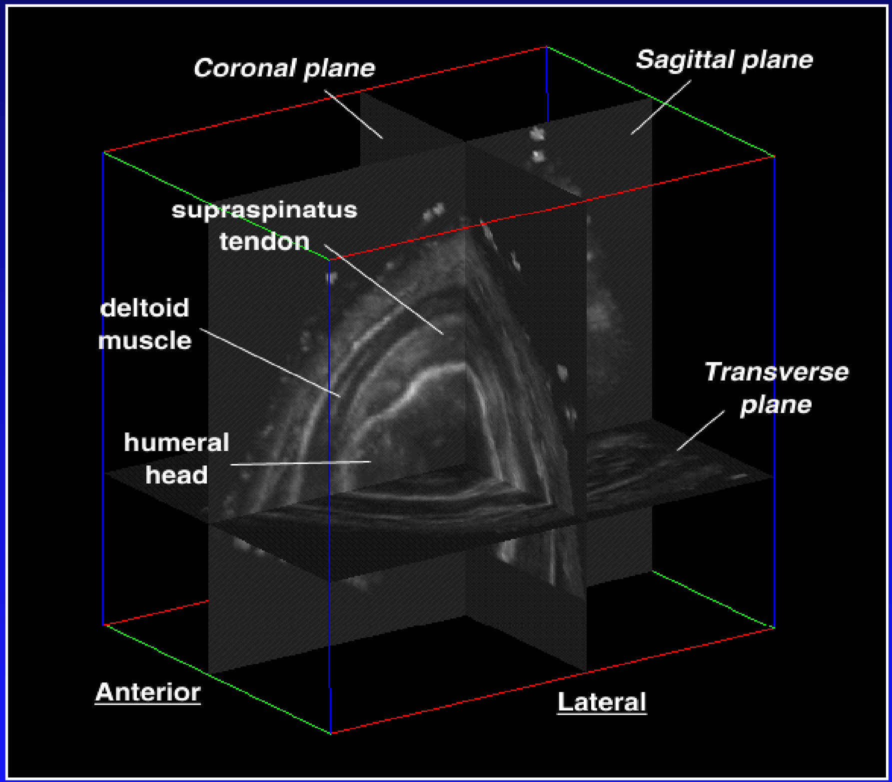
- Calculate volume bounds in transmitter coordinate system based on all images
- Specify voxel size in mm
- Calculate 3D location of pixel
- Calculate voxel number corresponding to pixel's 3D location
- Insert pixel value in voxel

Coordinate systems
- (*x*,*y*,*z*): magnetic transmitter
- (*r*,*c*): image
- (*i*,*j*,*k*): reconstructed volume

Shoulder Rotator Cuff



Images acquired from multiple windows



3D volume reconstruction

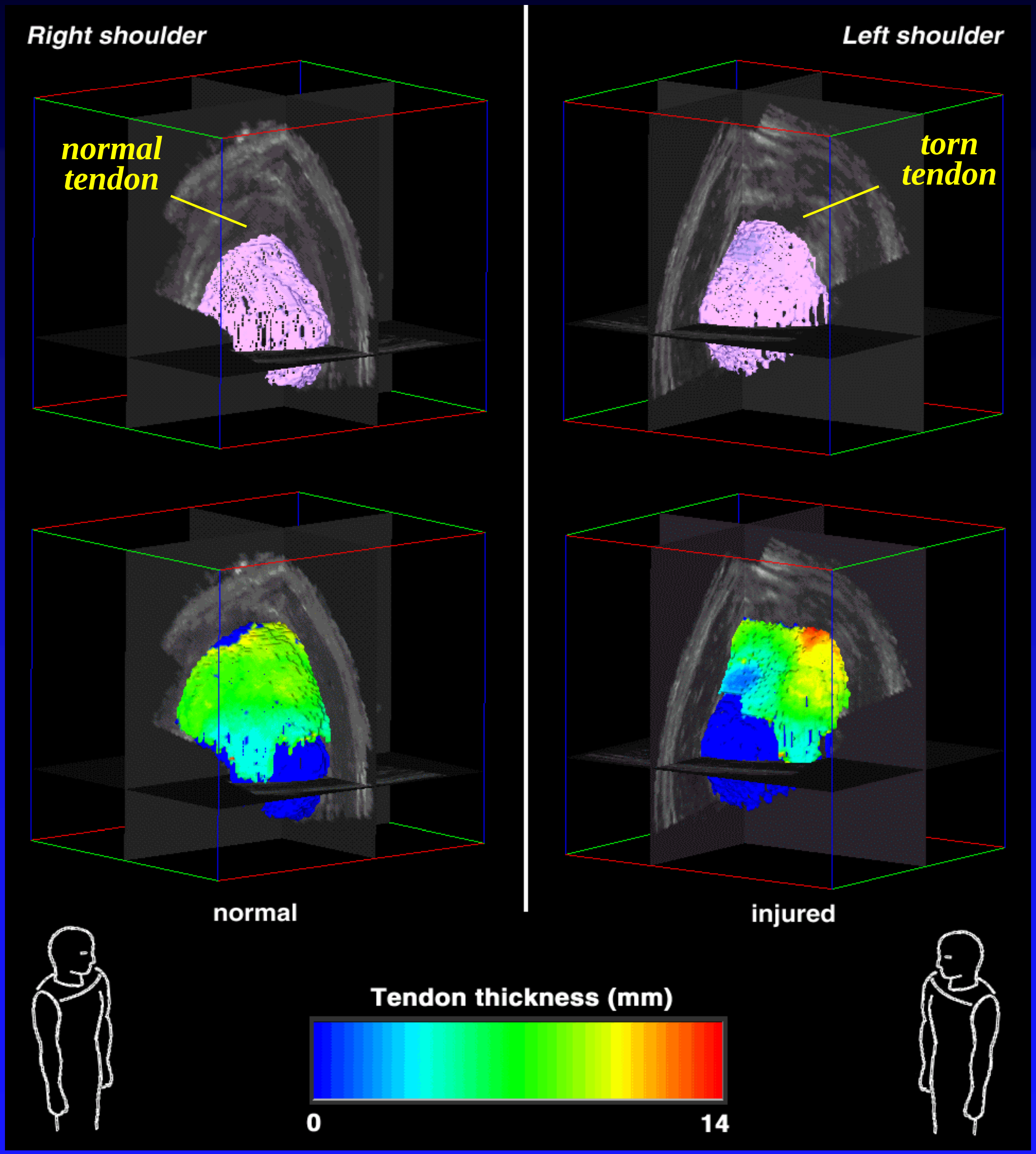
Rotator Cuff Thickness

• Tendon thickness measurements show changes in morphology in a subject with an acute tear of the left supraspinatus tendon. In particular note

– the nearly uniform thickness of the normal tendon (≈ 7 mm)

– the absence of the tendon on the anterior side of the bone in the injured shoulder

– a pronounced bulge of the retracted end of the torn tendon (thickness ≈ 14 mm)



Leotta and Martin
UMB, 26:509, 2000

Surface Reconstruction

- The vessel is traced on each image
  - manual or semi-automated
- Outline points are connected to create a surface for visualization and measurement

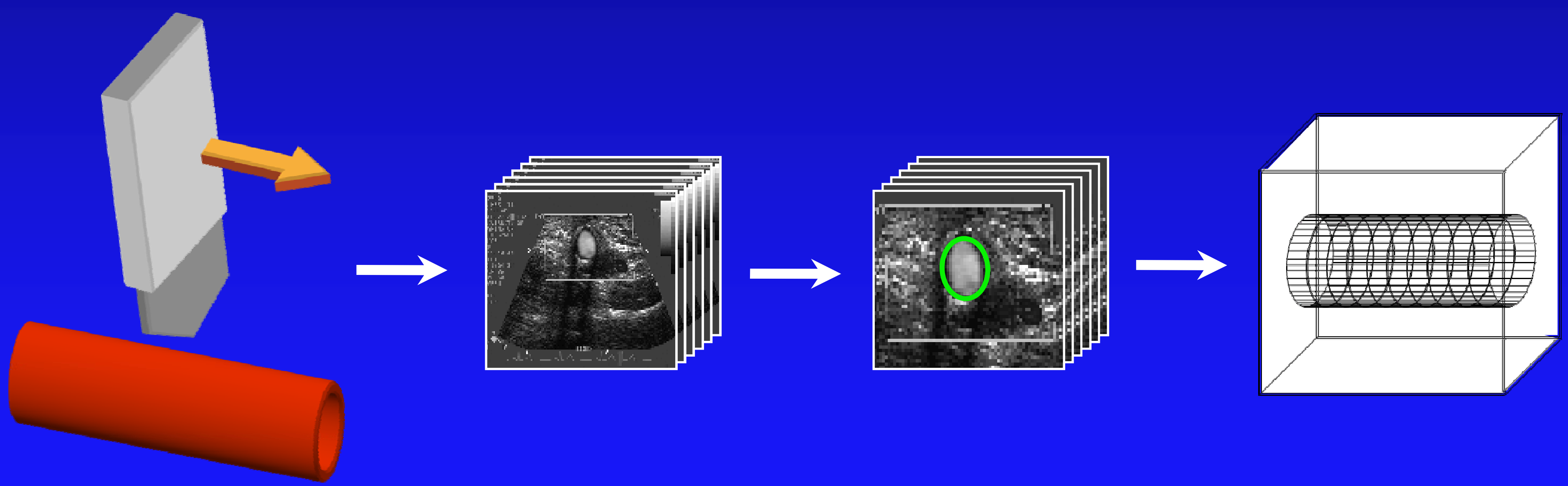
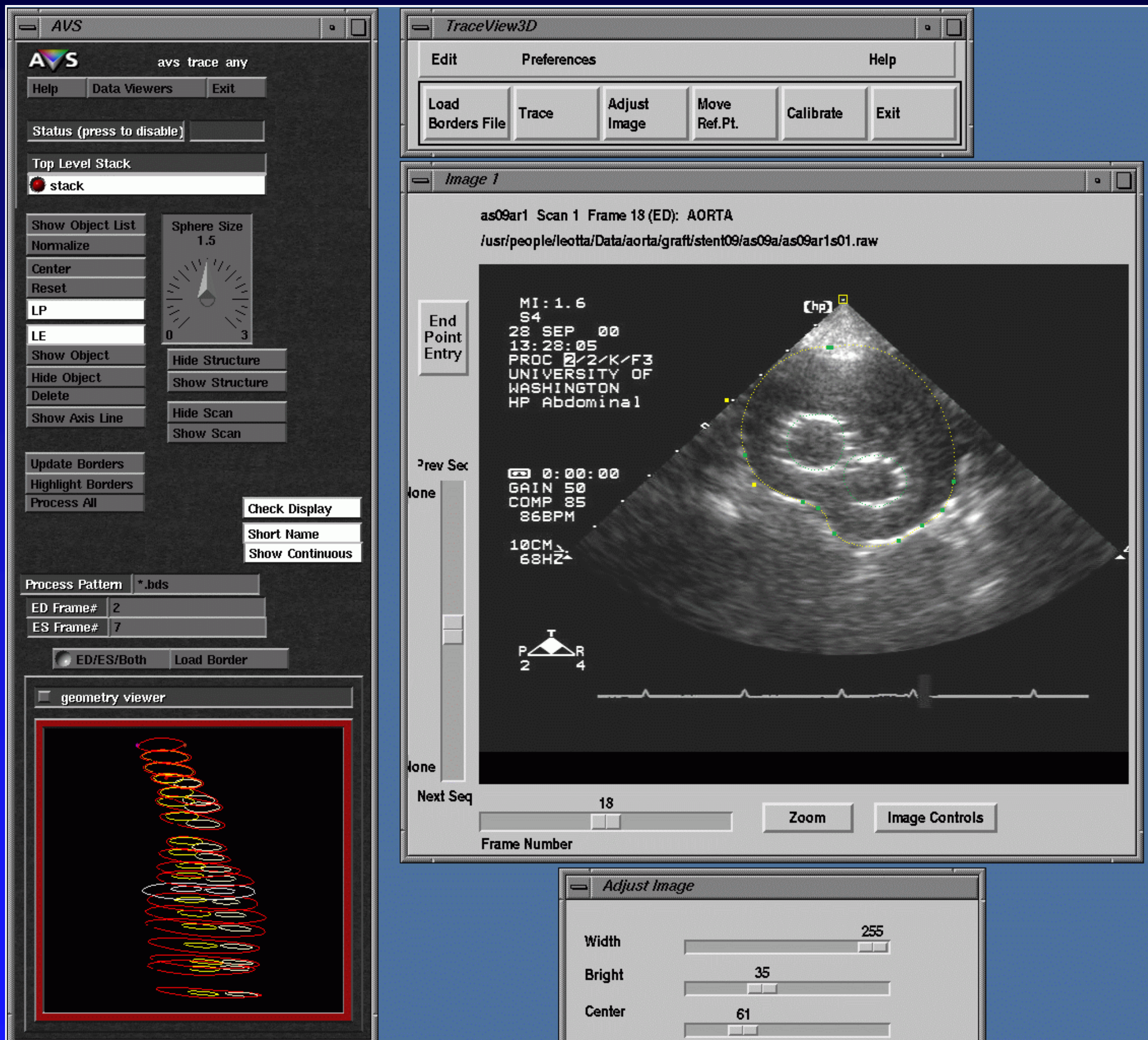


Image acquisition | Image stack | Outlines | 3D mesh/surface

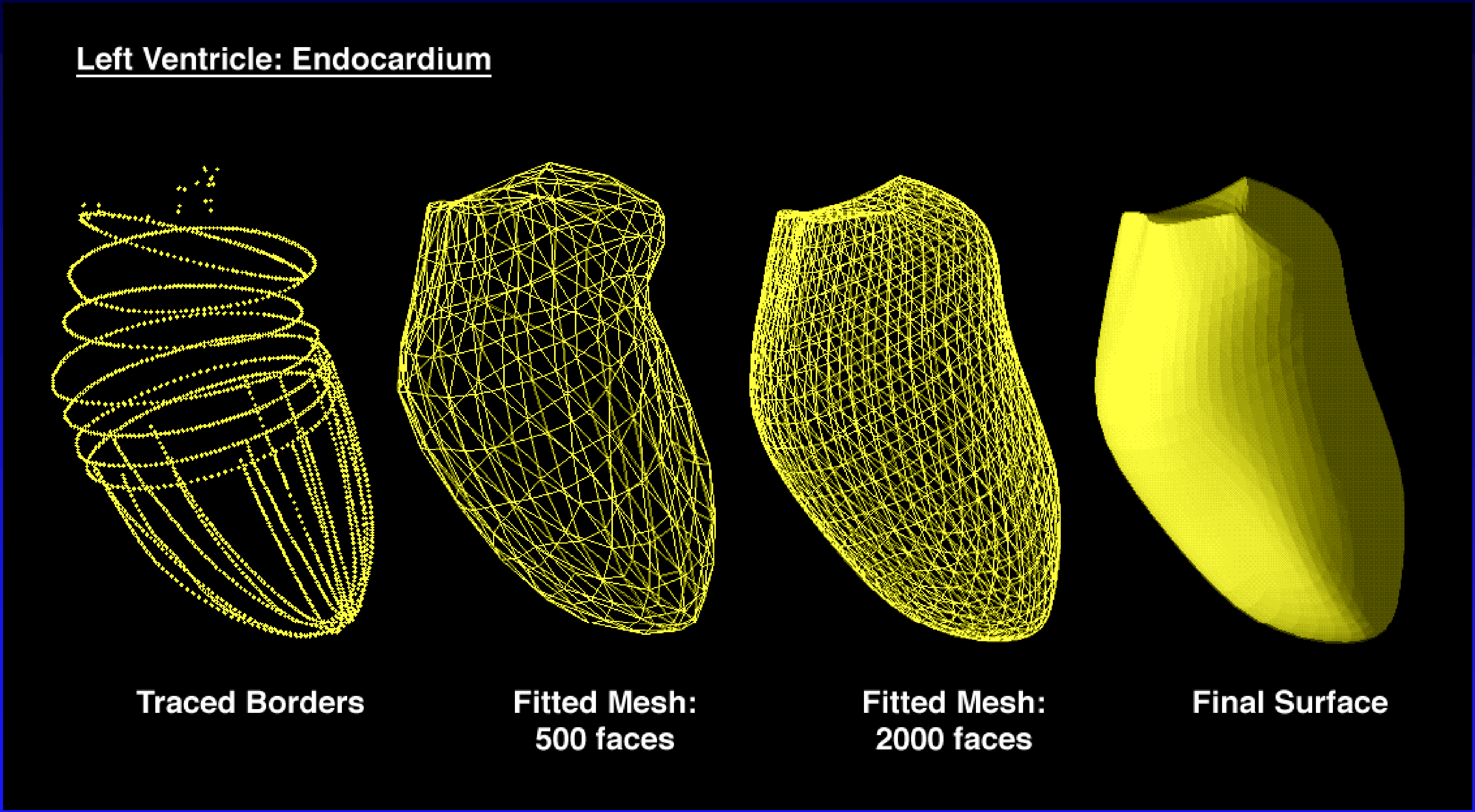
Surface Reconstruction



Cardiovascular Research Training Center, UW

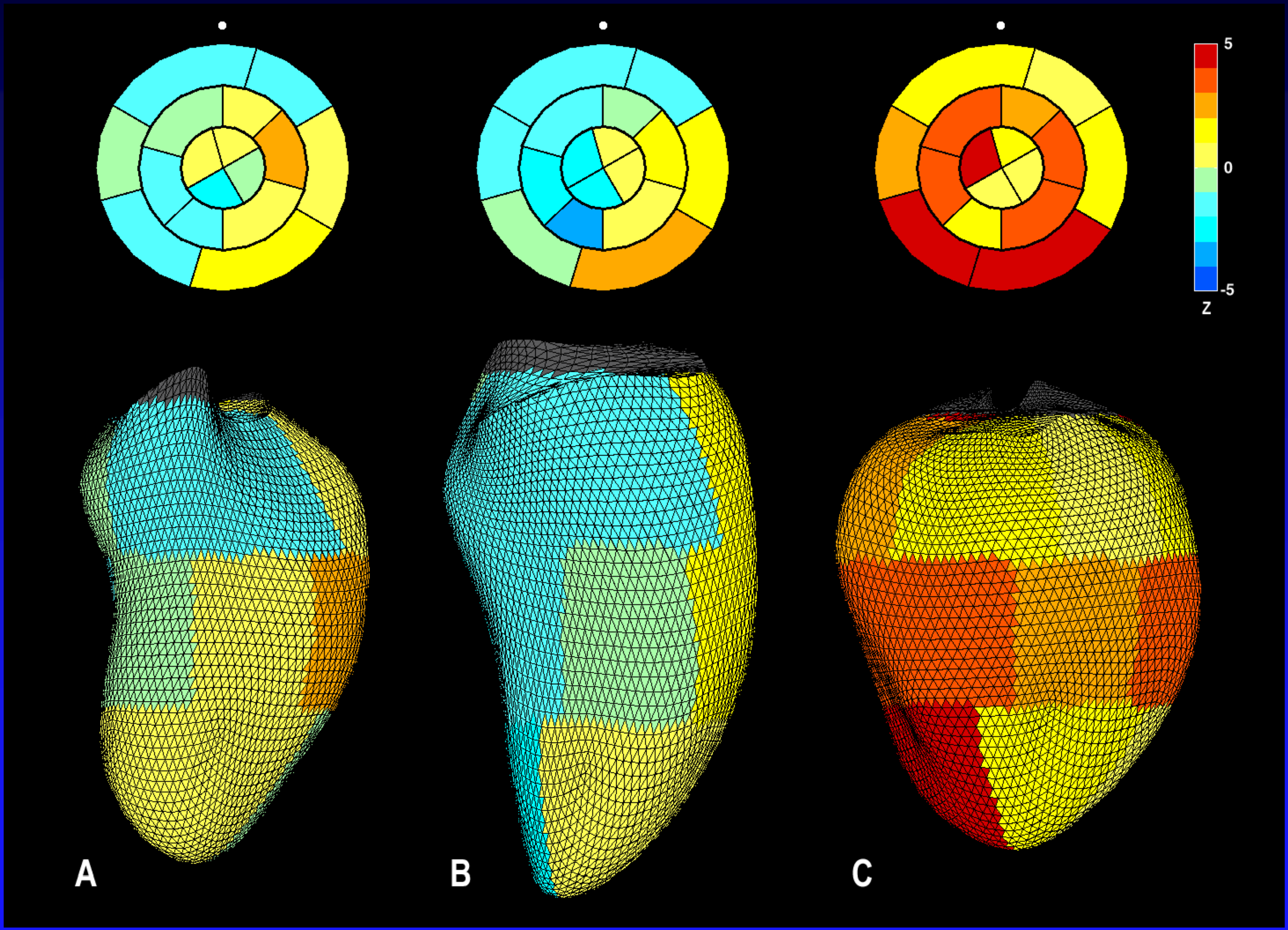
- Image outlining for structure measurement
- Segment multiple structures of interest
- 3D viewing window helps assess /guide tracing

Surface Reconstruction



Leotta et al.
J Am Soc Echocard, 1997

Cardiac Shape

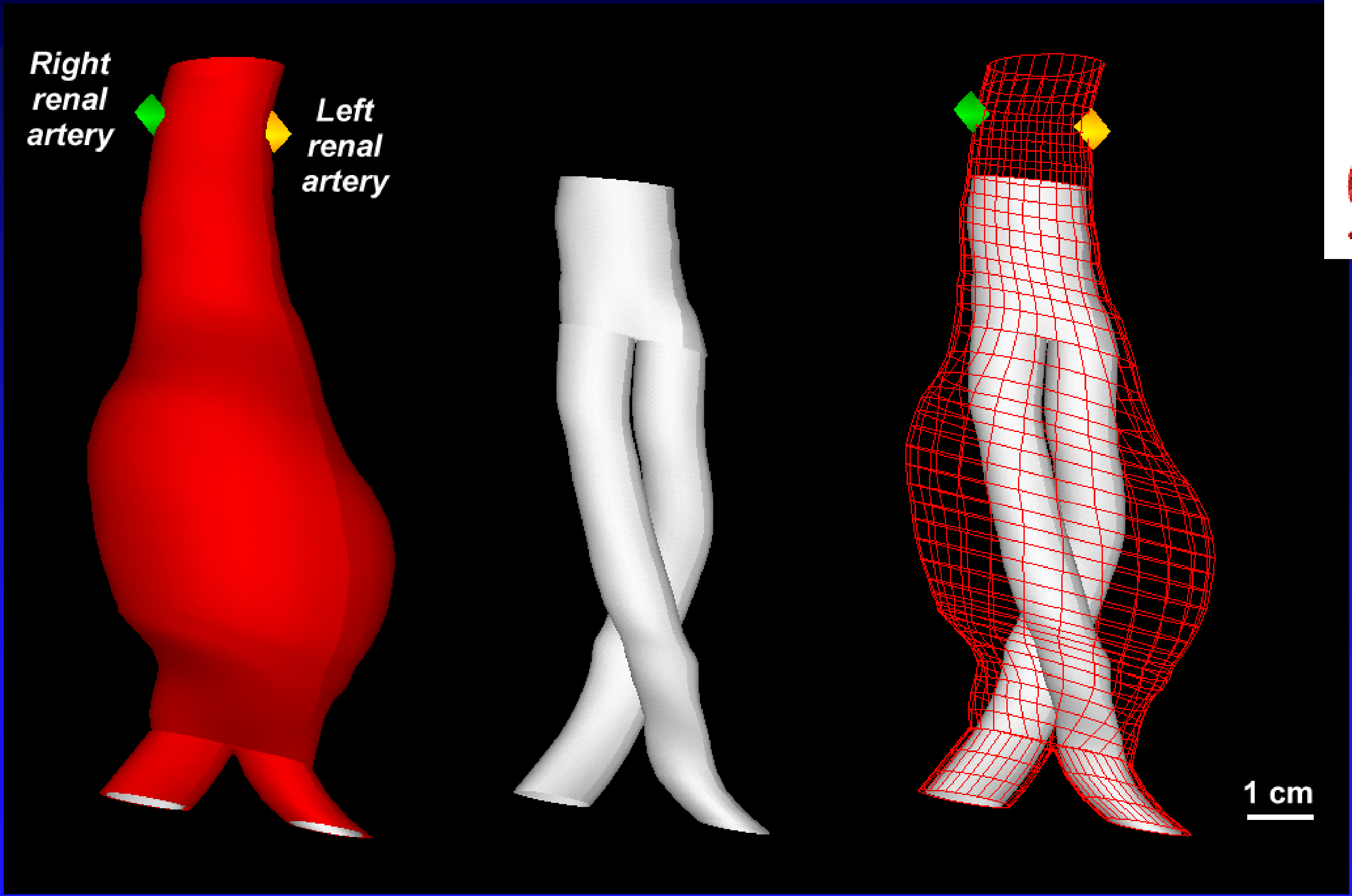


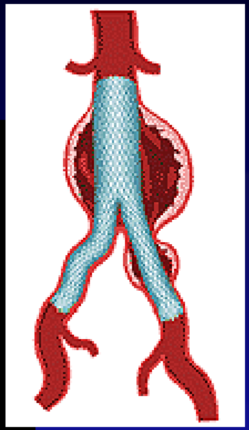
Normal Aortic stenosis Cardiomyopathy

- Compare individual reconstructed surfaces to a normal model derived from imaging of a representative population

Munt et al.
J Am Soc Echocard, 1998

AAA with Endovascular Graft



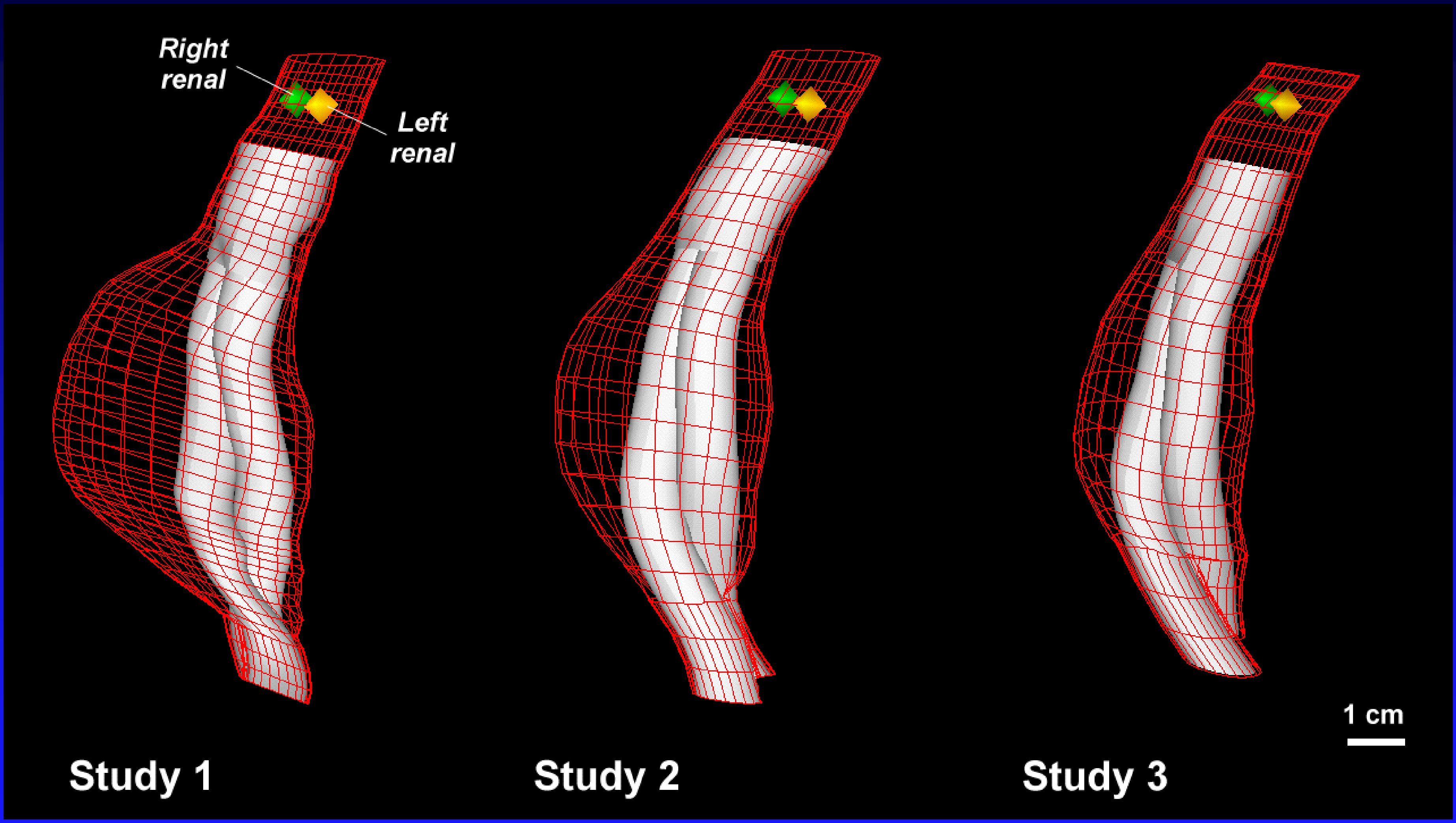


Abdominal Aortic Aneurysm

Anterior view of an aneurysm repaired by placement of an endovascular graft

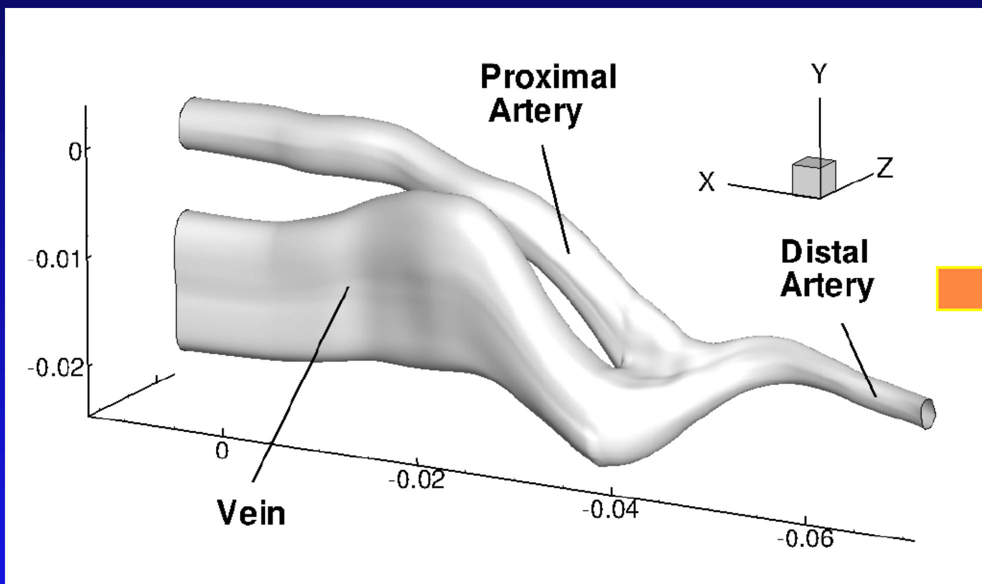
Leotta et al., J Vasc Surg,, 2001

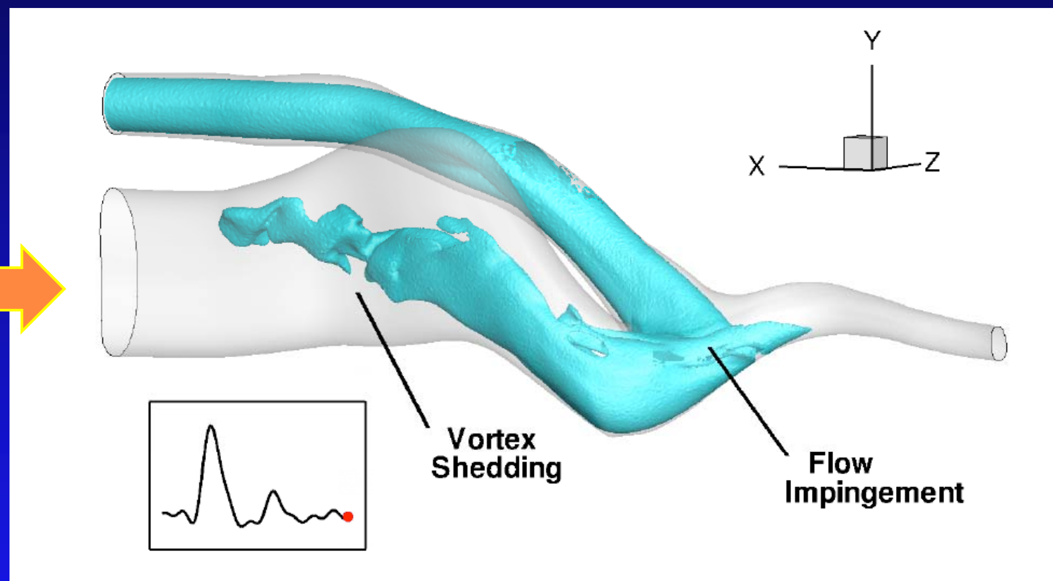
AAA Serial Study



Repaired aneurysm imaged 2 weeks (*left*), 6 months (*center*) and 1 year (*right*) after graft placement

Computational Flow Modeling for Dialysis Access Surgical Planning



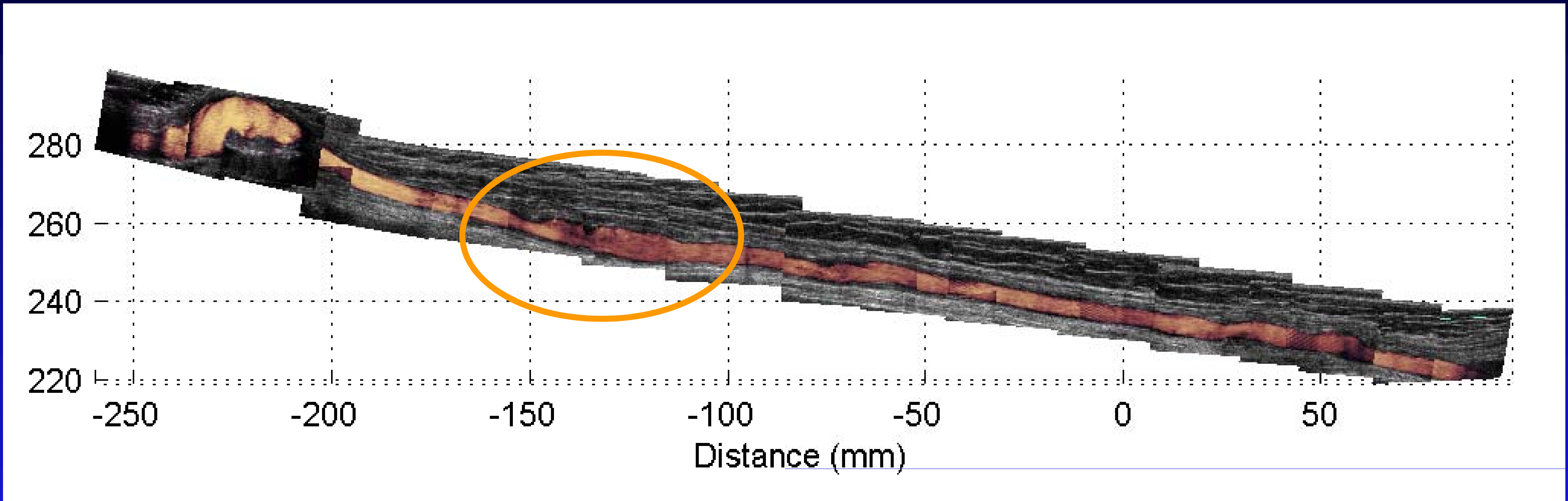


3D surface model

Computational flow model

Dialysis Access Arteriovenous Fistula

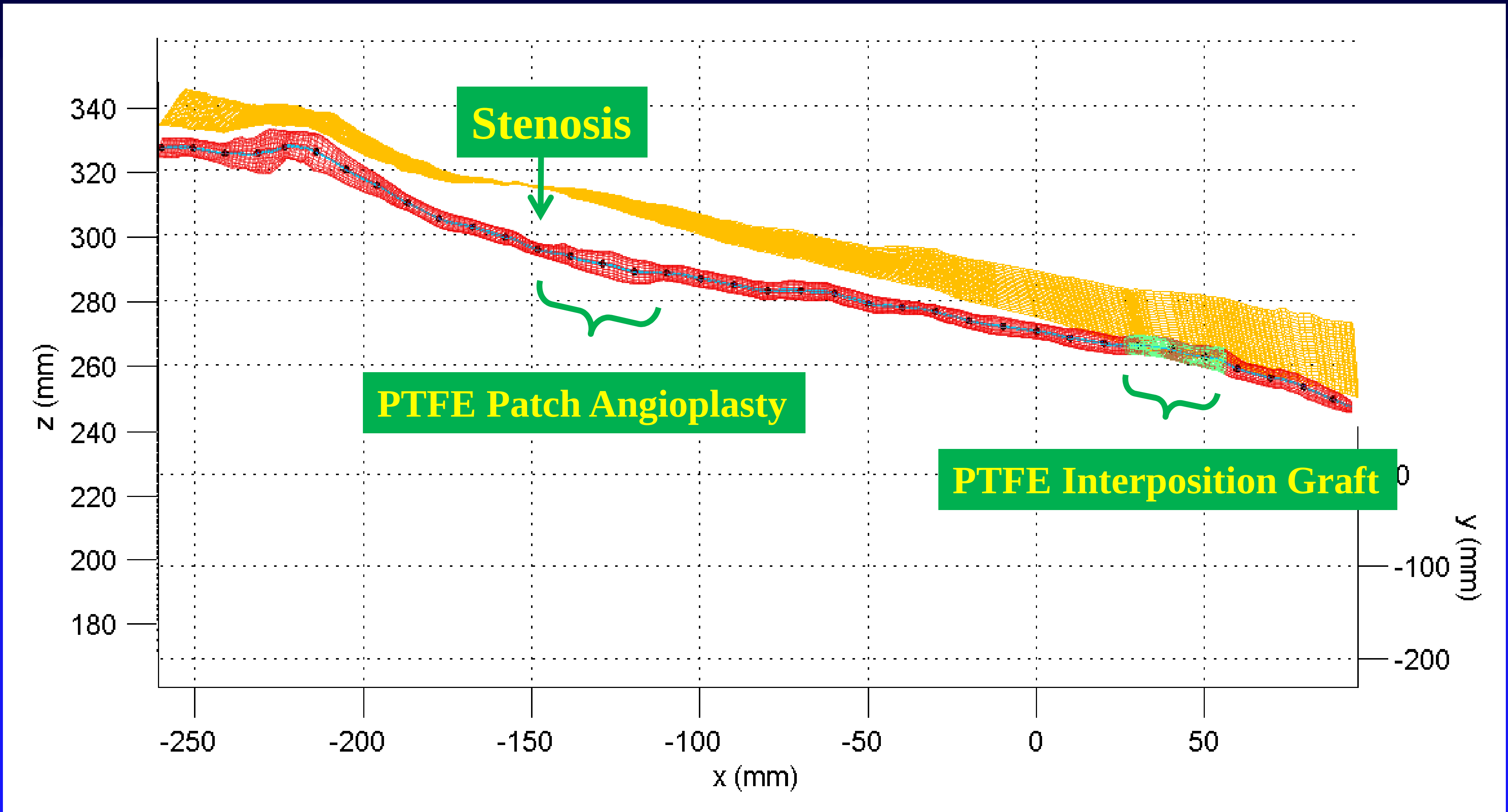
Serial Study: Vein Graft Revision



6 months post-revision

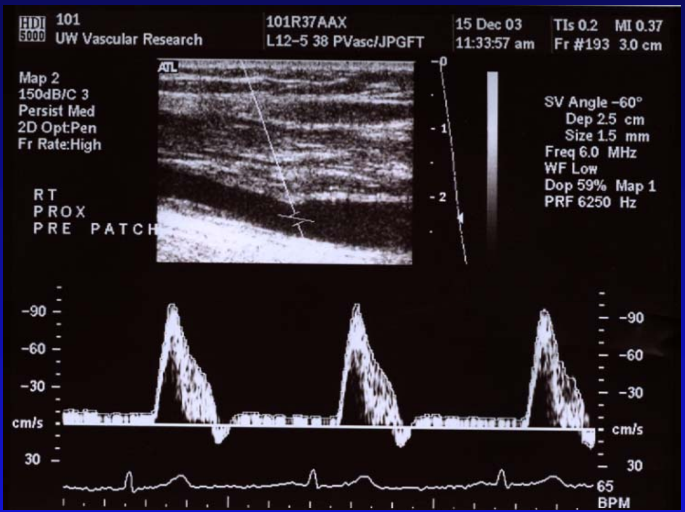
- **Vein graft with progressing stenosis at site proximal to a PTFE patch angioplasty repair**
  - **Femoral to above-knee popliteal reversed saphenous vein graft**
  - **Original graft: November 1996**
  - **Revision: November 2003**

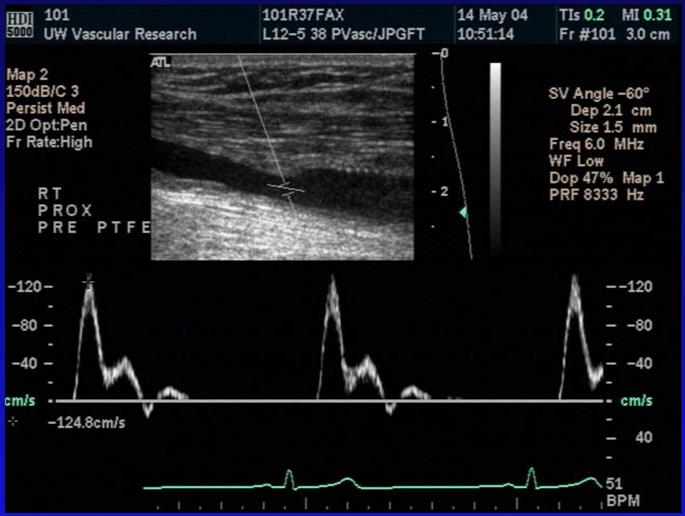
Serial Study: Vein Graft Revision

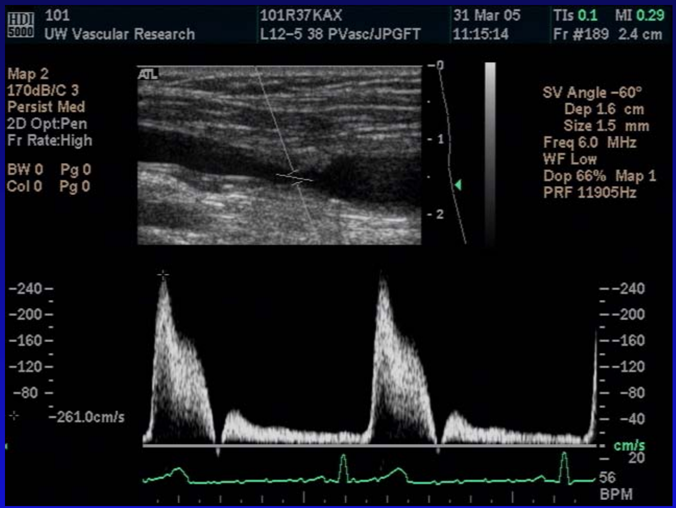


6 months post-revision

Serial Study: Vein Graft Revision

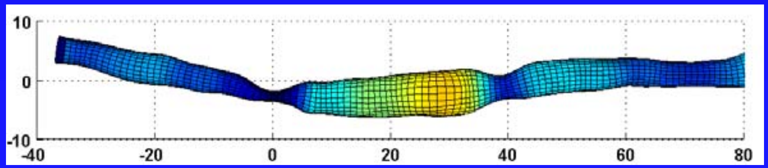




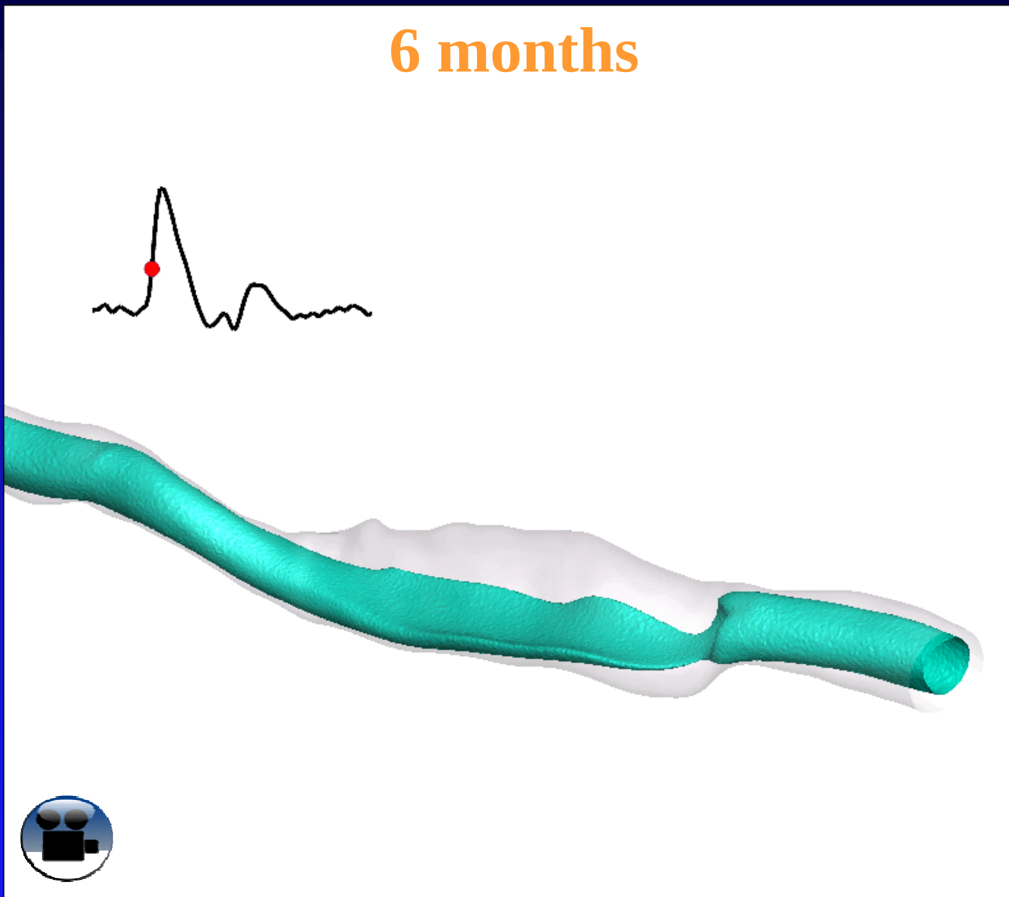


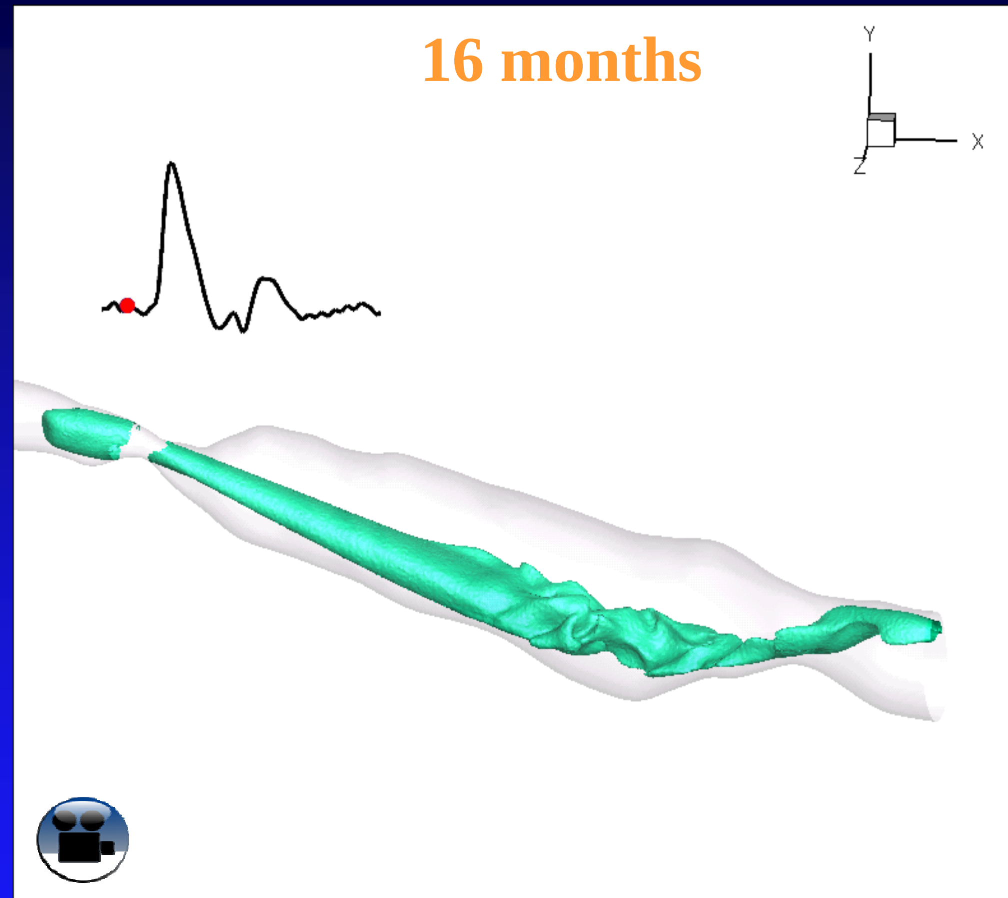
1 month **6 months** **16 months**

- **Vein graft with progressing stenosis at site proximal to a PTFE patch angioplasty repair**
  - **Femoral to above-knee popliteal reversed saphenous vein graft**
  - **Revision: November 2003**



Serial Study: Vein Graft Revision



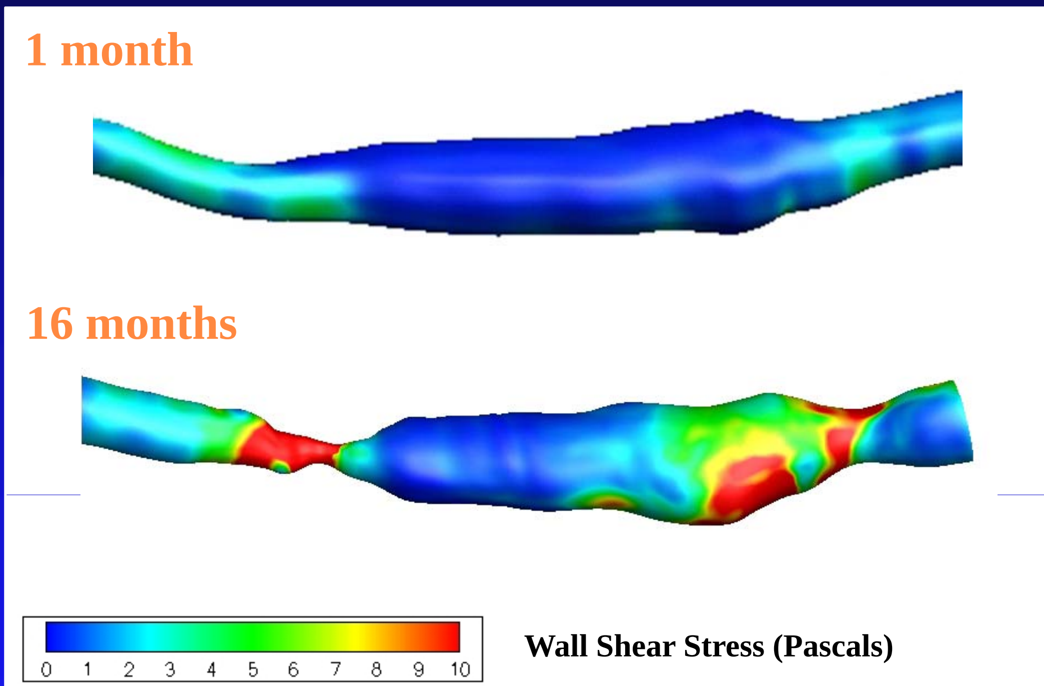


Axial Velocity over the Cardiac Cycle

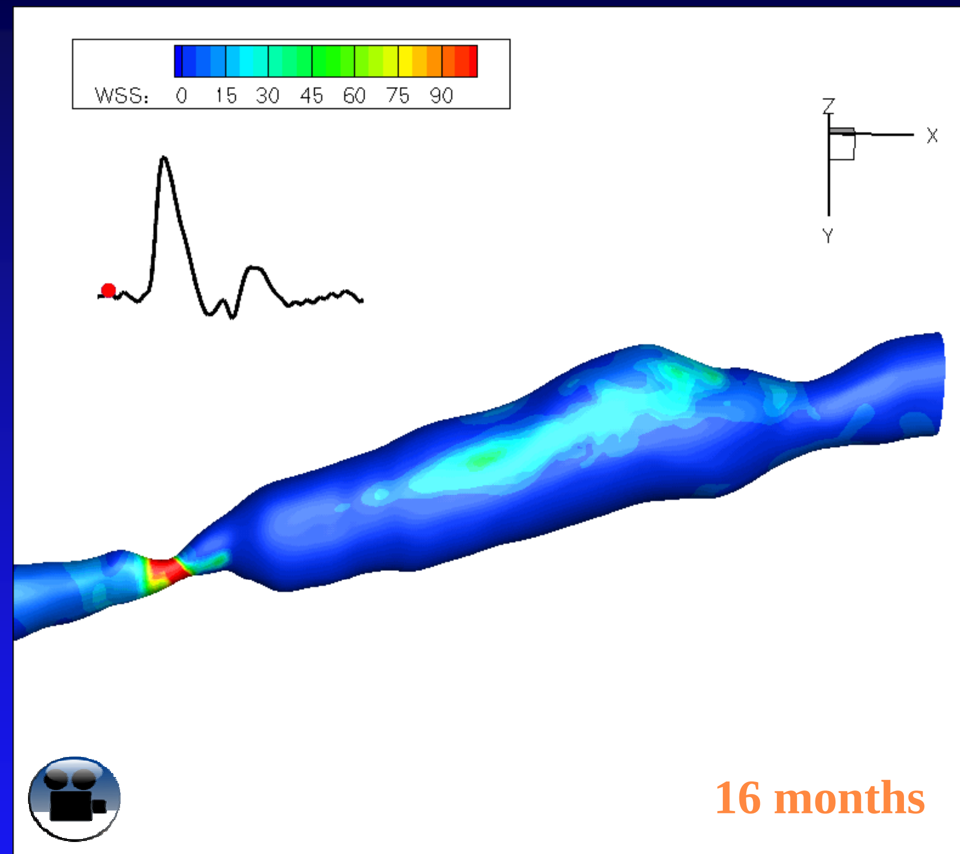
- **Turbulent jet impinging on vessel wall leads to dilation over time**

McGah et al., J Biomech Eng, 2011

Serial Study: Vein Graft Revision



Wall Shear Stress time-averaged over one cardiac cycle



Wall Shear Stress over the cardiac cycle

<u>Wall Shear Stress</u>: stress (force per unit area) that is applied parallel or tangential to a face of a material

McGah et al., J Biomech Eng, 2011