Basic Principles - 2D Echo and Doppler

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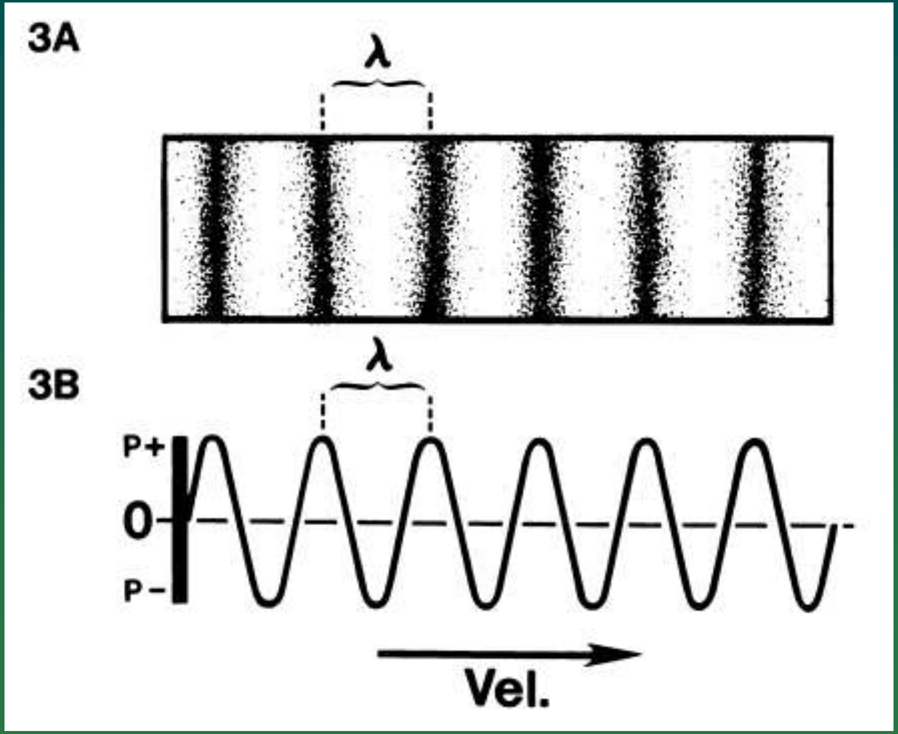
Acknowledging many illustrations from Weyman’s text and others.

Echo-Doppler Basic Principles

- Ultrasound physics
  - resolution - axial and lateral
  - attenuation
- Doppler ultrasound
  - Aliasing
  - Bernoulli principle
  - laminar and turbulent flow
  - continuity equation

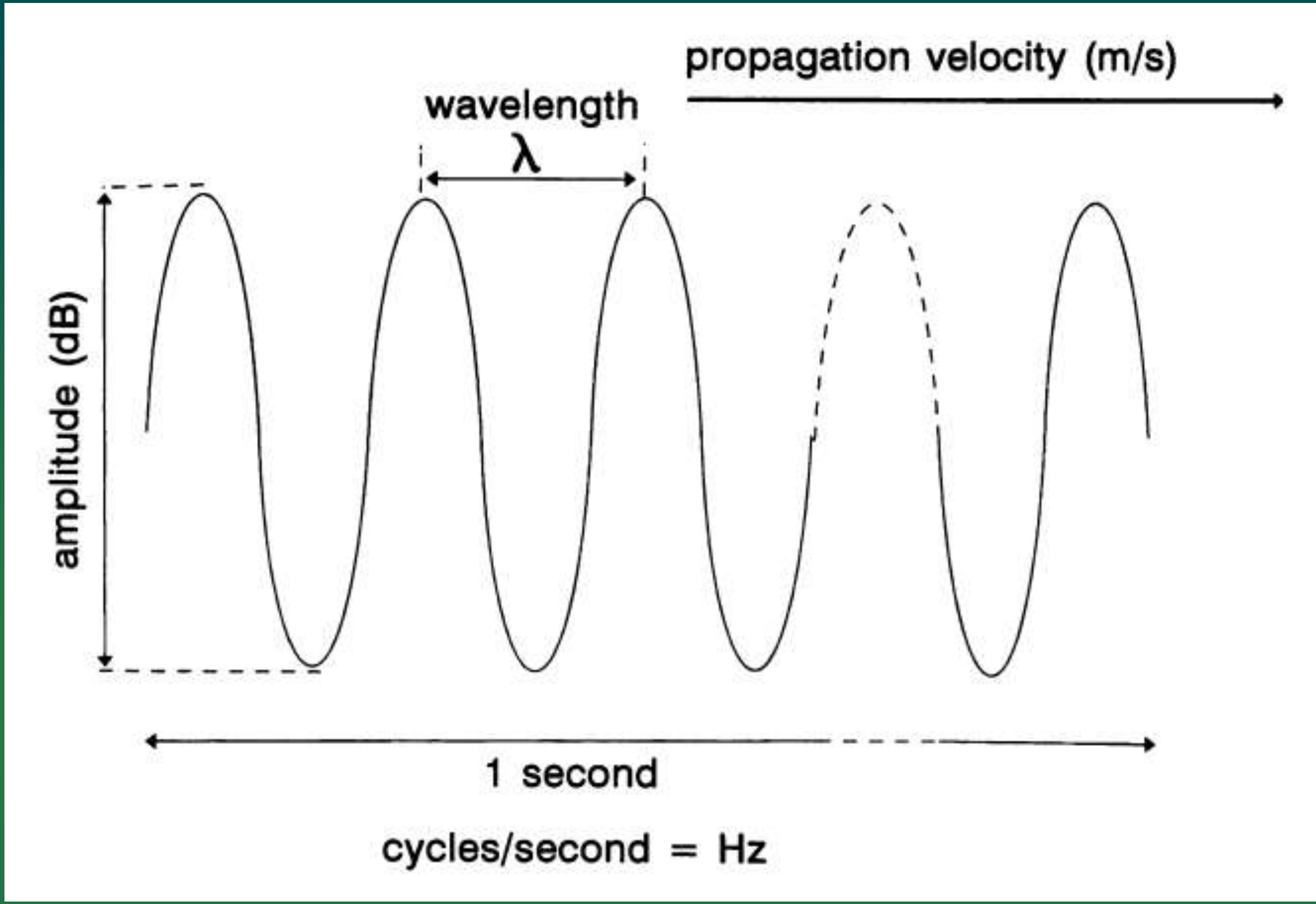
Ultrasound physics



- Sound - waves of compression and rarefaction propagated through a medium
- Ultrasound - sound frequency above 20kHz
- Cardiac ultrasound - 1 - 20 mHz
- Intensity - watts/cm2 (joule/sec/cm2)

Sound Wave Diagram

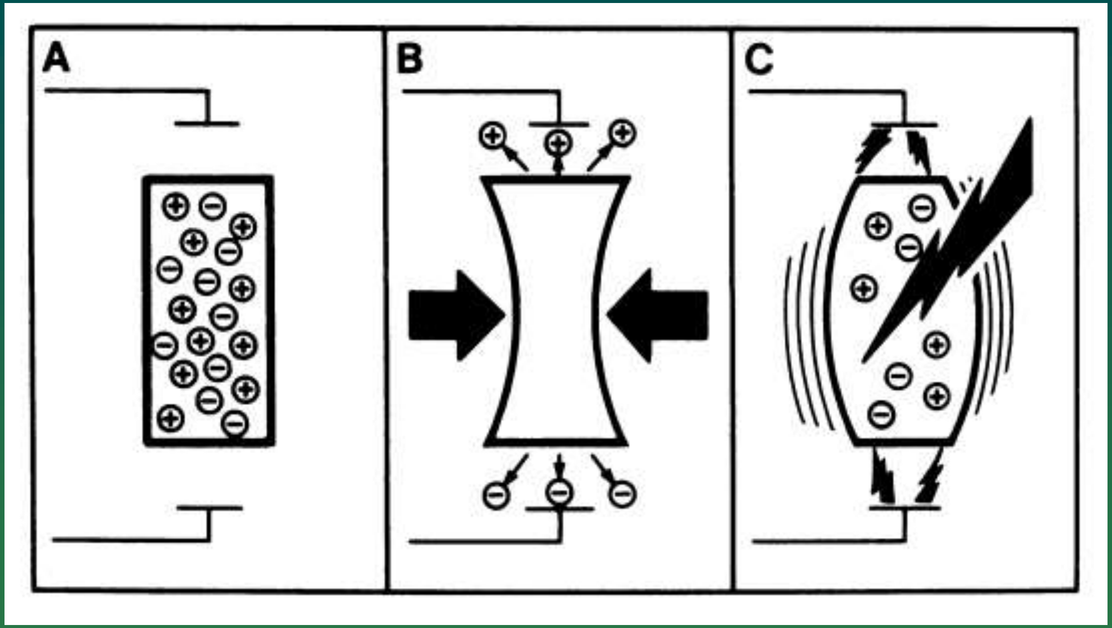
$c=\lambda*f$



Wavelength times frequency equals propagation velocity
$c=\lambda*f$, and c=1540 m/s, so λ (mm)= 1.54/f (MHz)
λ for 2 MHz is 0.75mm
λ for 4 MHz is 0.38mm

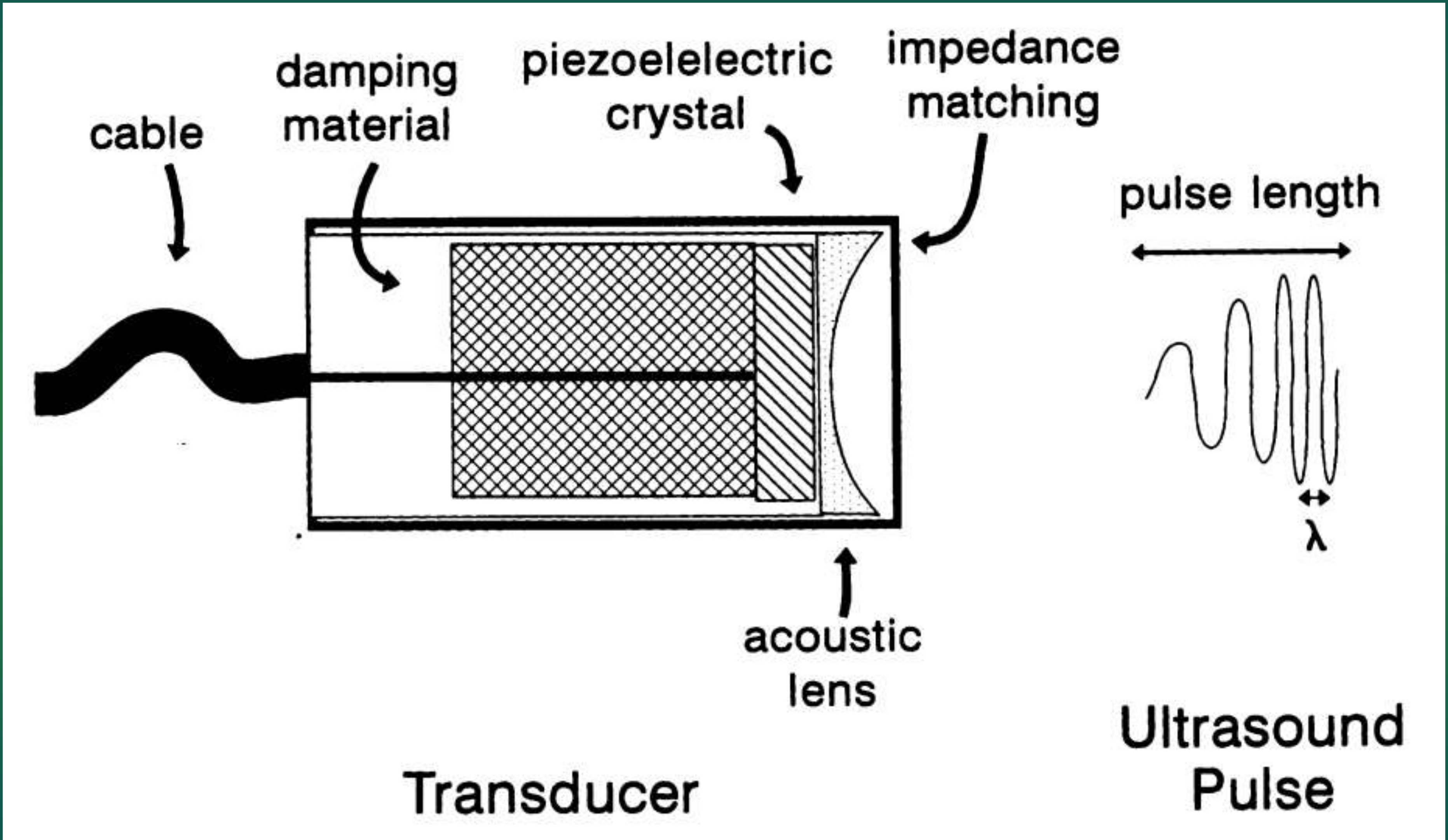
Otto CM. Textbook of Clinical Echocardiography. 2000.

Piezoelectricity



- Mechanical stress applied to a crystal causes electrical charges bound in the crystal to shift to the surface where they can be measured as a voltage
- Electric current applied to a crystal changes the crystal shape, alternating current can cause vibration of crystal, producing sound wave

Transducer Structure



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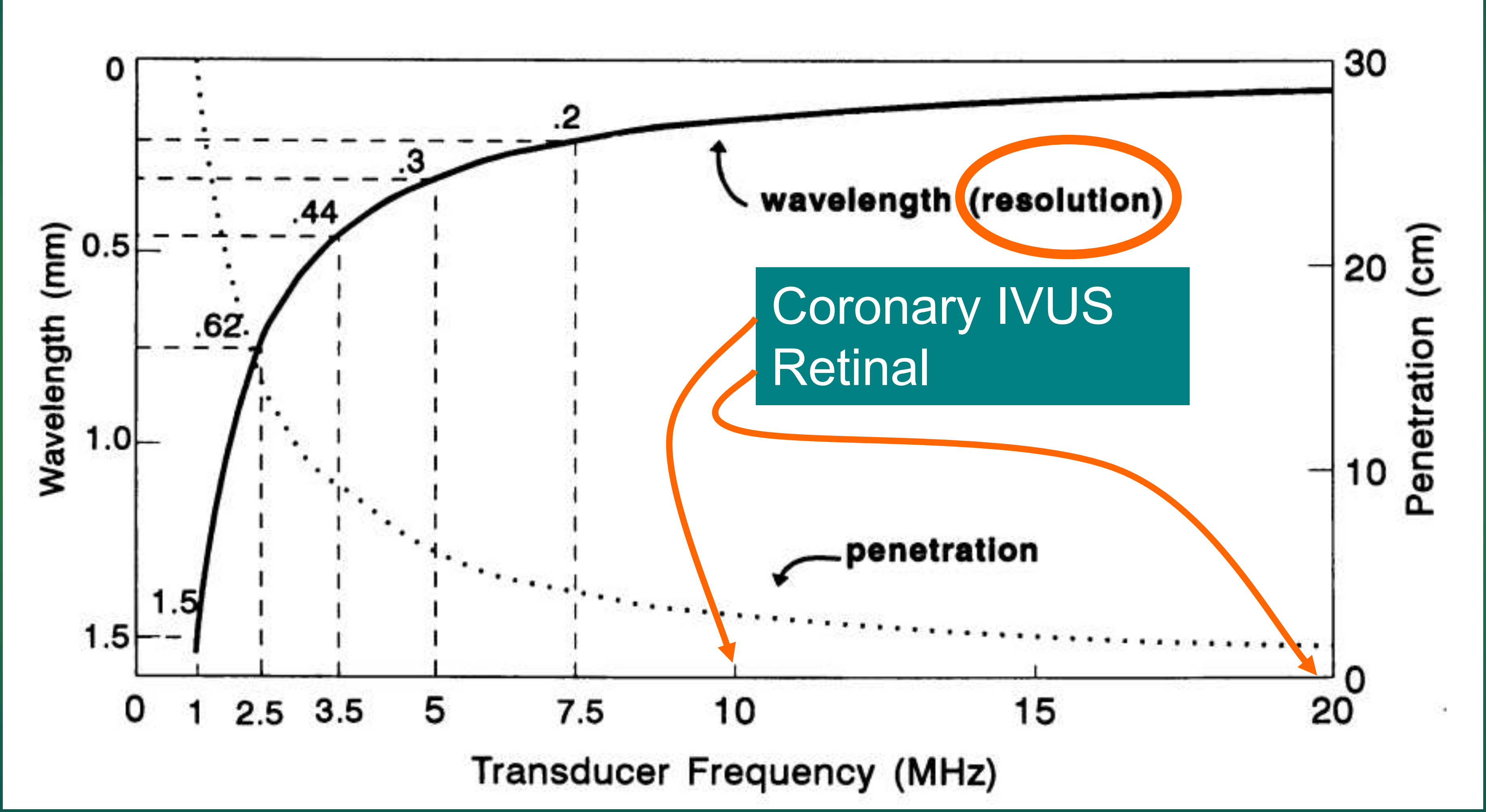
Wavelength versus Penetration



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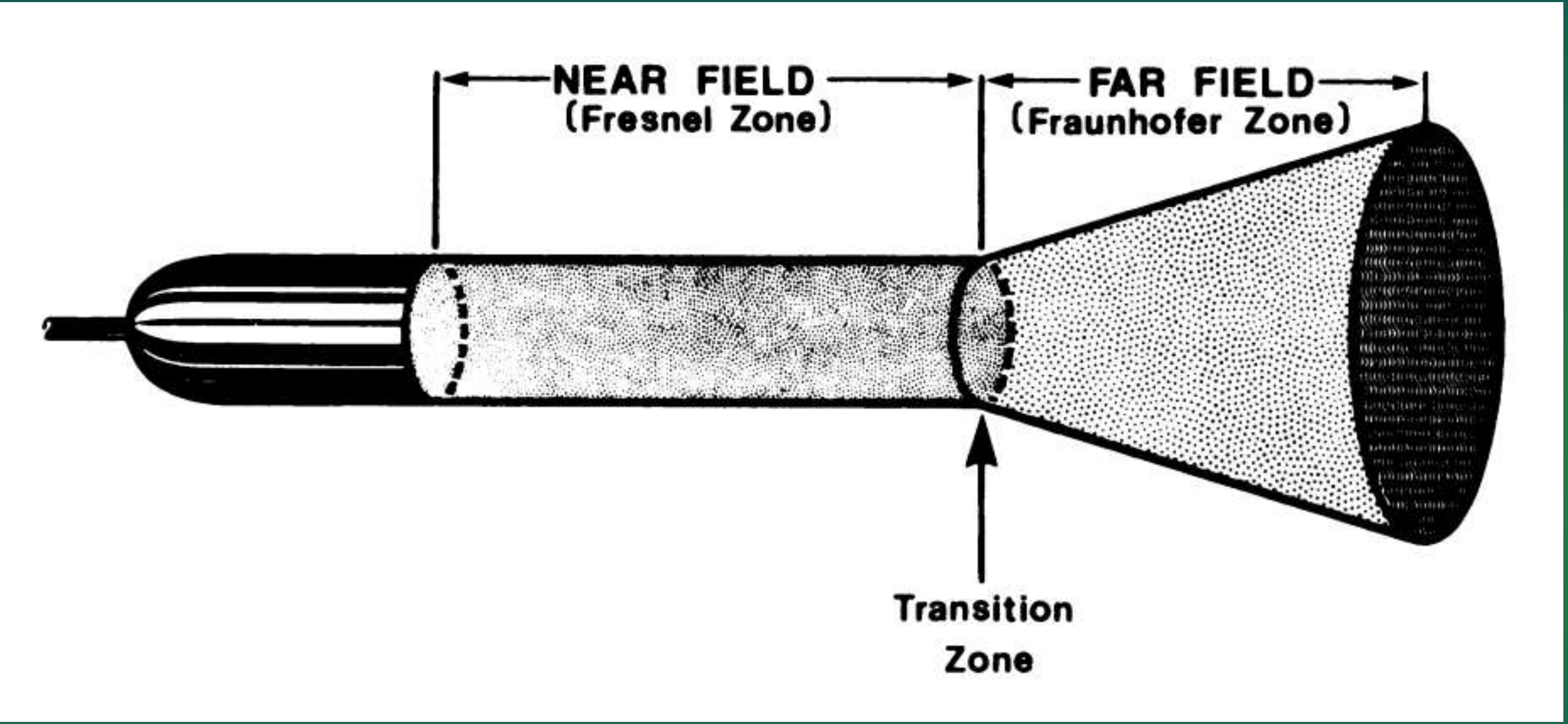
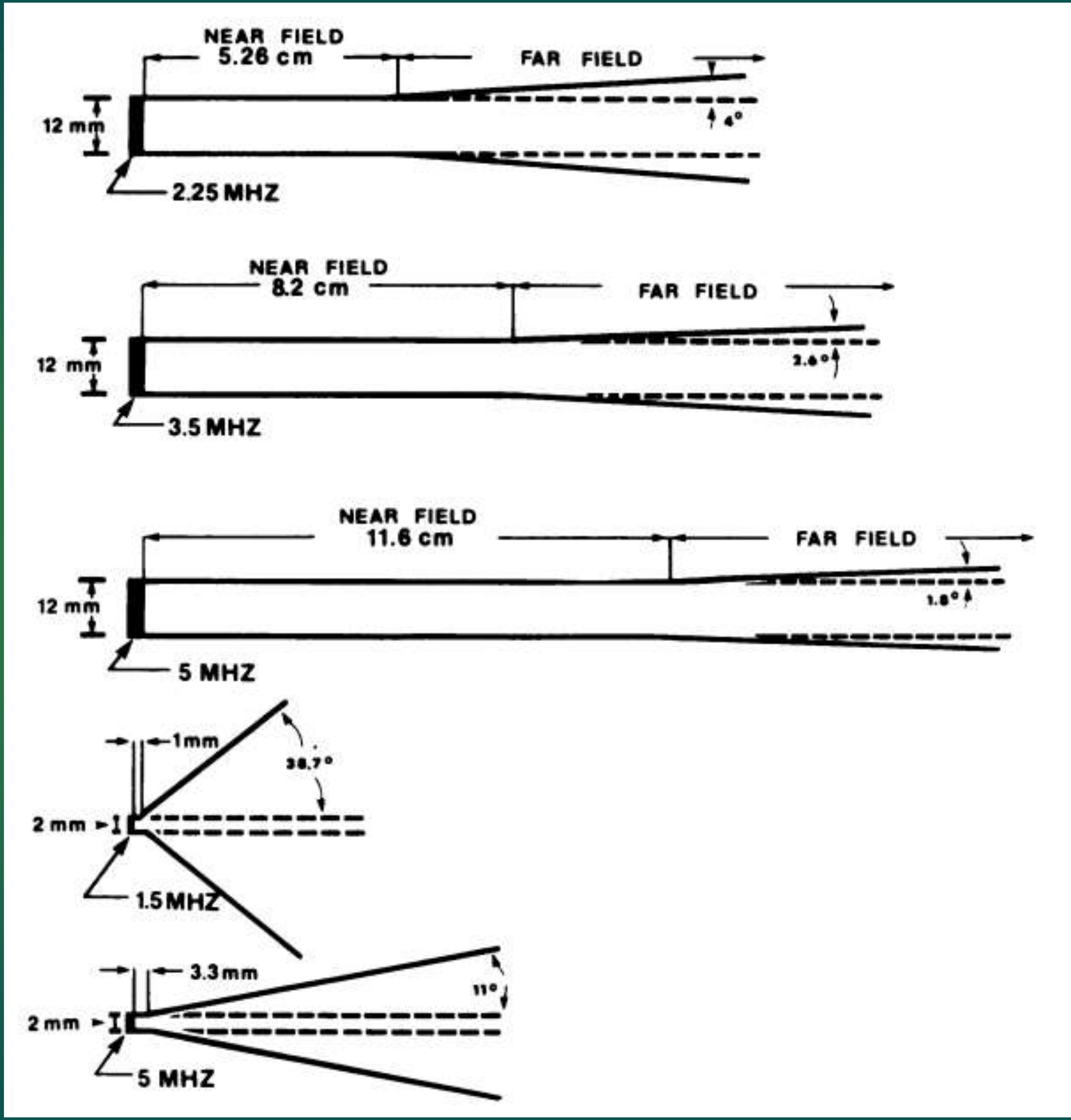
Echo Transducer



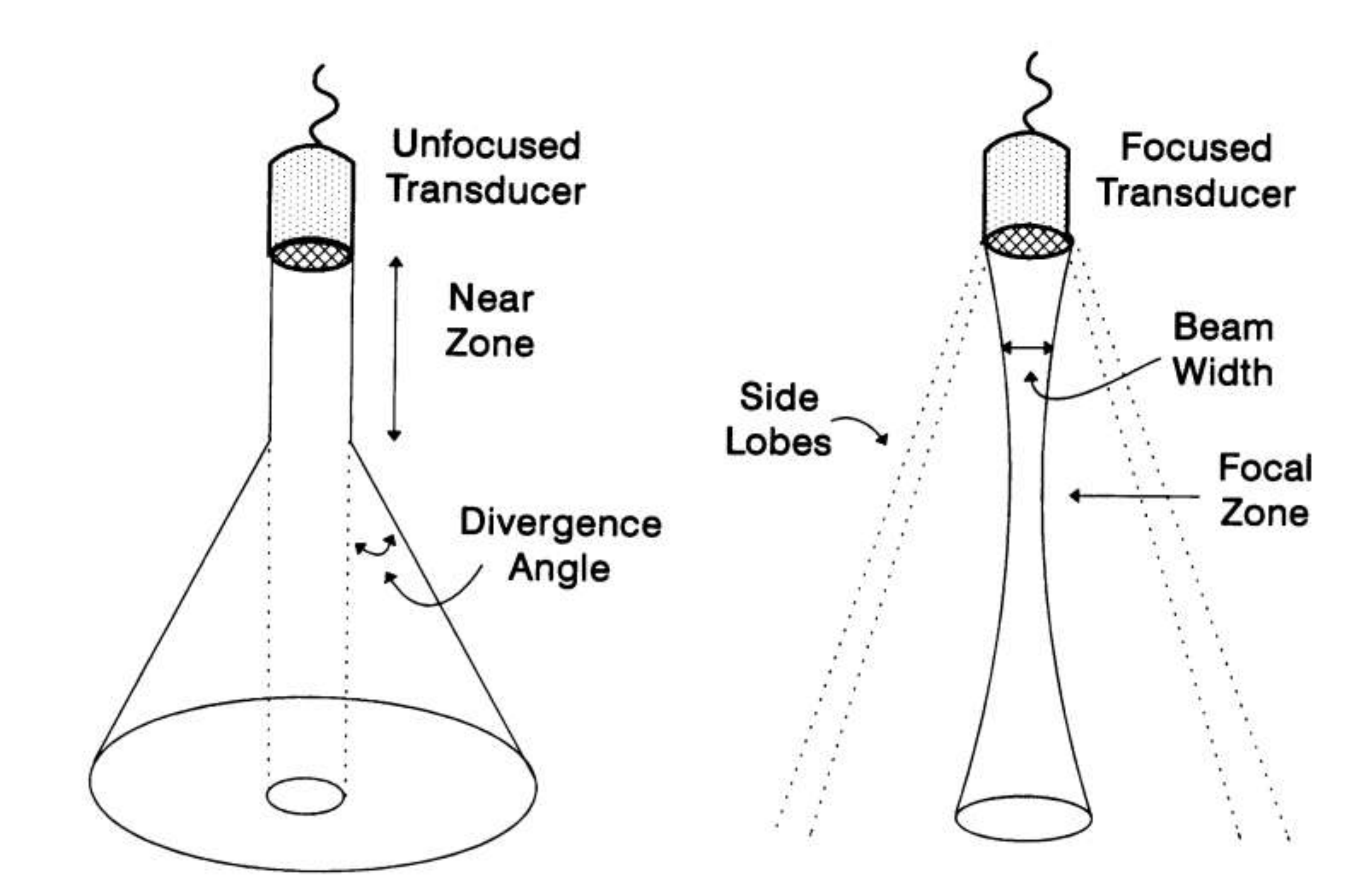
- Definitions of fields of transducer performance

Echo Transducer

- Single crystal
- Effect of crystal diameter and frequency on near field and far field
- Larger diameter and higher frequency give longer near field and less divergent far field



Transducer Beam Zones



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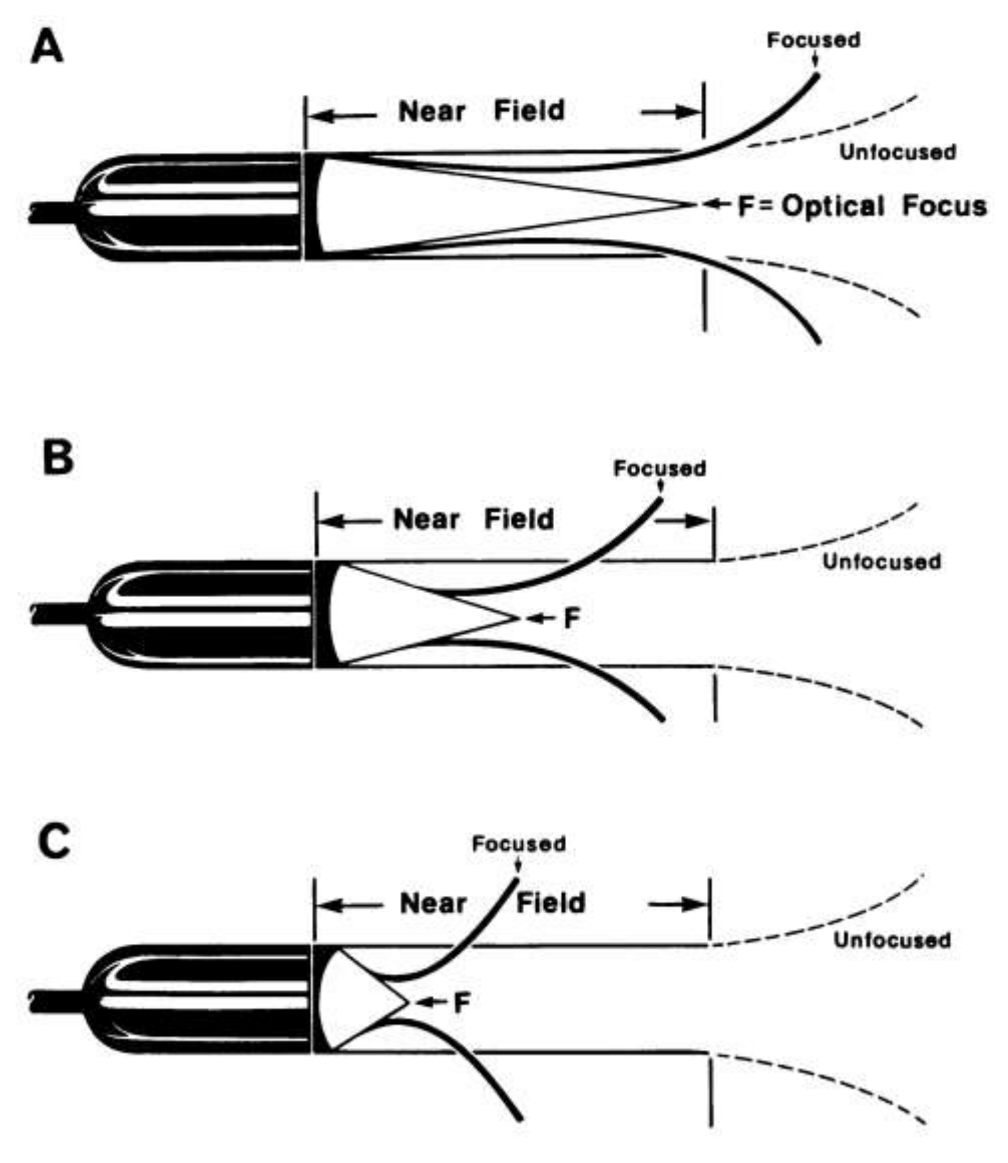
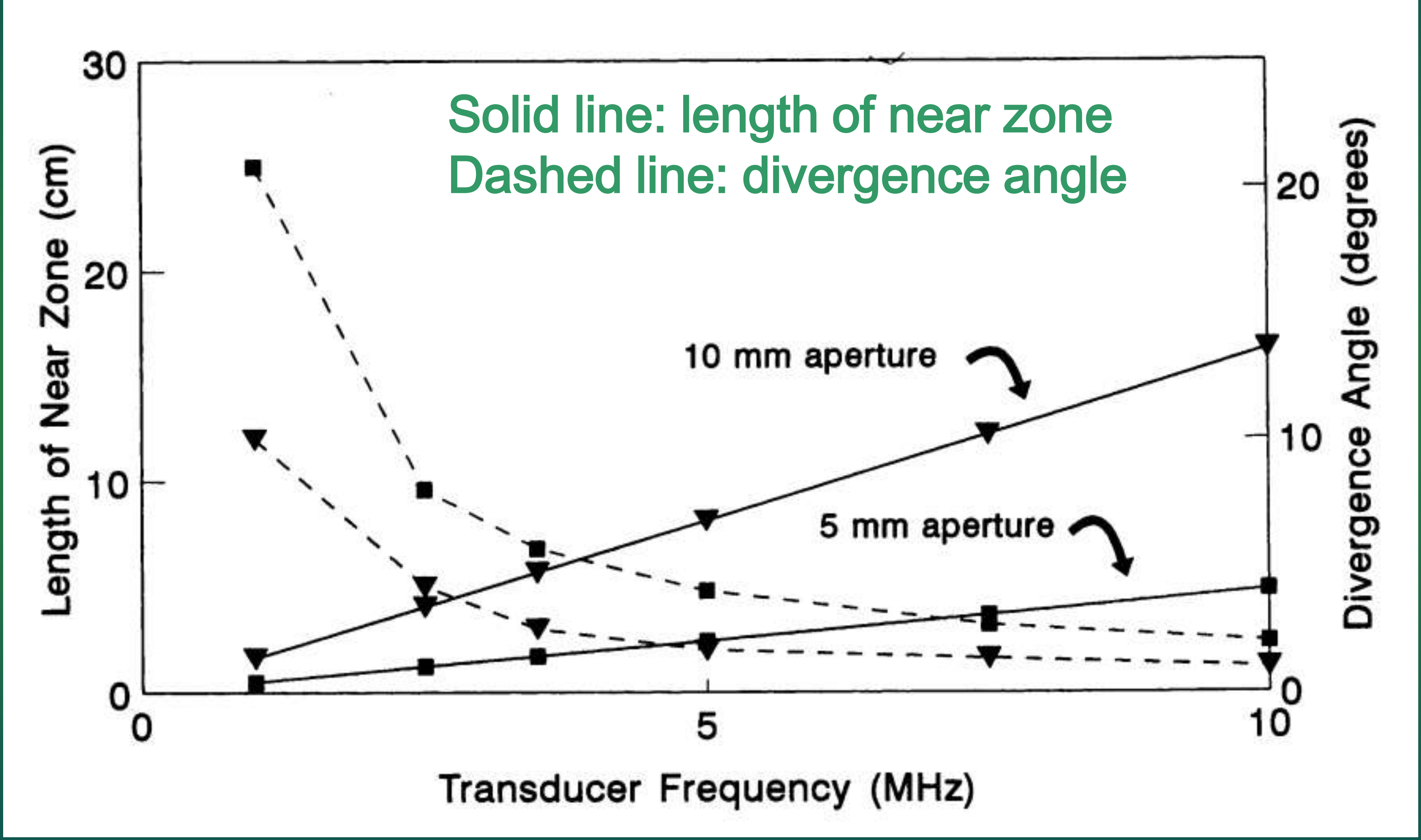
Echo Transducer

- Effect of focal length and focusing on near field and far field



Graph: Transducer Beam Zones



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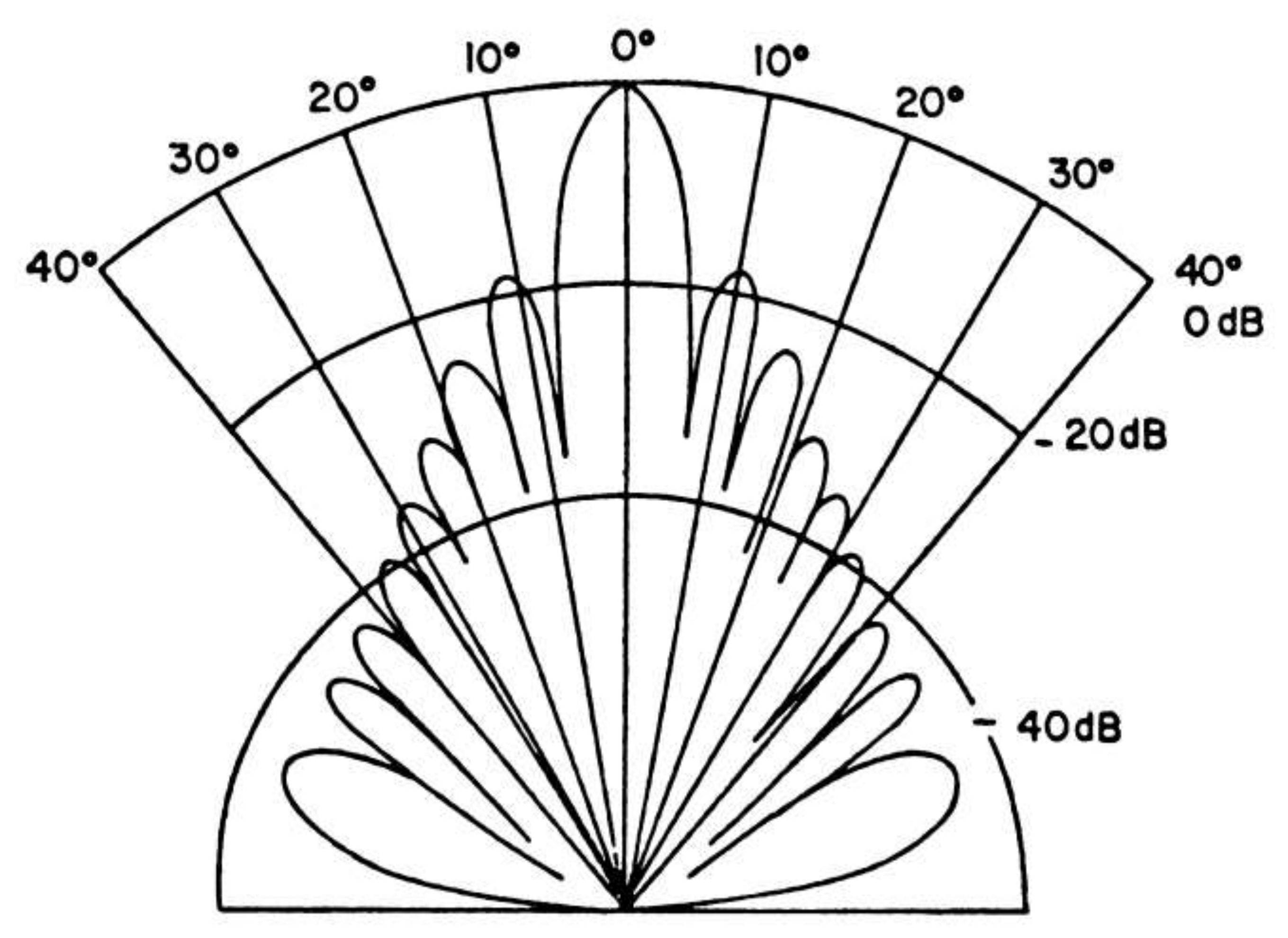
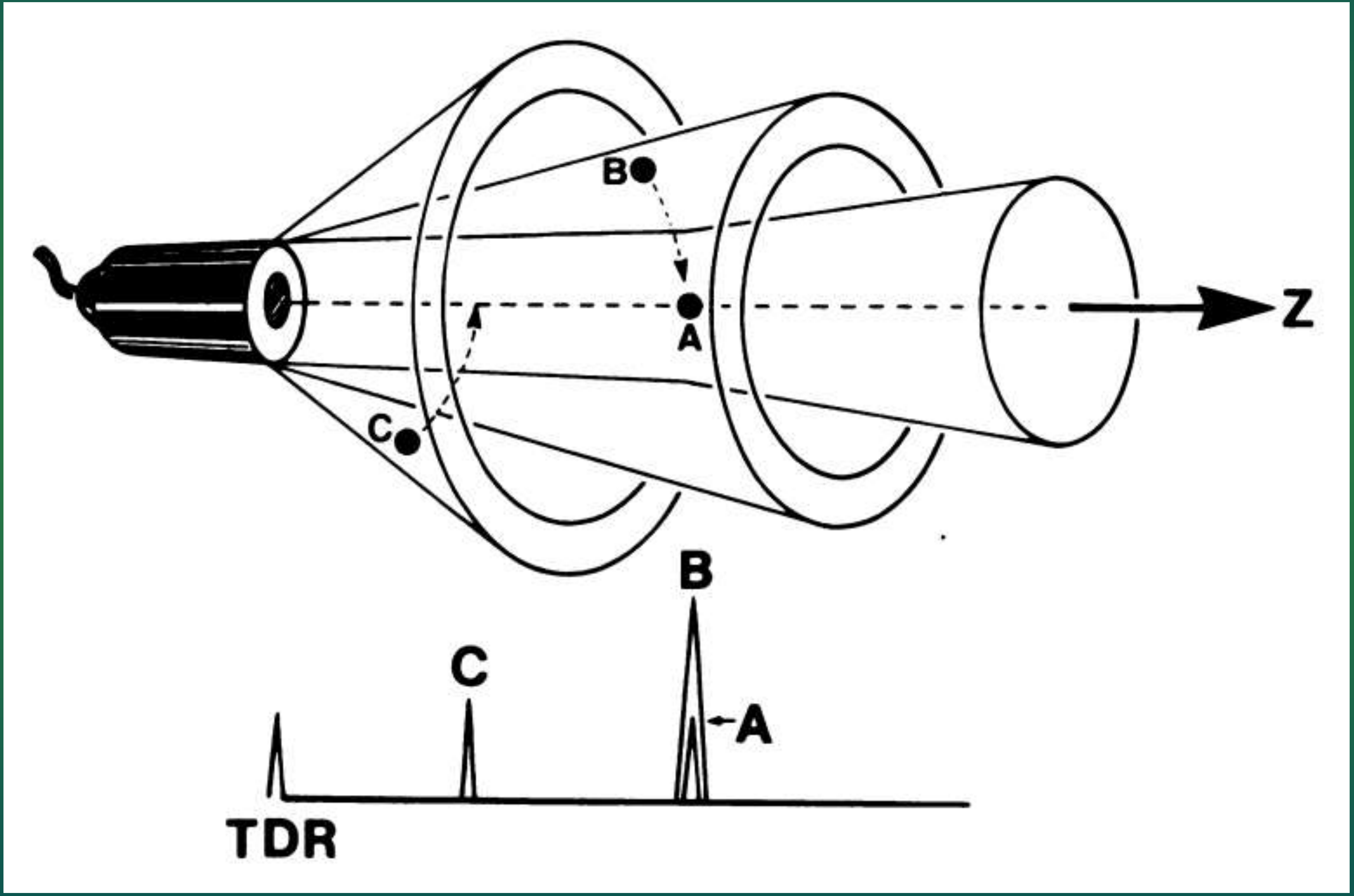
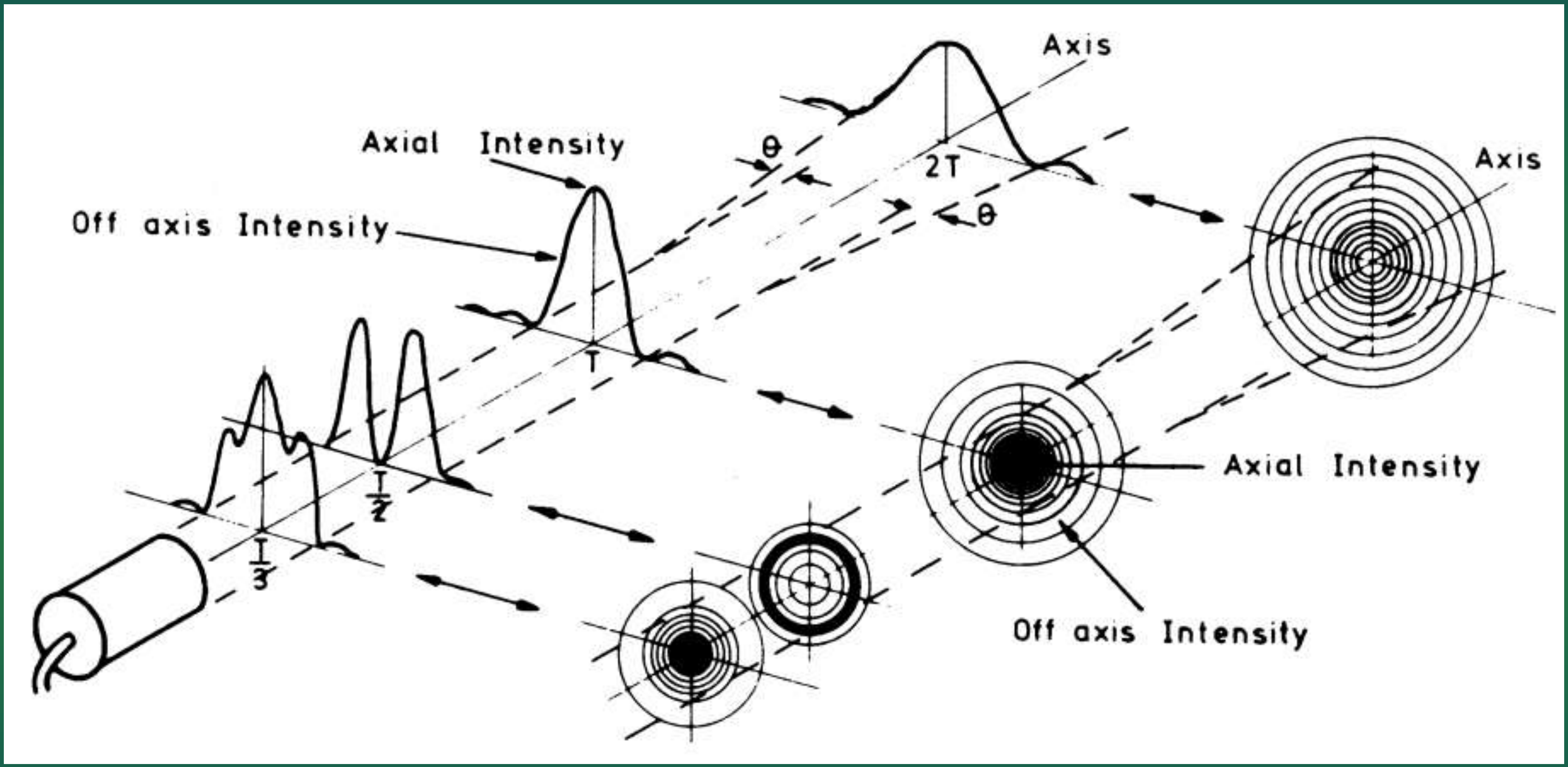
Side Lobe Artifacts



Side Lobe Artifacts

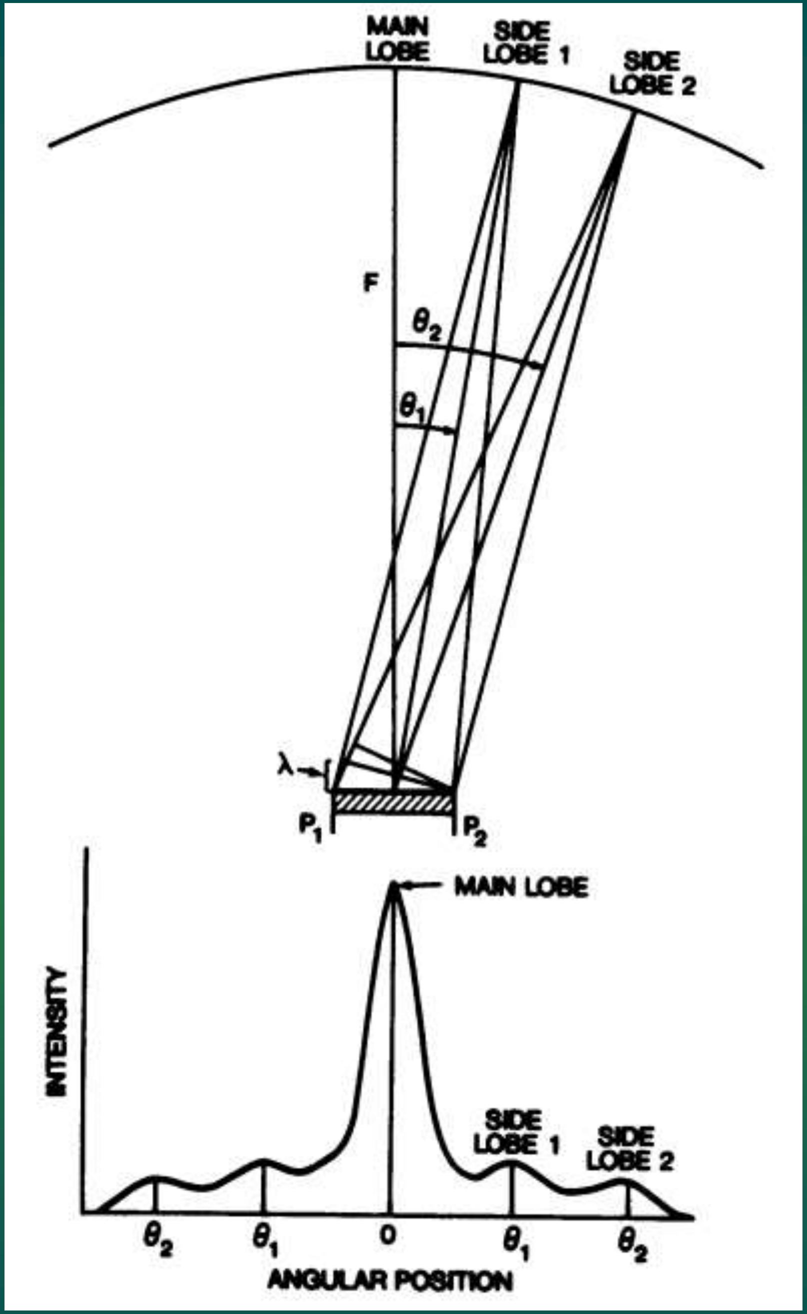


Echo Transducer Lateral Variation



Side Lobe Artifact in Single Crystal Transducer

Position of side lobes at locations where the distances from each edge of the crystal face differ by one wavelength



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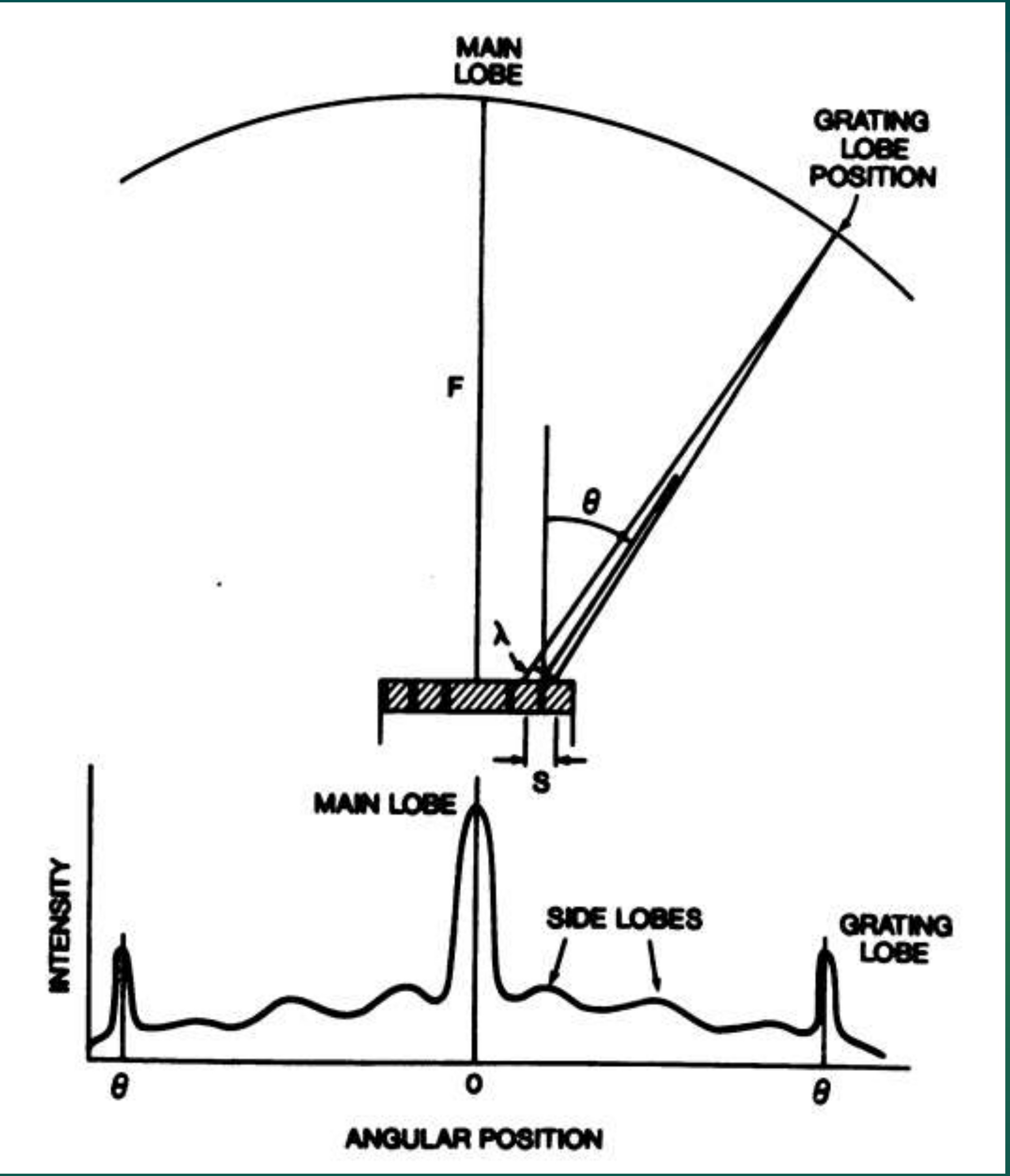
Grating Lobe Artifact in Phased-Array Transducer

Position of grating lobes is determined by spacing between centers of independent crystal elements in the transducer.

S=spacing between elements

F=focal length

λ=wavelength



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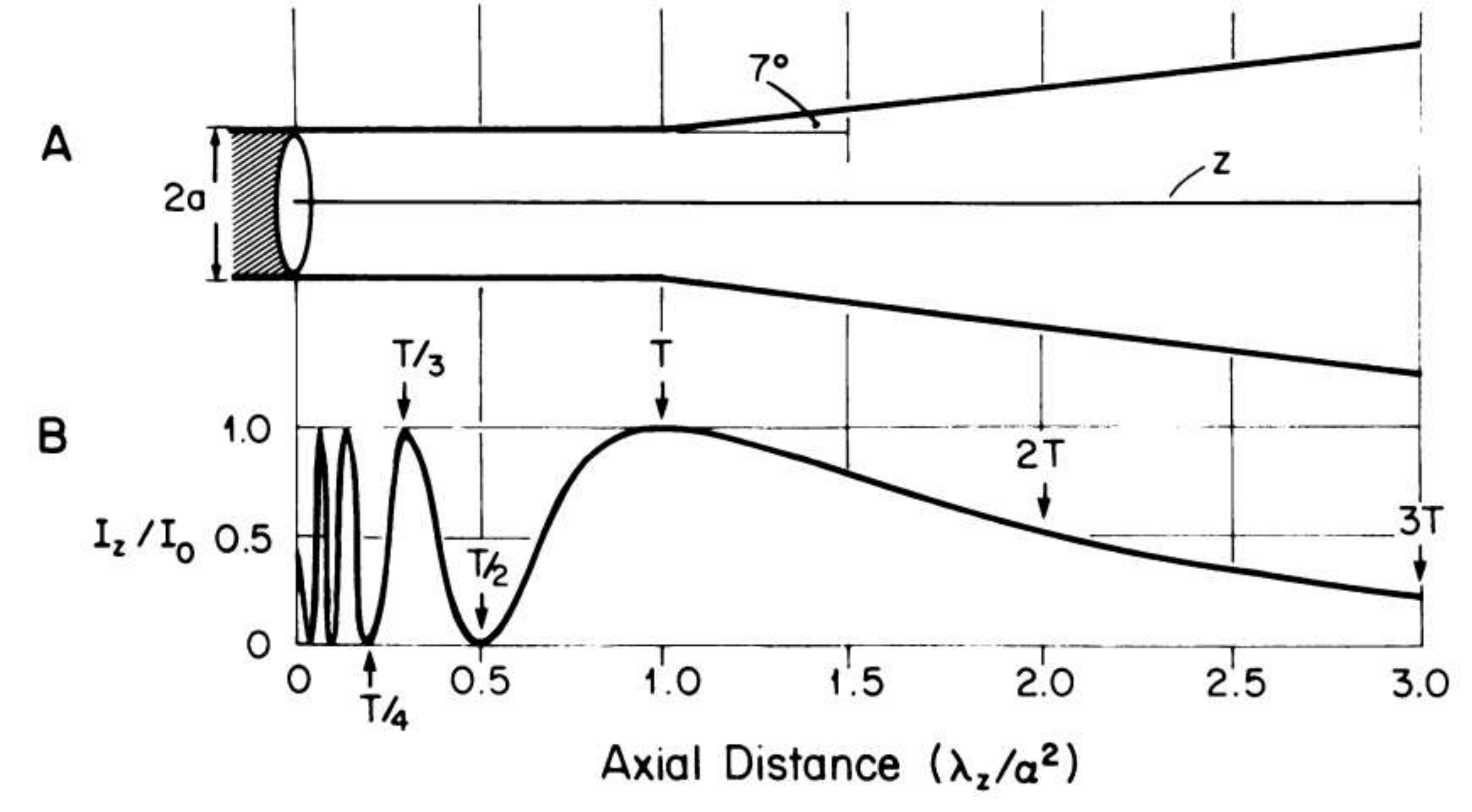
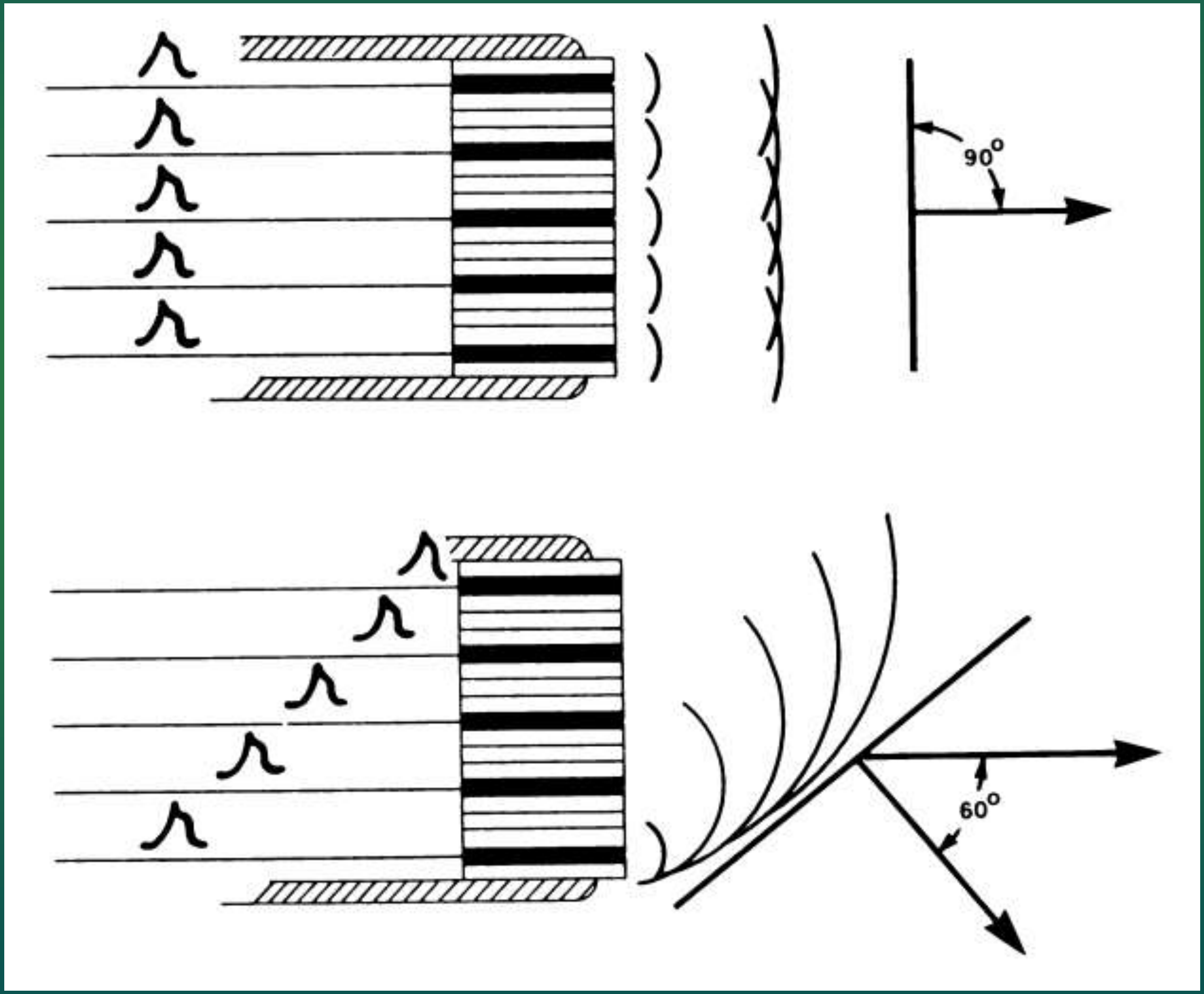
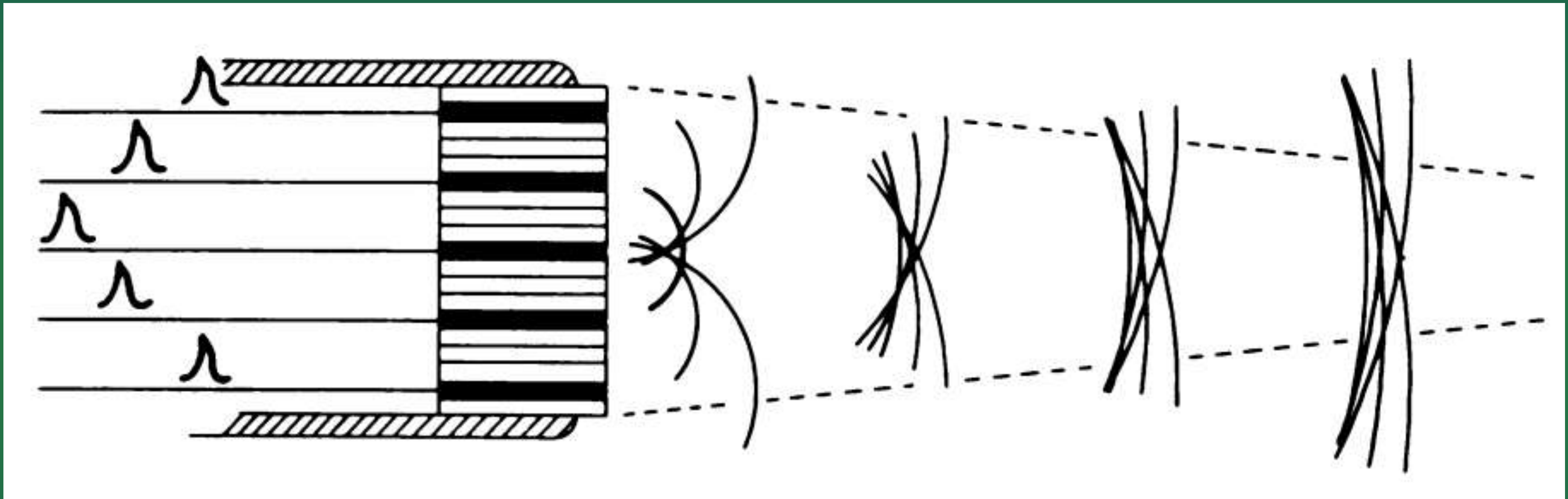
Echo Transducer Axial Variation



Phased Array Echo Transducer

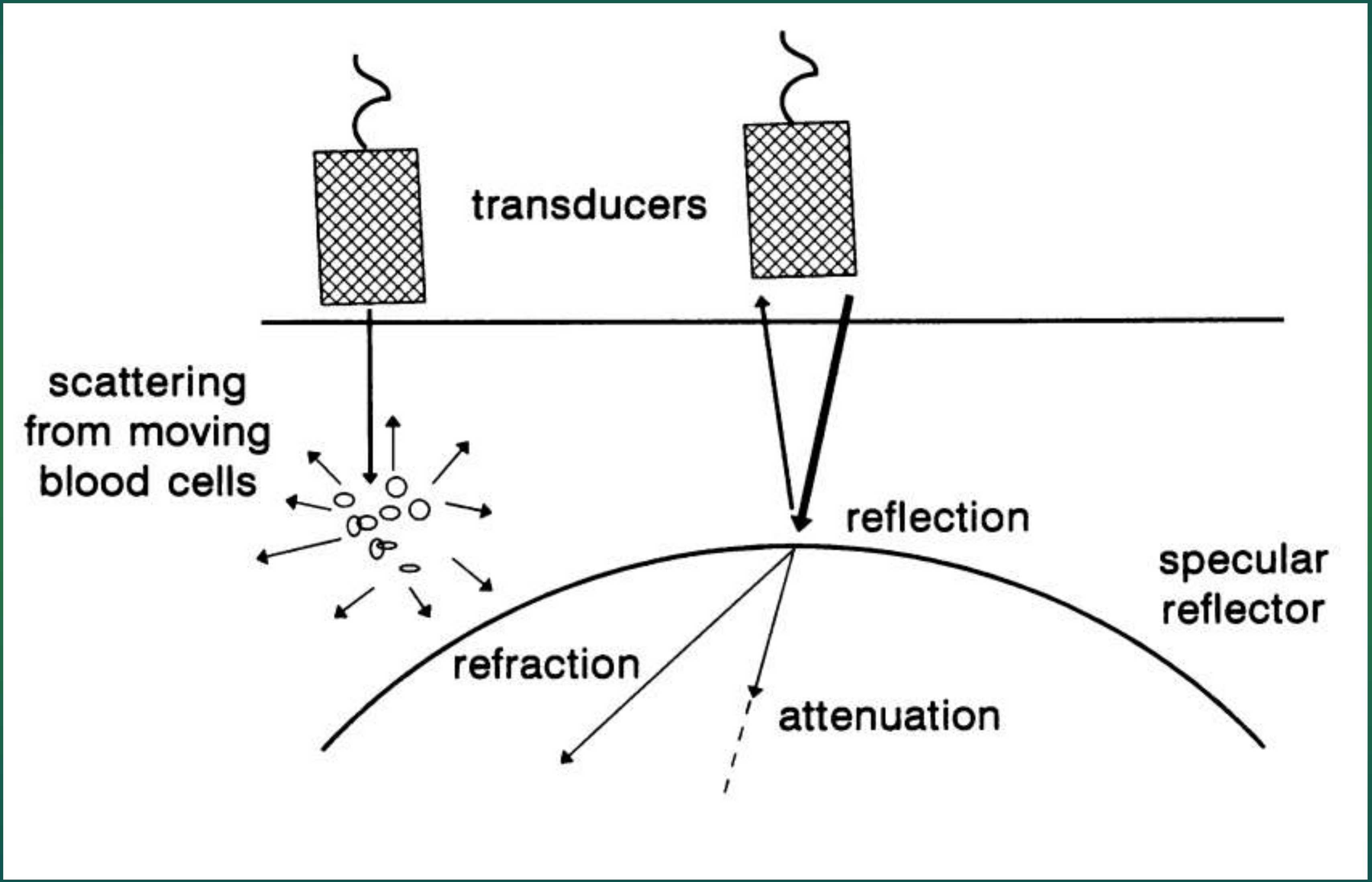


Phased Array Echo Transducer



- Effect of electronic focus

Destiny of Sound Wave



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Velocity of Sound in Air and Various Tissues

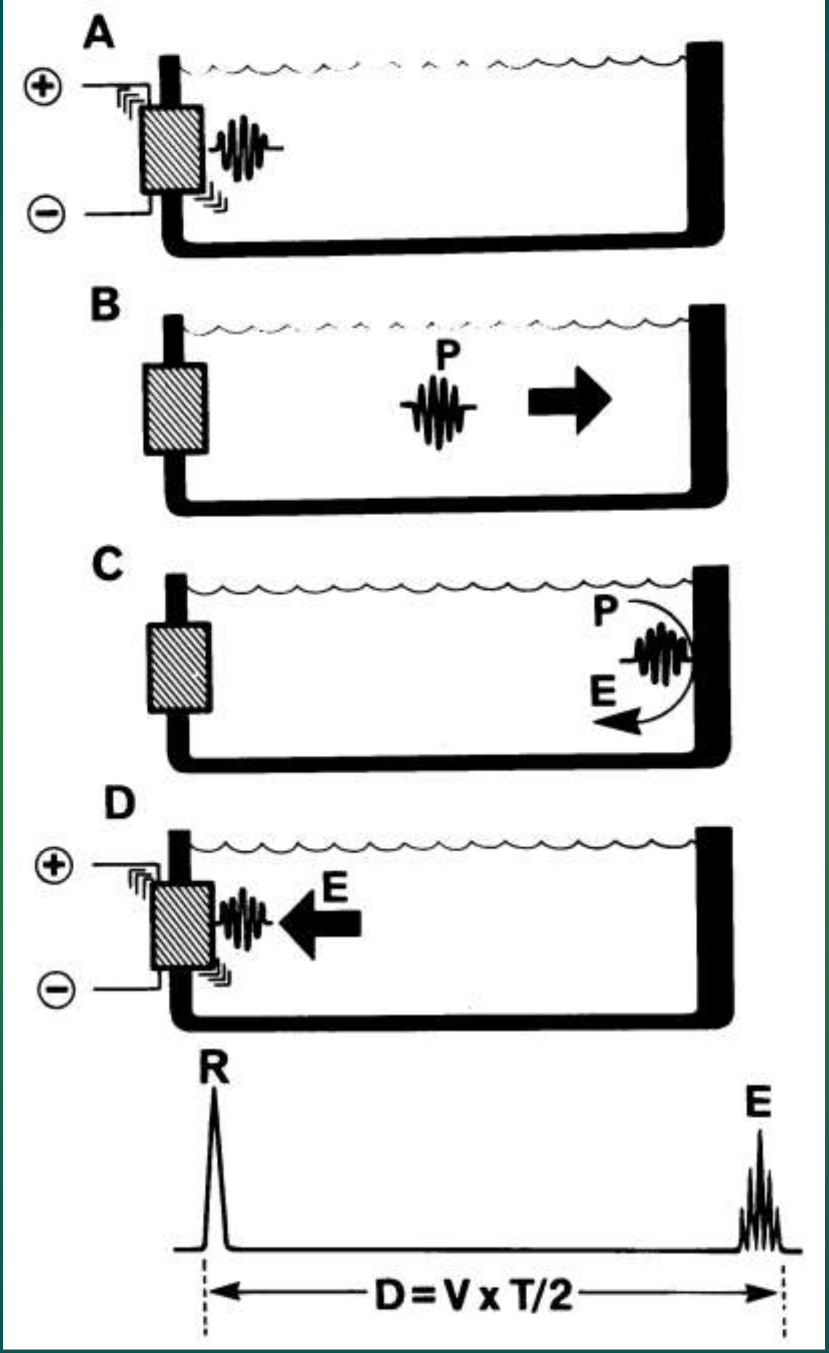
• Air	330 m/s
• Fat	1450 m/s
• Water	1480 m/s
• Soft tissue	1540 m/s
• Kidney	1560 m/s
• Blood	1570 m/s
• Muscle	1580 m/s
• Bone	4080 m/s

Clinically, use 1500 m/s

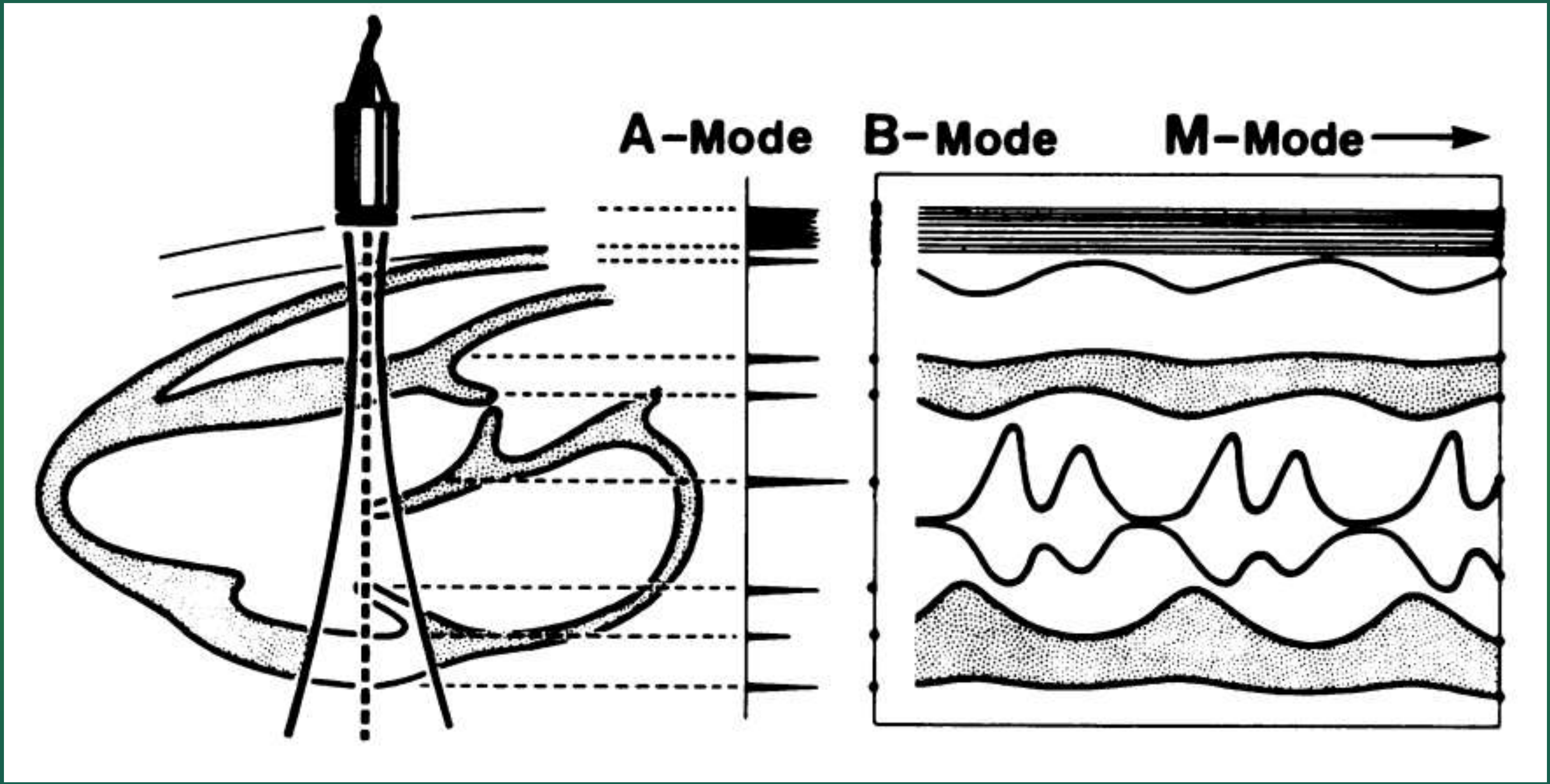
Feigenbaum, 6th ed, 2005, p. 13.

Velocity and Time Relation in Echocardiography

- Distance = rate * time
- Time = distance/rate
- Distance/rate (for 15 cm)=
  - 0.15m/1500m/sec =
  - .0001 sec, or 1/10,000 sec for one way trip
  - .0002 sec or 2/10,000 sec for 2 way trip
  - .0003 or 3/10,000 sec for 2 way trip of 20 cm



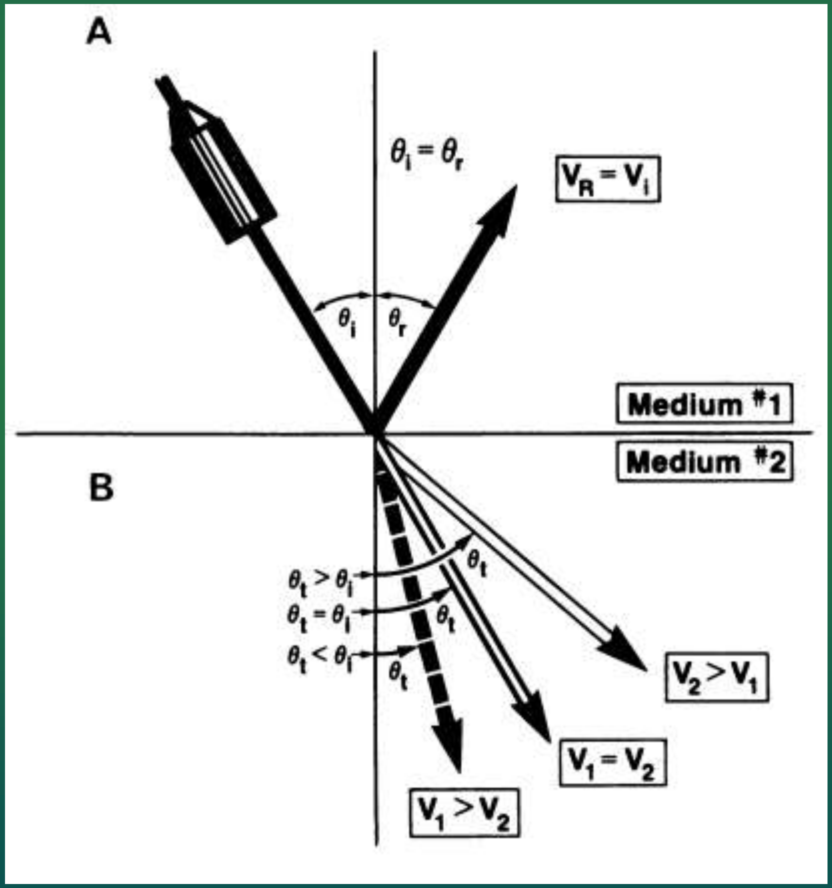
Display of Echo Signal



A = amplitude B = brightness M = motion

Echo Interfaces

- Velocity of sound in a medium depends on density (denser is faster) and elasticity of the medium
- Human tissue - 1540 m/sec, faster in bone
- Acoustic mismatch or change in acoustic impedance of an interface causes a reflection
- Interface perpendicular to beam is strongest



Echo Reflection

- **Specular reflection** - reflector is large and smooth relative to ultrasound wavelength - responsible for the echo images, angle of incidence is important
- **Scattered reflection** - reflector is small and rough relative to ultrasound wavelength - responsible for some images and critical for Doppler

Echo Resolution and Attenuation

- **Axial resolution** - better with higher frequency and fewer cycles/pulse (packet size) 3.5MHz=0.43mm wavelength
- **Lateral resolution** - varies with transducer size, shape, frequency, and focusing
- **Attenuation** - worse with high frequency
- **Attenuation** - half-value layer (35cm in blood, 3.6 cm in muscle)

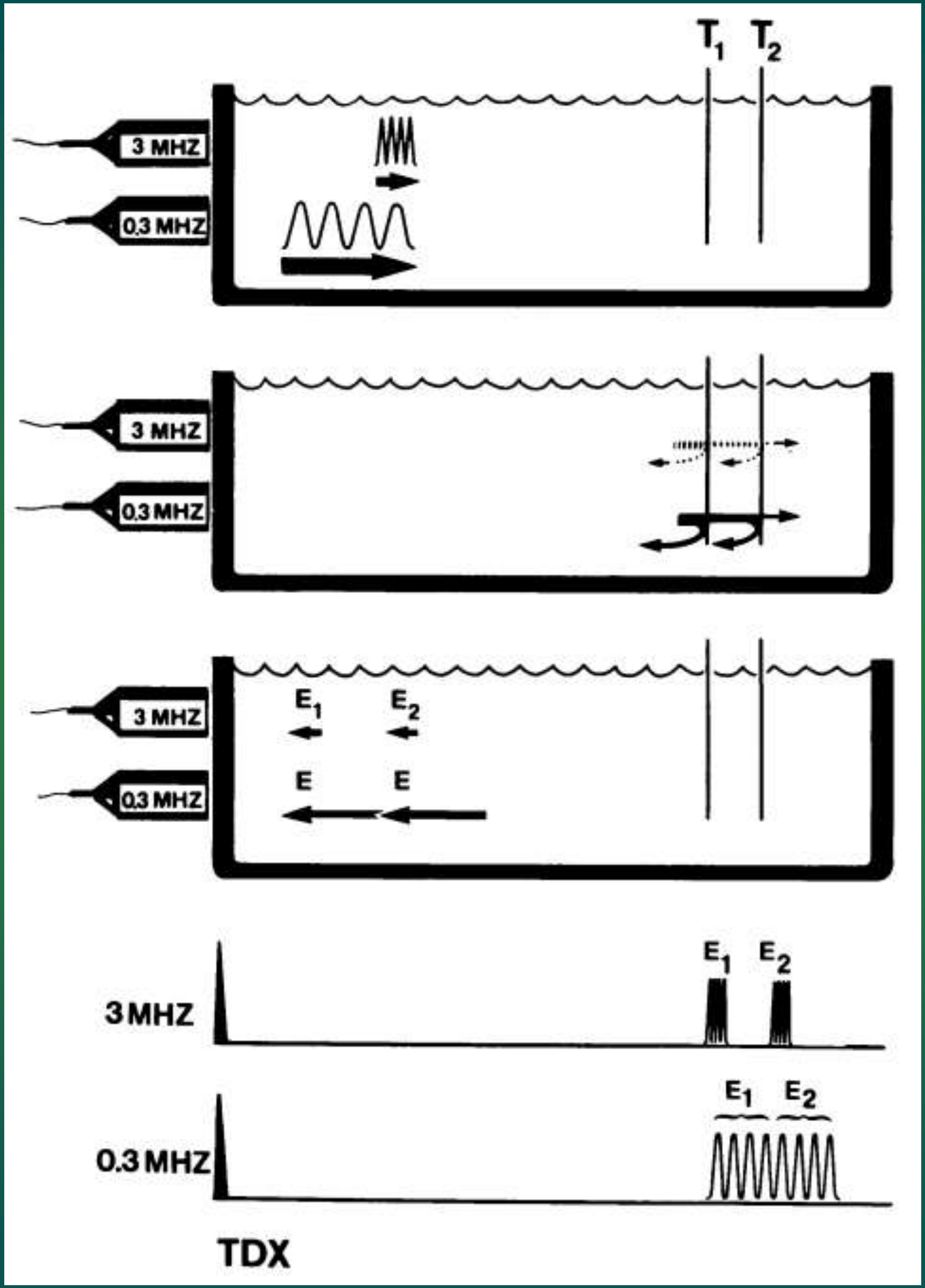
Echo Attenuation – Half-power Distance

• Water	380 cm
• Blood	15 cm
• Soft tissue (except muscle)	1-5 cm
• Muscle	0.6-1 cm
• Bone	0.2-0.7 cm
• Air	0.08 cm
• Lung	0.05 cm

Feigenbaum, 6th ed, 2005, p. 14.

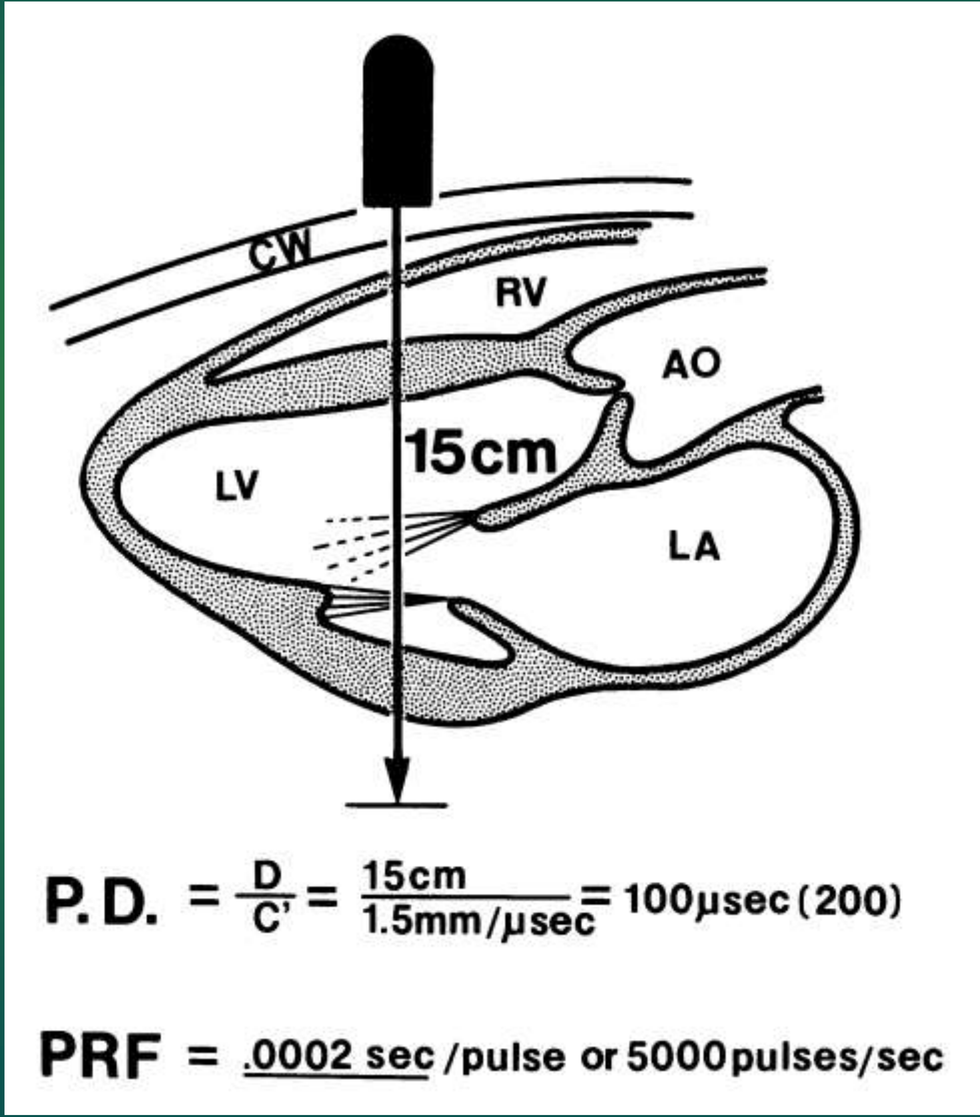
Axial Resolution

- Better with:
  - high frequency
  - short packet length

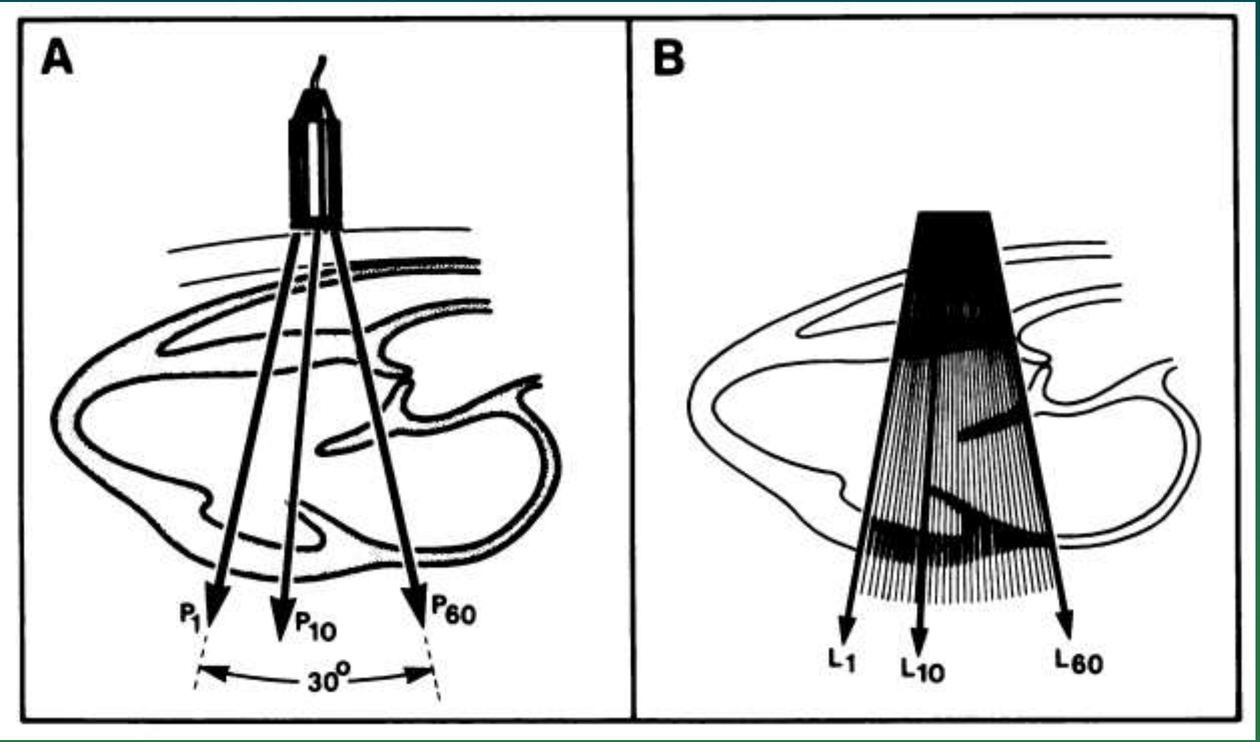


Time Constraint in Echo-Doppler

- 1540 m/sec in tissue
- 20 cm depth is 40 cm round trip
- 3850 round trips/sec (M-mode)
- 150 round trips for one image
- 25 images/sec
- Trade-off: temporal resolution, spatial resolution (line density) and depth

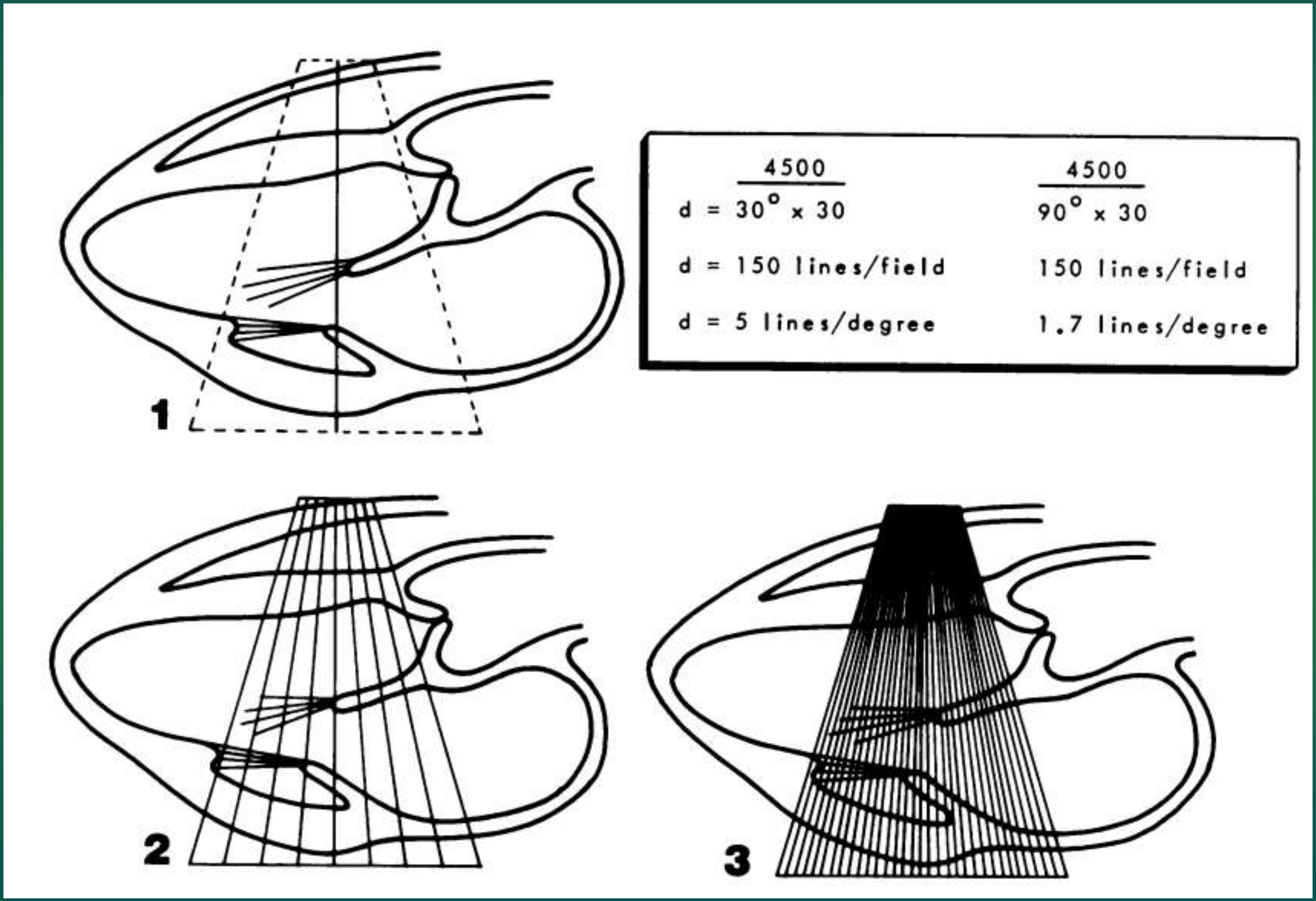


Time Constraint in Echo-Doppler

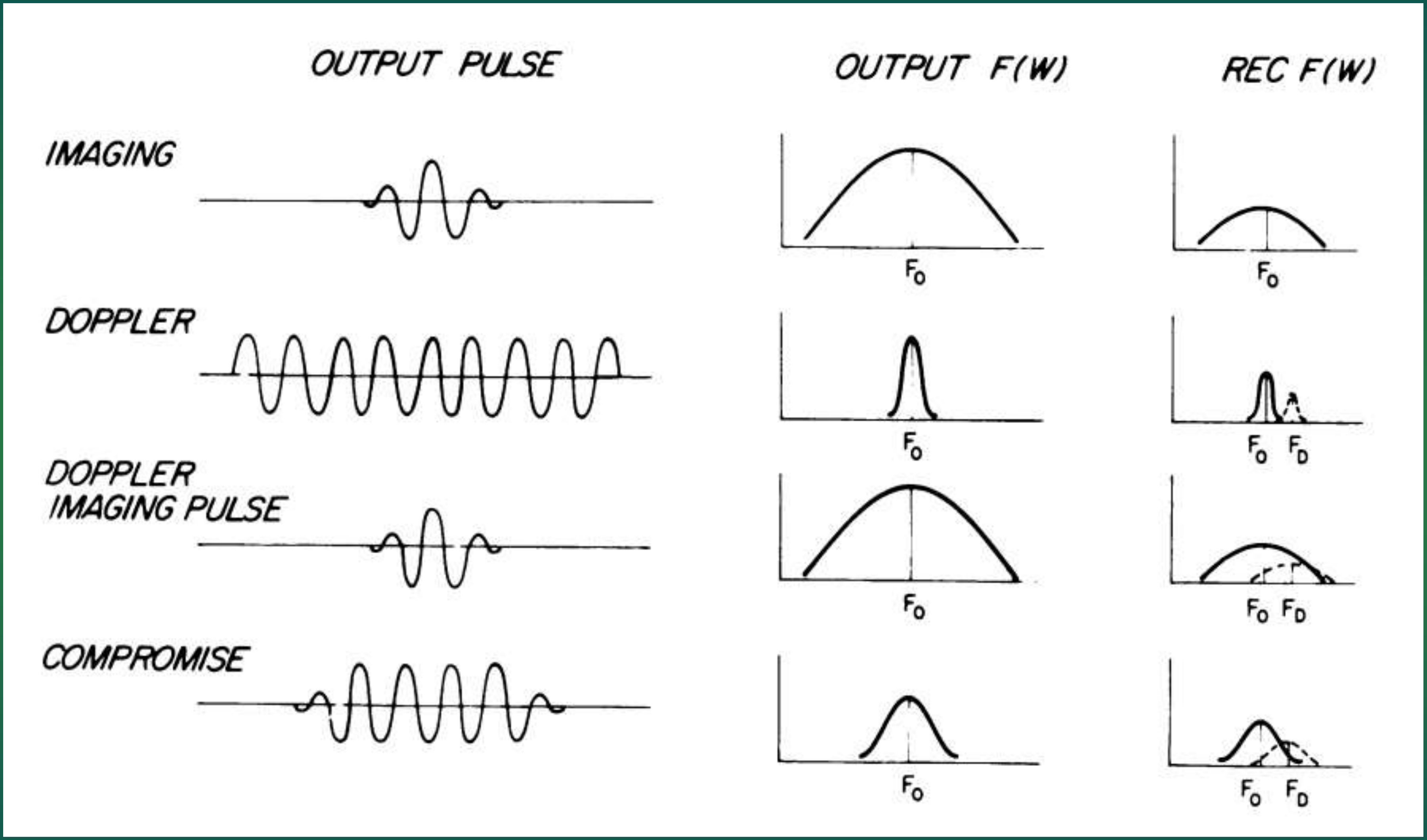


- 1540 m/sec in tissue, 20 cm depth is 40 cm round trip, 3850 round trips/sec (M-mode), 150 round trips for one image, 25 images/sec
- Trade-off: temporal resolution, spatial resolution (line density) and depth
- <u>In the "Res" mode, there is improved temporal resolution</u>

Time Constraint in Echo-Doppler



Constraints in Echo-Doppler



The larger packet size, the better Doppler discrimination, the worse the echo discrimination

Time Constraint in Echo-Doppler

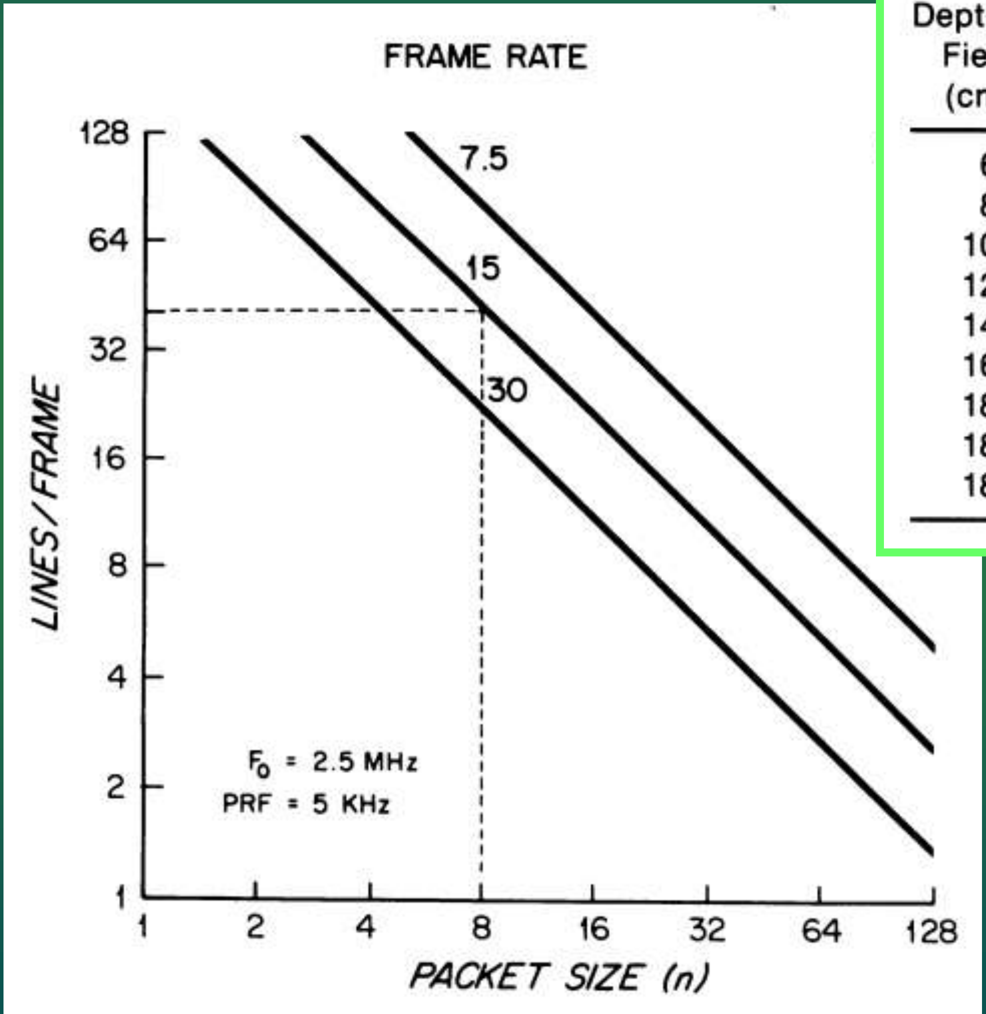
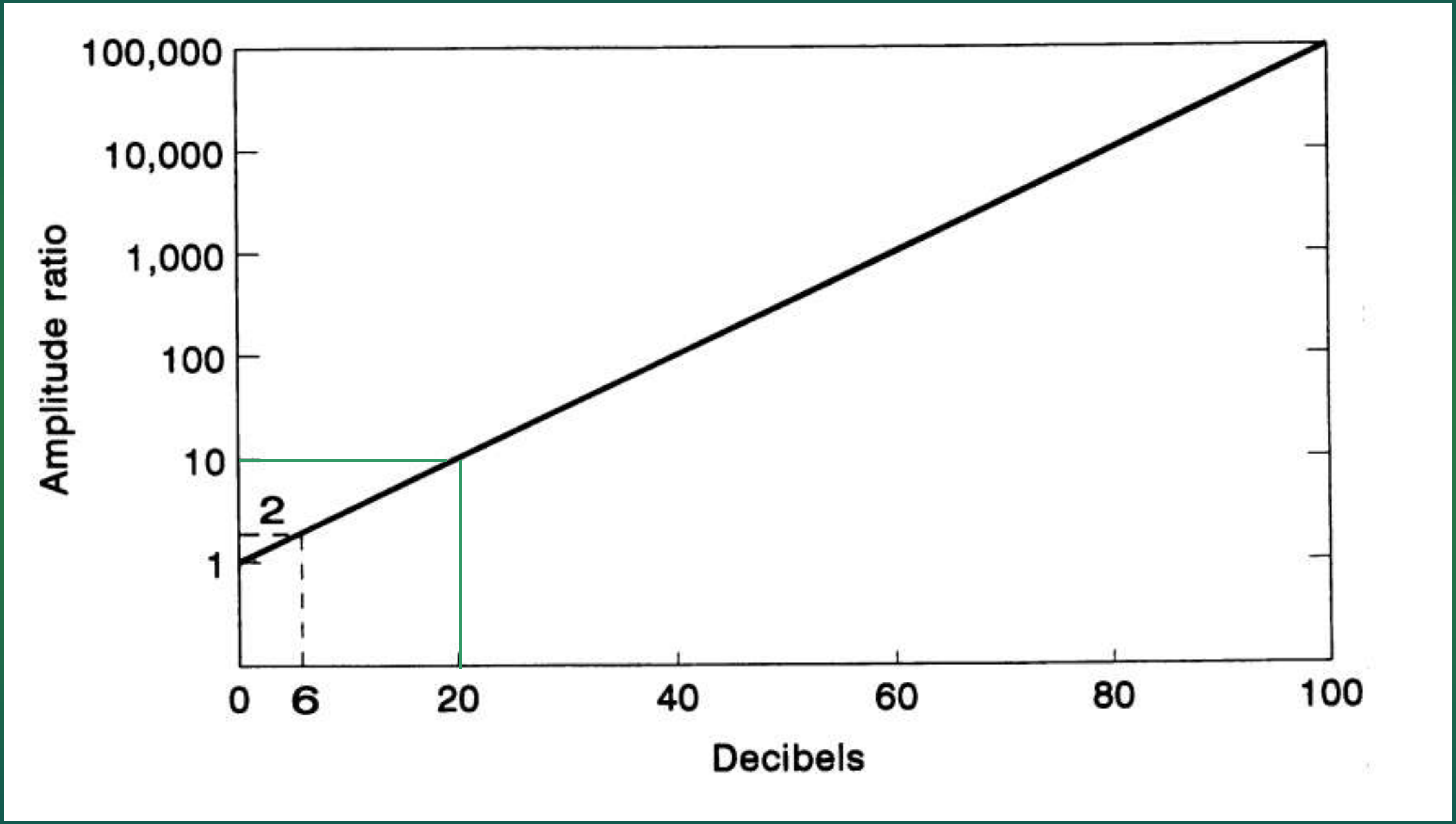


Table 11–2. Typical Combinations of Depth of Field, Packet Size, Lines per Frame, and Frame Rate for Color Flow Images

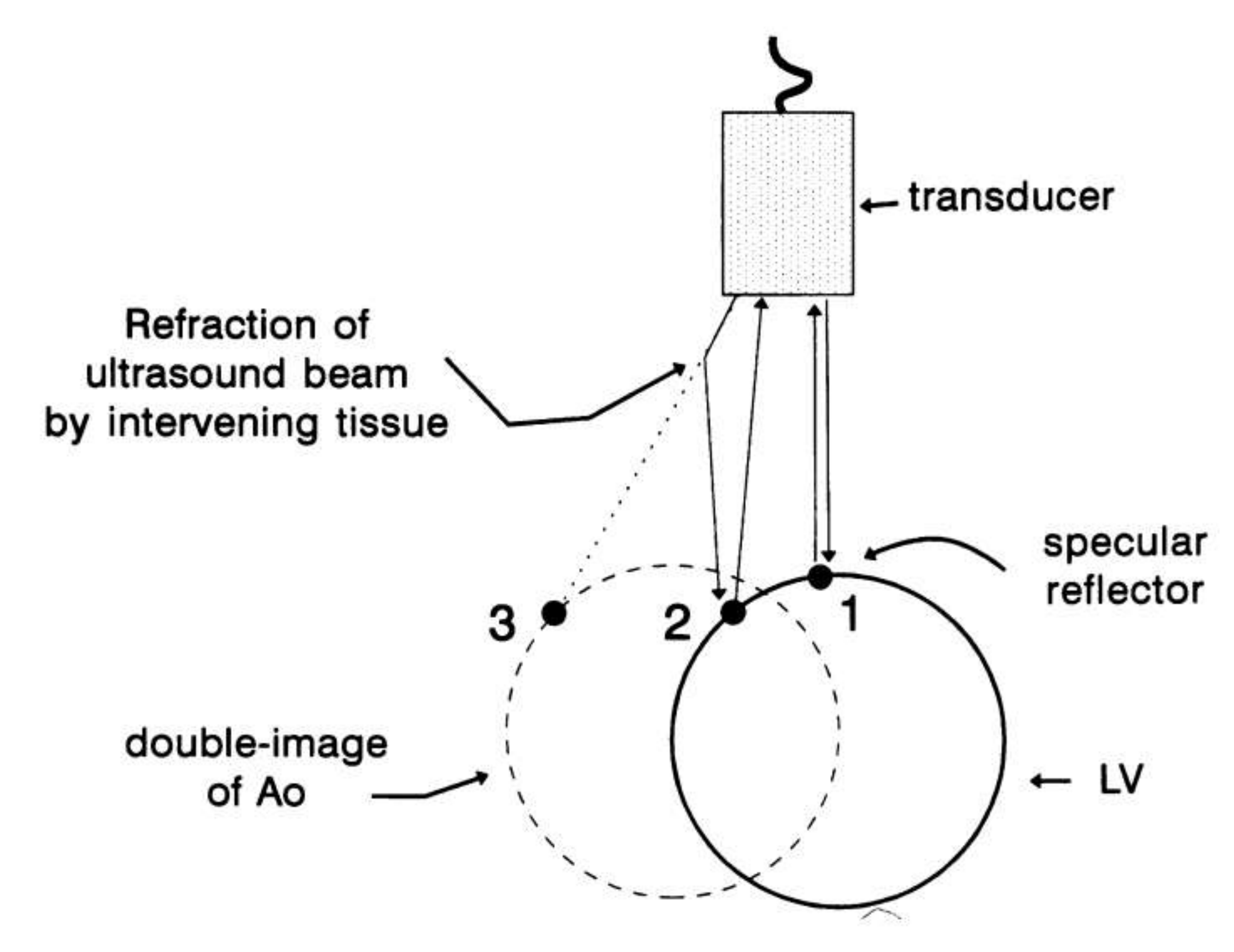
Depth of Field (cm)	Packet Size	Lines per Frame	Frames per Second
6	4	30	30
8	8	45	20
10	8	30	30
12	4	45	30
14	8	30	20
16	8	45	15
18	4	30	30
18	4	45	20
18	8	45	12

Decibel Graph



Otto CM. Textbook of Clinical Echocardiography. 2000.

Artifact From Scatter



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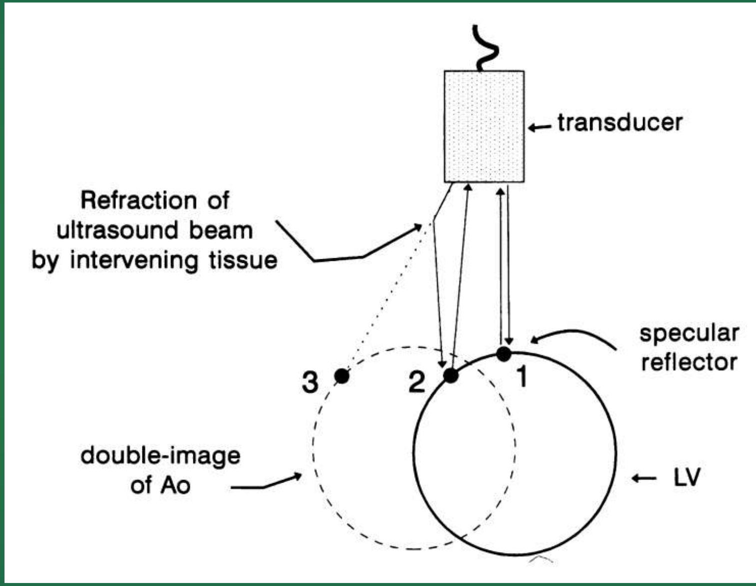
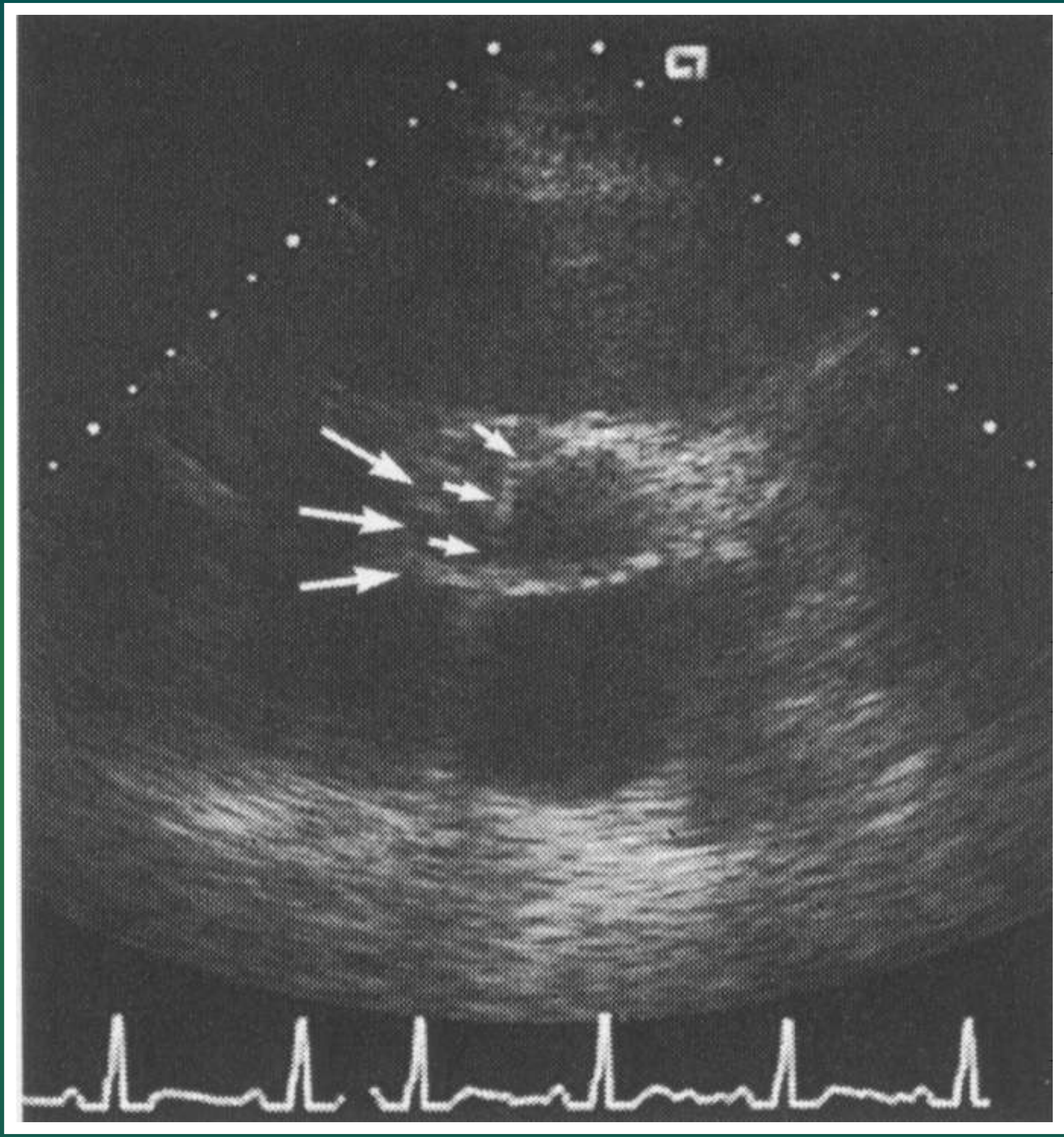
Artifact From Scatter





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Breakthrough: Harmonic Imaging

- Since about 1997 with the introduction of harmonic imaging, there has been a dramatic improvement in image quality
- Today, essentially all images are obtained with harmonic imaging
- Sound travels a little faster at the peak of the sound wave (more compressed) than the trough, so with each subsequent waveform a small amount of harmonic is generated (1962), similar to the breaking of the crest of a wave at the beach

Thomas JD Rubin DN. JASE 1998; 11:803.

2 Types of Harmonic Imaging

- Harmonic energy in reflection can occur with echo-contrast agents which resonate with ultrasound stimulation and produce harmonic emission
- Harmonic energy in transmission occurs due to the compressibility of the tissue

Thomas JD Rubin DN. JASE 1998; 11:803.

Harmonic Imaging

Classical ultrasound theory:

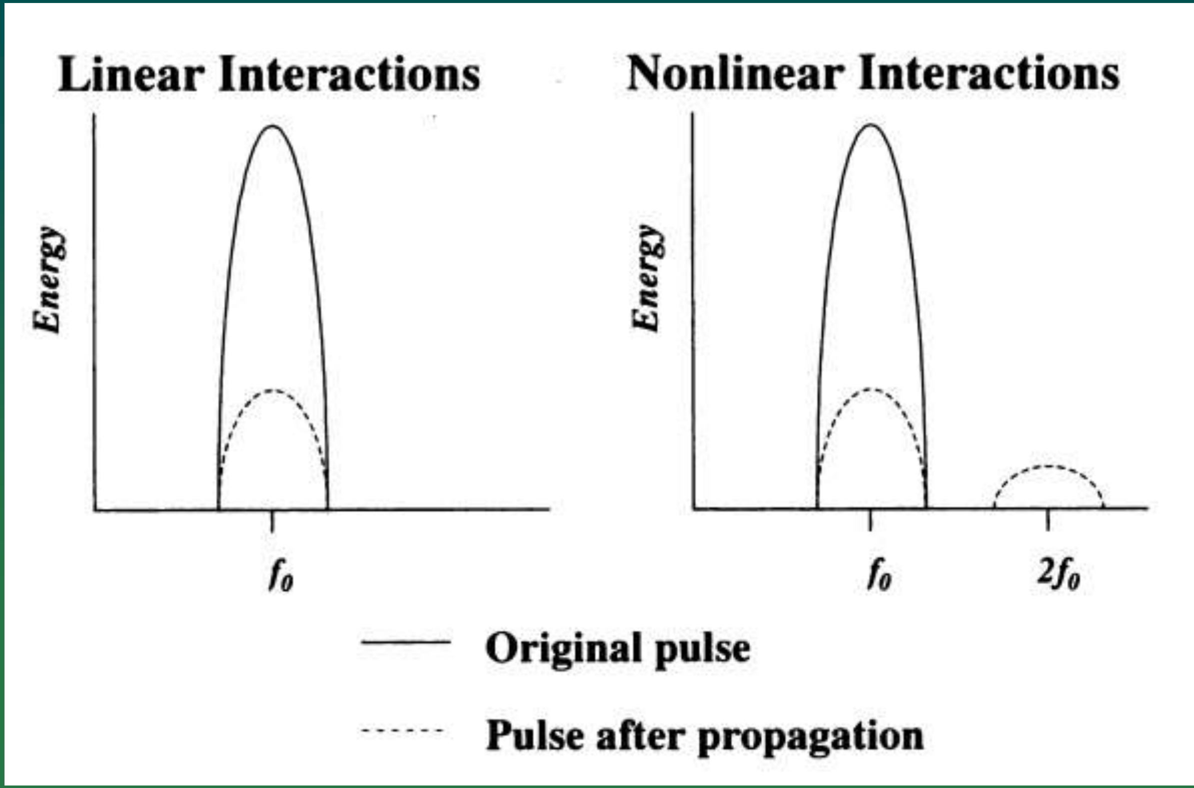
- Energy propagation is linear
- Different frequencies travel at the same speed in the same medium
- New frequencies should not appear
- Attenuation only reduces amplitude

Harmonics:

- Some objects in the path of the beam may resonate and emit higher frequencies
- Transmission of ultrasound through a compressible medium yields harmonics
- Signal grows with distance
- Signal generation is nonlinear
- Reduced backscatter

Thomas JD Rubin DN. JASE 1998; 11:803.

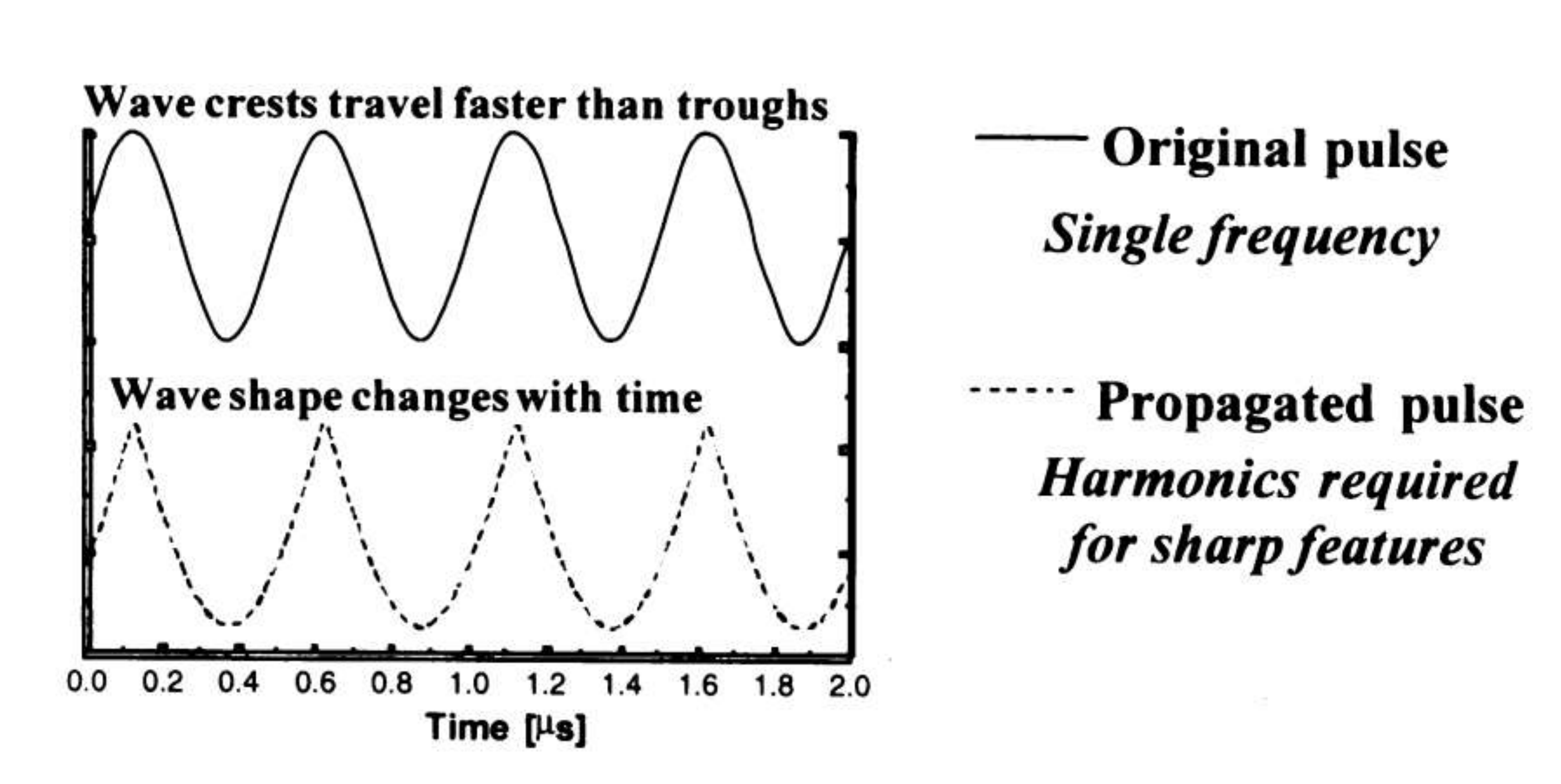
Harmonic Imaging



- The reflected pulse has harmonic information

Thomas JD Rubin DN. JASE 1998; 11:803.

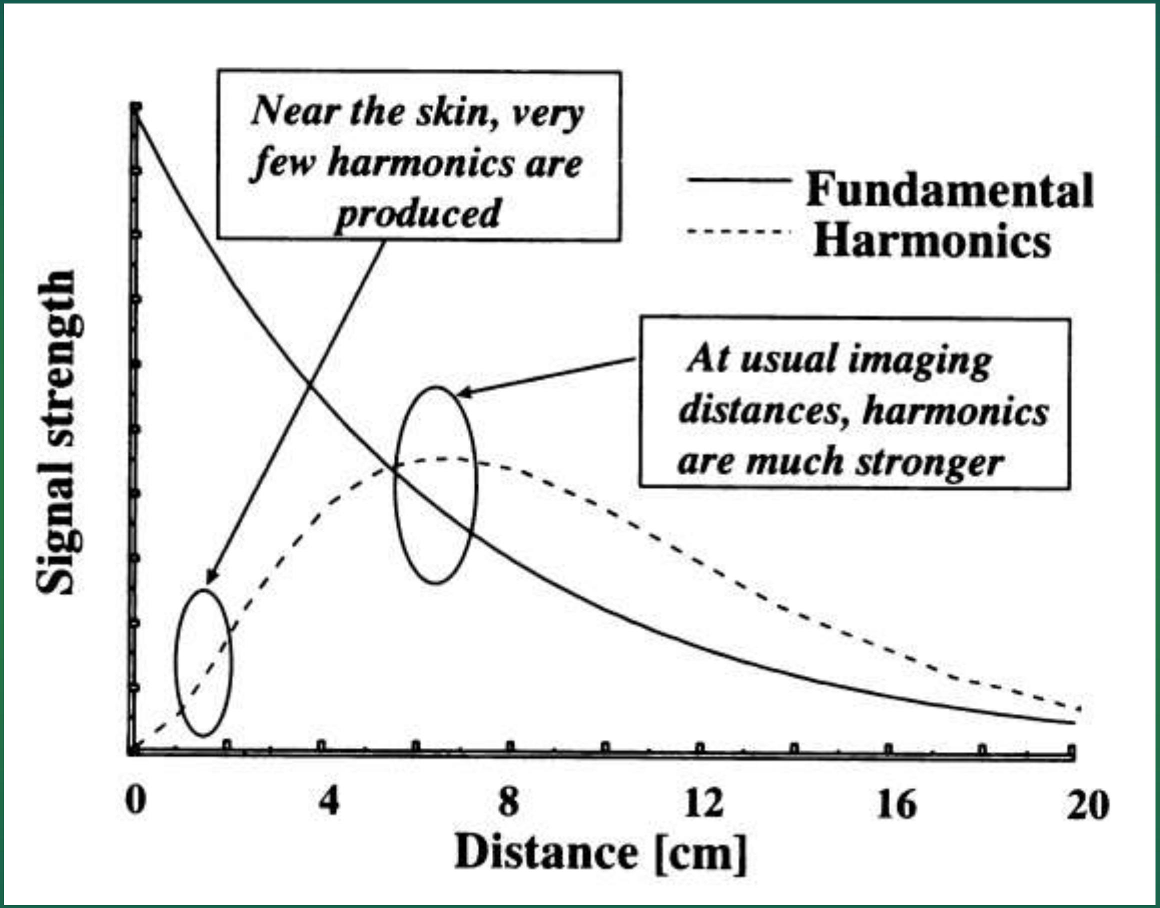
Harmonic Imaging



- The propagated pulse is distorted by the addition of harmonic energy

Thomas JD Rubin DN. JASE 1998; 11:803.

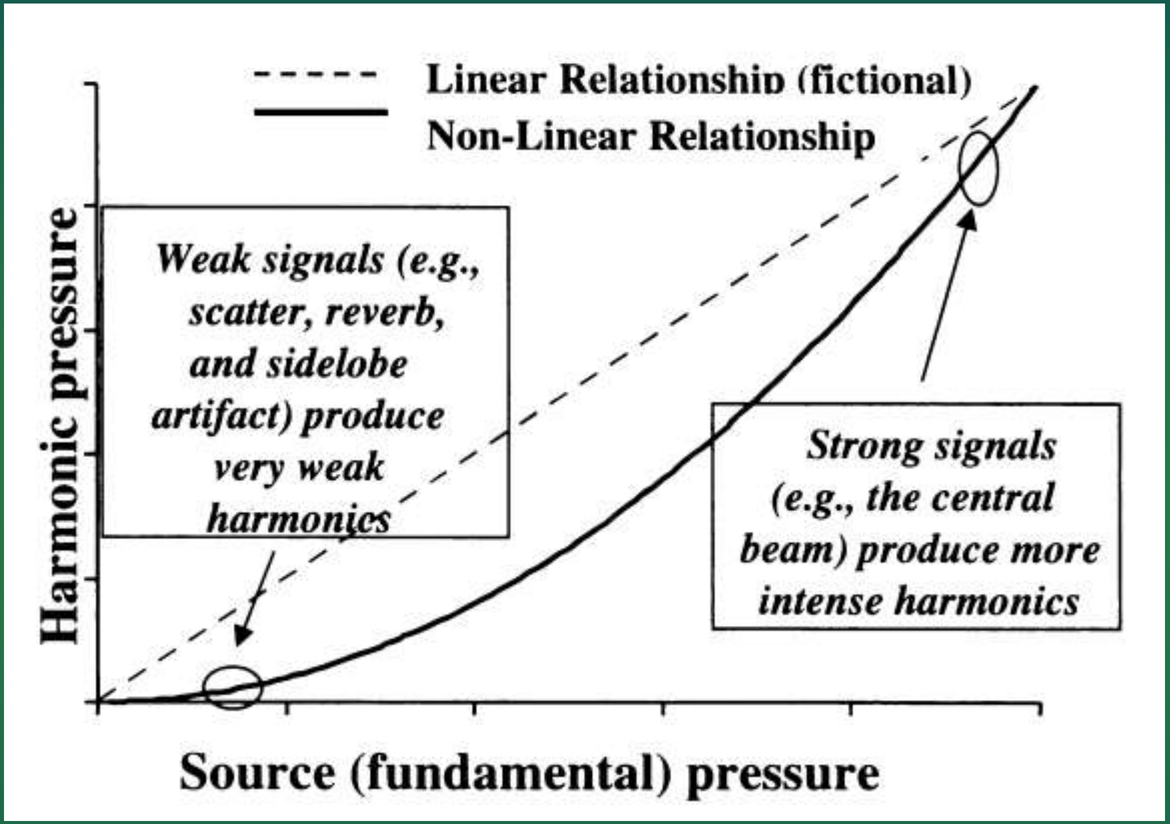
Harmonic Imaging



- The harmonic energy increases with depth

Thomas JD Rubin DN. JASE 1998; 11:803.

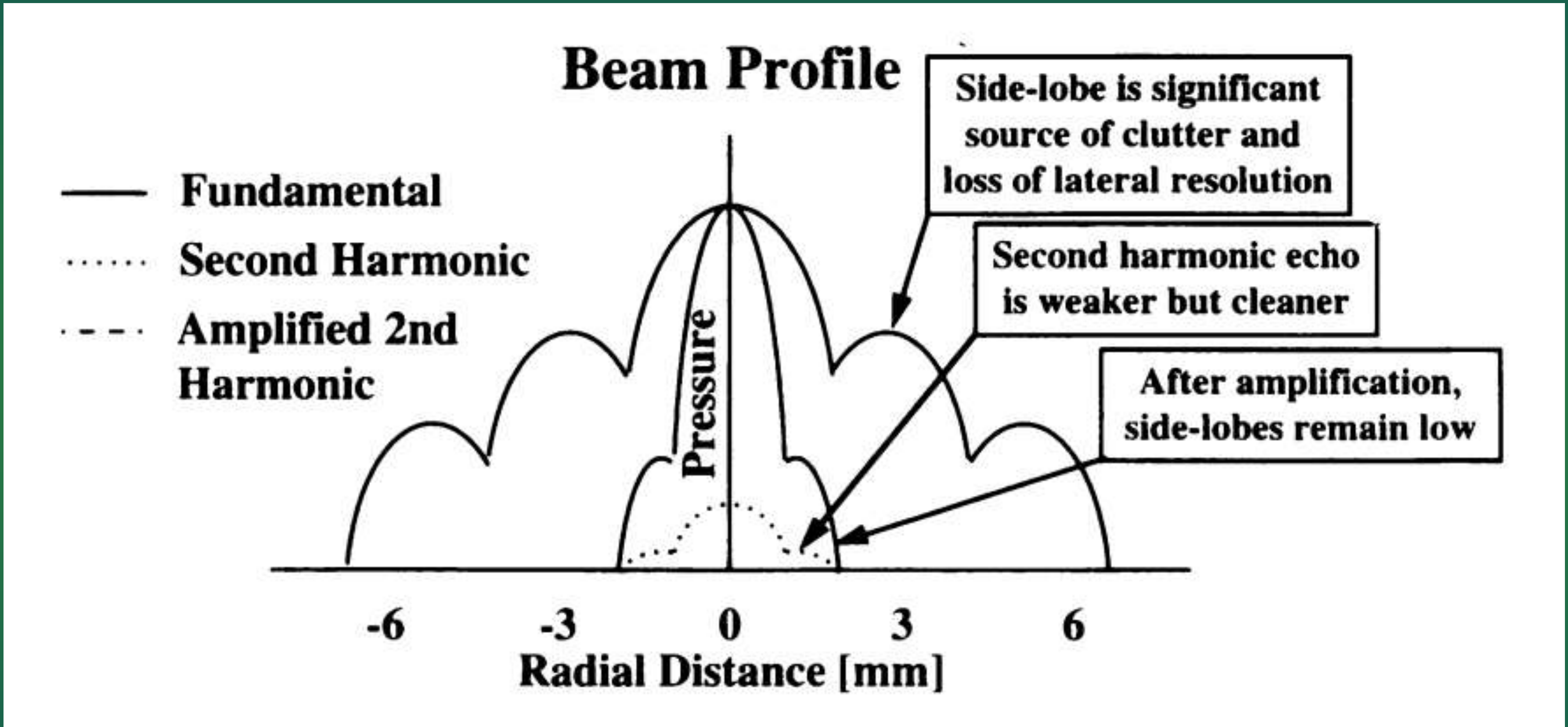
Harmonic Imaging



- Strength of harmonics increases as square of source energy, so weak reflections produce poor harmonics

Thomas JD Rubin DN. JASE 1998; 11:803.

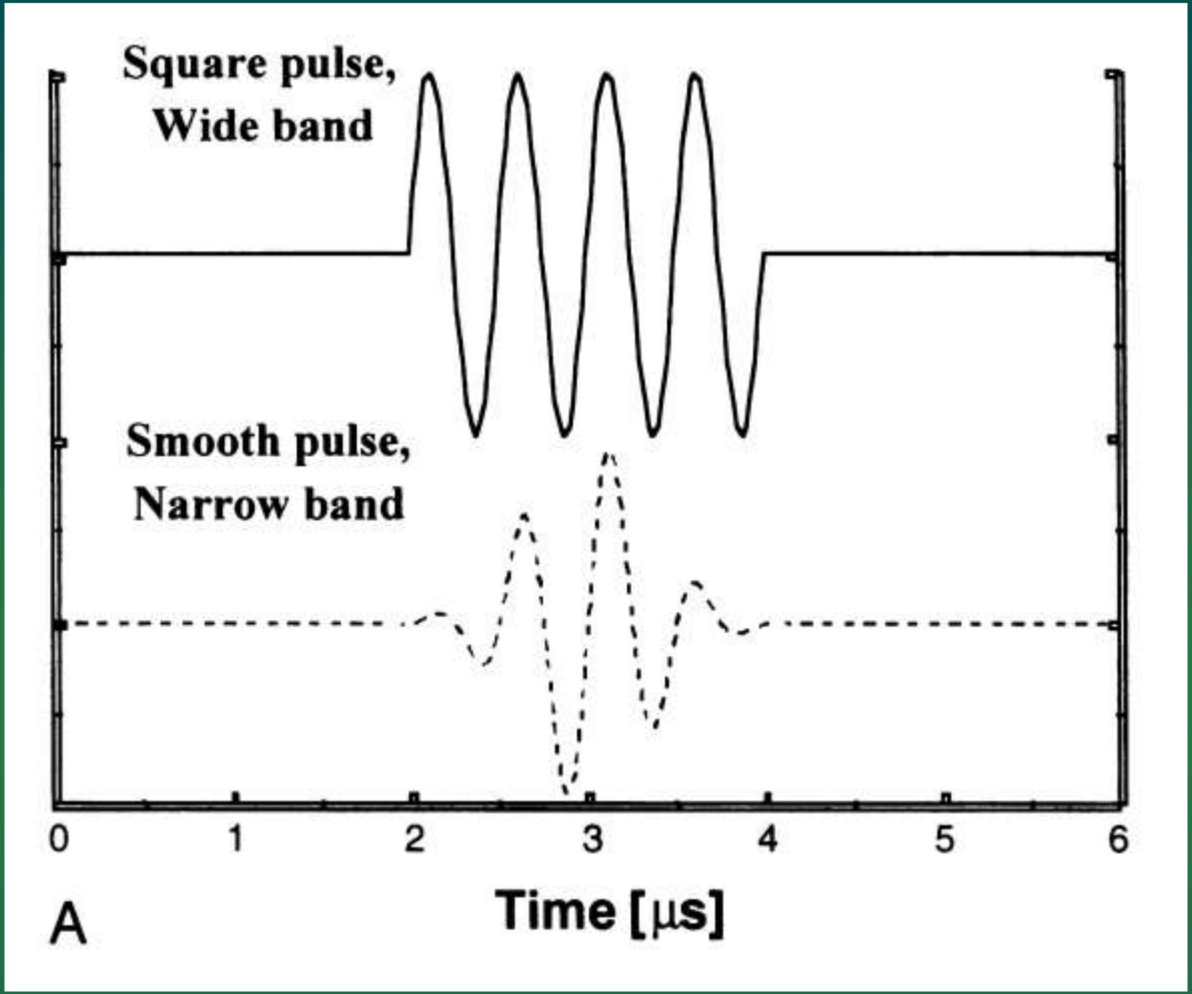
Harmonic Imaging



- Boosting of harmonic information is required

Thomas JD Rubin DN. <u>JASE</u> 1998; 11:803.

Harmonic Imaging

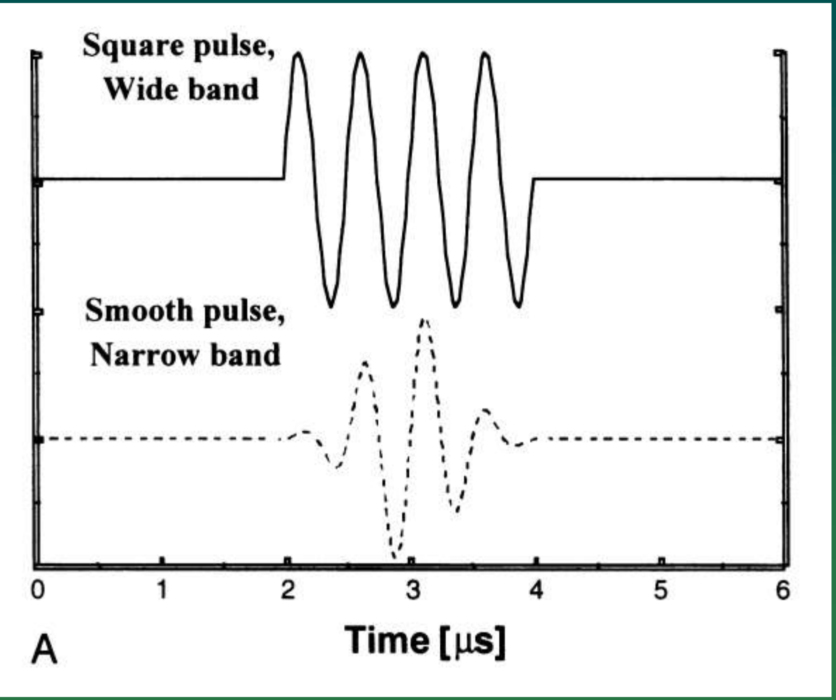


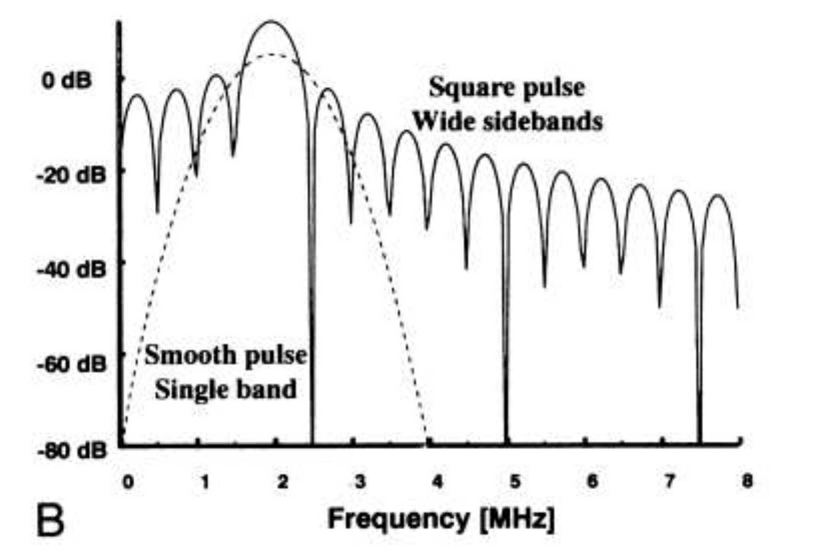
- Narrow band pulse is smoother in initiation and termination

Thomas JD Rubin DN. JASE 1998; 11:803.

Harmonic Imaging

- The square pulse (wide band) has significant energy at many frequencies, including at the harmonic frequency
- In contrast, the smooth pulse (narrow band) has almost no energy at the harmonic frequency





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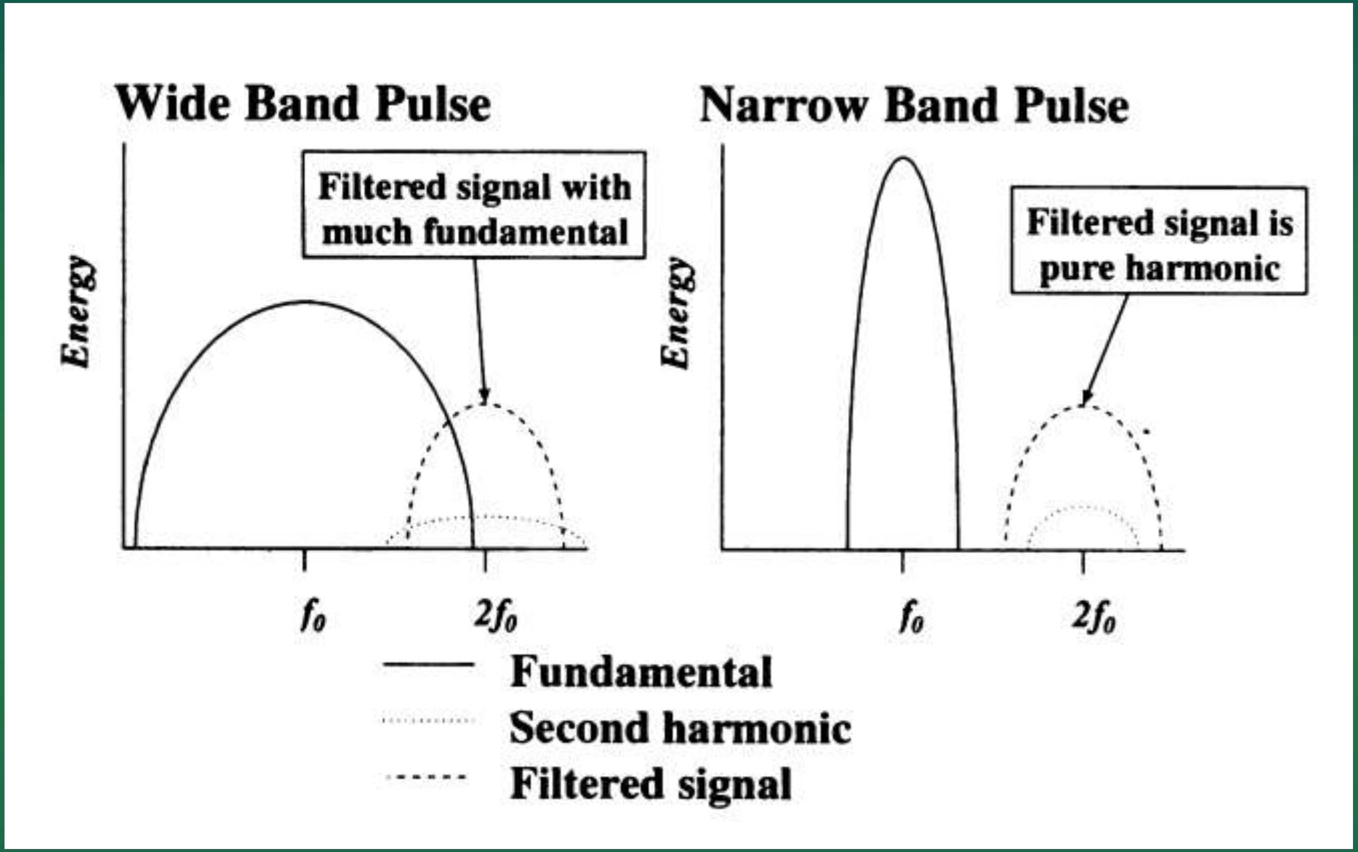
Harmonic Imaging



- Narrow band pulse allows filtered signal to be essentially free of fundamental frequency reflection

Thomas JD Rubin DN. JASE 1998; 11:803.

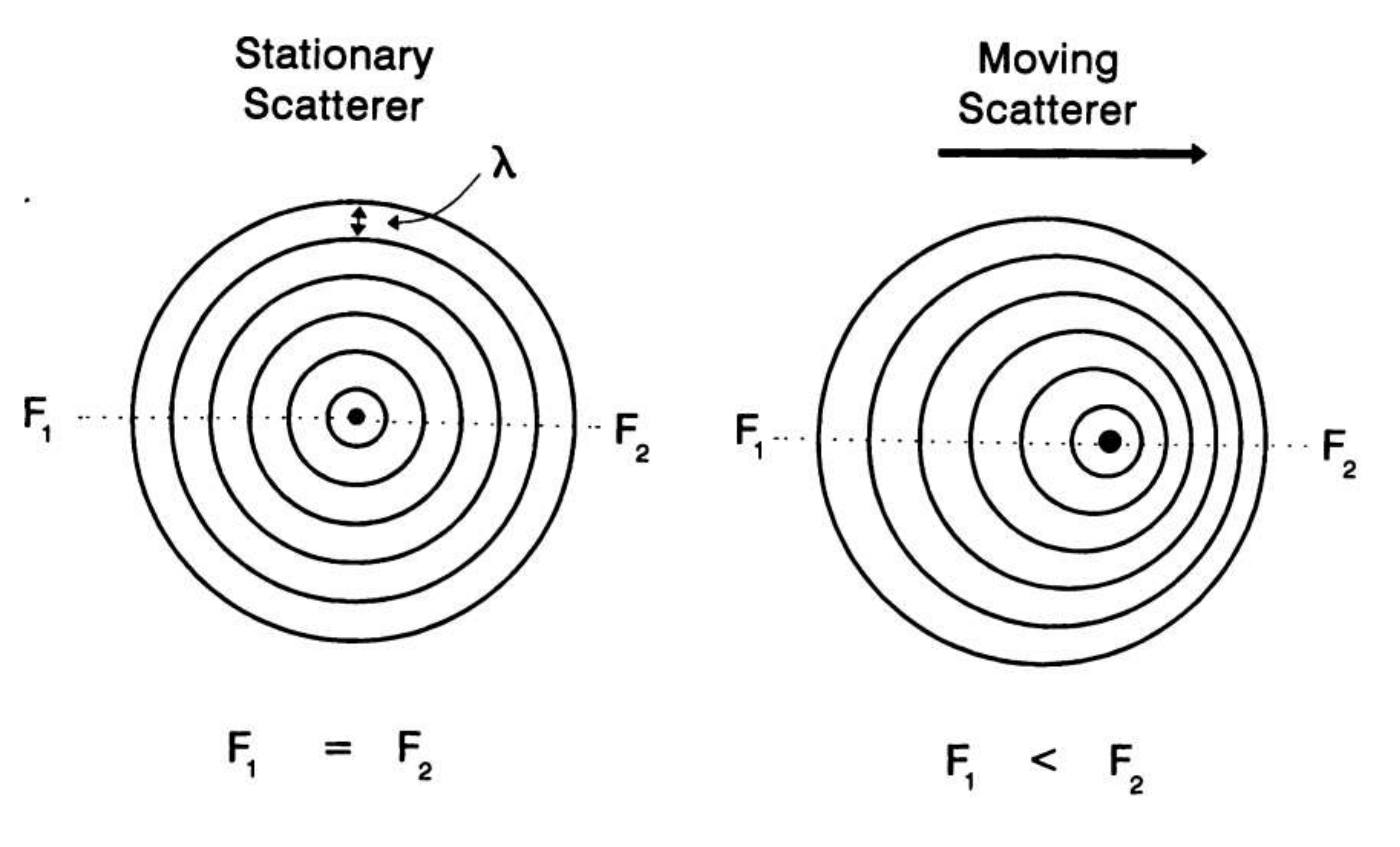
New Instrumentation: High PRF Equipment

- The classical time constraints explained earlier are rendered nonconstraining by the process of multiple simultaneous scan line analysis
- I don’t understand it

Doppler Cardiography

- Primary source of ultrasonic reflection - RBC
- Scattered reflector
- Motion with respect to transducer causes shift in frequency of sound wave, the measurement of which is fundamental to the Doppler signal

Doppler Shift



Otto CM. Textbook of Clinical Echocardiography. 2000.

Doppler Shift

$$\text{Doppler shift} = (F_S - F_T)$$

F_T is frequency transmitted, and F_S is scattered frequency received

Doppler Equation

$$v = \frac{c\,(F_S - F_T)}{2\,F_T(\cos\theta)}$$

V is velocity of blood
C is speed of sound in blood (1540m/s)
Θ is intercept angle between beam and blood flow
2 is factor to correct for 2 trips

Otto CM. Textbook of Clinical Echocardiography. 2000.

Doppler Equation

$$v = \frac{c\,(F_S - F_T)}{2\,F_T(\cos\theta)}$$

V is velocity of blood
C is speed of sound in blood (1540m/s)
F_S-F_T is Doppler shift
Θ is intercept angle between beam and blood flow
2 is factor to correct for 2 trips

V=[1540m/s*(2.009-2.000)MHz]/2*2.000MHz(1.0)=
1540*.009/4=3.465 m/sec

Otto CM. Textbook of Clinical Echocardiography. 2000.

Doppler Equation

$$v = \frac{c\,(F_S - F_T)}{2\,F_T(\cos\theta)}$$

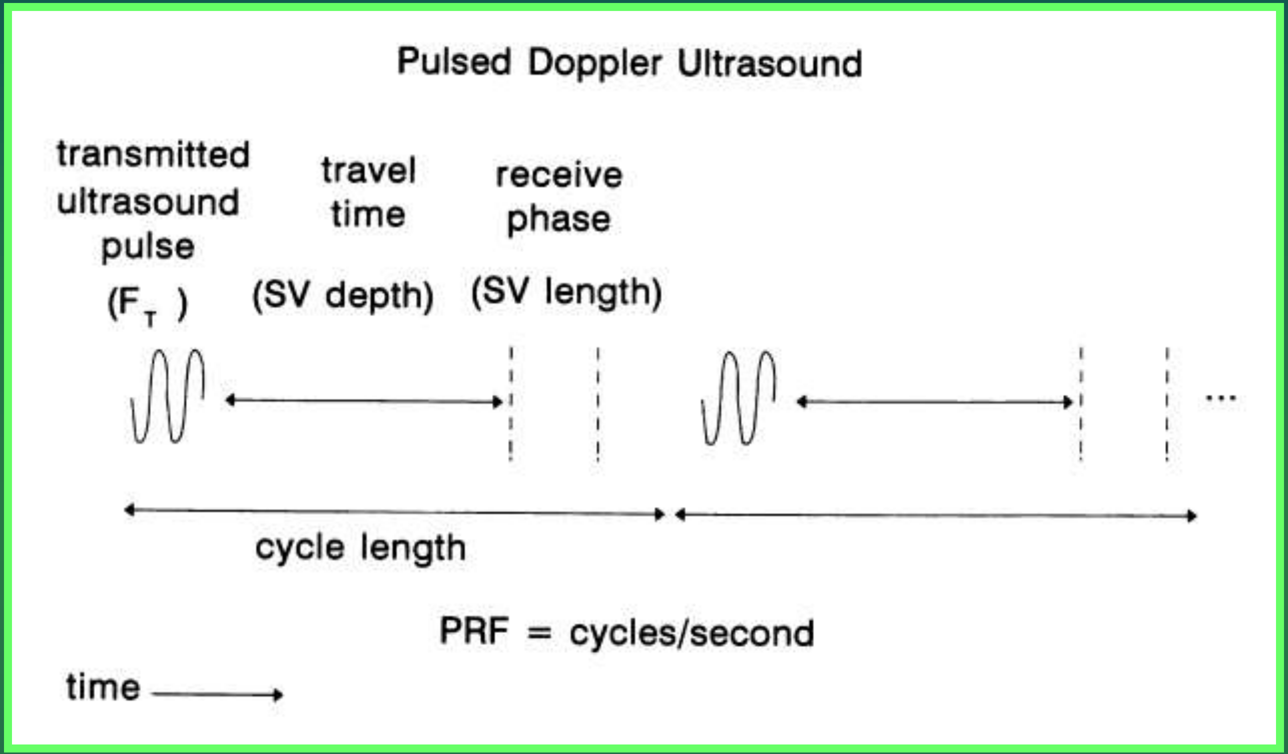
V is velocity of blood
C is speed of sound in blood (1540m/s)
F_S-F_T is Doppler shift
Θ is intercept angle between beam and blood flow
2 is factor to correct for 2 trips

Ignoring the speed of sound and cosine components, a direct relationship exists between Doppler shift & transmit frequency & blood velocity:
1.3KHz shift for 1.0 MHz per 1.0 m/sec velocity
2.6KHz shift for 1.0 MHz per 2.0 m/sec velocity
2.6KHz shift for 2.0 MHz per 1.0 m/sec velocity
5.2KHz shift for 2.0 MHz per 2.0 m/sec velocity

Otto CM. Textbook of Clinical Echocardiography. 2000.

Timing in Pulsed Doppler: The PRF

PRF is mainly limited by depth in Pulsed Doppler Ultrasound

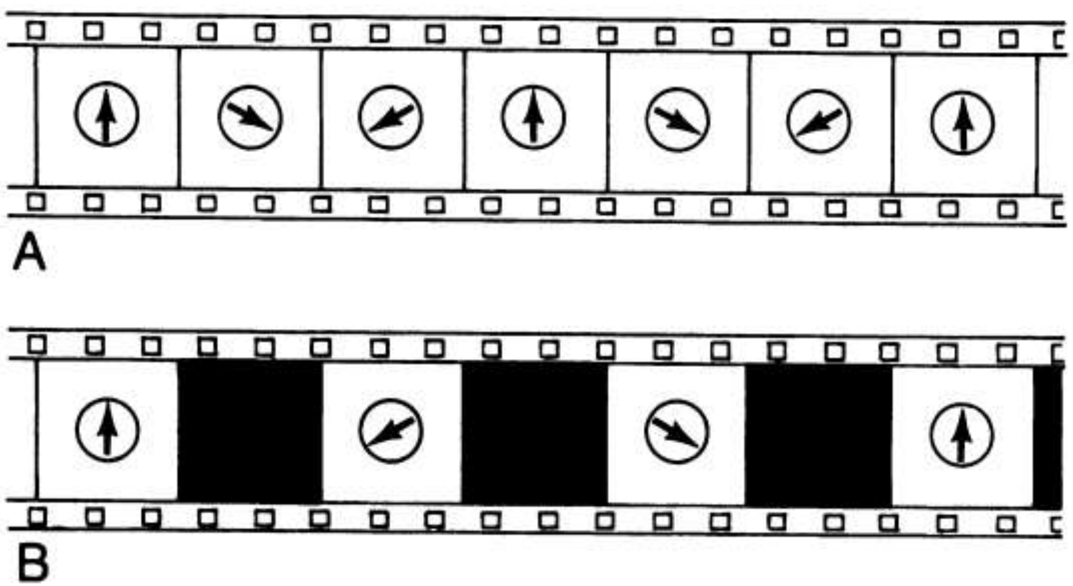


<u>Pulse Cycle Consists of Three periods</u>:

- Transmission (duration affects velocity resolution)
- Travel time (duration determines depth)
- Reception (duration determines sample volume)

Otto CM. <u>Textbook of Clinical Echocardiography</u>. 2000.

Timing in Pulsed Doppler



A: Sampling at three times the cycle rate, apparent direction is clockwise

B: With sampling at less than twice the cycle rate, apparent direction is counterclockwise

A waveform must be sampled at least twice in each cycle for accurate determination of wavelength.
Therefore, the maximum detectable <u>frequency shift</u> (the Nyquist Limit) is one-half the PRF. But the maximal detectable <u>velocity</u> depends on the equation.

Otto CM. <u>Textbook of Clinical Echocardiography</u>. 2000.

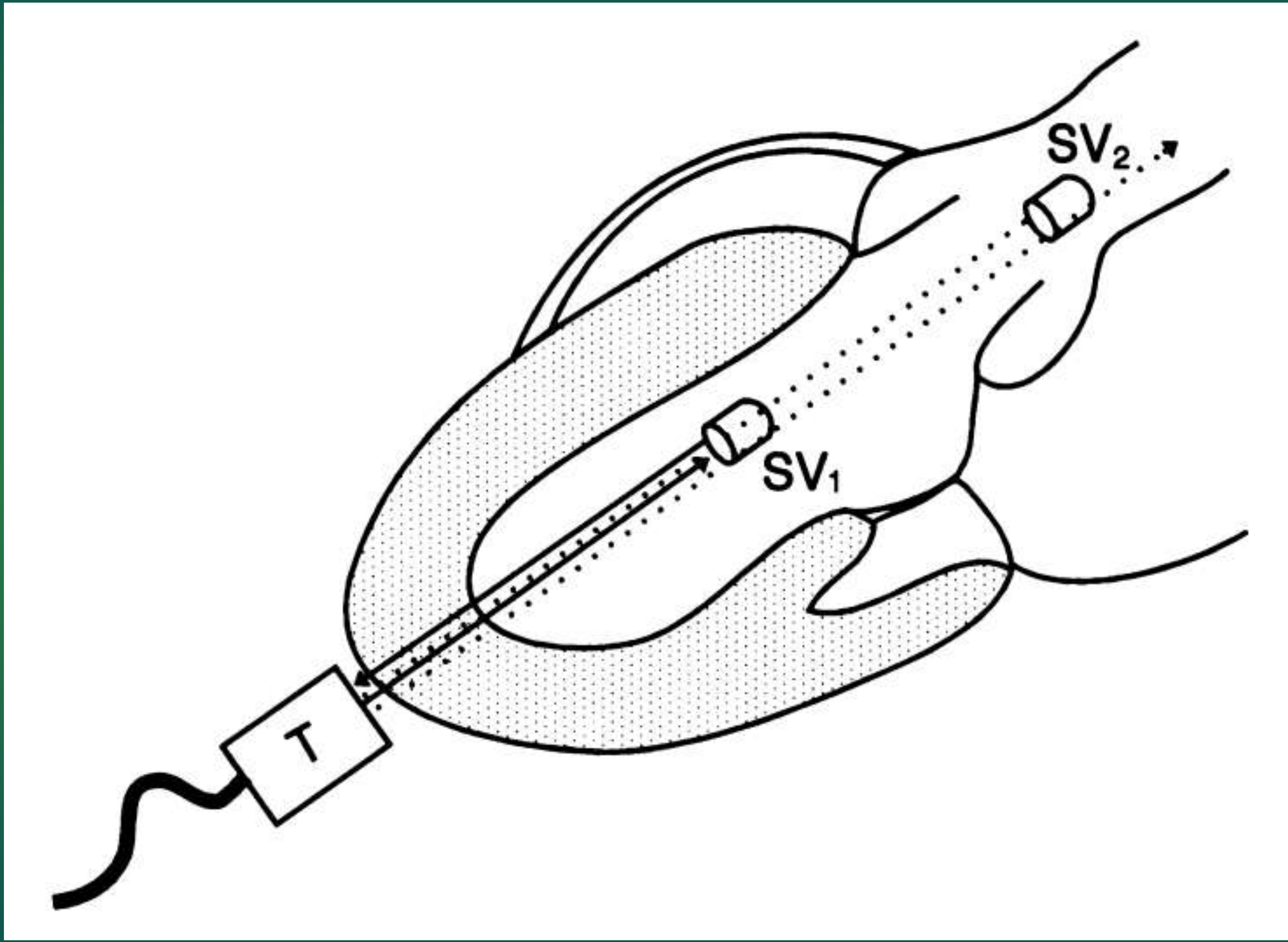
Nyquist Limit

- The maximum detectable frequency shift (the Nyquist Limit) is one-half the PRF. But the maximal detectable velocity depends on the equation.

$$v = \frac{c\,(F_S - F_T)}{2\,F_T(\cos\theta)}$$

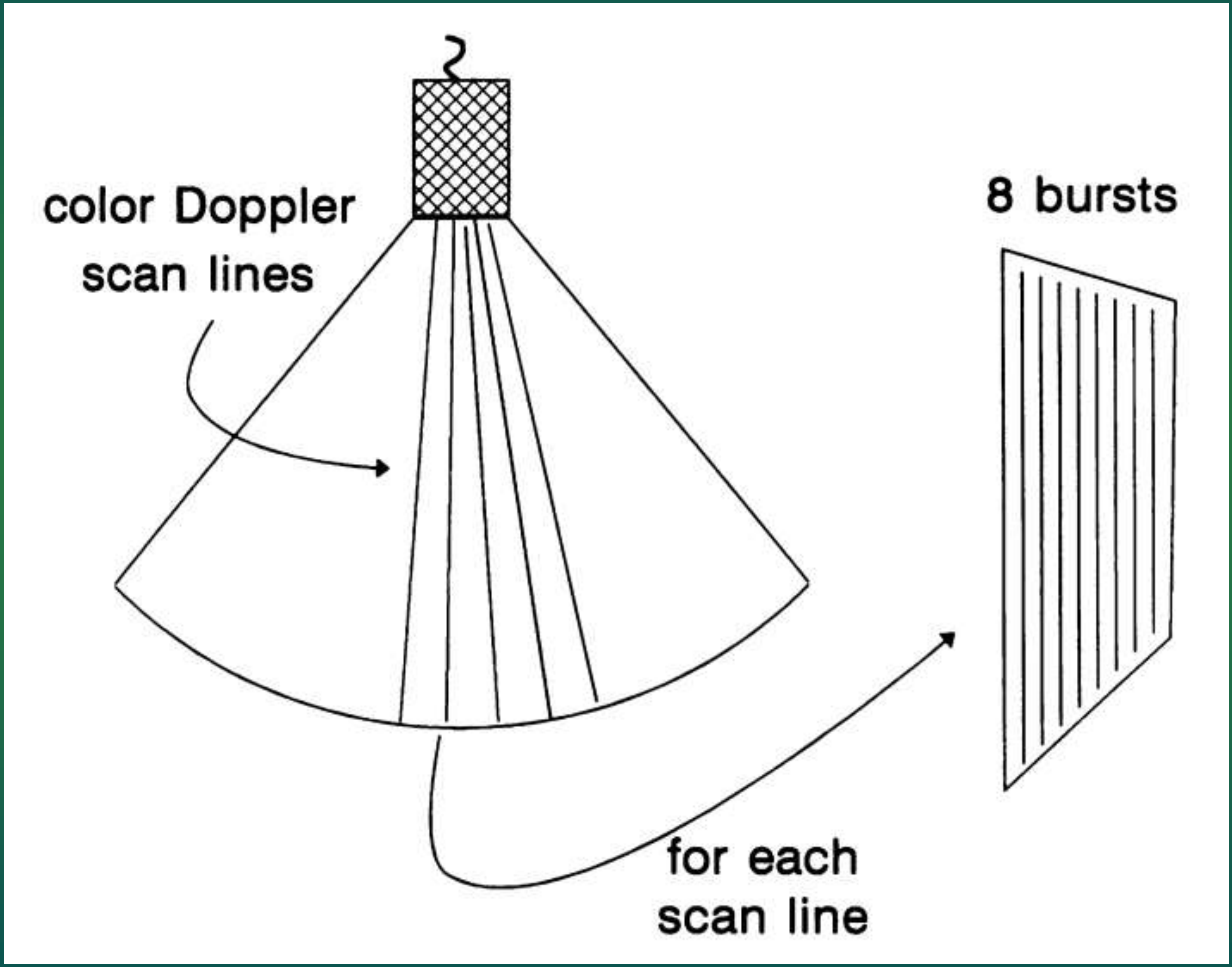
Let’s say the PRF is 5000.
Nyquist limit= (2500Hz or 2.5KHz=.0025MHz)=
[1540m/s*(PRF/2)MHz]/2*2.000MHz(1.0)=
[1540m/s*(0.0025)MHz]/2*2.000MHz(1.0)=
1540*.00025/4=0.9625 m/sec= 96cm/sec

High PRF Doppler



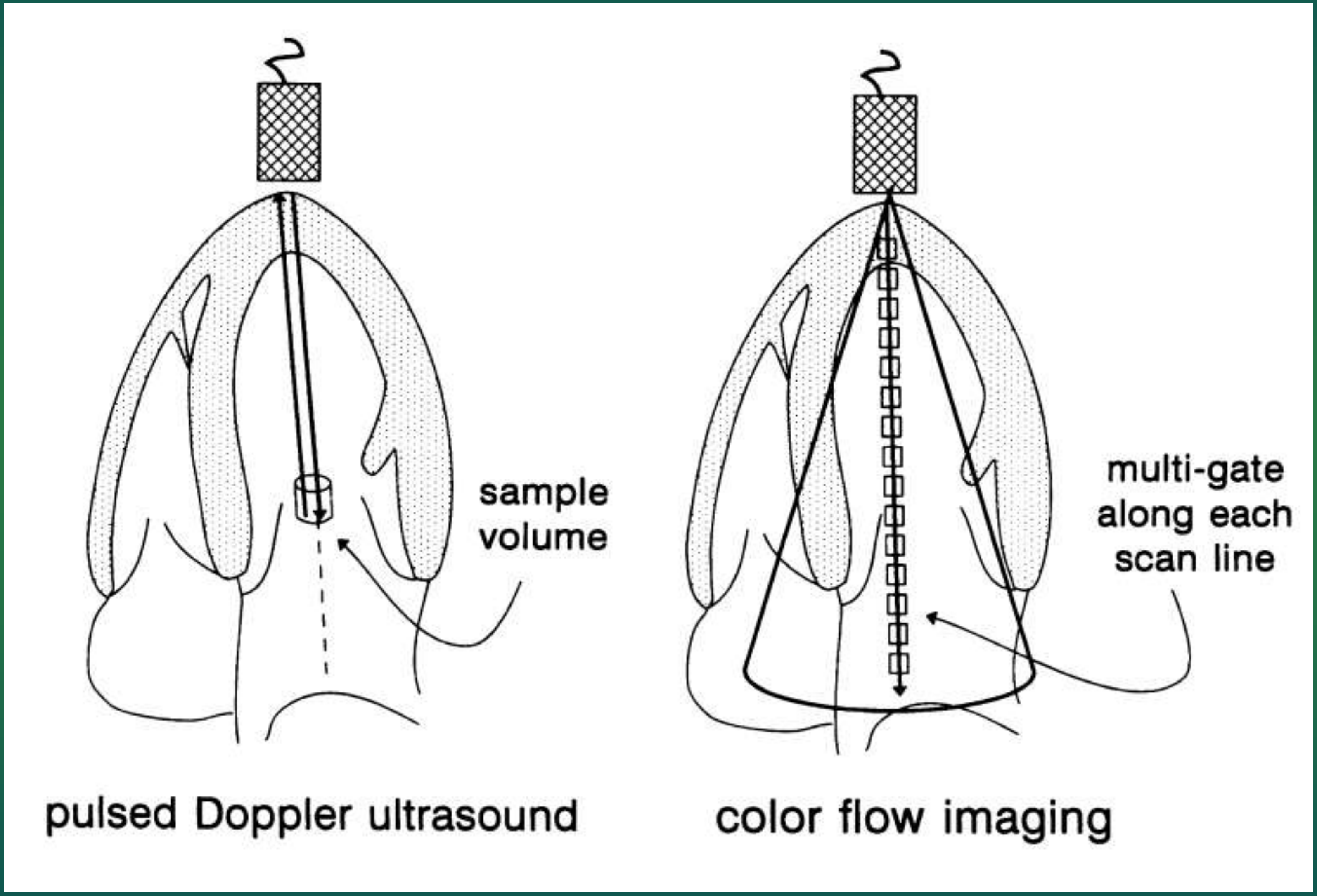
Otto CM. Textbook of Clinical Echocardiography. 2000.

Color Doppler Diagram



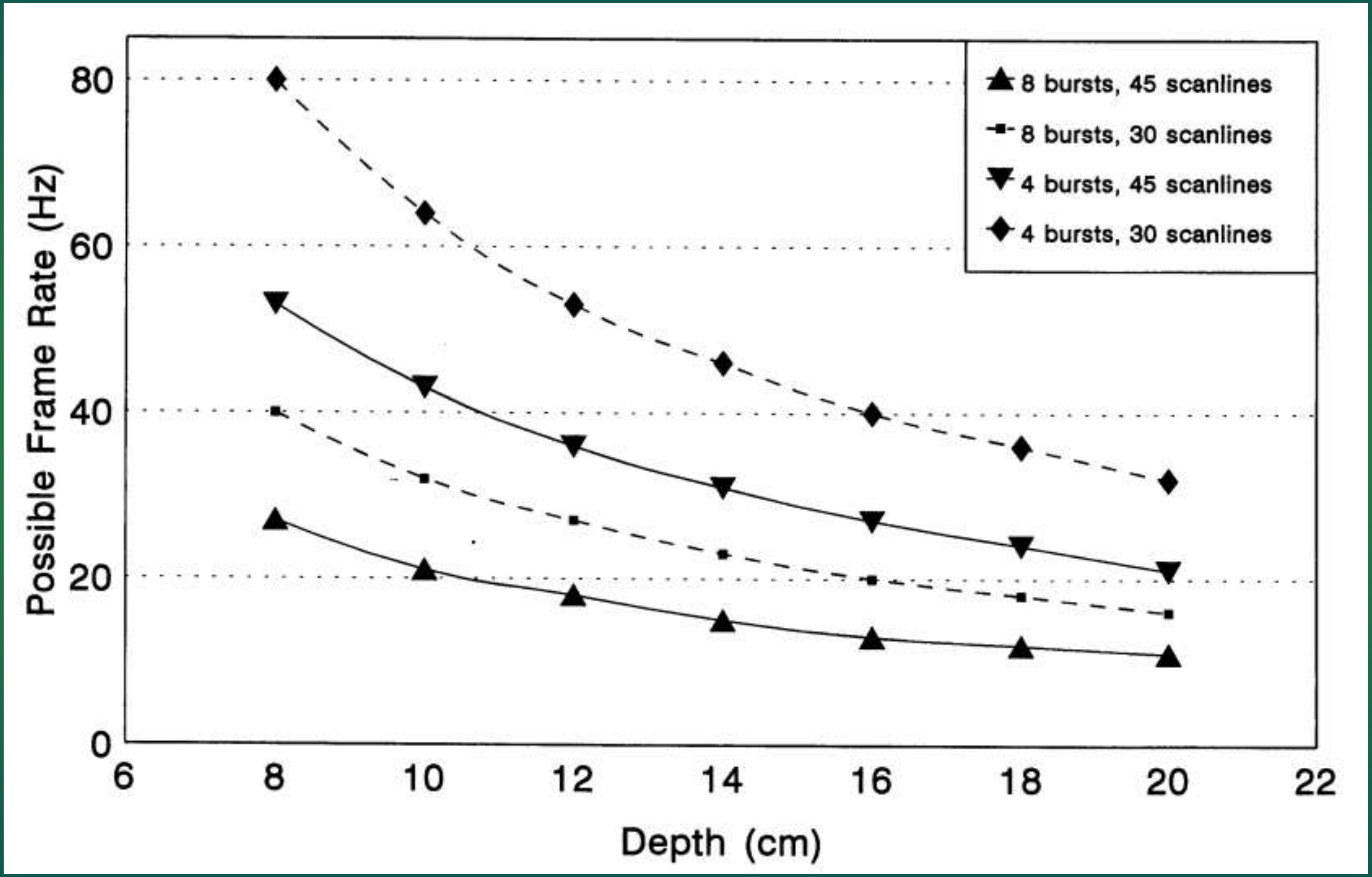
Otto CM. Textbook of Clinical Echocardiography. 2000.

Color Doppler Diagram



Otto CM. Textbook of Clinical Echocardiography. 2000.

Color Doppler Diagram



Otto CM. Textbook of Clinical Echocardiography. 2000.

Type of Doppler Signal

- Pulsed wave Doppler
- Continuous wave Doppler
- Color flow Doppler (a form of pulsed wave Doppler)
- (High pulse repetition frequency pulsed wave Doppler is intermediate between pulsed and continuous wave)

Types of Doppler Signal

	Pulsed wave	Continuous wave	High PRF
Aliasing	Yes	No	Yes, better
High velocity	No	Yes	Yes
Range specific	Yes	No	Some ambiguity
Laminar Resolution	Yes	No	Yes, somewhat

Principles of Imaging

- Reducing scan angle can increase line density
- Side lobe artifacts can be generated by the transducer
- Signal processing can alter relationship between strength of received signal and display strength

Principles of Imaging - 2

- Persistence of image on screen smoothes discontinuities (temporal image processing)
- Signal processing is complex
- Connection with video monitor is not trivial, and can result in signal loss

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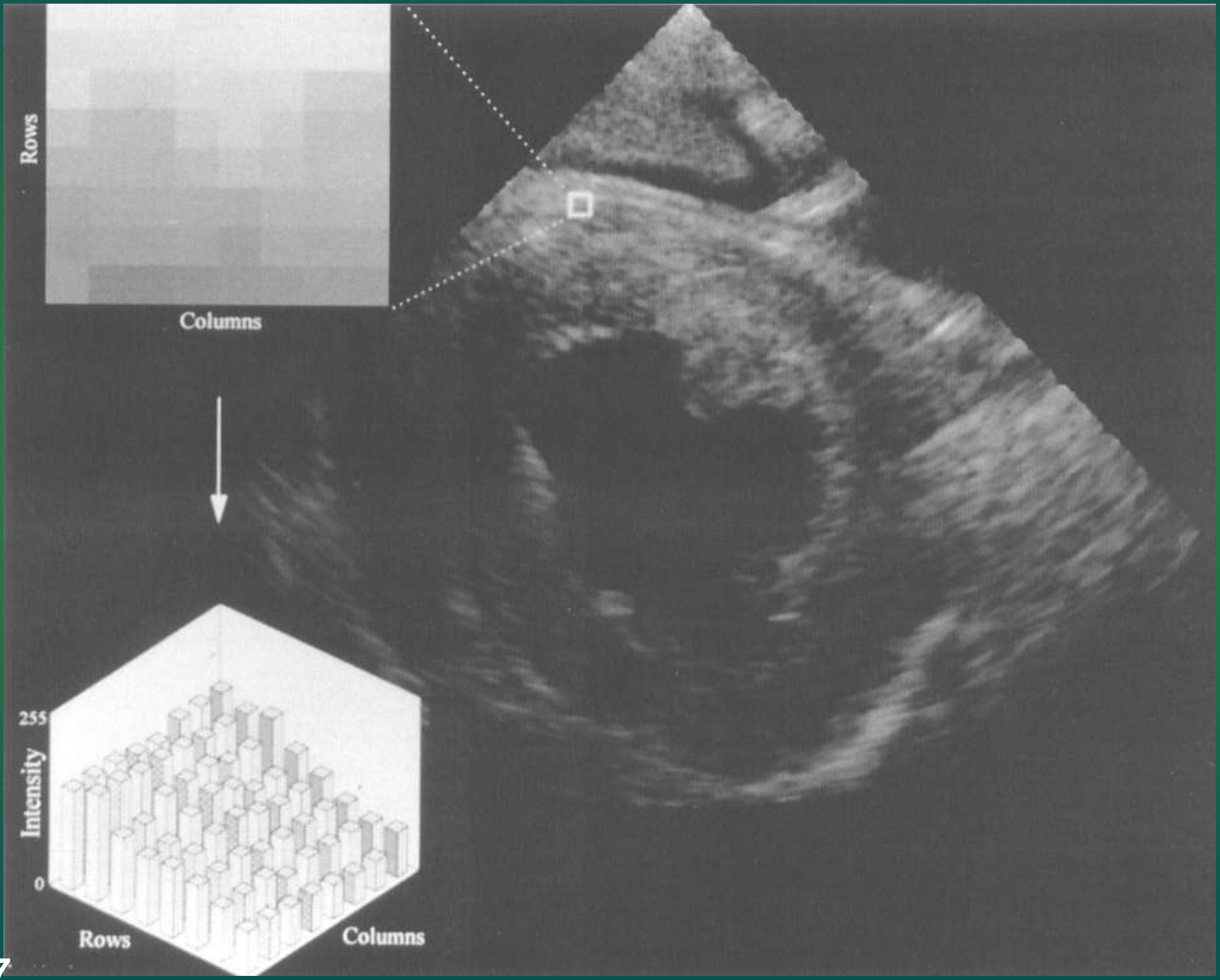


Big Otto, 1997

Principles of Imaging - Contrast

- Contrast enhancement with specular reflectors
  - air microbubbles
  - protein microparticles
- Contrast location
  - bloodstream
  - myocardial (experimental)

Imaging Advances

- Imaging traditionally looks for the frequency transmitted
  - Human tissue naturally causes increase in frequency of returned signal, and can be imaged at twice the transmitted frequency
  - Native tissue harmonic imaging
- Tissue characterization

Echo Artifacts

- Incorrect persistence on screen - bright object may last into subsequent frames
- Point spread function in far field
- Internal reverberations, projected at twice the real distance
- Reverberations from highly reflective interface may be a series of echoes
- Shadowing behind a strong reflector

Uses of Doppler Information

- Analysis of velocity
- Analysis of turbulence
- Analysis of valve area
  - MV - pressure half time
  - Continuity Equation
- Analysis of pressure difference, instantaneous or mean

Blood Flow

- The original and still most commonly used source of Doppler information
- Flow is generally either laminar or turbulent depending on whether Reynold's number is exceeded (about 5,000-10,000)
- Reynolds: 2RVp/n where R=radius, V=velocity, p=density, n=viscosity
- Flow is usually at least somewhat pulsatile

Murgo, JP. J Am Coll Cardiol. 1998;32:1596-1602, Weyman AE text 1994, p. 191

Doppler Pressure Gradient

$$\Delta P = 1/2\rho(v_2^2 - v_1^2) + \rho\int_1^2 \frac{d\vec{v}}{dtd\vec{s}} + R(\vec{v})$$

Convective acceleration plus flow acceleration plus viscous forces

Weyman AE Text, 2nd ed. 1994, p. 195.

Bernoulli Equation

$3.972 \approx 4$

$$\Delta P = \underbrace{1/2\rho(v_2^2 - v_1^2)}_{1.} + \underbrace{\rho\int_1^2 \frac{d\vec{v}}{dt\,d\vec{s}}}_{2.} + \underbrace{R(\vec{v})}_{3.}$$

1. <u>Convective acceleration</u> – Velocity squared

    Pressure energy → kinetic energy

2. <u>Flow acceleration</u> – Derivative of velocity

    Energy to impart momentum

3. <u>Viscous forces</u> – Velocity

    Energy losses from friction between neighboring fluid elements, more with turbulence

Bernoulli Equation

1. Convective acceleration
2. Flow acceleration
3. Viscous forces

$$\Delta P = 1/2\rho(v_2^2 - v_1^2) + \rho\int_1^2 \frac{d\vec{v}}{dtd\vec{s}} + R(\vec{v})$$

Doppler Pressure Gradient

- At peak velocity, flow acceleration is zero
- Viscous forces are negligible when flow is high and orifice is small
- So, the pressure gradient is by convective acceleration alone, and by substituting appropriate constants and neglecting proximal velocity, $\mathbf{\Delta P=4v^2}$

Pulsed Doppler Limit

- Nyquist limit: Aliasing occurs when the frequency of the Doppler shift exceeds 1/2 the PRF
  - Doppler shift is proportional to transducer frequency
  - More with greater depth of sample volume
  - Decrease effect by shift of baseline
  - Decrease effect by increasing angle of signal

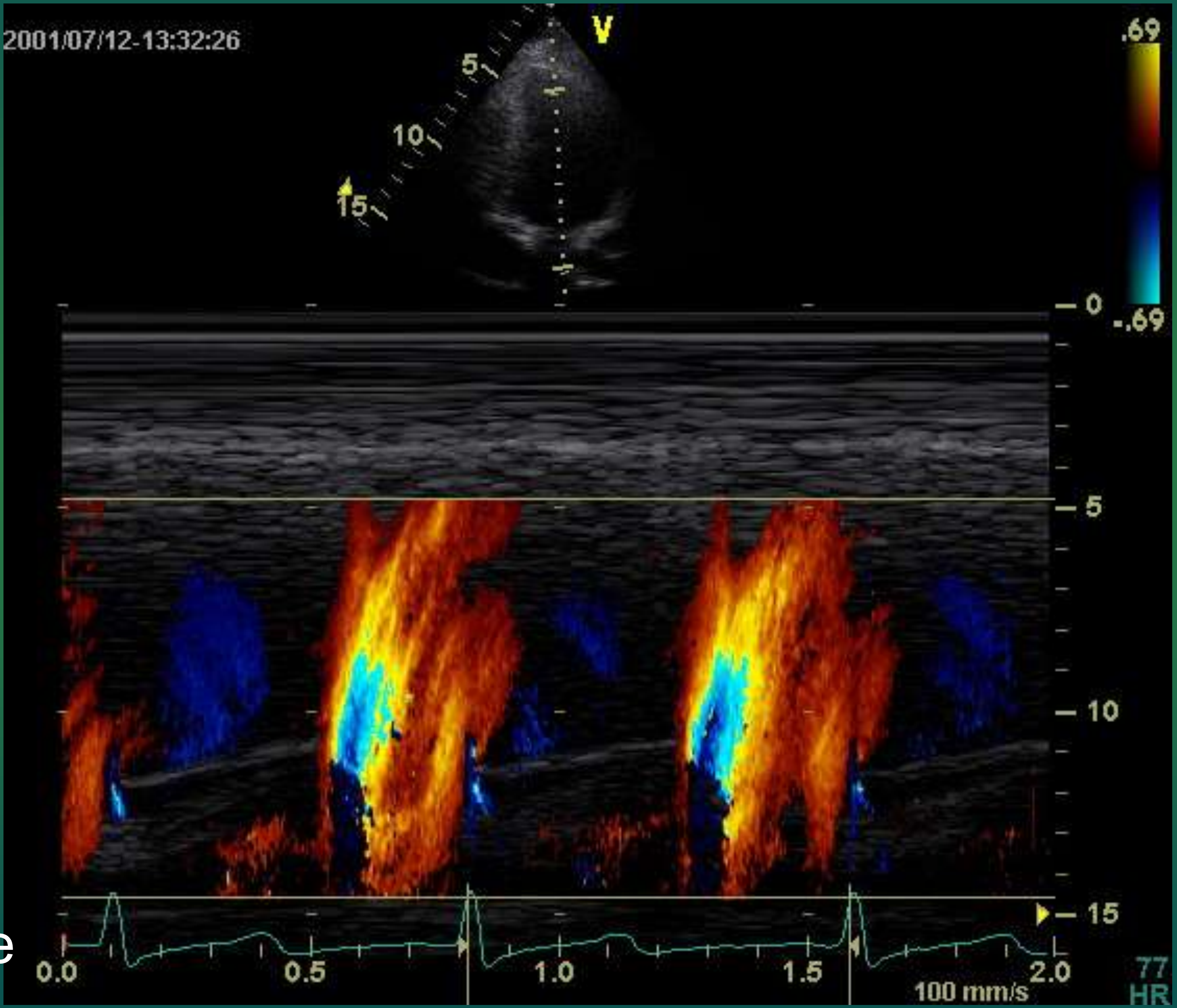
Doppler Artifacts

- Aliasing (also range ambiguity)
- Mirroring - when Doppler shift is displayed as equal frequency and opposite direction (solve by decreasing gain)
- Display of external audible noise as Doppler
- Signal loss by data sharing
- Beam width artifact

Color Doppler Artifacts

- Color aliasing
- Reverberations
- Effects of wall shadowing
  - usual suppression
  - suppression by strong echo signal
- Effects of flow angle in the scan plane

Color Doppler M-Mode



From GE website

Doppler Advances

- Contrast Doppler for myocardial perfusion
  - Intracoronary
  - Intravenous (perfluorocarbons)
- Doppler analysis of tissue
  - Wall motion
  - Strain

Tissue Doppler Background

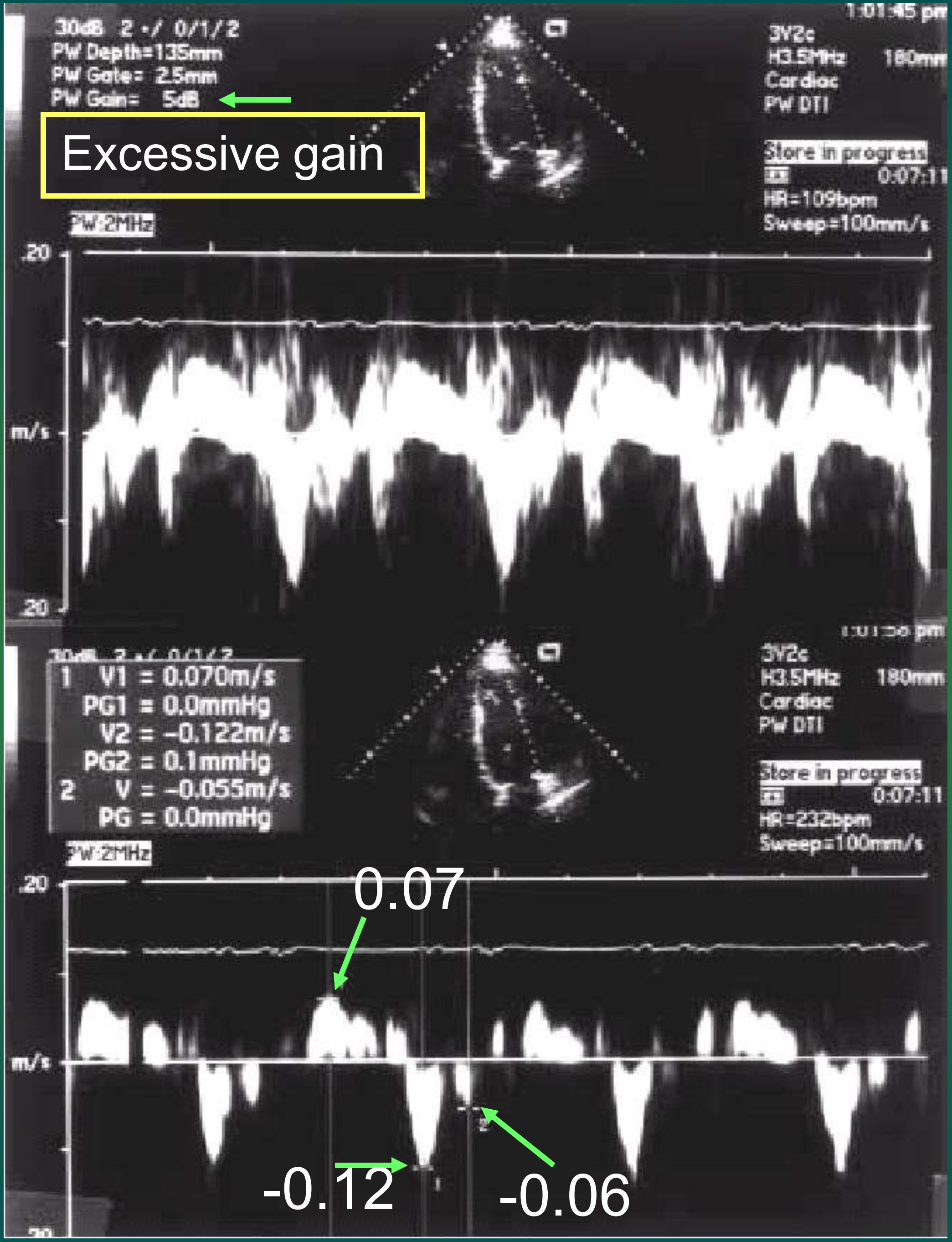
- Signal from tissue is different than that from blood pool
  - Blood flow is 10 times faster than wall motion
  - Blood flow signal is much weaker (40dB) than wall motion signal

Tissue Doppler Imaging Techniques

- High pass wall filter is disabled
- Gain amplification for low velocity or reduction for myocardium
- Expanded scale (peak less than 25 cm/s)
- Small (2 mm) sample volume ("gate")

Waggoner AD et al.
J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

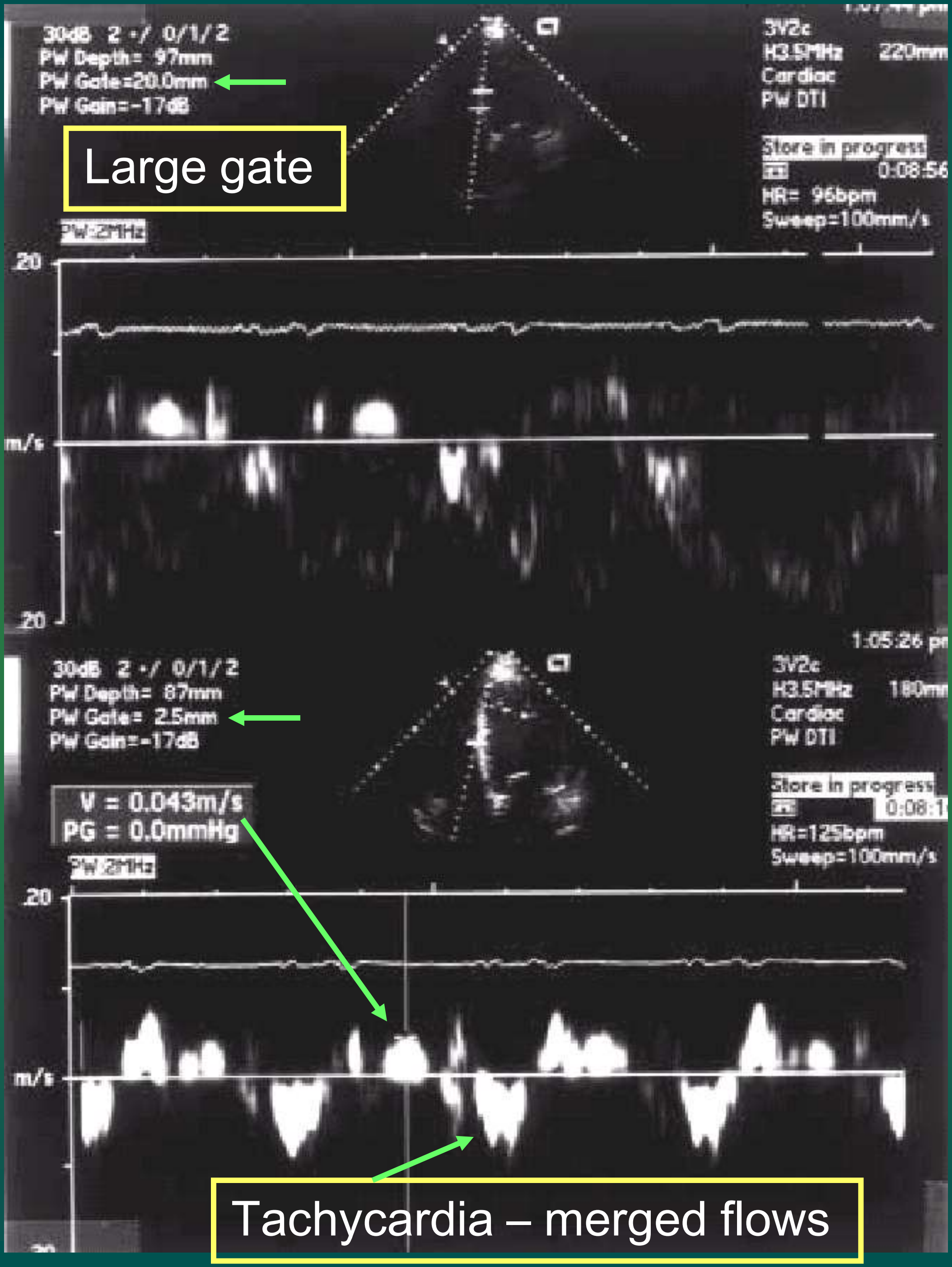


- Gate: lateral LV base in A4C
- V1 – systolic
- V2 – early diastolic
- V – late diastolic

Waggoner AD et al.
J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

- Gate: midventricular septum in A4C
- Large gate (20mm) gives inadequate recording



Waggoner AD et al.
J Am Soc Echocardiogr 2001;

Tissue Doppler Imaging

Table 1 Advantages and limitations of color tissue Doppler imaging (TDI) versus pulsed wave (PW) TDI

	Advantages	Limitations
Color TDI	• Spatial orientation of myocardial velocities can be seen superimposed on the real-time 2-dimensional image • Myocardial velocities can be displayed in a manner similar to that of conventional color flow imaging • Representation of mean myocardial velocity	• Poor temporal resolution caused by longer processing times involved with autocorrelation analysis • Typically requires off-line analysis for quantification of the myocardial velocity color maps
PW TDI	• Real-time velocity interrogation with improved temporal resolution • Ability to quantitate peak rather than mean myocardial velocities • Does not require off-line analysis • Provides instantaneous temporal display of the Doppler spectral information • Objective assessment of regional function, which is especially useful in dobutamine stress echocardiography	• Only regional quantification of myocardial velocities can be done at selected sites reducing spatial resolution • Sampling cannot be localized to the endocardial or epicardial layers • Alignment of the beam parallel to the heart muscle movement may be difficult in some patients • No correction for normal cardiac translation and rotation during sampling

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

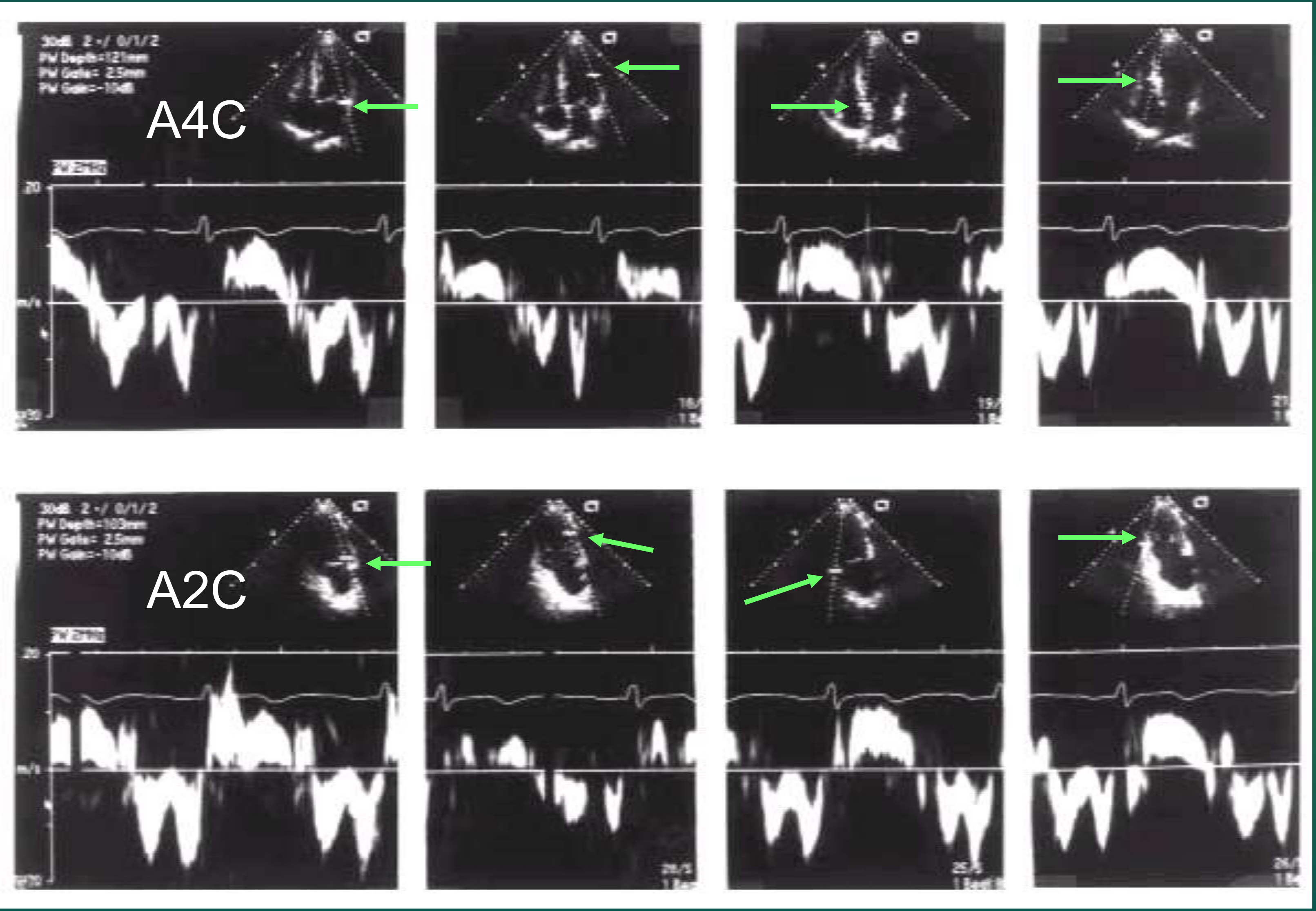
Table 2 Normal values in cm/s ± 1 SD for the basal segments of the left ventricle using pulsed TDI[9,23,29,41,44]

	Sm	Em	Am	Em/Am velocity ratio
Lateral	10.6 ± 2.3	13.3 ± 3.3	11.3 ± 2.9	1.5 ± .6
Septal	9.9 ± 1.7	11.5 ± 2.6	9.5 ± 2.4	1.0 ± .7
Anterior	9.2 ± 1.8	11.7 ± 3.4	10.3 ± 2.9	1.2 ± .7
Posterior	10.4 ± 2.5	14.3 ± 3.6	11.6 ± 2.6	1.3 ± .7

Am, Late diastolic myocardial velocity; *Em*, early diastolic myocardial velocity; *Sm*, systolic myocardial velocity.

Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

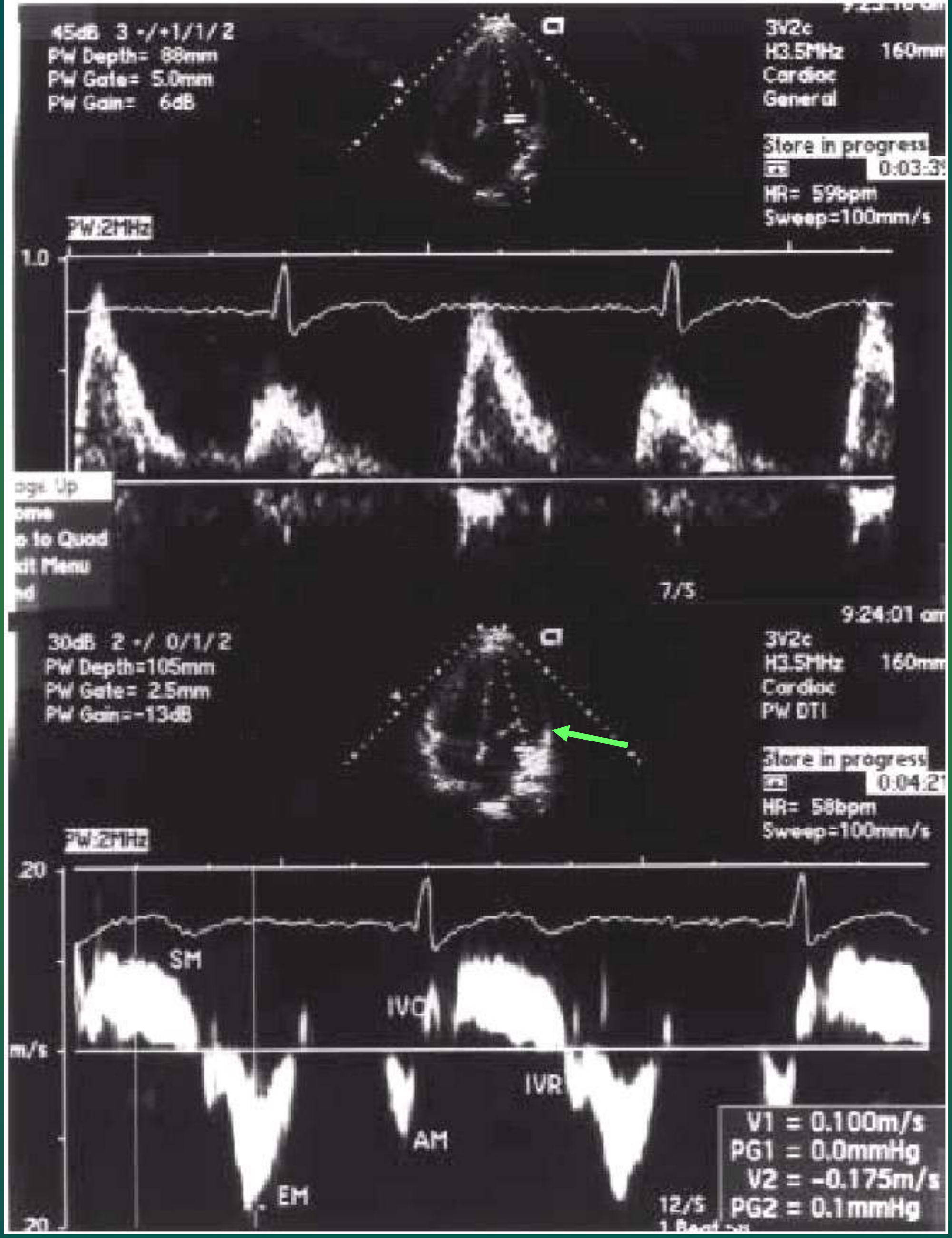
Normal Tissue Doppler Imaging



Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

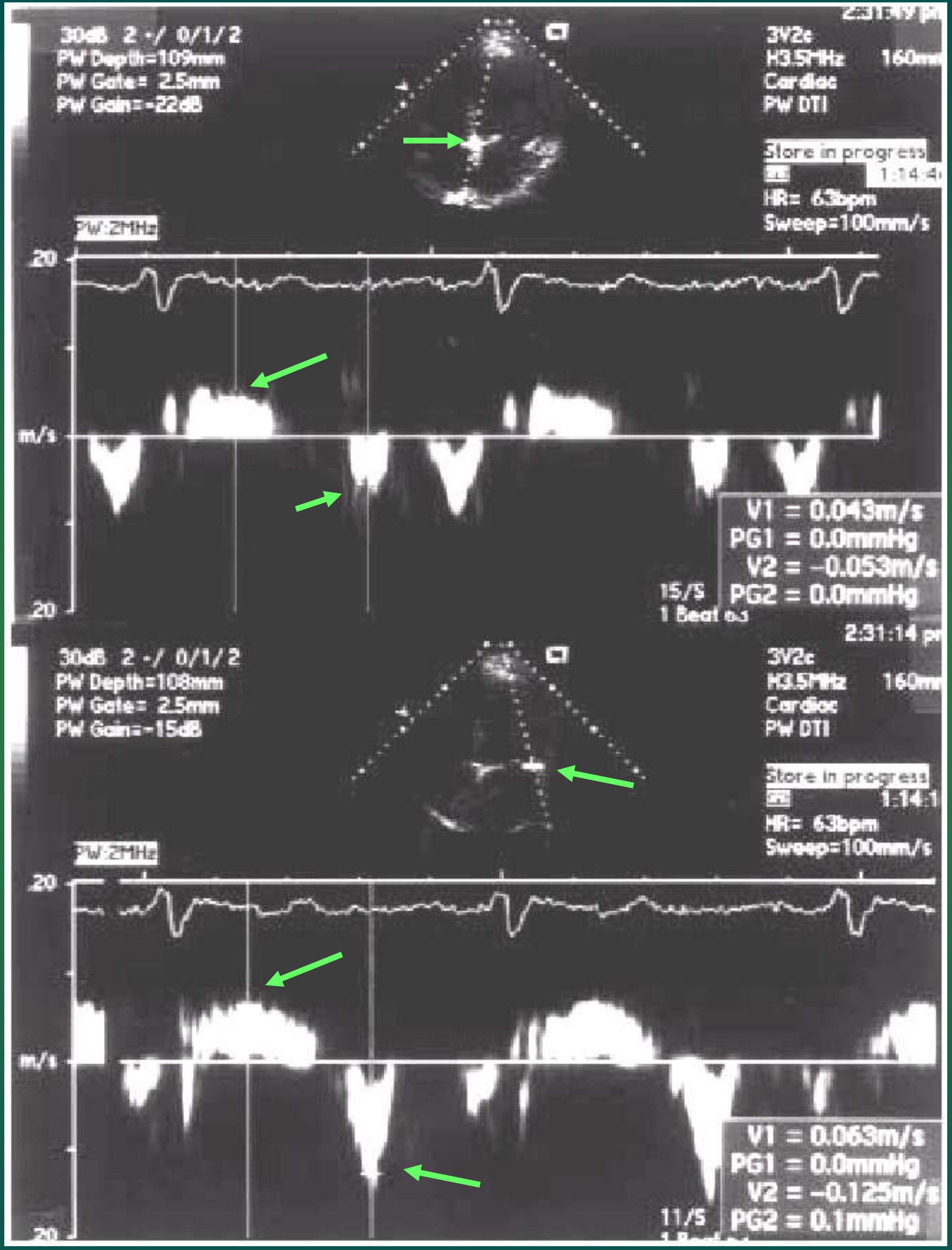
- Normal tissue Doppler signal from basal lateral wall in A4C



Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

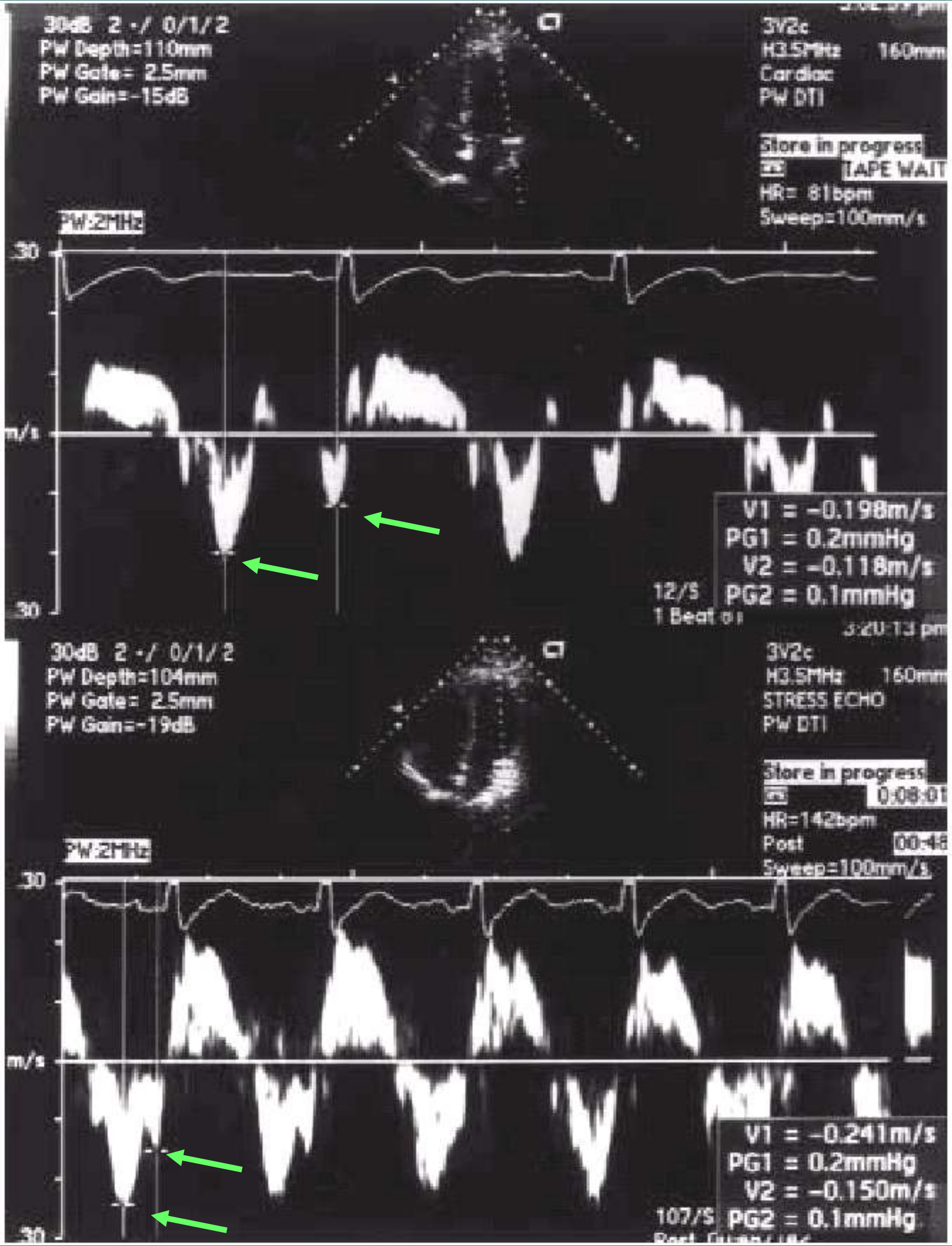
- Patient with prior anteroseptal MI
  - Top: septal base
  - Bottom: lateral base
- V1 systolic velocity
- V2 diastolic velocity



Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

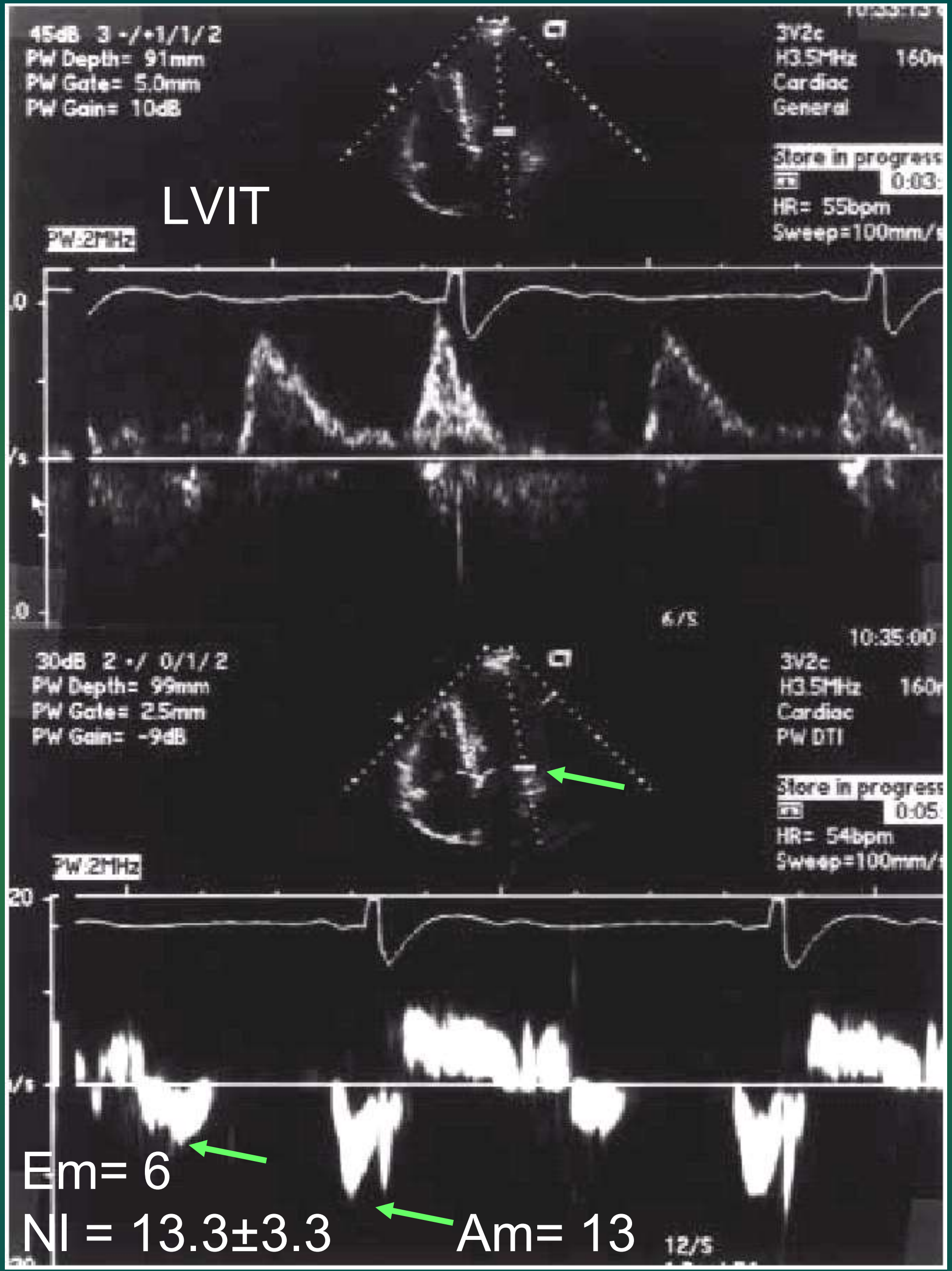
- Top: rest
- Bottom: exercise
- Increases in systolic and diastolic velocities



Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Tissue Doppler Imaging

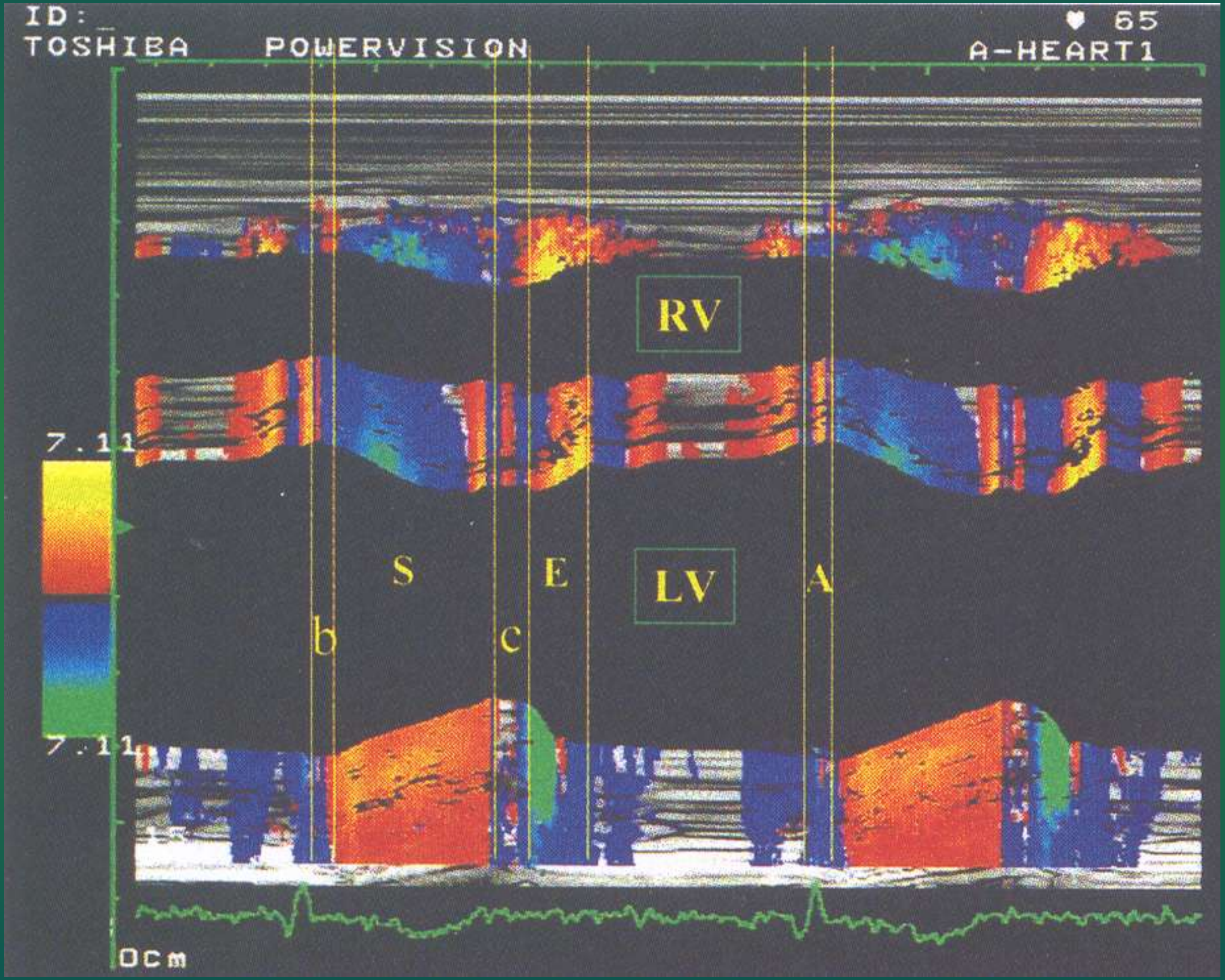
- Htn and LVH
- Pseudonormalized LVIT pattern
- Abnormal TDI of lateral base with decreased early diastolic velocity and normal late diastolic velocity



Waggoner AD et al. J Am Soc Echocardiogr 2001; 14:1143

Color Tissue Doppler Imaging

b=regional isovolumic contraction time
c=regional isovolumic relaxation time
s=systole
E=rapid filling
A=atrial contraction

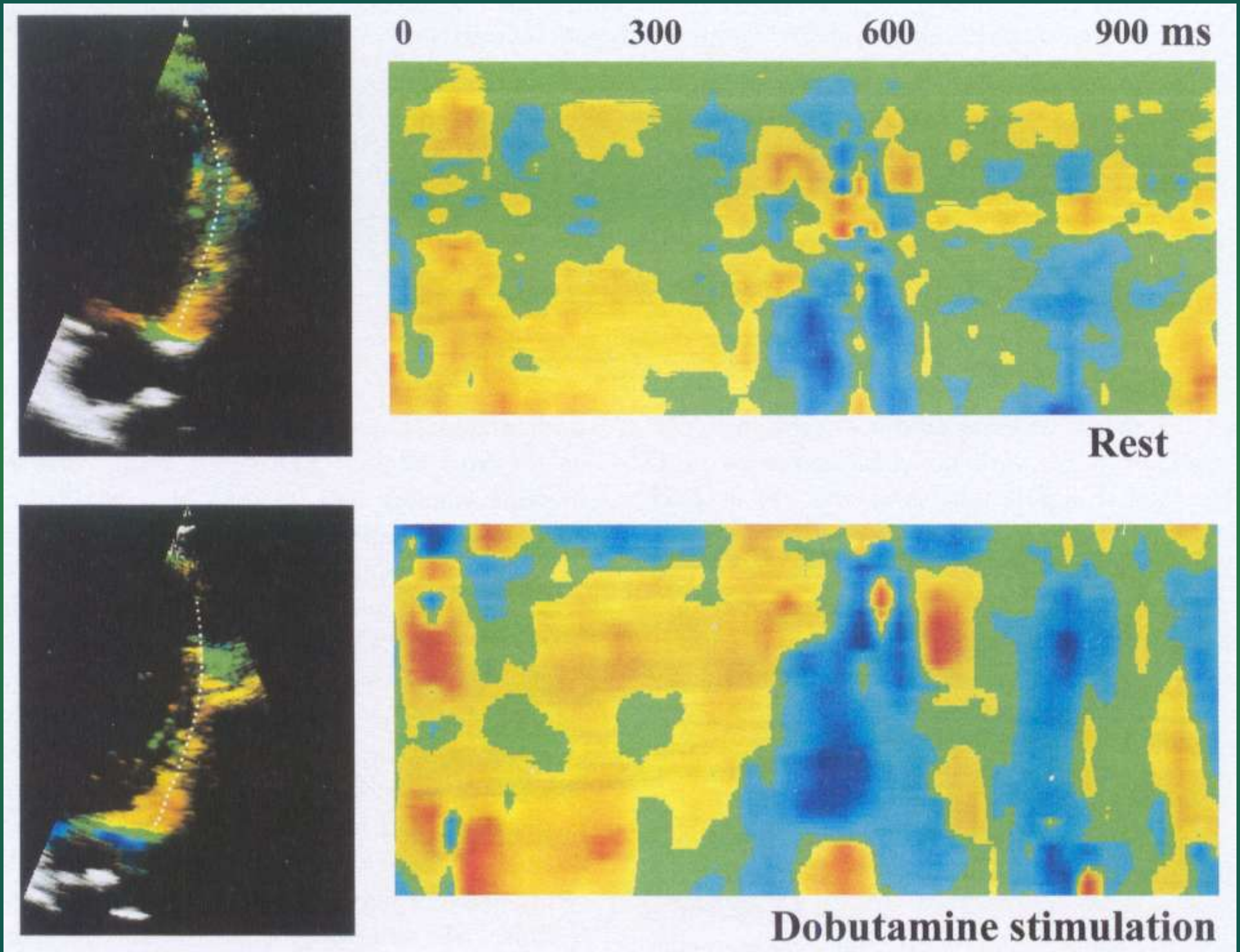


Trambaiolo P et al. JASE 2001; 14:85

Strain Rate Imaging

Septal wall

More yellow in the mid-anterior segment indicates increase in contractility, indicating viability



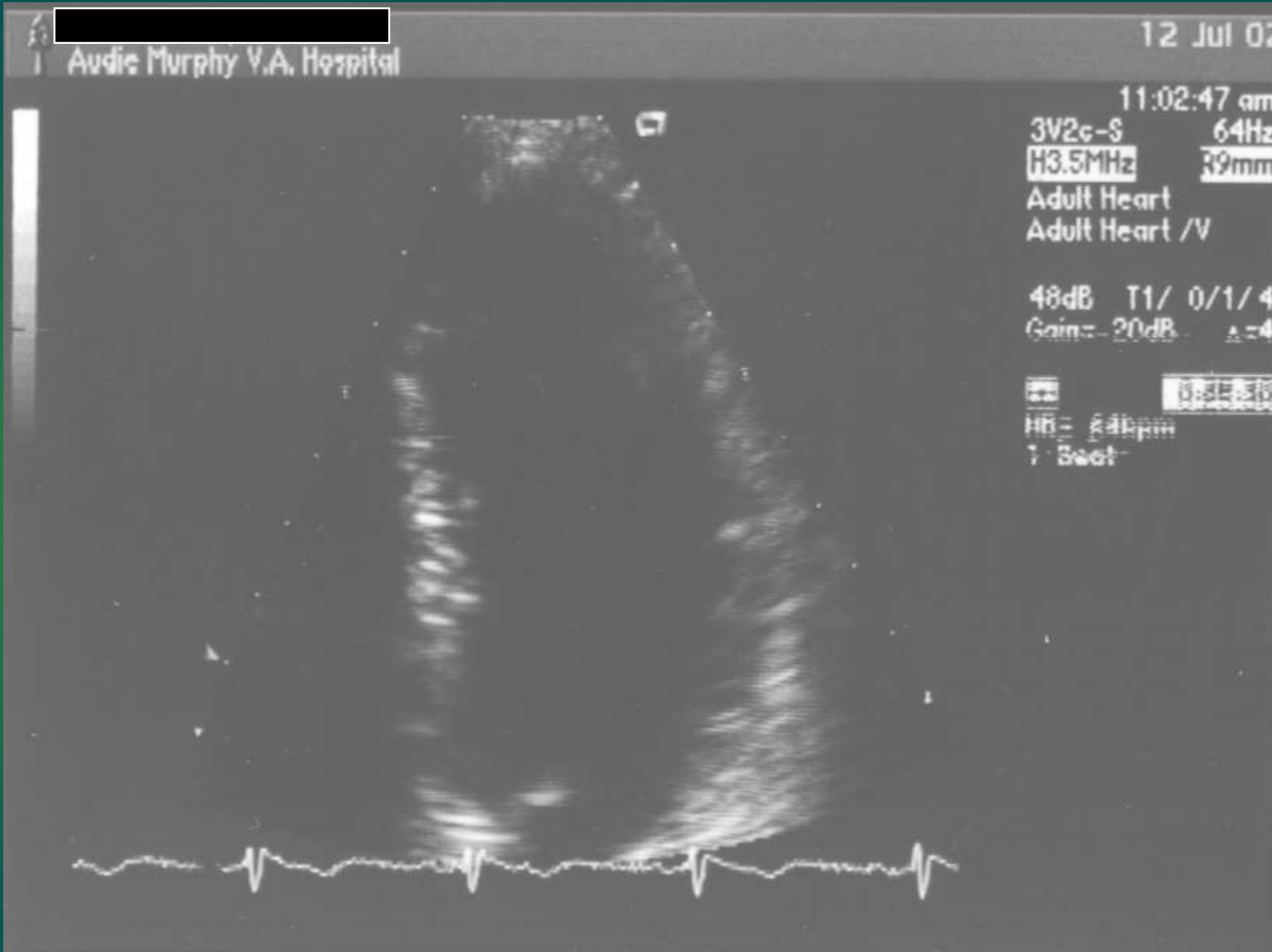
Hoffmann R et al. JACC 2002;39:443

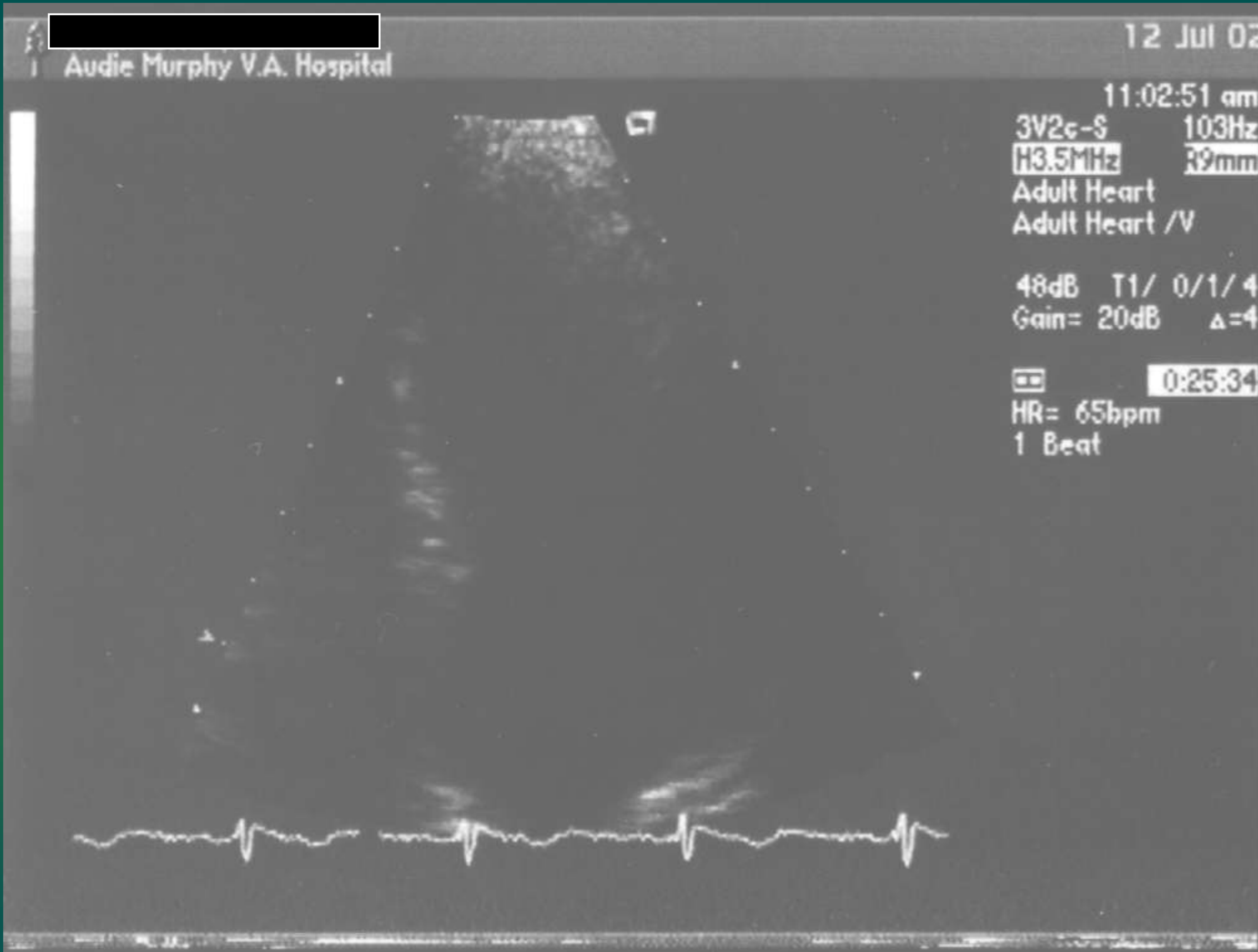
Tissue Doppler Applications

- Not widely used
- Diastolic function
- Regional systolic function
- Strain measurement using tissue Doppler displayed as tissue velocity imaging in color

For images to view and clips to see, try GE Medical Systems website
http://www.gemedicalsystems.com/rad/us/products/vivid_7/msuvivid7img.htm


[redacted]
Audie Murphy V.A. Hospital
12 Jul 02
3V2c-S
H3.5MHz
26Hz
140mm
Adult Heart
Adult Heart
T1/-2/ 0/VV:1
1/2
CD:2.0MHz
CD Gain = 50
0:26:21
HR= 64bpm
1 Beat
.80
.80


[redacted]
Audie Murphy V.A. Hospital
12 Jul 02
11:02:47 am
3V2c-S
64Hz
H3.5MHz
R9mm
Adult Heart
Adult Heart /V
48dB T1/ 0/1/ 4
Gain= -20dB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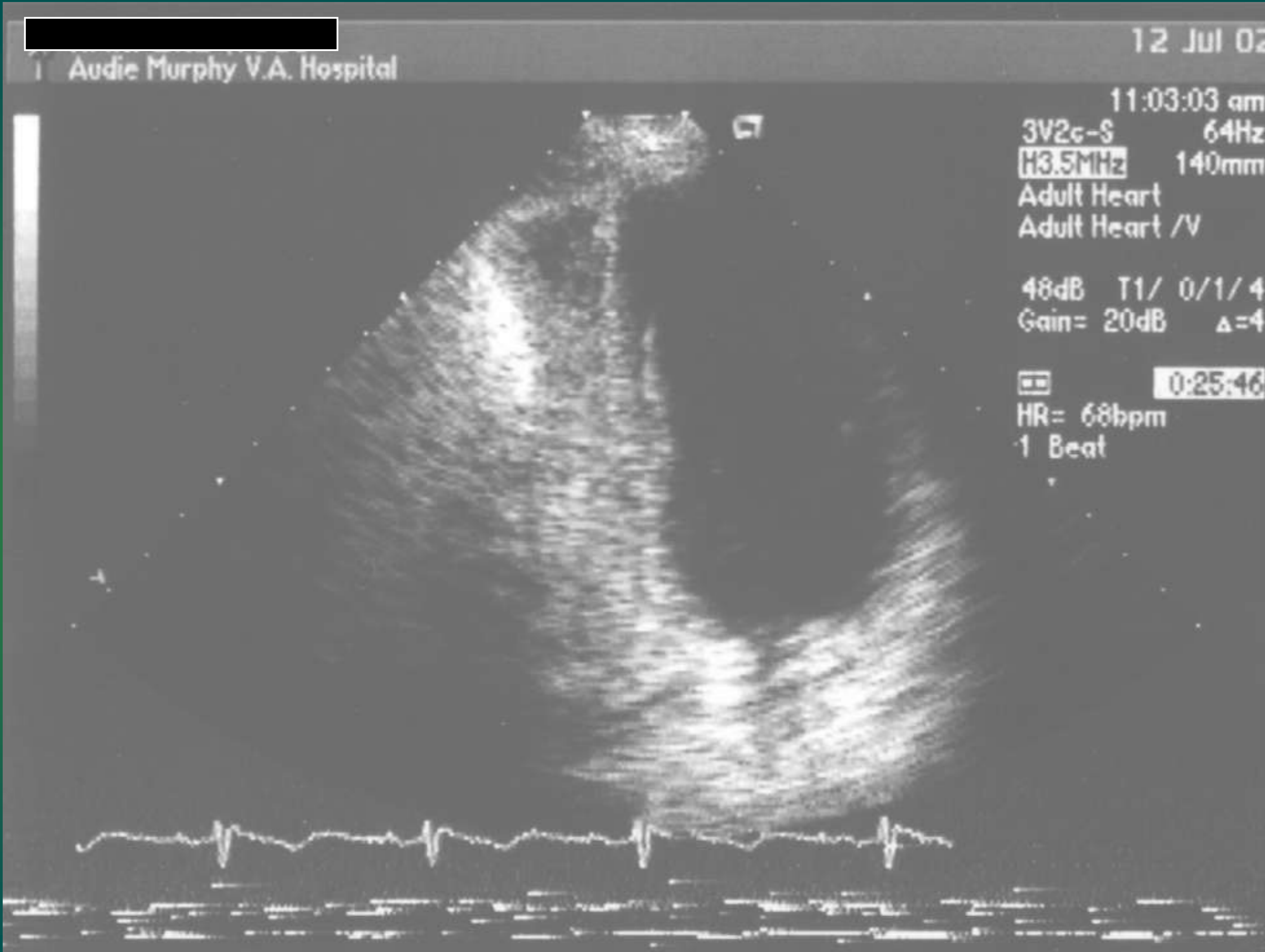
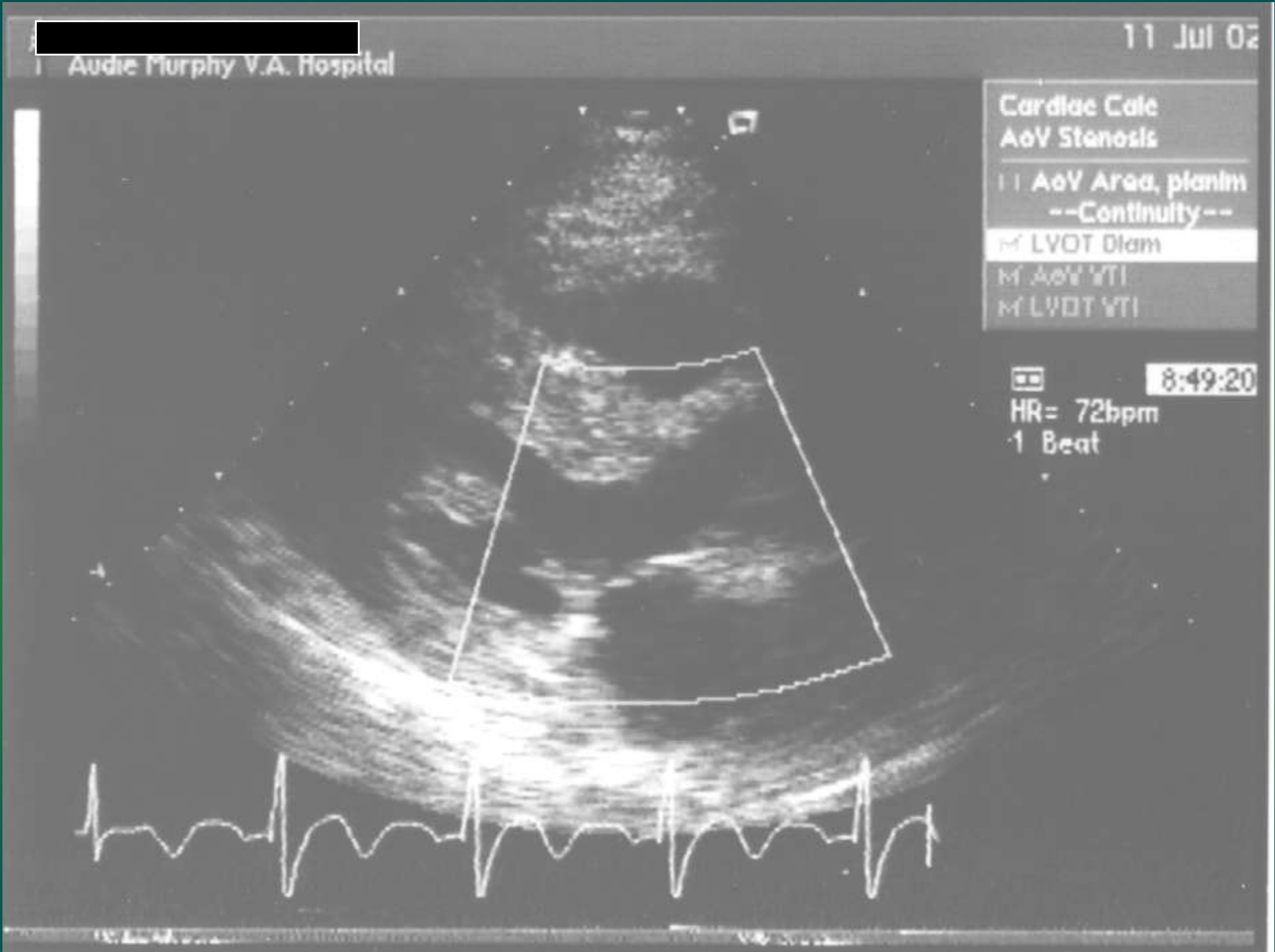
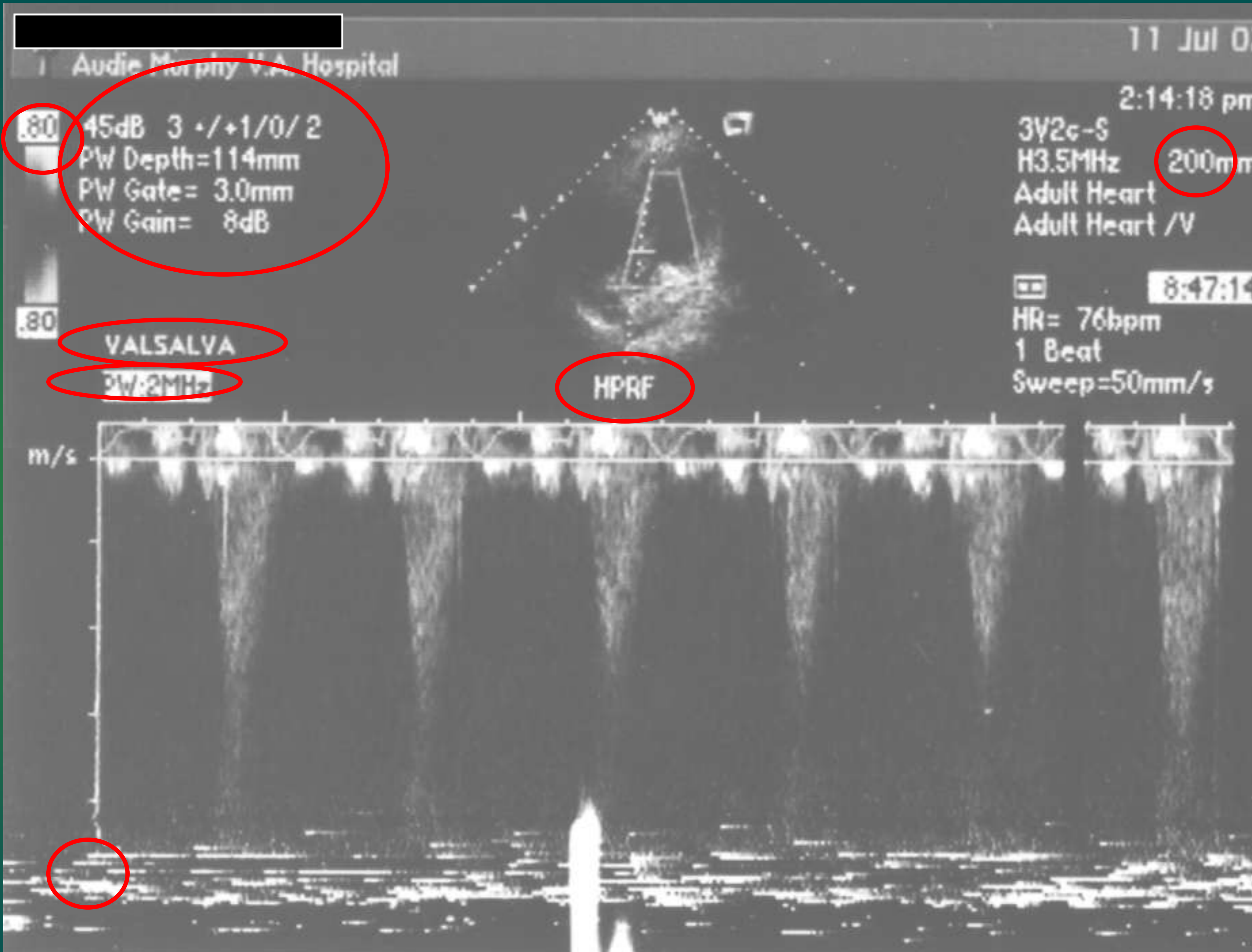
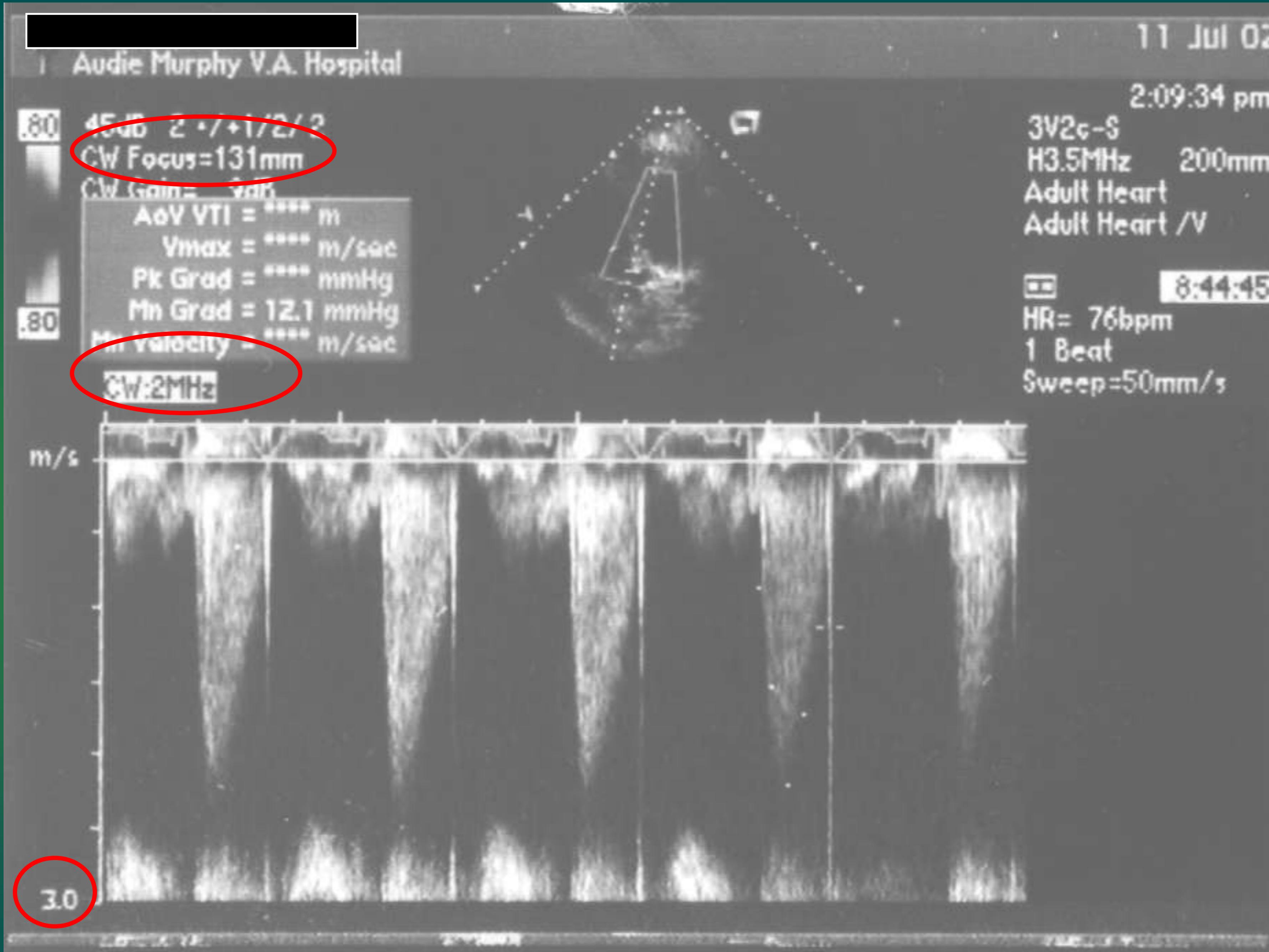
Audie Murphy V.A. Hospital
12 Jul 02
11:02:51 am
3V2c-S
103Hz
H3.5MHz
39mm
Adult Heart
Adult Heart /V
48dB T1/ 0/1/4
Gain= 20dB ∆=4
0:25:34
HR= 65bpm
1 Beat


[redacted]
Audie Murphy V.A. Hospital
12 Jul 02
11:03:03 am
3V2c-S
64Hz
H3.5MHz
140mm
Adult Heart
Adult Heart /V
48dB T1/ 0/1/ 4
Gain= 20dB Δ=4
0:25:46
HR= 68bpm
1 Beat


[redacted]
Audie Murphy V.A. Hospital
11 Jul 02
Cardiac Calc
AoV Stenosis
AoV Area, planim
--Continuity--
LVOT Diam
AoV VTI
LVOT VTI
8:49:20
HR= 72bpm
1 Beat


[redacted]
Audie Murphy V.A. Hospital
11 Jul 02
2:14:18 pm
.80
45dB 3 •/+1/0/ 2
PW Depth=114mm
PW Gate= 3.0mm
PW Gain= 8dB
3V2c-S
H3.5MHz
200mm
Adult Heart
Adult Heart /V
8:47:14
HR= 76bpm
1 Beat
Sweep=50mm/s
.80
VALSALVA
PW:2MHz
HPRF
m/s


[redacted]
Audie Murphy V.A. Hospital
11 Jul 02
2:09:34 pm
.80
CW Focus=131mm
AoV VTI = **** m
Vmax = **** m/sec
Pk Grad = **** mmHg
Mn Grad = 12.1 mmHg
Mn Velocity = **** m/sec
.80
CW:2MHz
3V2c-S
H3.5MHz 200mm
Adult Heart
Adult Heart /V
8:44:45
HR= 76bpm
1 Beat
Sweep=50mm/s
m/s
3.0