Utrasound Physics

2 – Instrumentation, Doppler Physics and Safety

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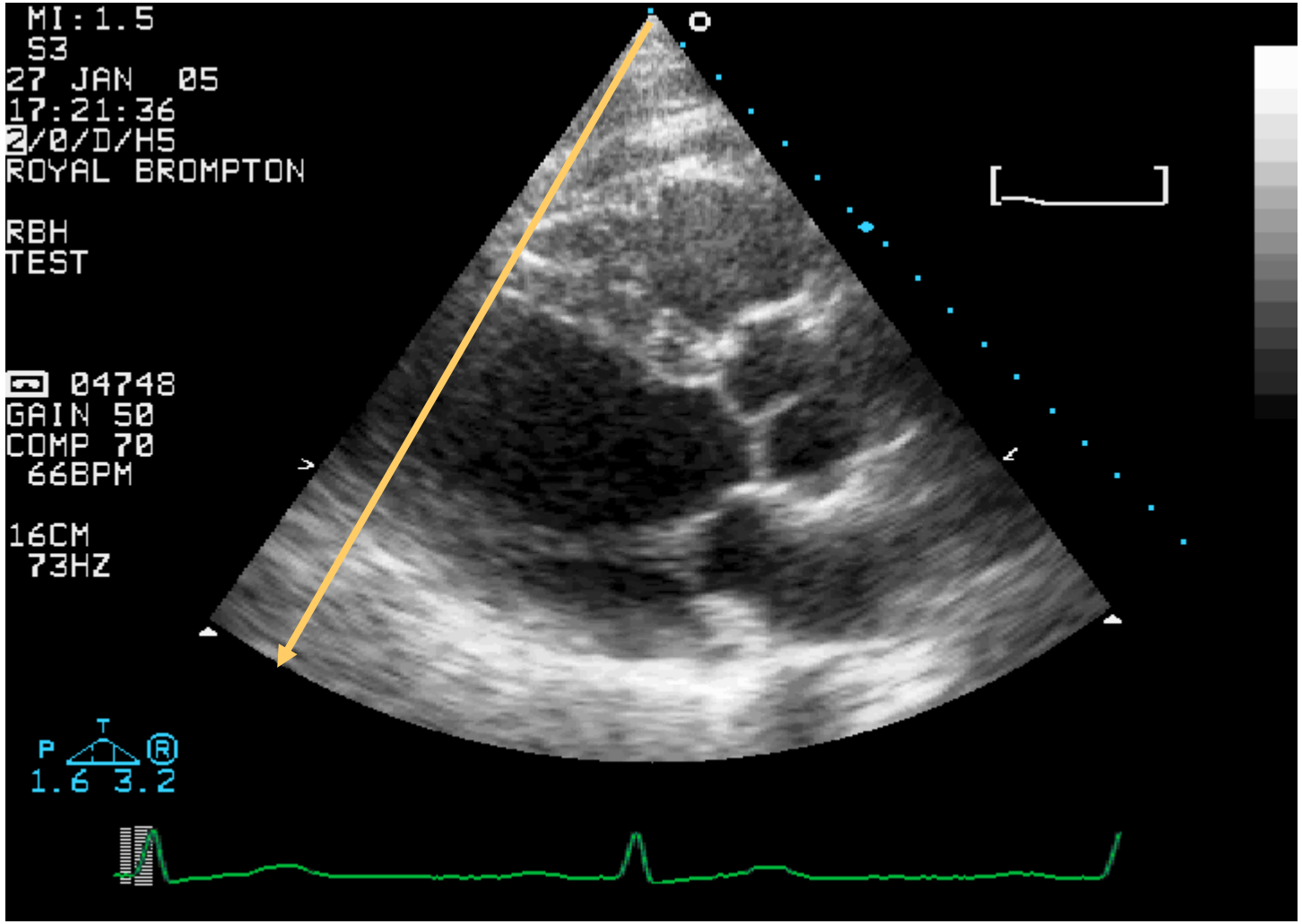
Signal Processing and Machine Controls

- Imaging modes
- Pre and Post Processing
- Grey Scale, Dynamic Range, Logarithmic Compression
- Transmitted Power and Receiver Gain
- Time Gain Compensation (TGC)
- Reject
- Spatial and Temporal Smoothing

Signal from one scan line



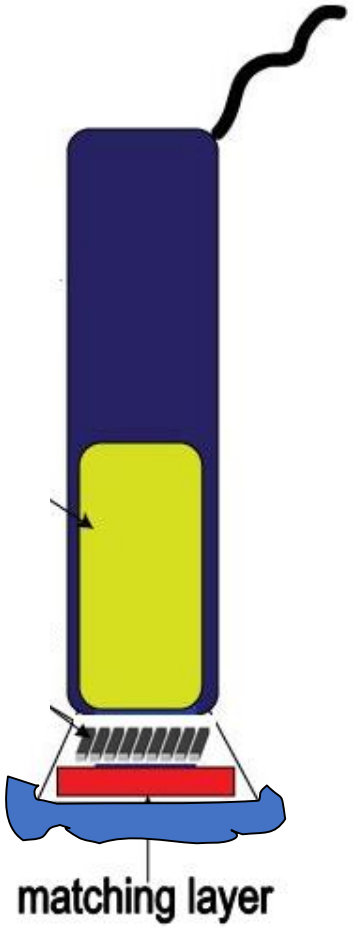
Maximising transmission

matching layer

thin layer between the piezoelectric elements and the skin

“accoustic matching”

reduces reflection → less attenuation and more energy transmitted

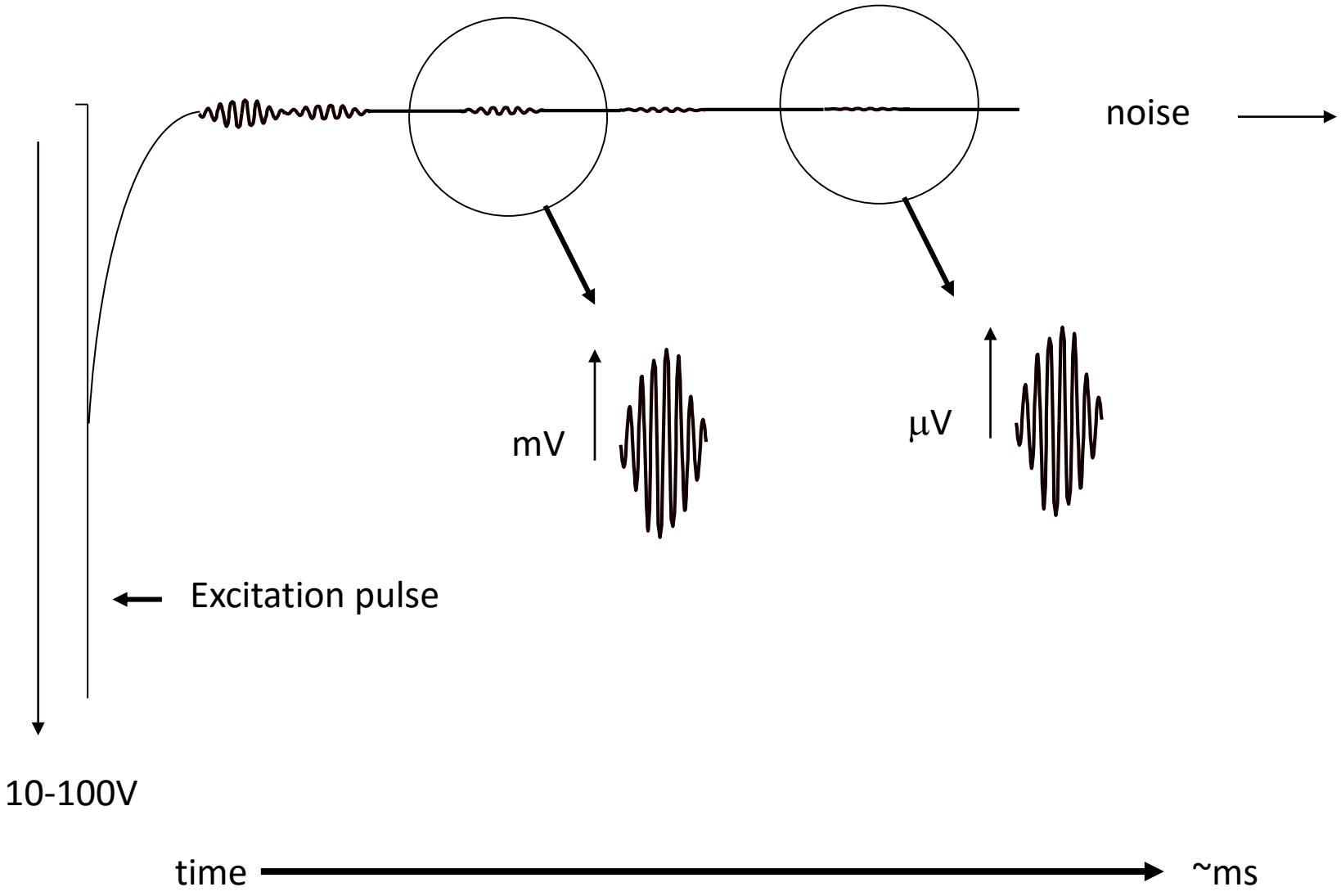


matching layer + gel

Additional intermediate “accoustic matching”

reduces reflection → less attenuation and more energy transmitted

At the transducer – one scan line

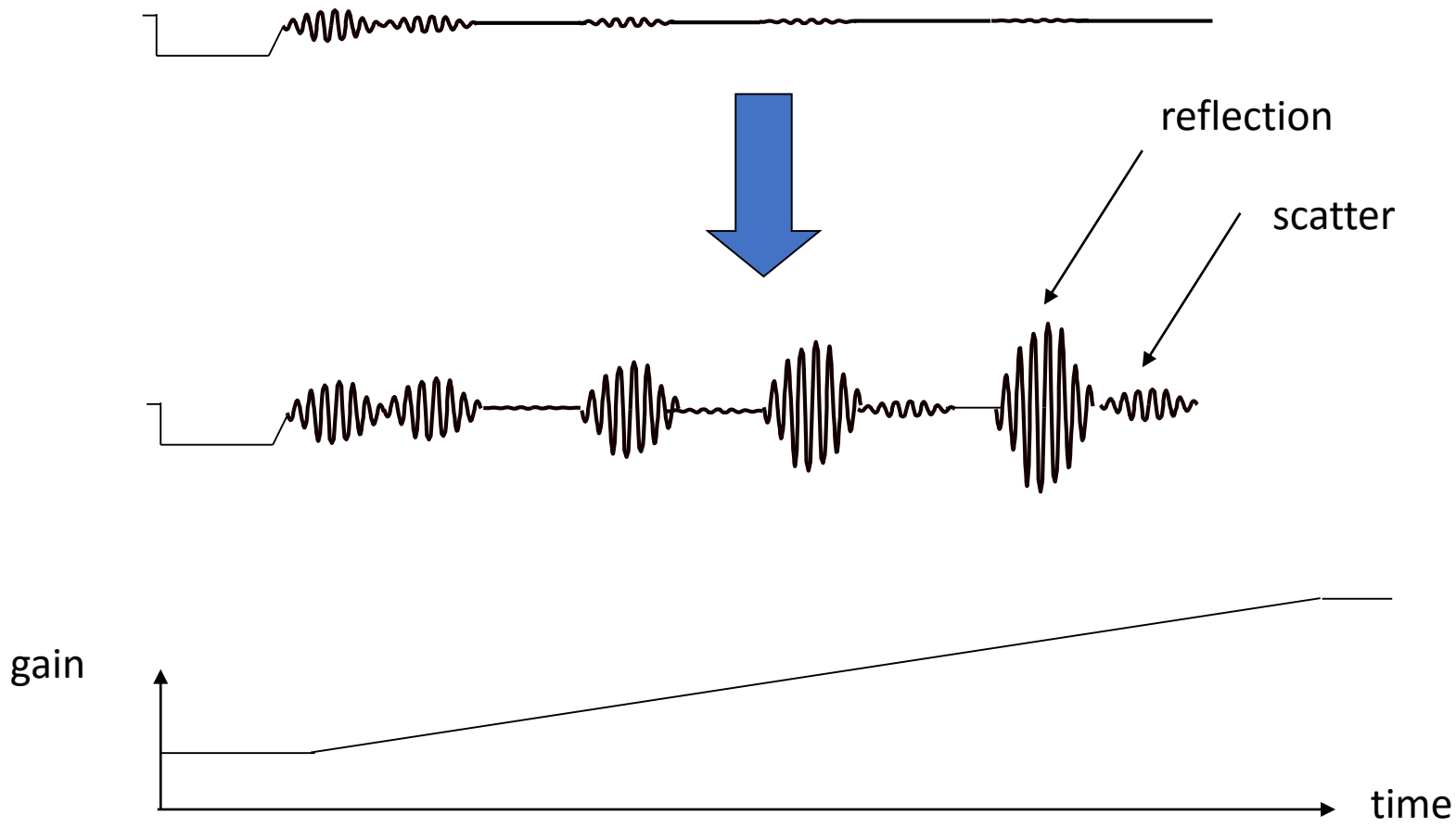


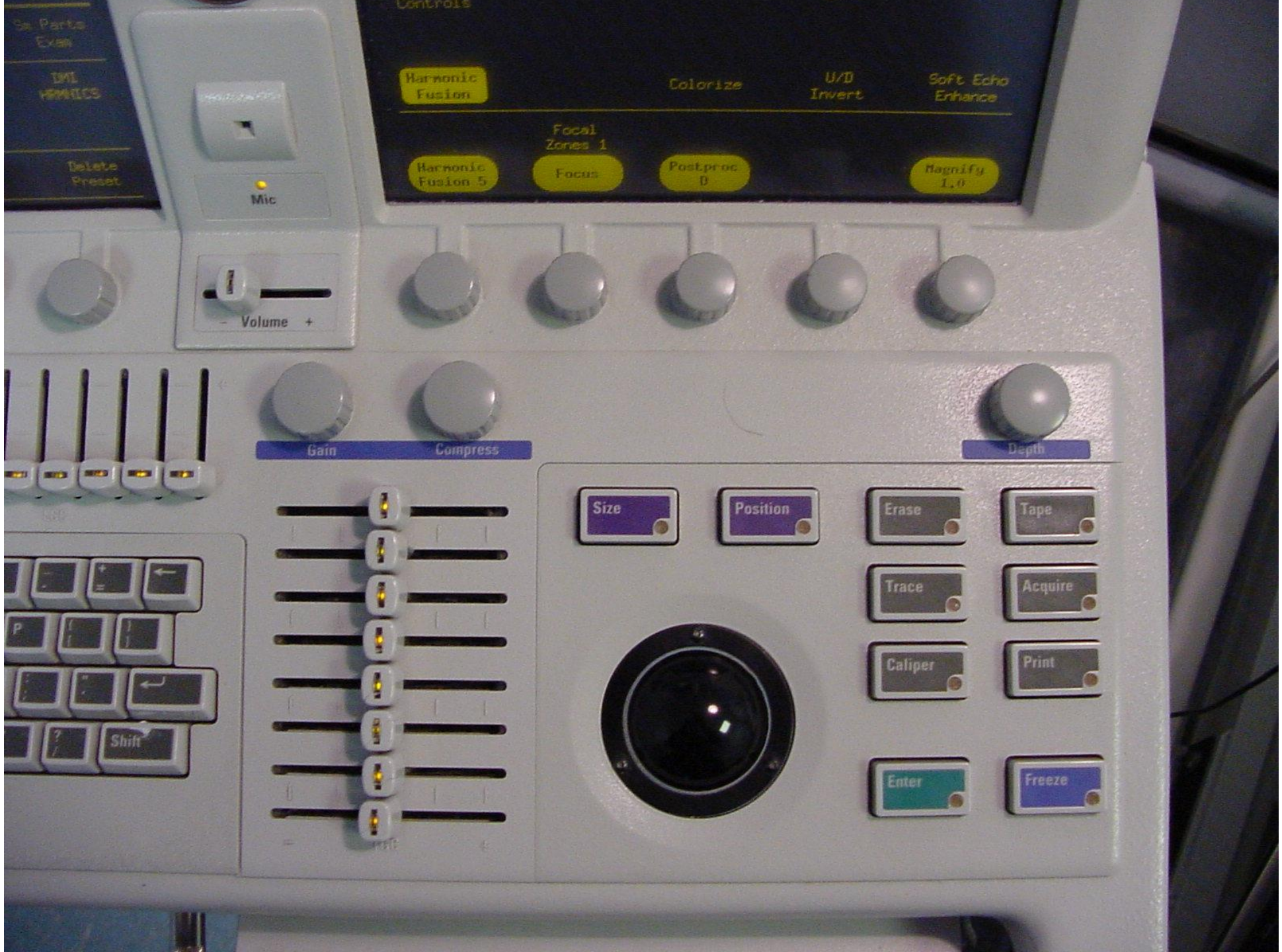
This has to be turned into an image line and displayed with a dynamic range of around 30 dB – How?

Clipping
Depth gain compensation
Dynamic range compression
Reject

Tine gain (or Depth gain) compensation

Changing the gain of the receiver along each line with time compensates for losses in signal strength as the ultrasound comes from greater depths.

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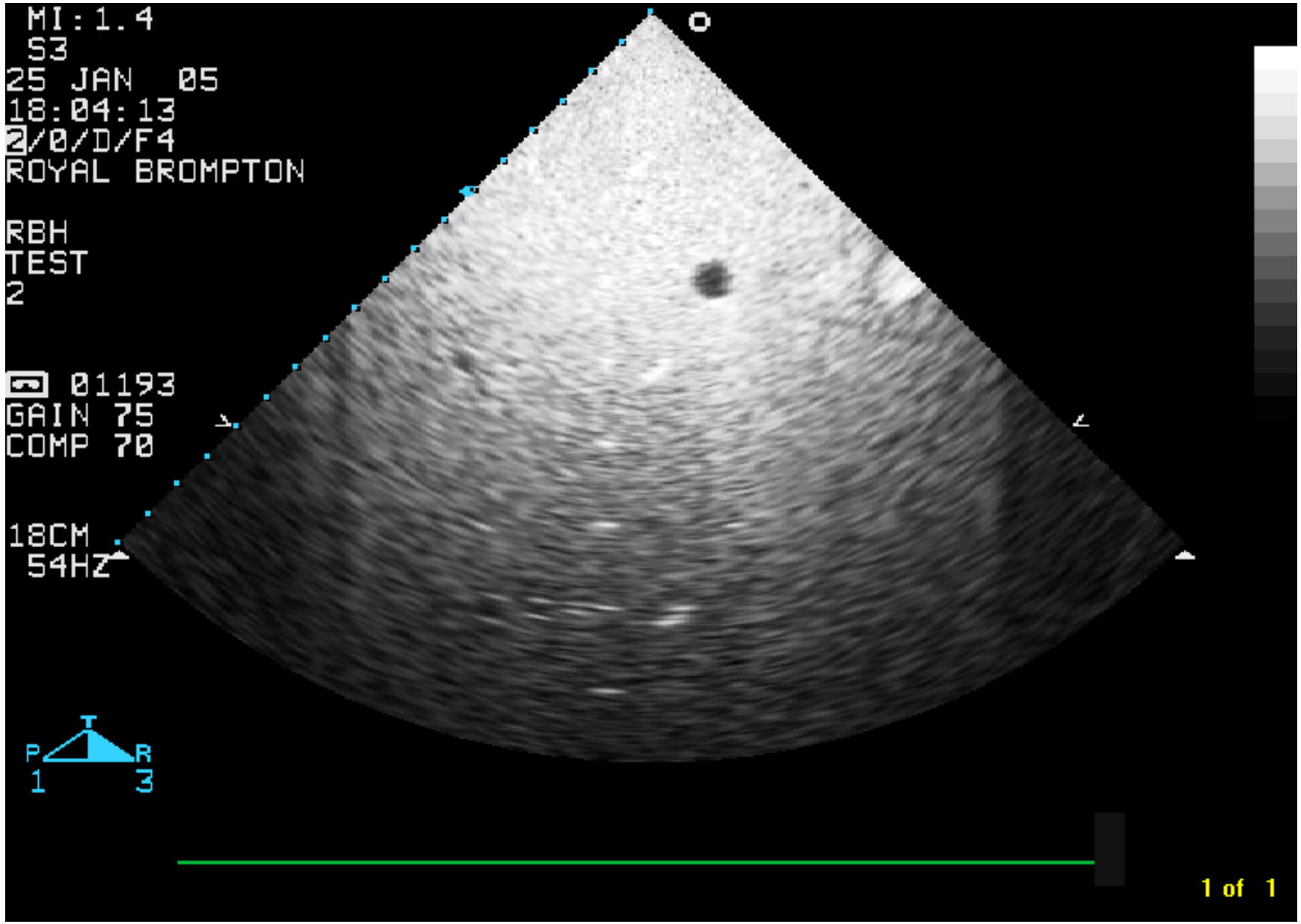
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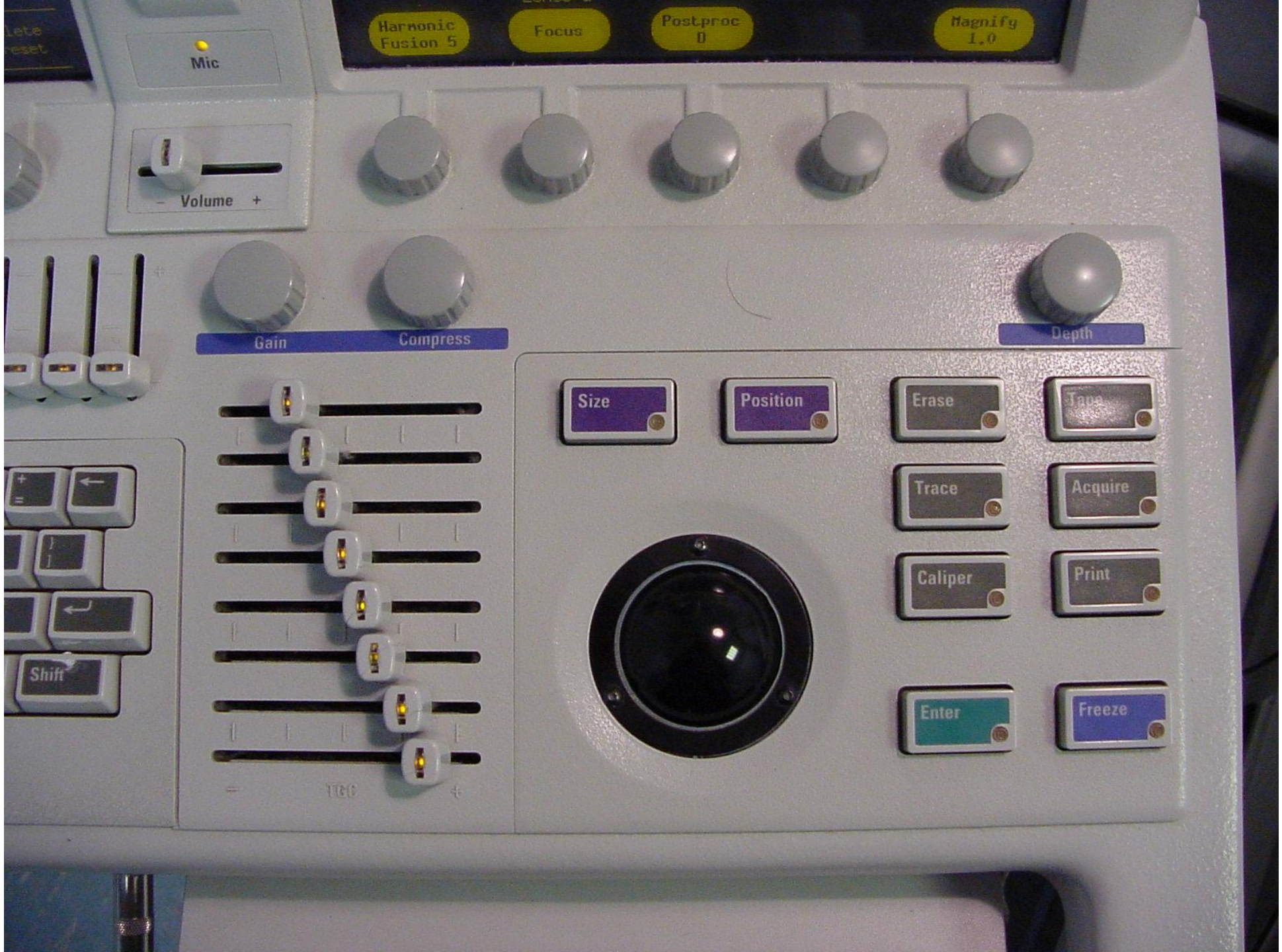
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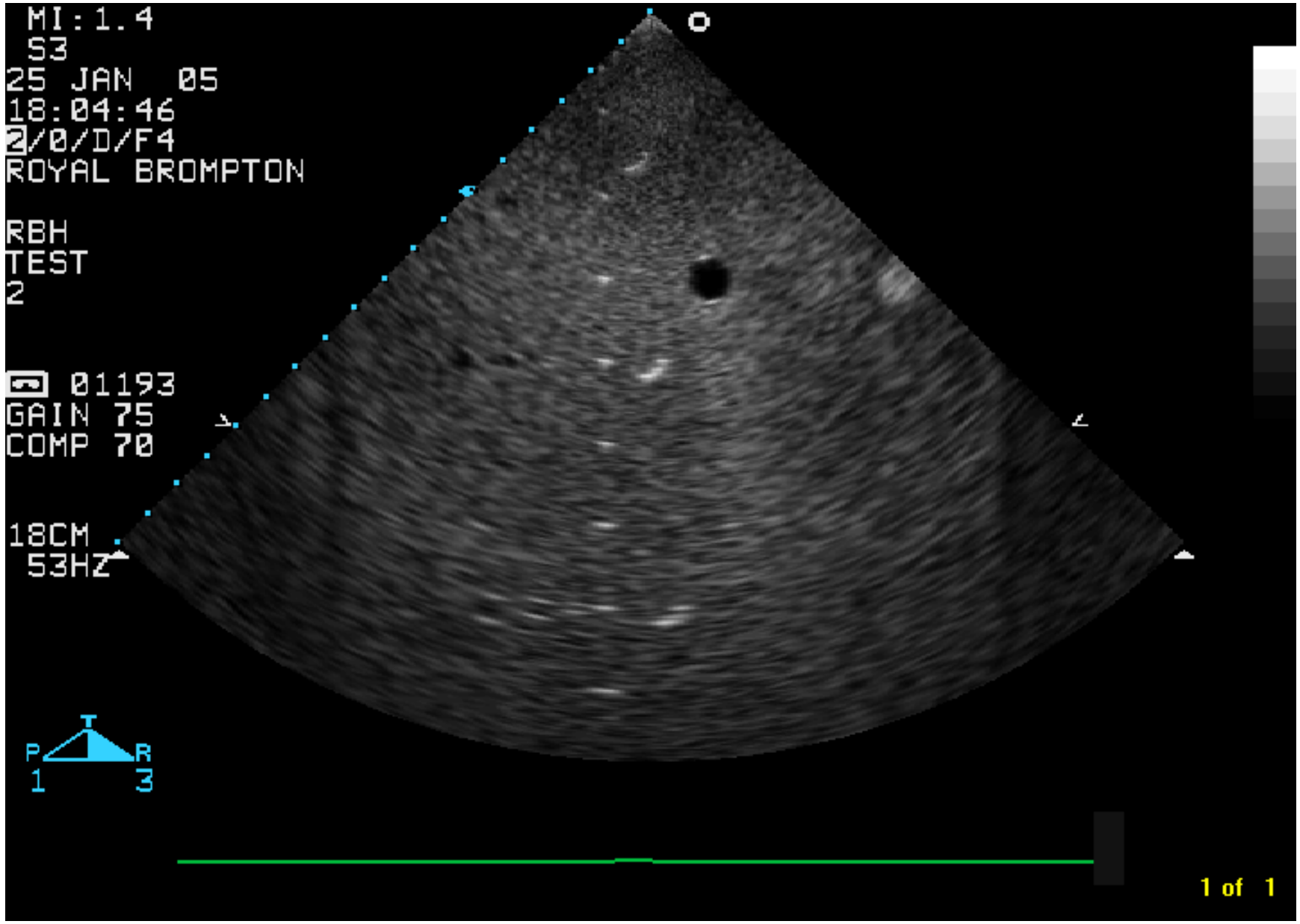
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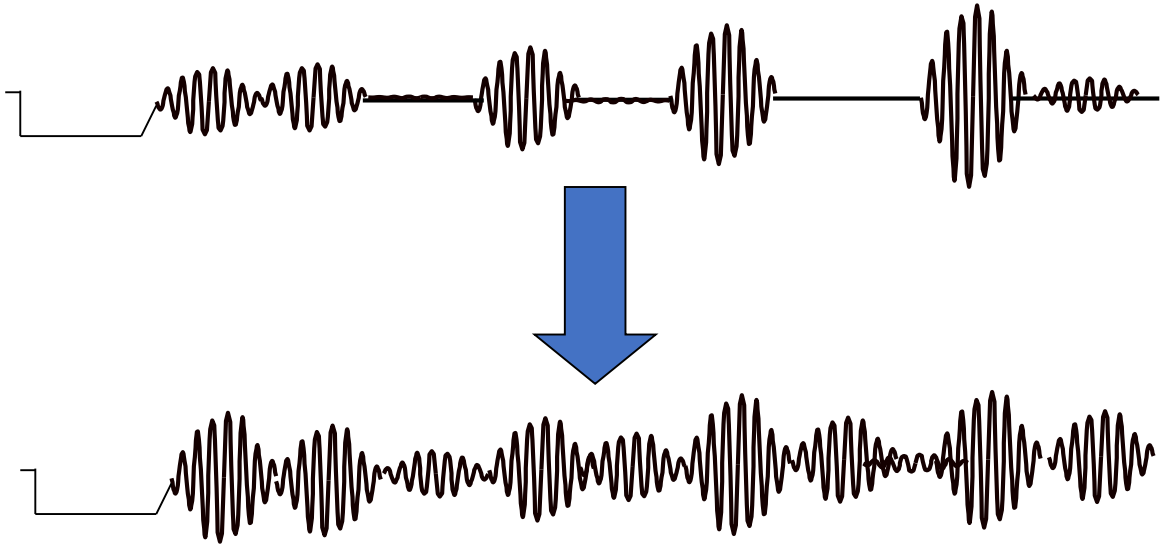
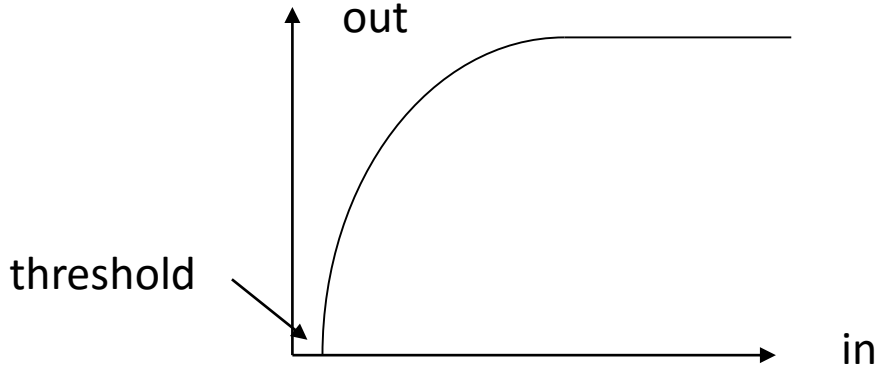
Non-linear (logarithmic) amplification



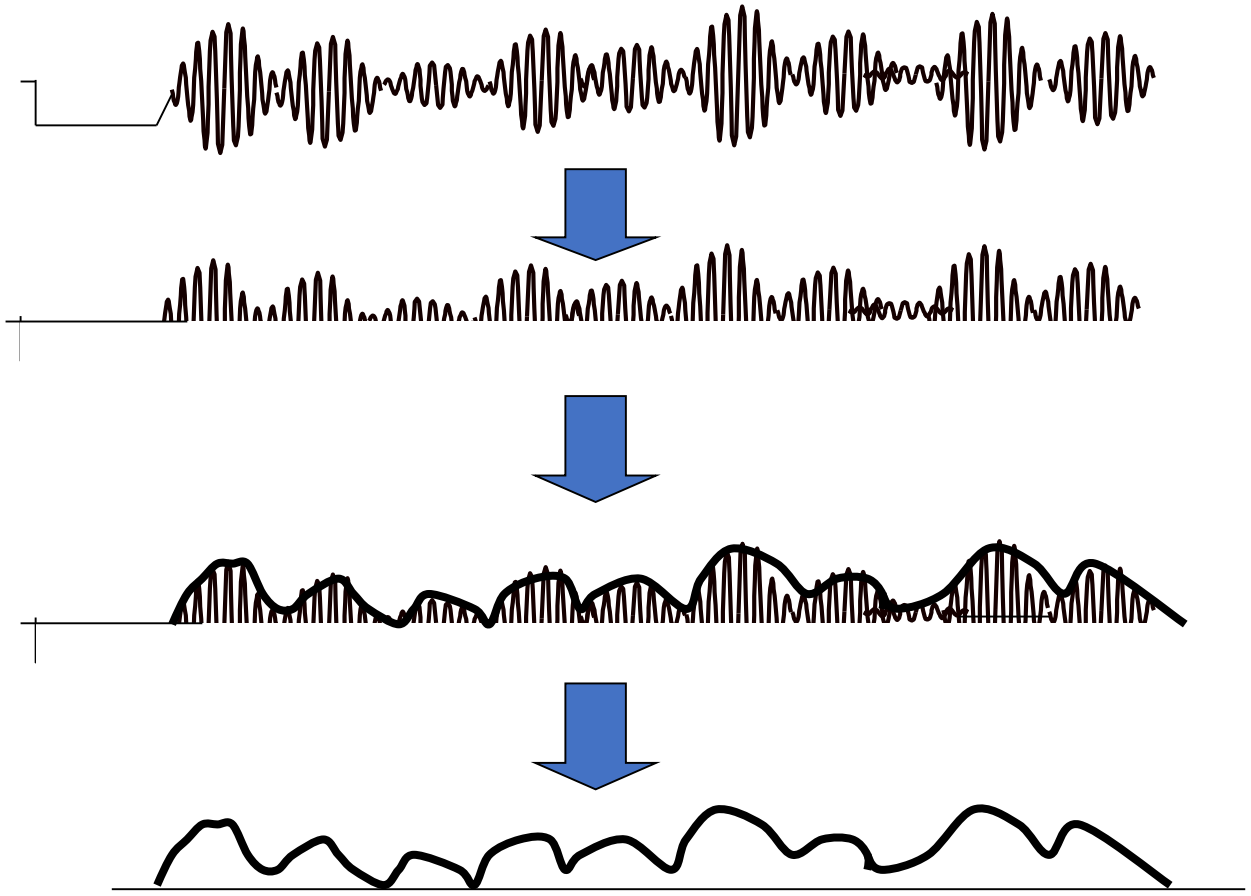
Small echoes amplified more

- Scatter and weak echoes brought up to same order as strong reflections; very strong reflections reduced



"Reject" removes very small signals entirely – these are just noise

Demodulation

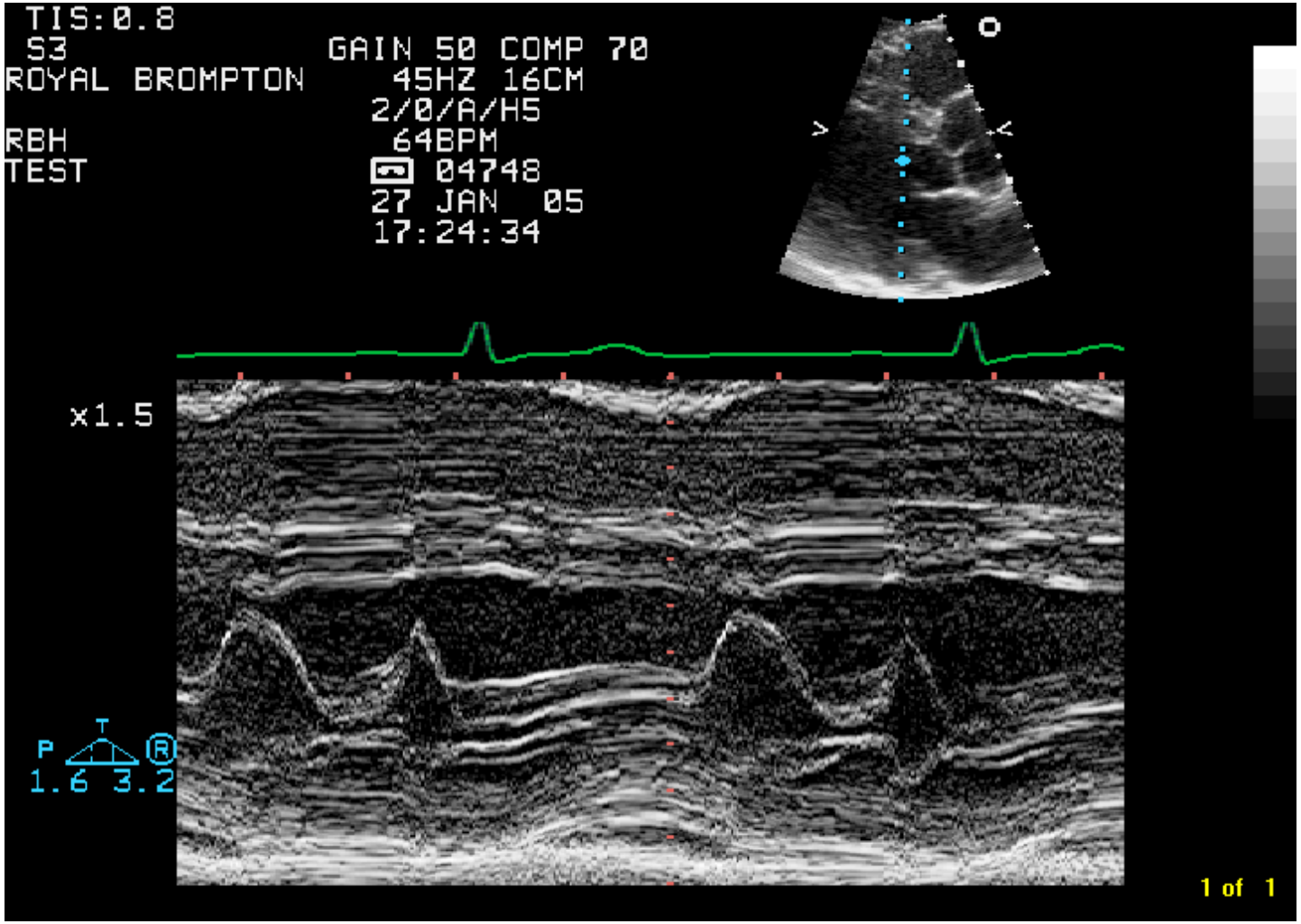


The high frequency oscillating pulses received are turned into a more slowly-varying “envelope” which can be used to place an image on the screen.

The image uses a “grey scale” where the brightness of a point on the screen is related to the amplitude of an echo

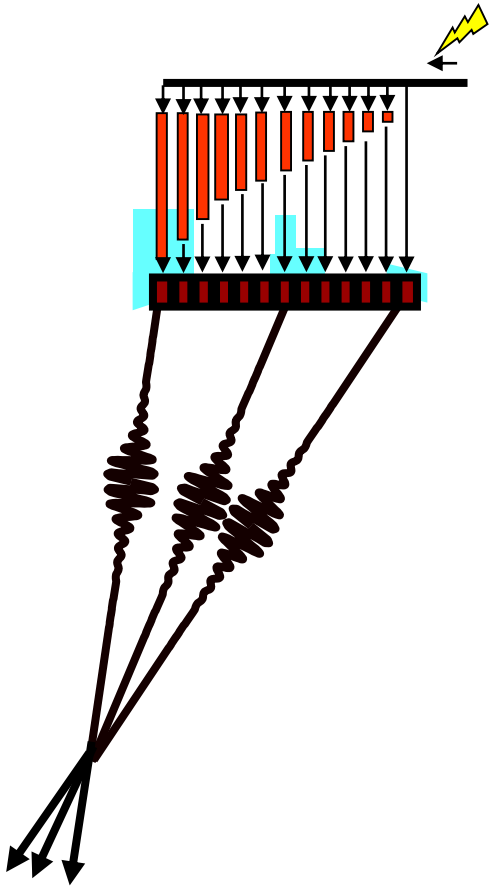
However the relationship between the echo amplitude and the brightness can be changed by the user, to change the brightness and contrast of the image to highlight desired areas of the grey scale.

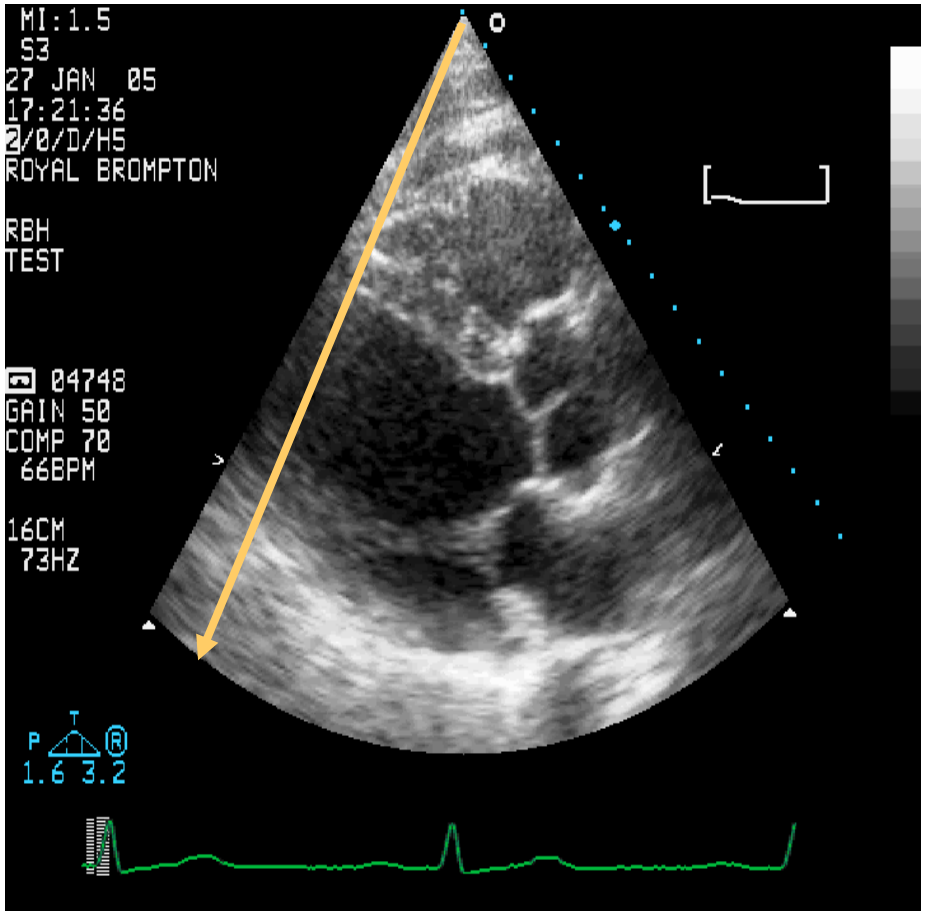
M Mode image – the image consists of a single line only, swept with time to show how the positions of the reflectors change



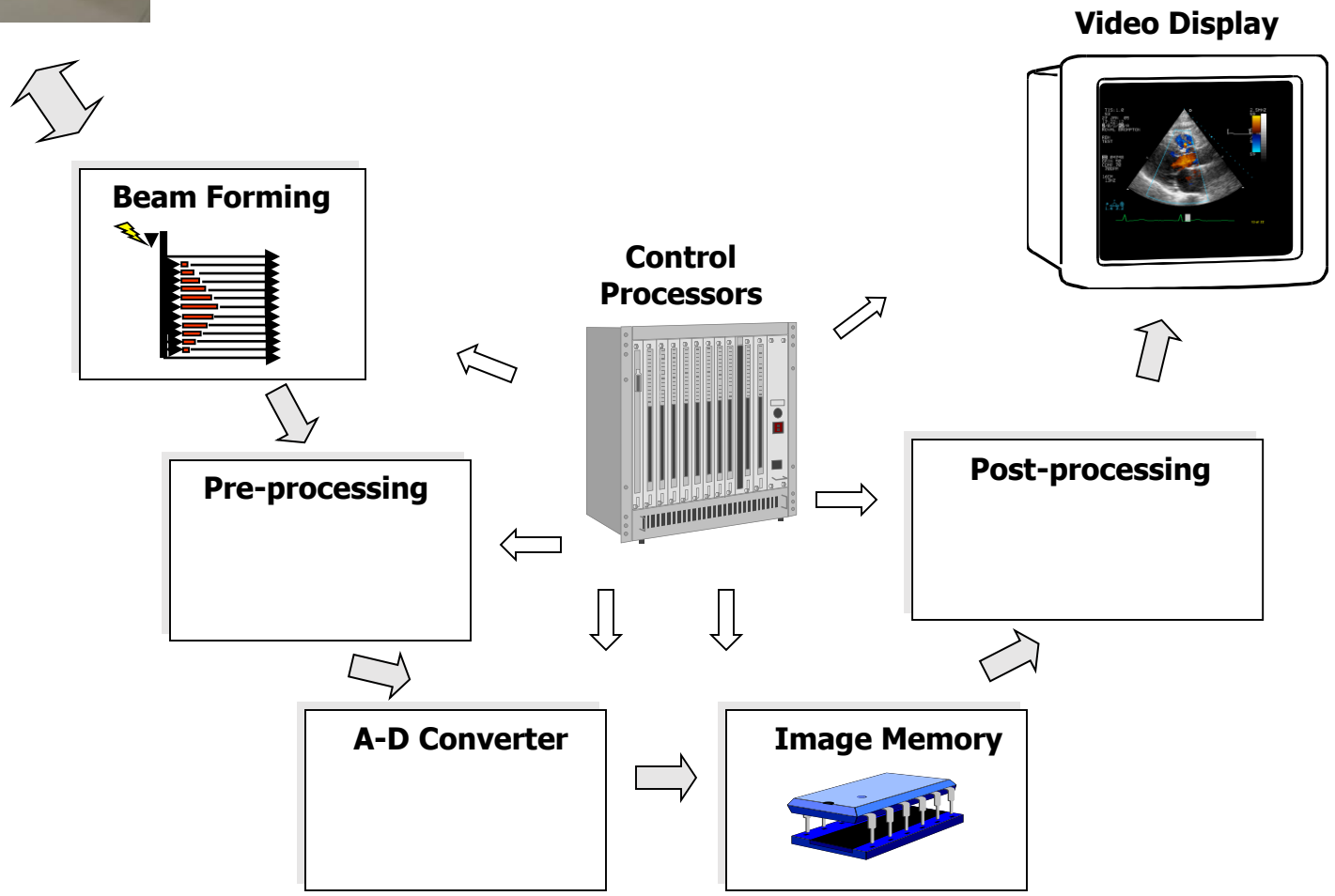
B-Mode Image This signal forms one line of a 2-D image.

The scan line line is placed in image memory with coordinates given by the scan format

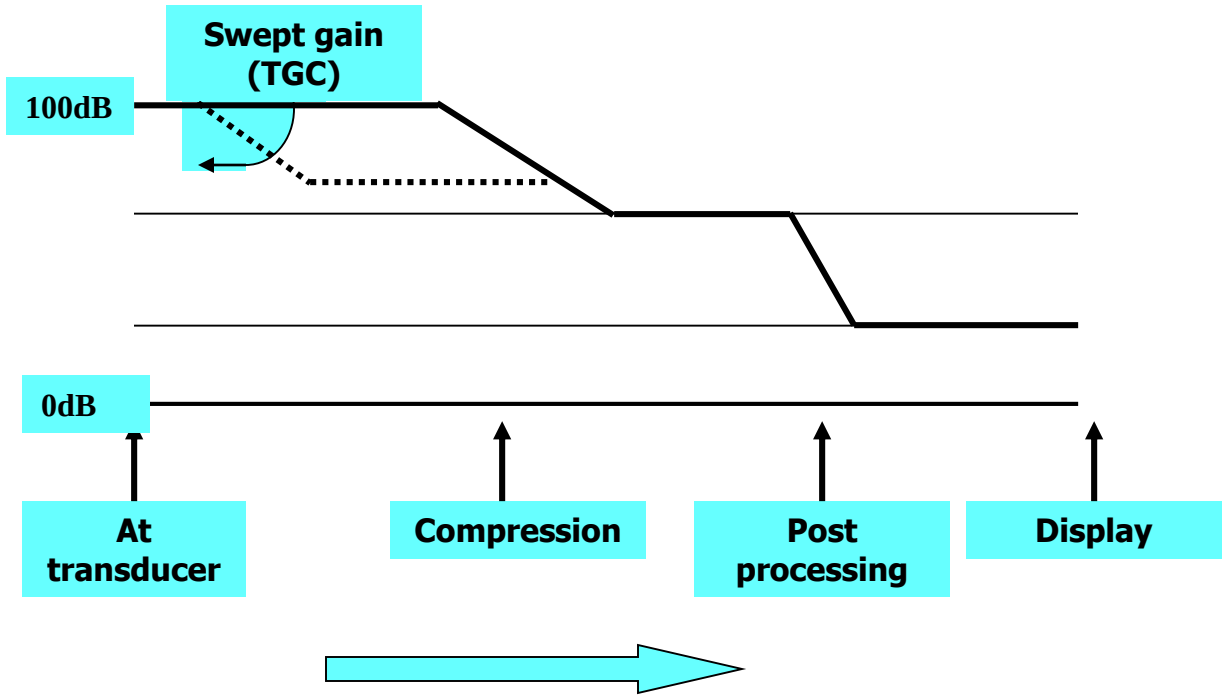


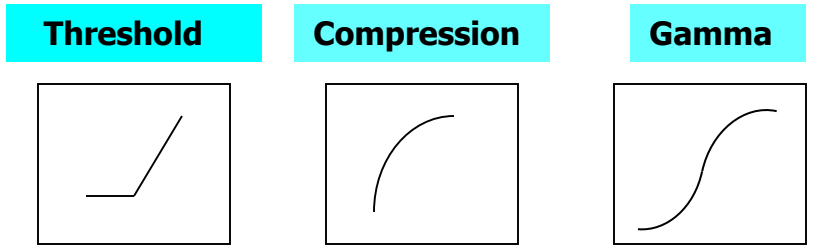


Scanner Architecture

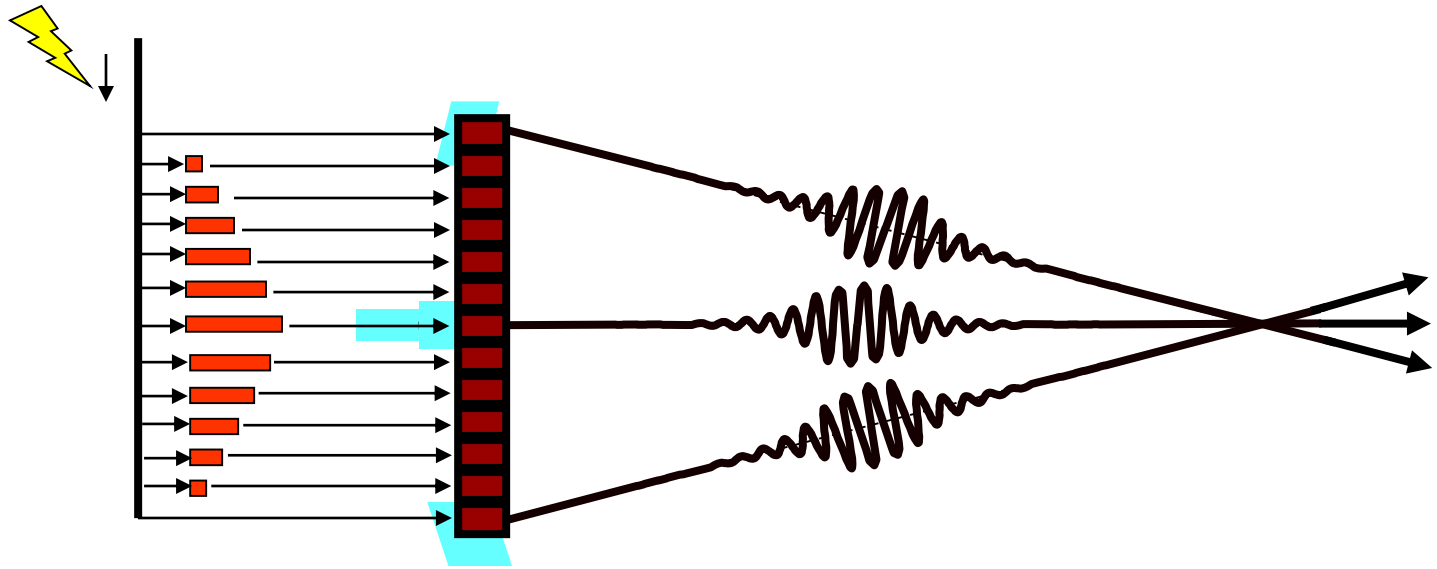


Signal Processing





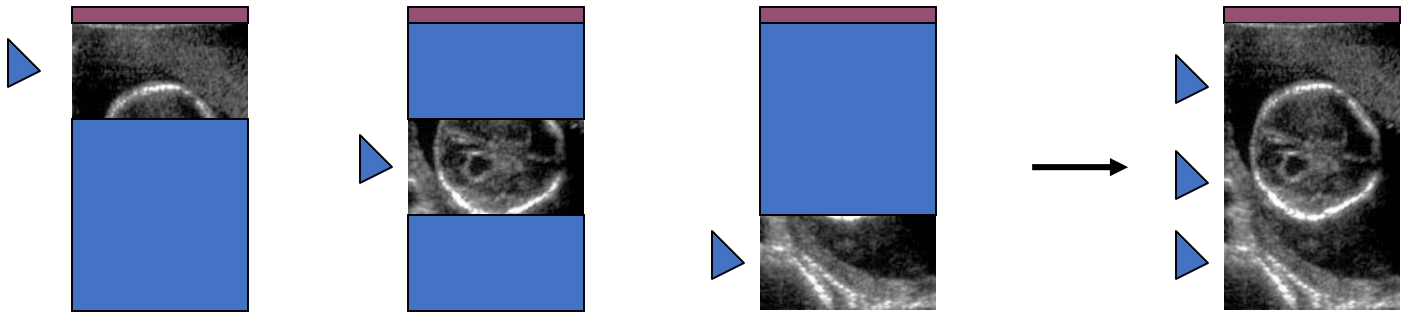
Electronic Focussing



Transmit focusing

- Cannot continually change transmit pulse

  once sent – not easy to modify
- Can however optimise focal zone of transmit pulse for a given depth
- Best to make separate images, each with different focal zone placement, and generate montage of best bits from each

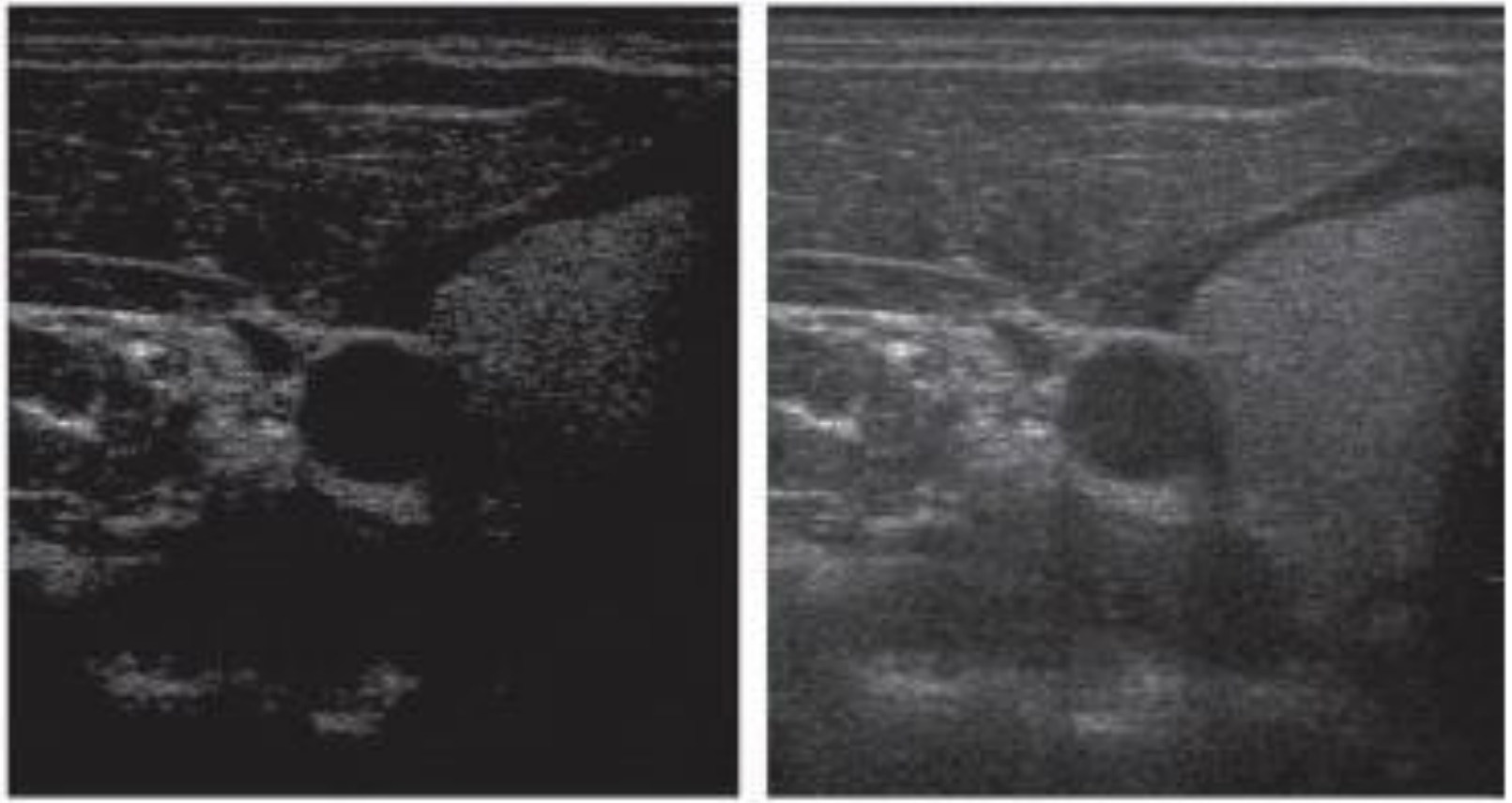


Pre processing and post processing

Pre-processing takes place before the echos are stored in memory and cannot be undone.

Post-processing occurs after storage and can be varied at will

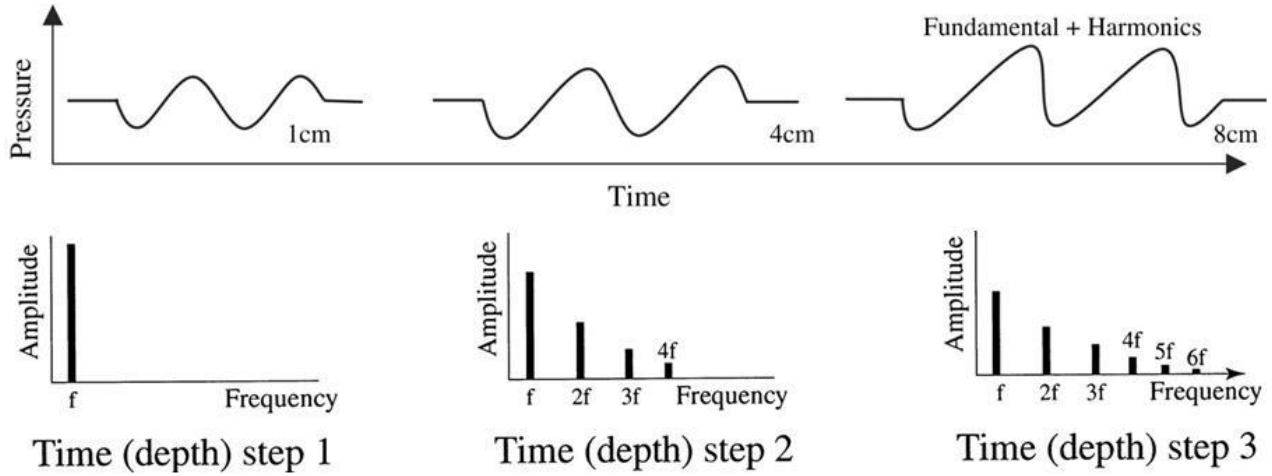
Image Formation showing compression



Harmonic Imaging

When a high amplitude ultrasound disturbance passes through an elastic medium it travels faster during the higher density compression phase than the lower density rarefaction phase causing harmonic distortions.

Progressively stronger harmonic component with distance travelled.



PRO: reduction in artifacts, improved signal-to-noise ratio and slight improvement in lateral resolution.

CON: reduced axial resolution due to longer initial pulse length

RESOLUTION

IMAGE MEMORY-where storage of digitized information contained in the pulse waveforms occurs

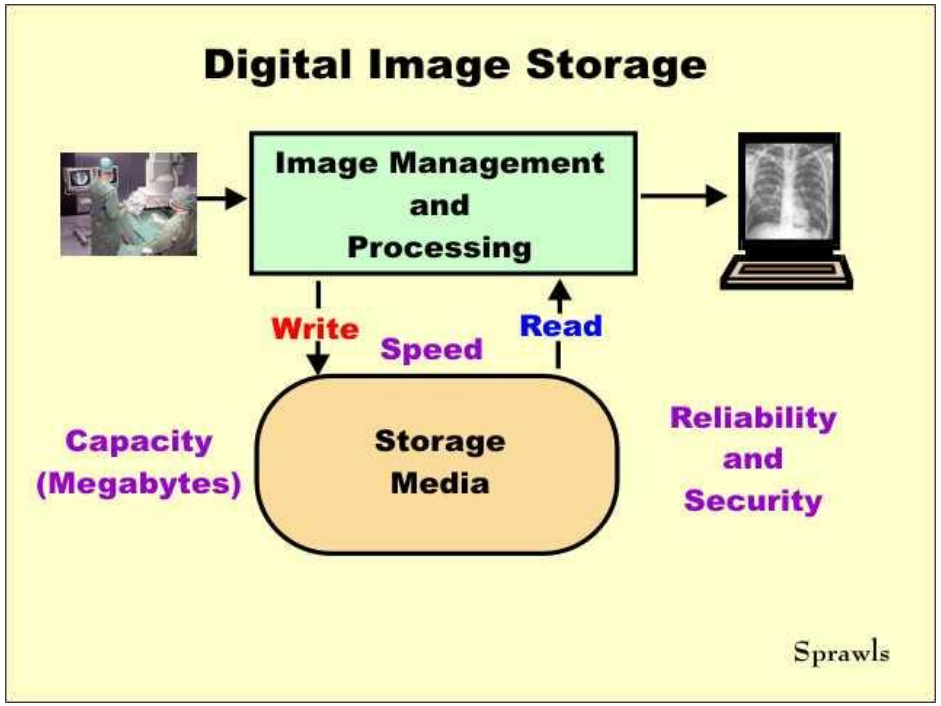
each part of the image memory called a pixel (picture element)

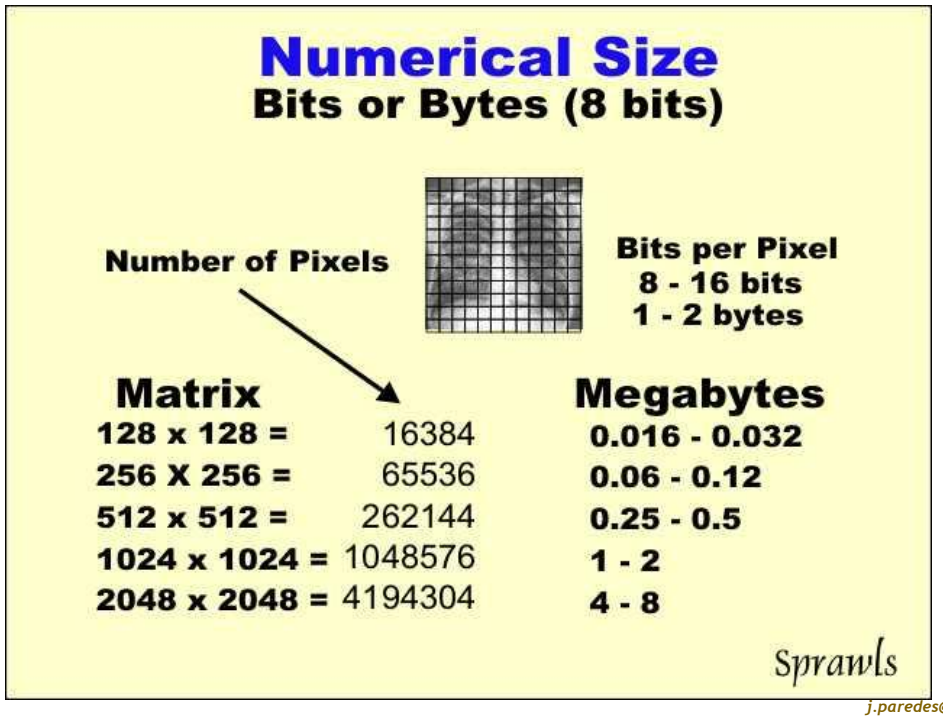
must have sufficient bits (binary digits) as possible to enable various shades of grey to be visualised

Must have sufficient size to store all pixels for high resolution images

Must have sufficient capacity (number of images that can be stored), speed (time required to write/record and read/retrieve images)

NB these considerations also apply to long-term storage systems outside the scanner..





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Doppler ultrasound and blood/tissue velocity estimation
The origins and processing of the Doppler ultrasound signal and how it is used to provide velocity estimation

Doppler Modes

- Continuous Wave Doppler (CW)
- Pulsed Wave Spectral Doppler (PW)
- Colour Doppler
- (TDI)
- Power Doppler

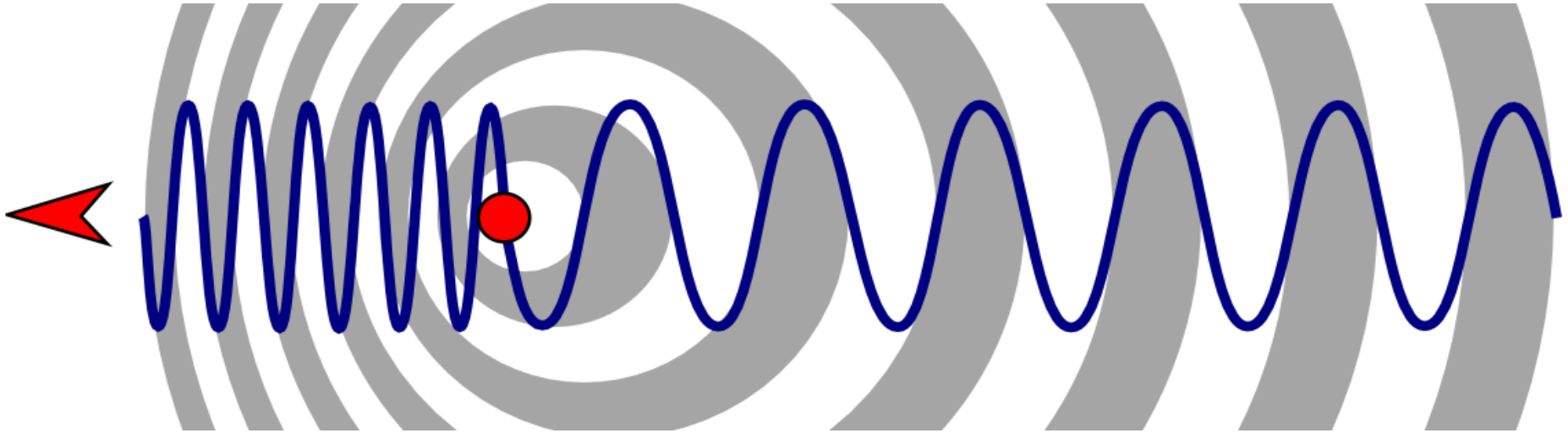
Doppler Effect

- Perceived shift in the frequency emitted by a source due to relative motion between the source and an observer
- Caused by changing distance between source and observer changing time of travel of wave between them.

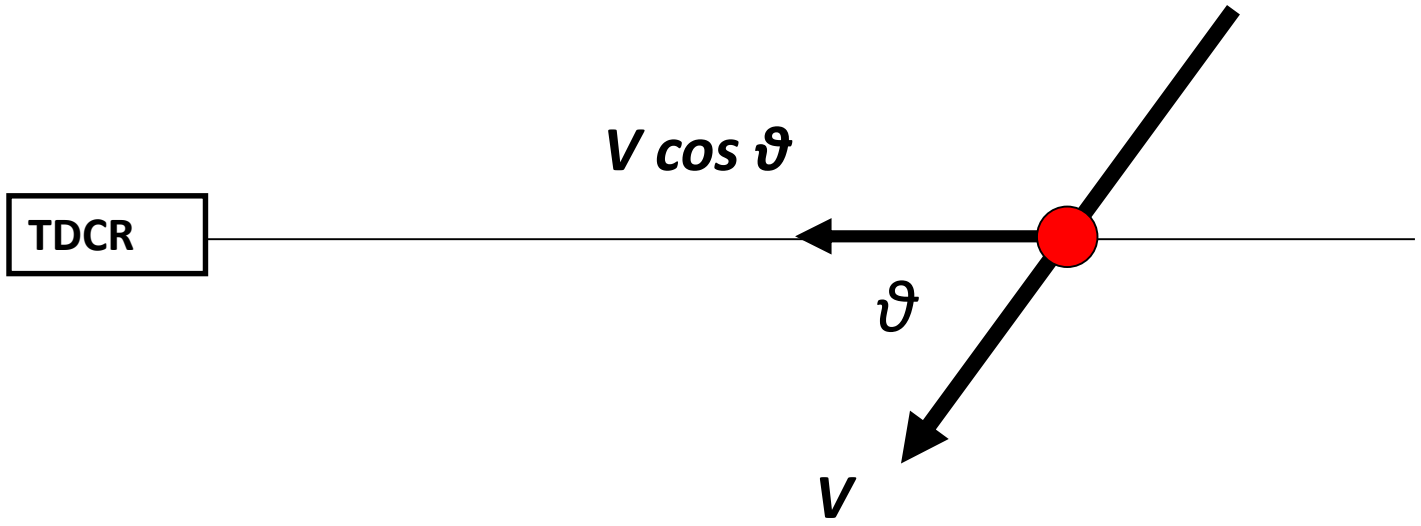
  i.e. changing path length causes successive wave fronts to arrive at the observer sooner than in stationary case if source is moving towards observer (or later if moving away).

The Doppler Effect

When ultrasound interacts with a moving object (i.e. red blood cells) the reflected frequency changes. If the cells are traveling towards the transducer the ultrasound wave is ”squashed” $\downarrow\lambda$ and $\uparrow f$ giving a positive Doppler shift. If RBC’s are traveling away the wave is “stretched” $\rightarrow\uparrow\lambda$ and $\downarrow f$

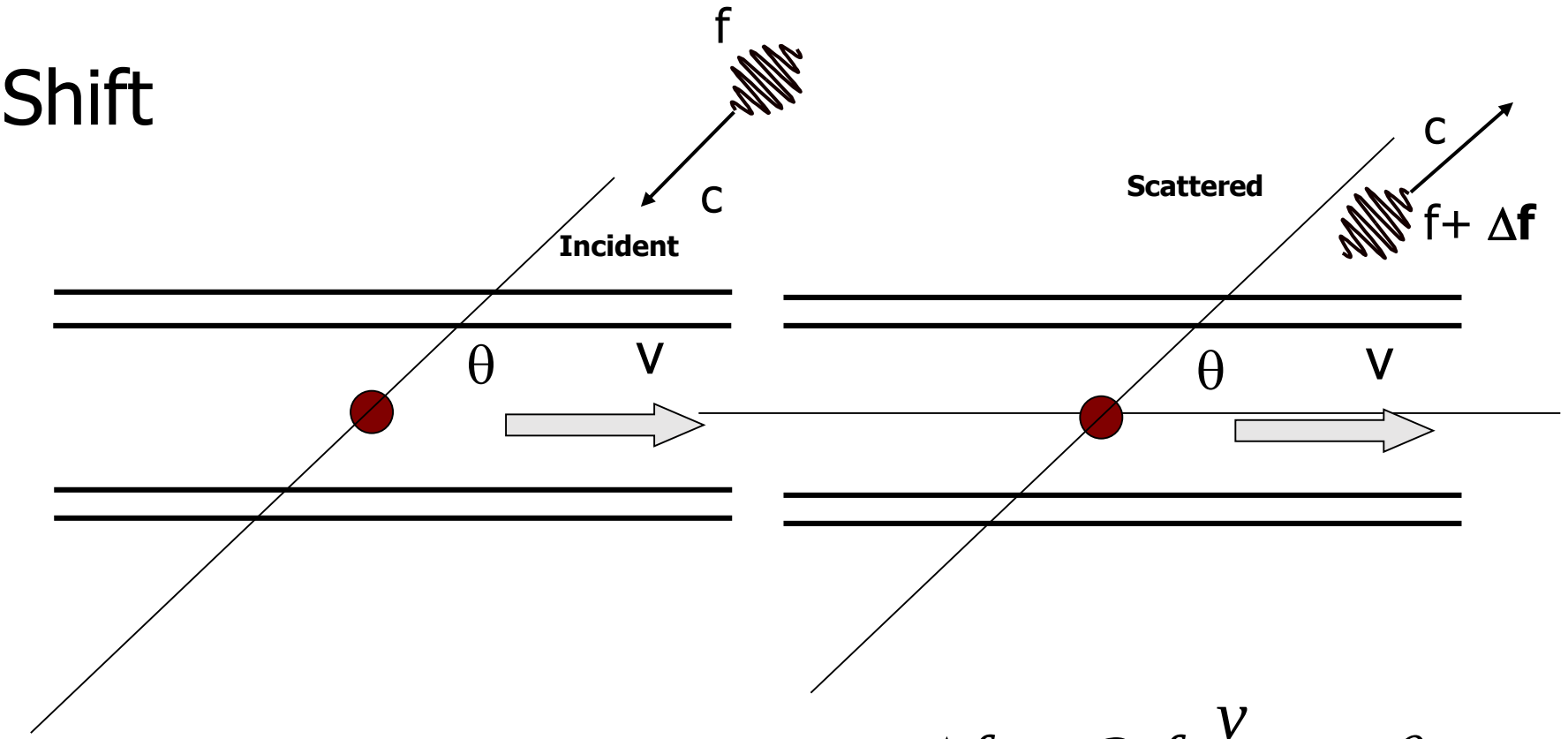


The received Doppler shift is the velocity component towards the observer – if the angle is unknown the frequency shift cannot be corrected to represent the actual velocity



When the angle is zero the cosine tends to 1. This gives the maximum received Doppler shift. In the heart we often measure along the direction of blood flow e.g. in four chamber view. Errors in angle measurement affect the cosine much more as the angle becomes larger, so it is good to keep the angle as small as practicable. At around 50 degrees the error in cosine is equal to the error in the angle.

The Doppler Shift



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

The received Doppler shift Δf is proportional to the ratio of the velocity v of the scatterer to the velocity c of ultrasound.

There is a factor of two because the change in position of the scatterer results in a change in distance in both transmitted and received paths.

Magnitude of the Doppler shift - example

Say $v_{blood} = 0.5\ ms^{-1}$

$c_{blood} = 1.5 \times 10^3\ ms^{-1}$

$\vartheta = 0$; $Cos\vartheta = 1$

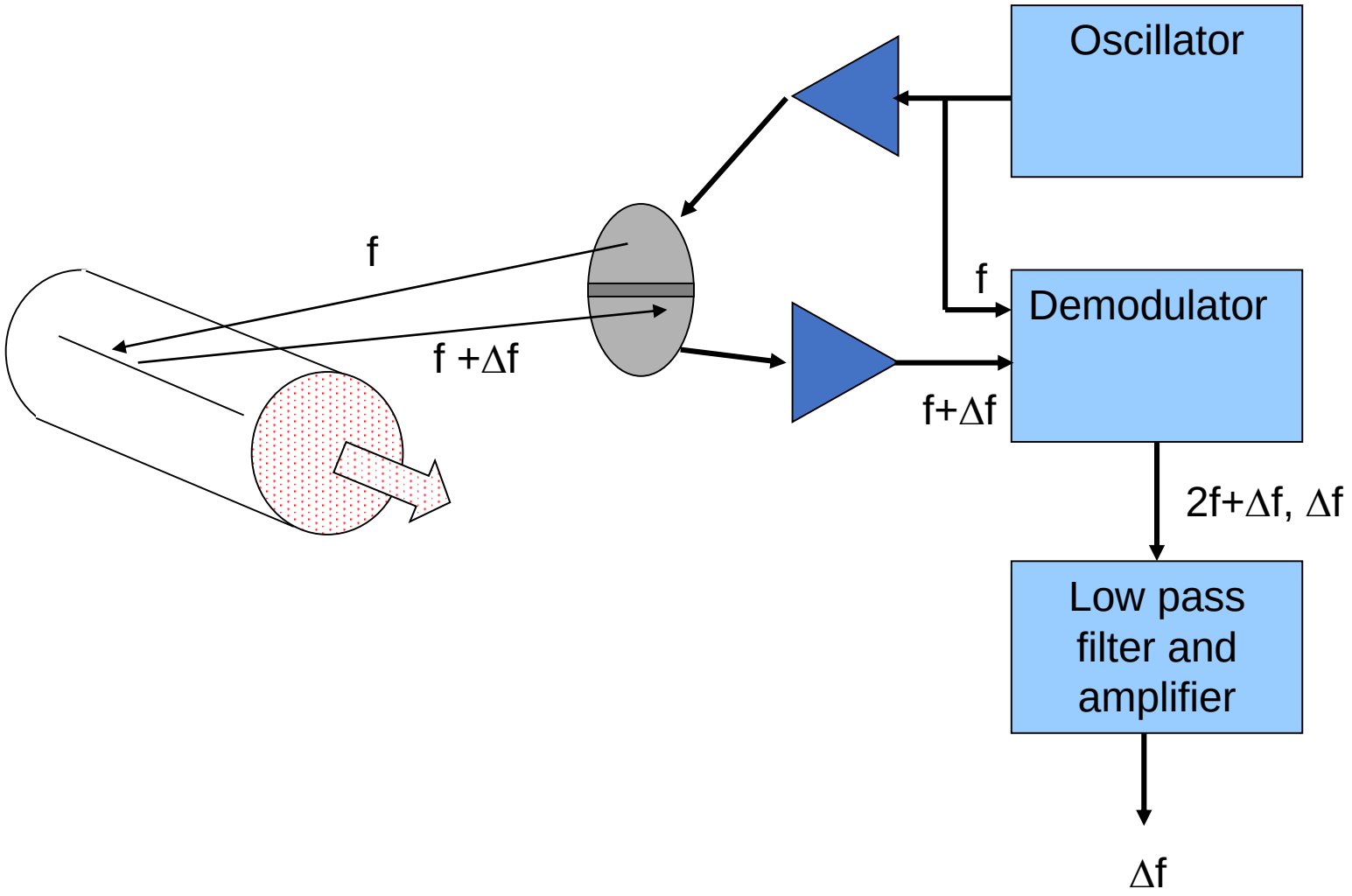
$f = 3 \times 10^6$ Hz

Then $\Delta f = 2\, f\, Cos\vartheta\, v/c$

$= 2 \times 3 \times 10^6 \times 1 \times 0.5 / 1.5 \times 10^3$

$= \underline{\mathbf{2 \times 10^3\ Hz}}$

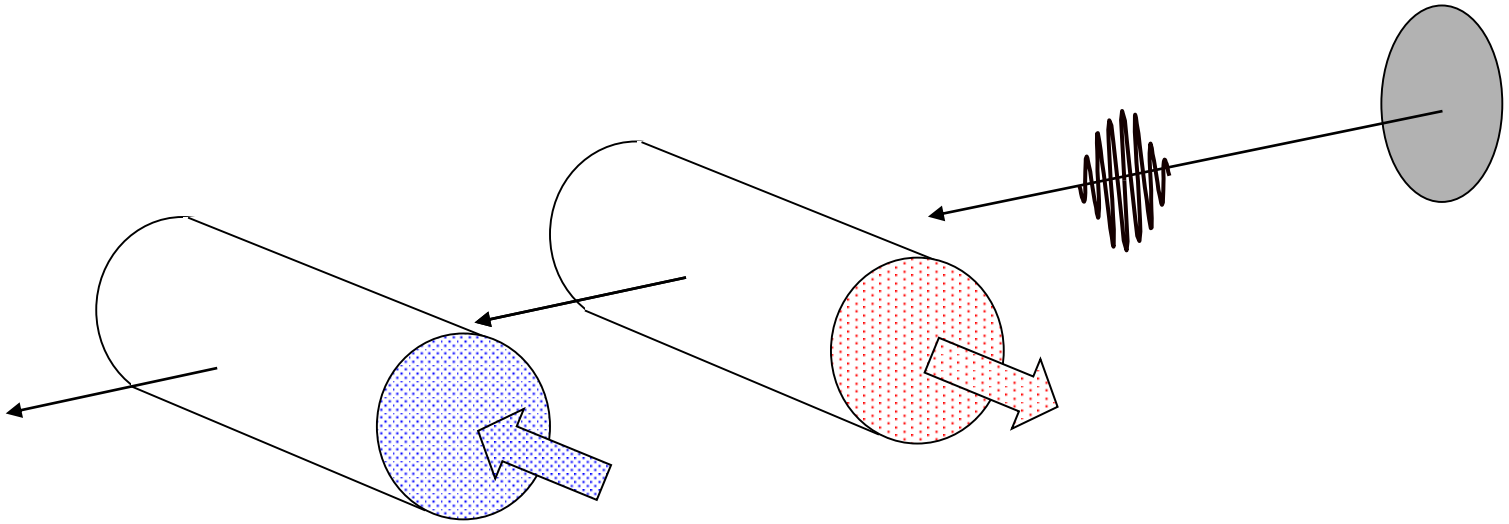
Continuous wave



Continuous wave Doppler

- No distance (depth) discrimination
- Doppler shift produced by all scatterers anywhere within the ultrasound “beam”
- So, prone to interference from unwanted vessels and moving structures
- However no “aliasing”

Pulsed Doppler

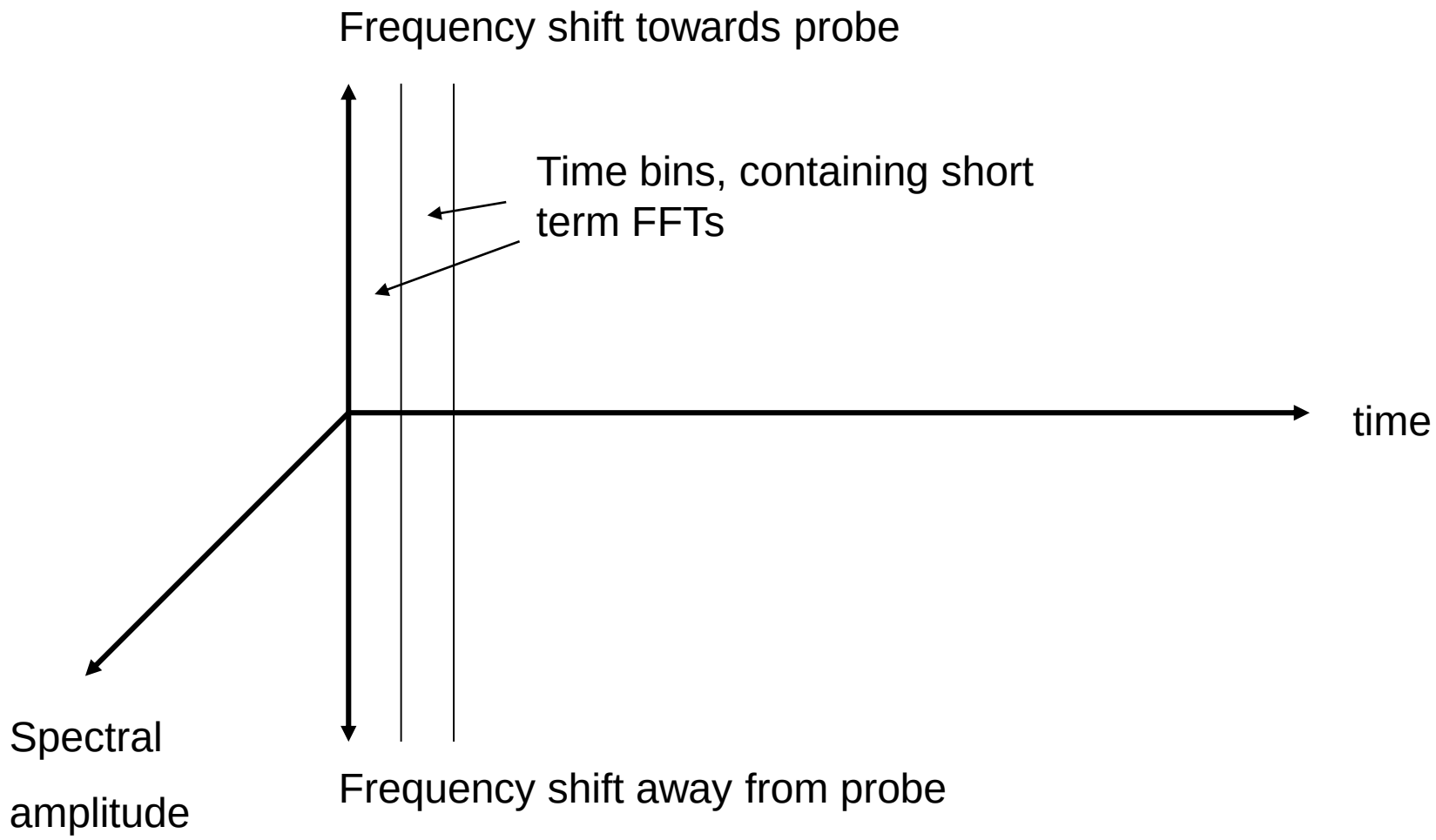


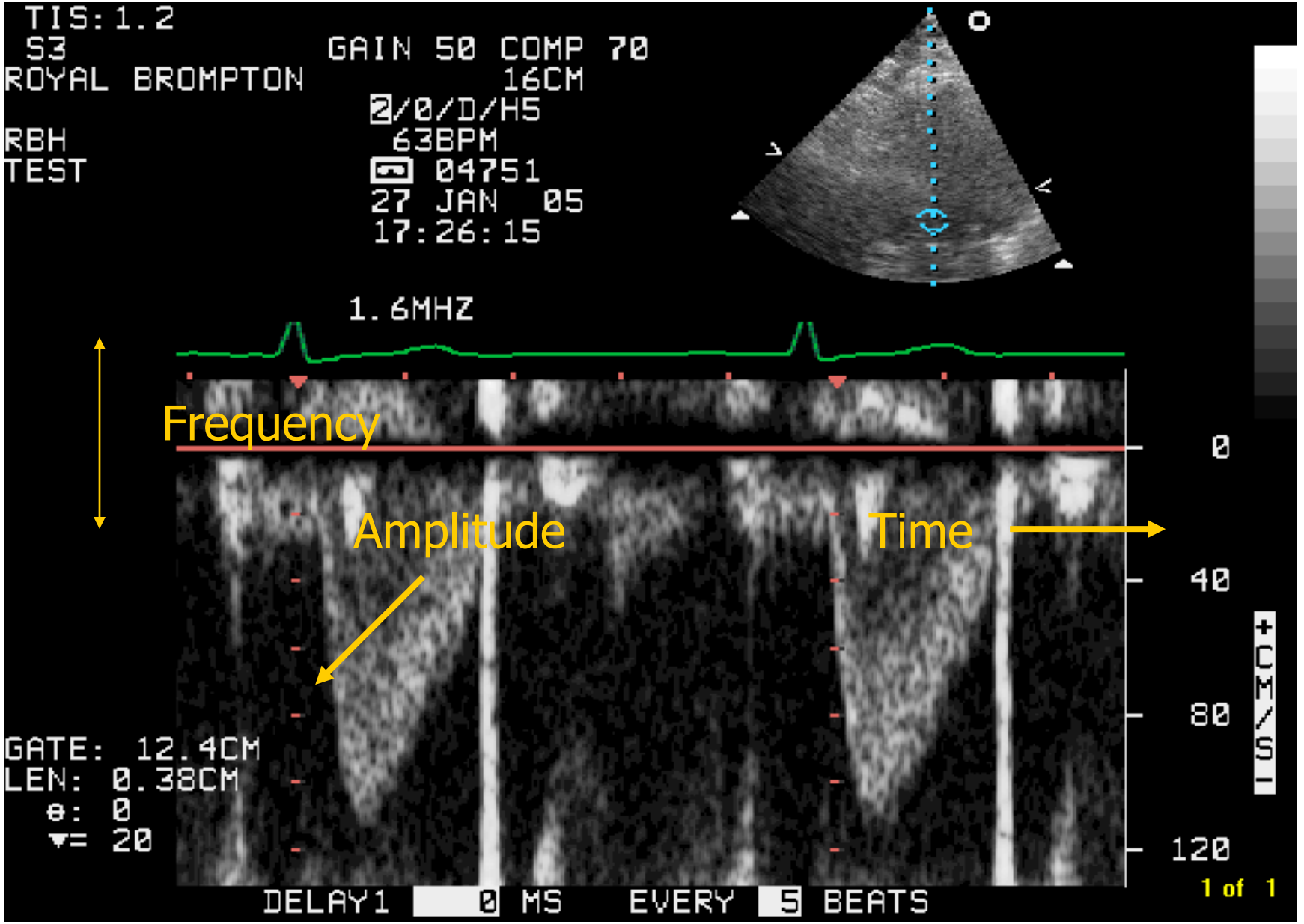
Pulsed Doppler

- Send out short samples (bursts) of a continuous wave signal at regular intervals
- Make the receiver sensitive only to echoes arriving between certain times after transmission i.e. from within a certain range of distances from the transducer
- Add ‘Sample & Hold’ to capture & maintain output between pulses
- Each ultrasound pulse is a sample of how much the wave has shifted i.e effectively how much the frequency has changed.

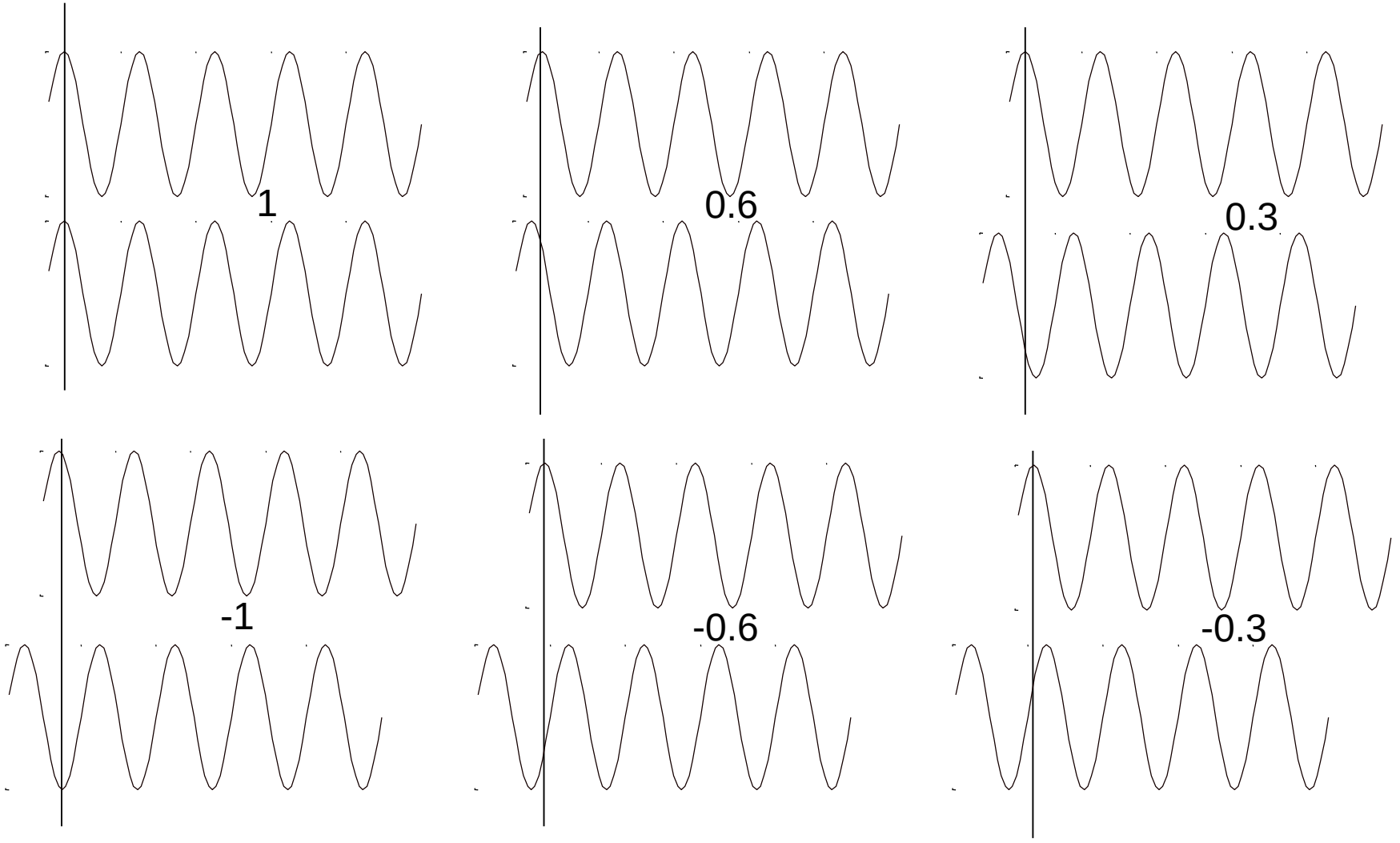
The spectrogram

The **Fast Fourier Transform** is used to mathematically process the Doppler signal to extract the amplitudes of all frequencies present in a short time segment.

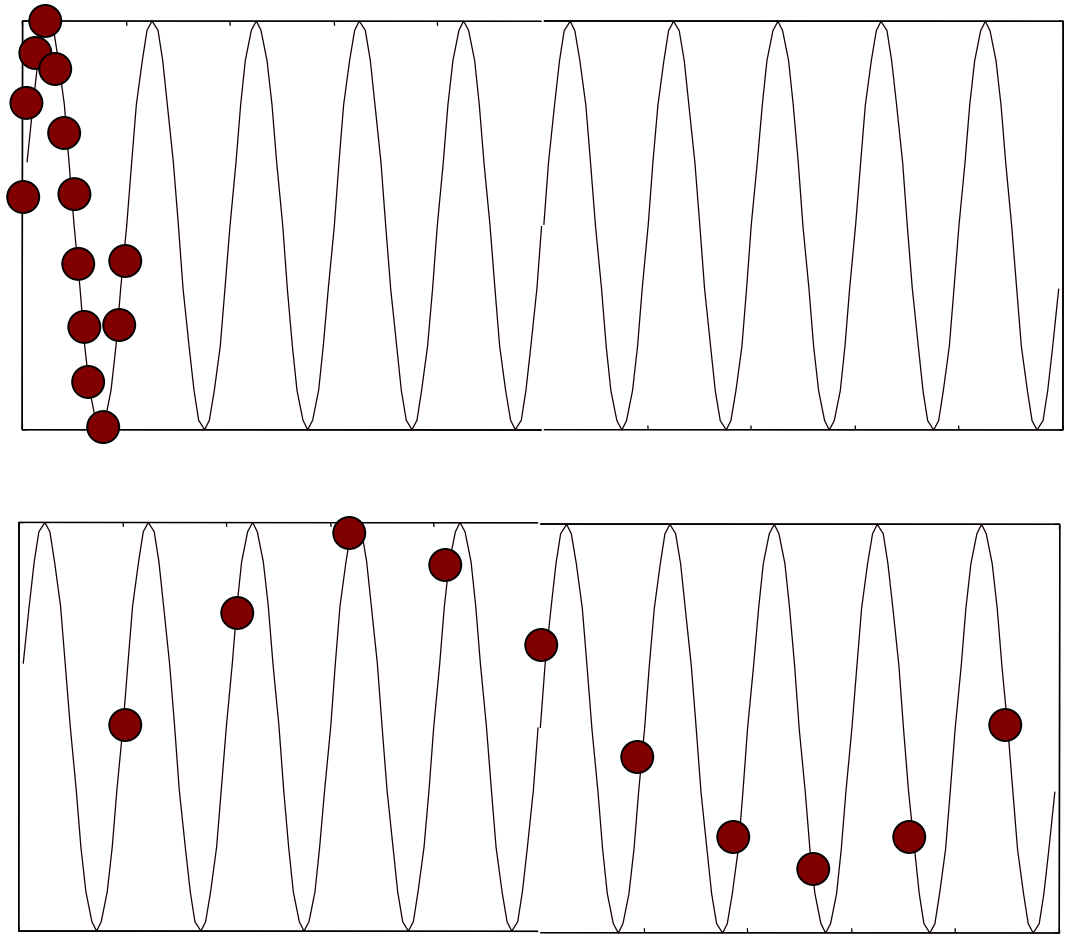


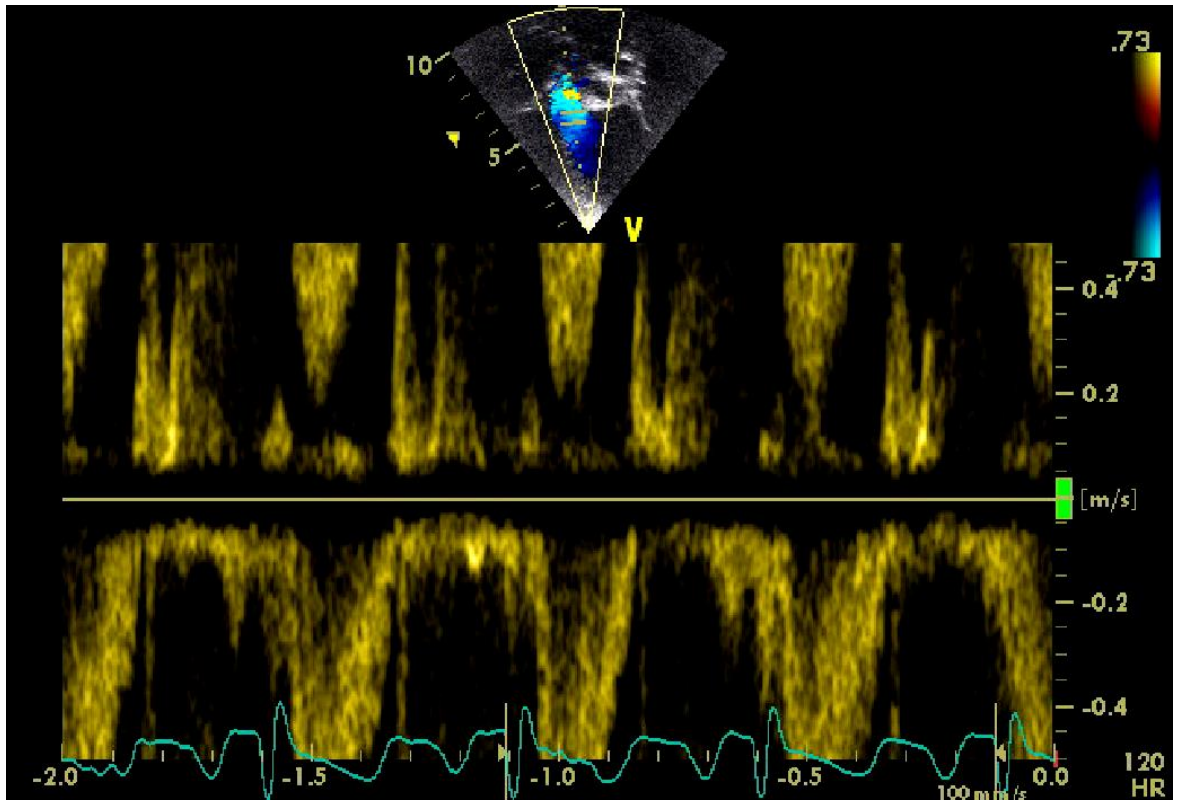

TIS:1.2
S3
ROYAL BROMPTON
RBH
TEST
GAIN 50 COMP 70
16CM
2/0/D/H5
63BPM
04751
27 JAN 05
17:26:15
1.6MHZ
Frequency
Amplitude
Time
0
40
80
120
GATE: 12.4CM
LEN: 0.38CM
θ: 0
▼= 20
DELAY1 0 MS EVERY 5 BEATS
1 of 1

Pulsed Doppler Sampling



Aliasing Occurs when sampling rate is too low compared with frequencies present in the Doppler shift signal



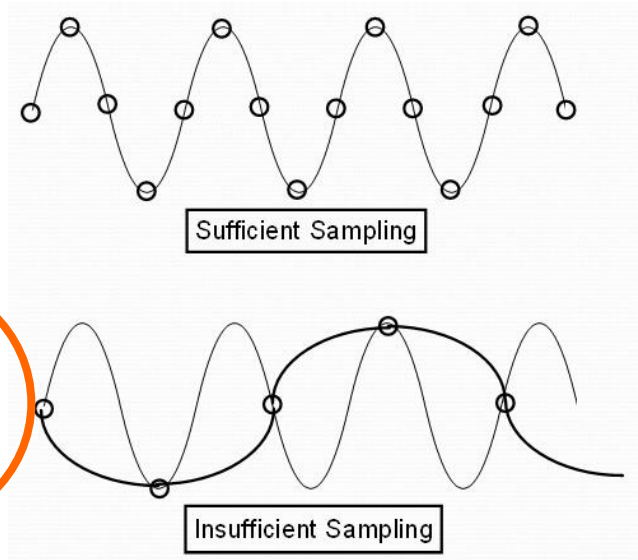


The Nyquist Limit (Aliasing)

The maximum Doppler shift (Δf_{max}) able to be displayed without aliasing.

Determined by the sampling rate (PRF).

$$\text{Nyquist Limit: } \Delta f_{max} = \frac{PRF}{2}$$



Recall that: $PRF_{max} = \frac{c}{2D}$ $\Delta f_{max} = \frac{c}{2x2D}$

High PRF

- Transducer sends out an additional pulse before the original pulse has returned.
- In effect it doubles the PRF and therefore doubles the Nyquist limit.
- The disadvantage is that the exact origin of the Doppler shift is not known.
- Potential for range ambiguity artifact (“depth confusion”)

Signal amplitude compared with soft tissue interfaces

Signal levels about 40db below those from soft tissue interfaces.

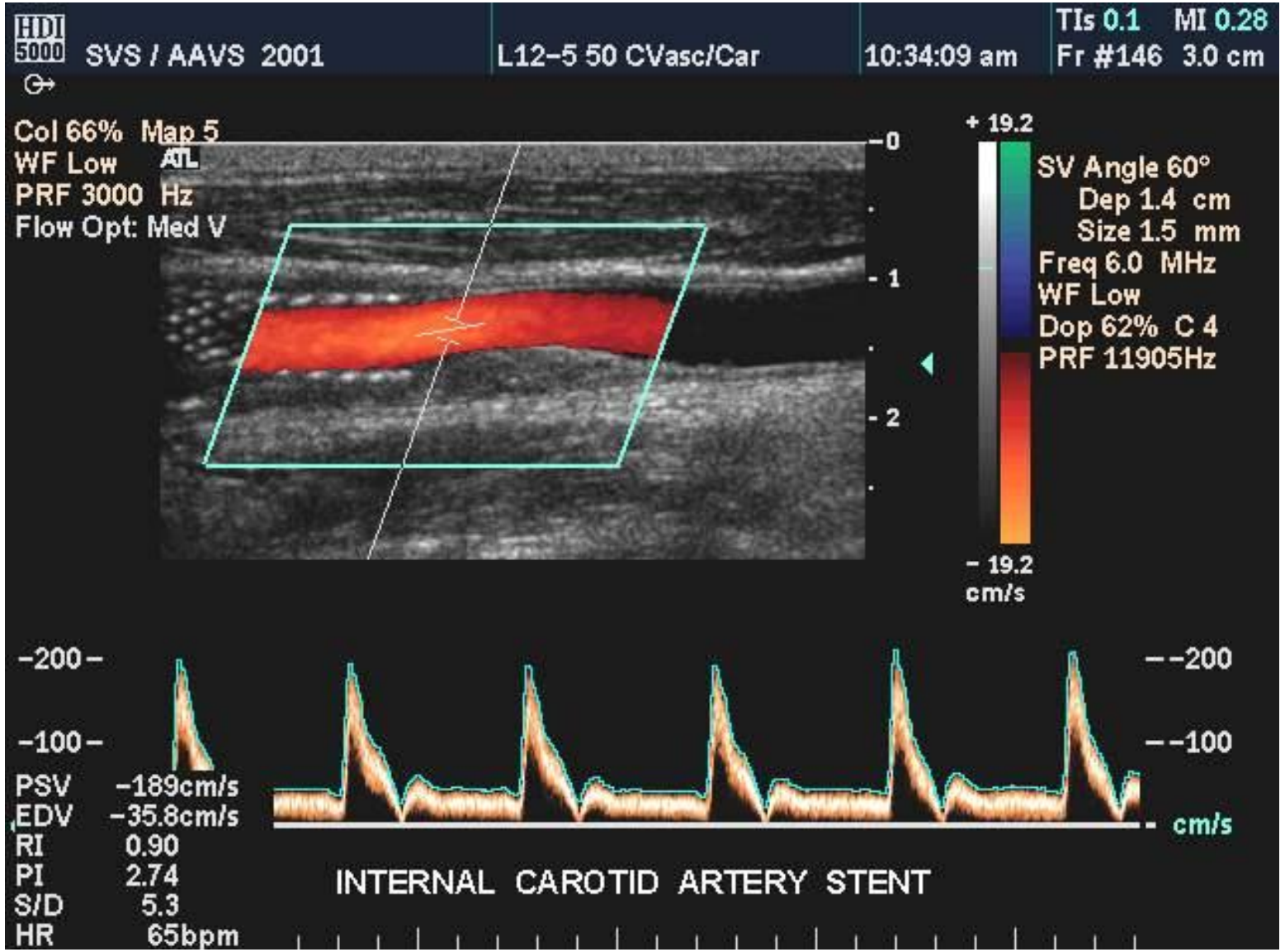
So “Doppler” signal requires much more amplification than “image” signal – so can pick up unintended moving tissue in sample volume

So we use a high pass filter to reject these high amplitude low velocity echos from the Doppler

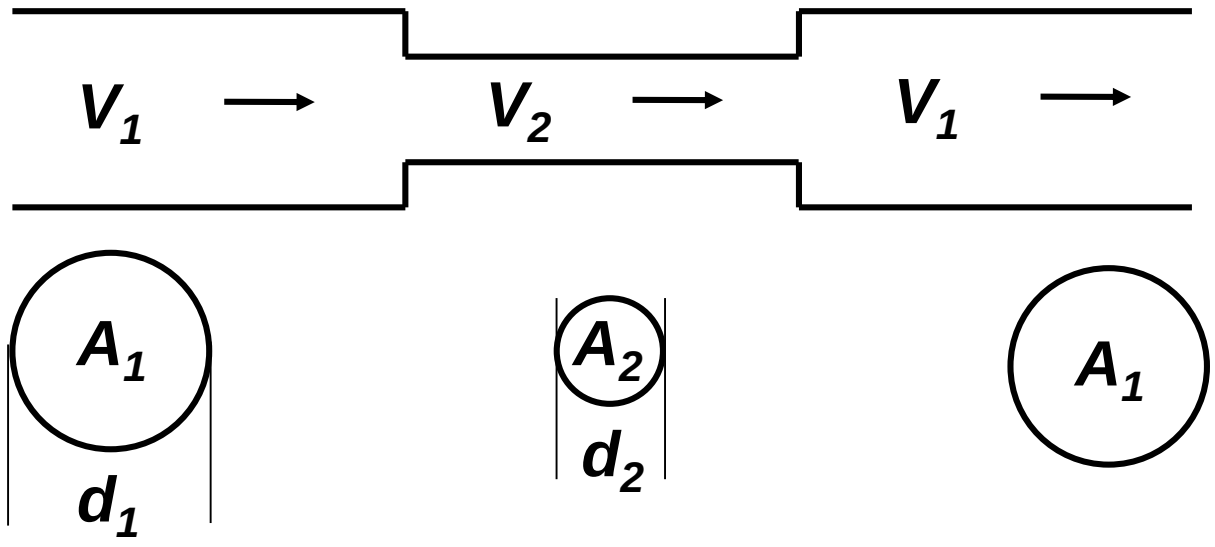
Fluid Dynamics

- Velocity at a Stenosis
- Volume Flow
- Flow Profiles and their associated mean velocities
- Spectral Doppler representation of Flow Profiles
- Transmitted Power and Receiver Gain
- Time Gain Compensation (TGC)
- Reject
- Spatial and Temporal Smoothing

Velocity at stenoses



Velocity at stenoses



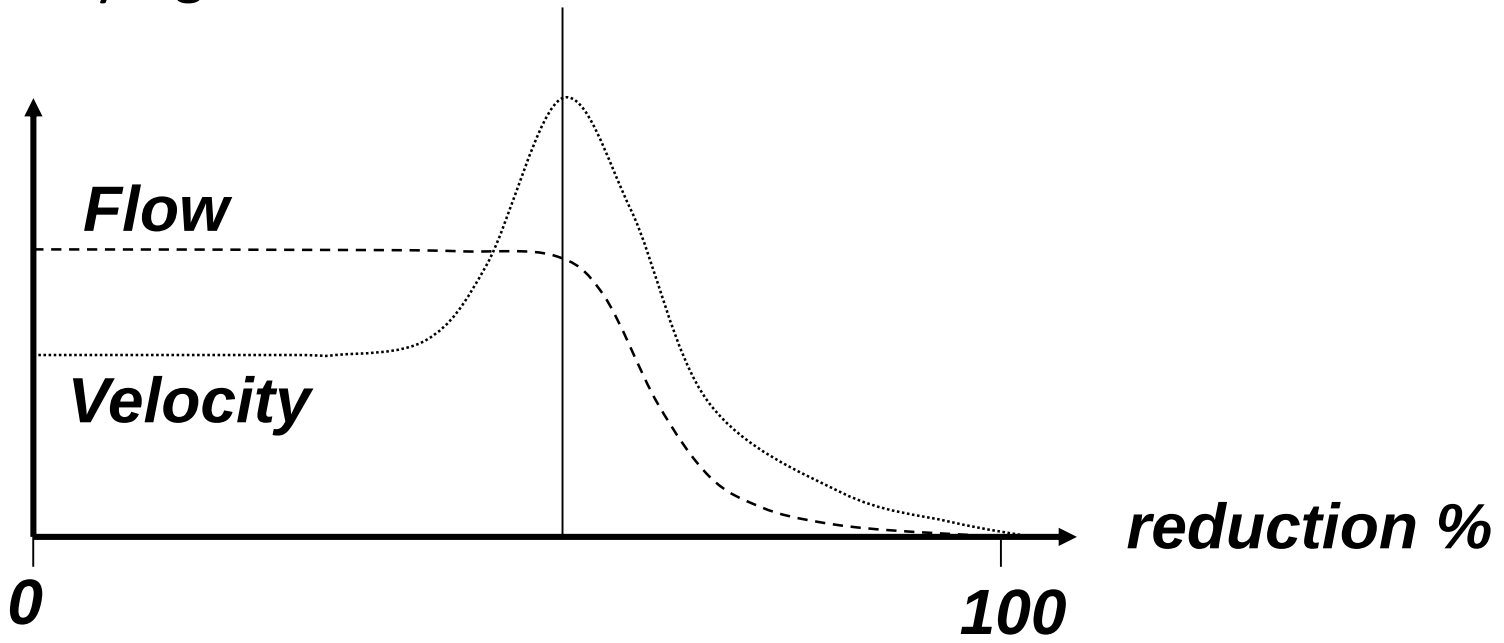
$A_1 V_1 = A_2 V_2$ (conservation)

Hence $V_2/V_1 = A_1/A_2 = d_1^2/d_2^2$

Velocity at stenoses (in vascular examination e.g. Carotid)

- **Diameter ↓ 30% ~ area ↓ 50%**
  - "haemodynamically significant"

- **Diameter ↓ 70% ~ area ↓ 90%**
  - "clinically significant"

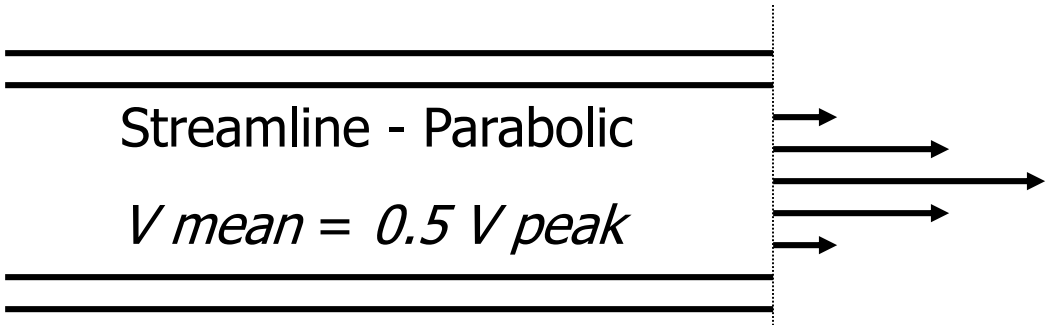


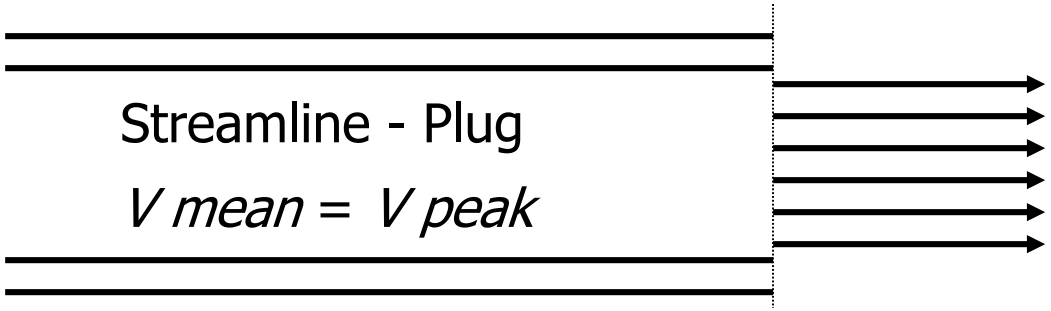
Simplified Bernoulli equation

Used to estimate pressure drop across stenosed valve

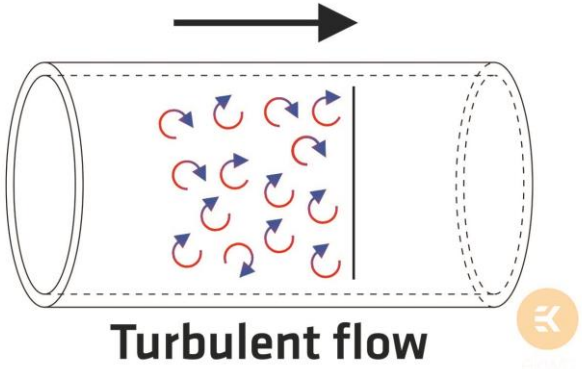
ΔP (peak pressure gradient in **mmHg**)= V^2 (V measured in **$m.s^{-1}$**)

Velocity Profiles





V mean = ???



Turbulence and Reynold's Number

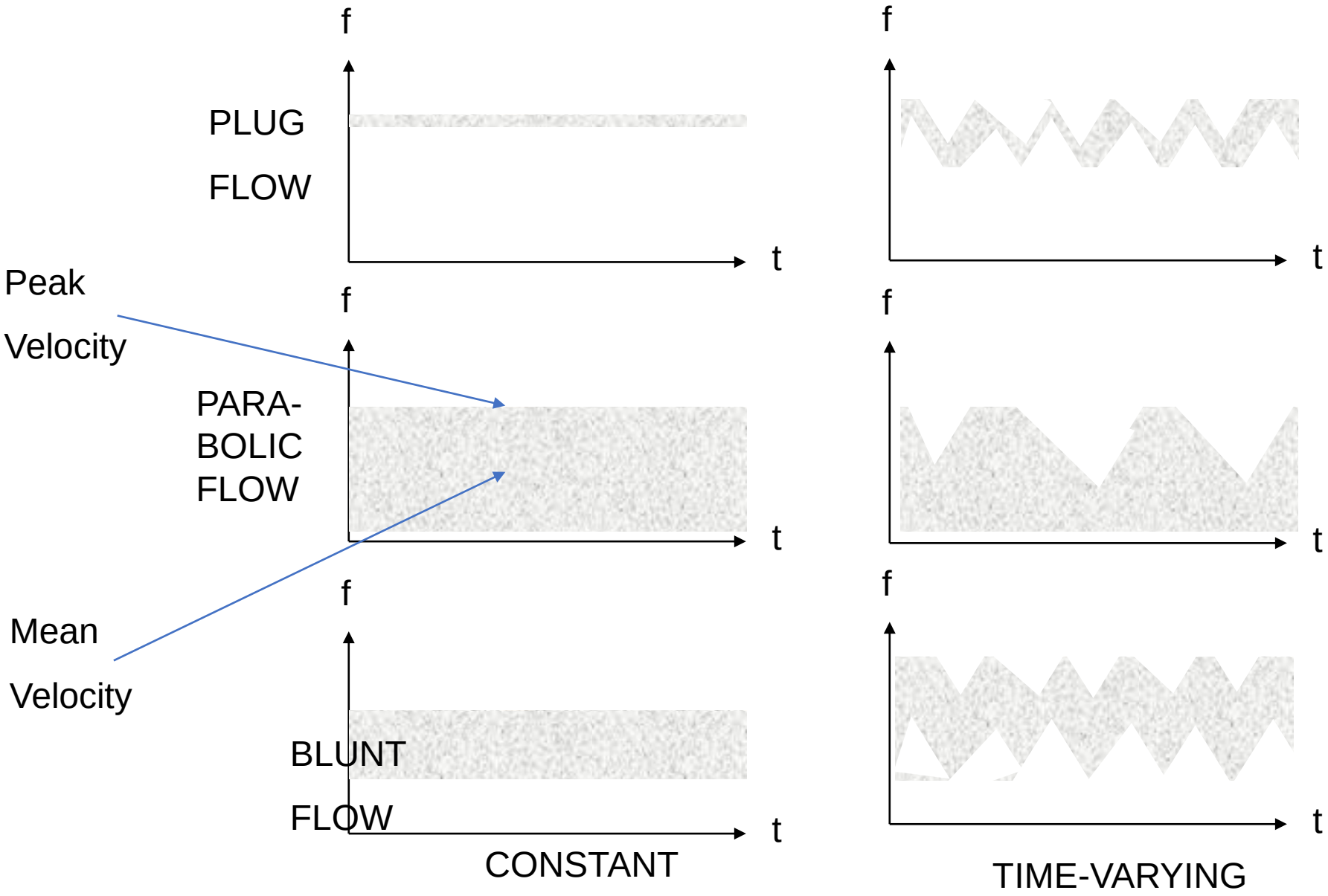
Whether or not flow is streamline (laminar) or turbulent depends on factors such as the dimensions of the flow containing vessel, the viscosity and density of the fluid, and its velocity.

The Reynolds number is a dimensionless quantity that expresses the likelihood of turbulence.

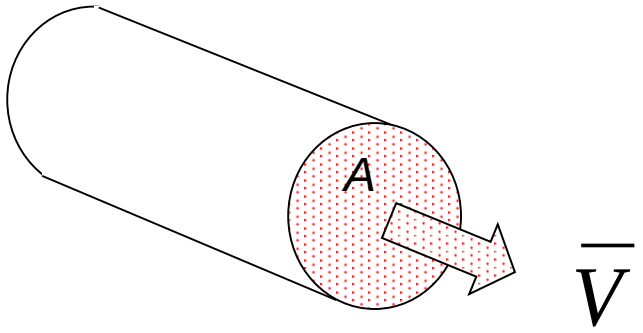
For flow in a pipe of diameter *D*, experimental observations show that laminar flow occurs when $Re_D < 2300$ and turbulent flow occurs when $Re_D > 2900$.

Turbulence can occur as fluid emerges from a jet into a wider vessel.

Notional spectrograms for various flow types



Mean flow velocity



$$volume\ flow = \overline{V}\,A$$

Colour flow Doppler

Effectively a multi-sampled PW from multiple sites (100-400) superimposed on a 2D image→ low FR!!!

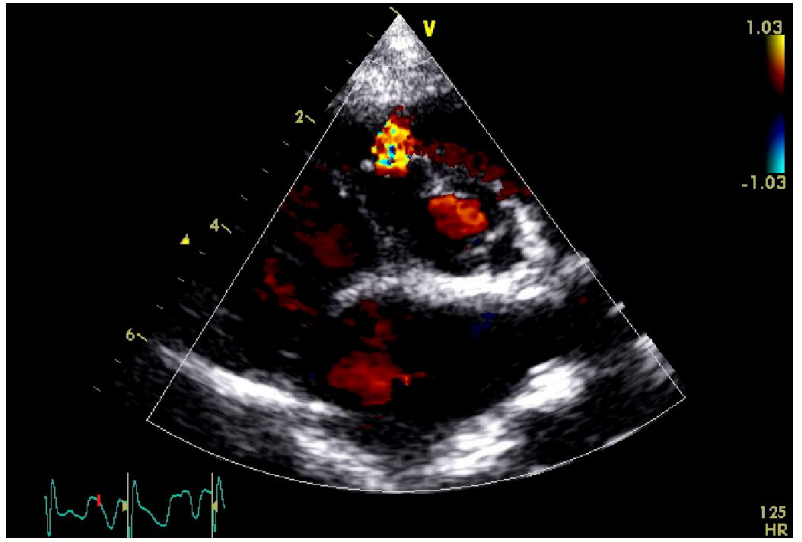
Each area sampled minimum of 3 times to calculate a Doppler frequency shift and estimate mean velocity.

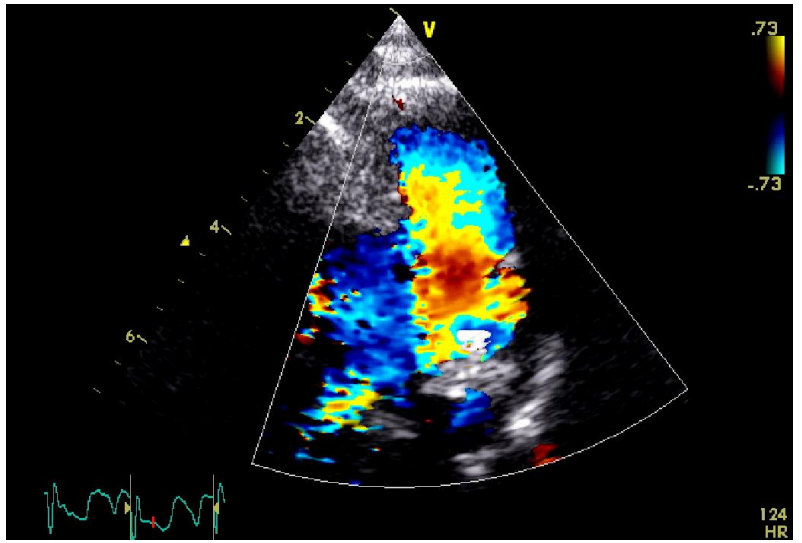
<u>Frame rate determined by:</u>

- **Sector size** ↓width/depth↑FR

- **Packet size:** The packet size is the number of pulses transmitted per line. ↓packet size ↑FR

Same limitations as PW Doppler (i.e. Nyquist limit), however as it is detecting mean velocity the Nyquist limit is lower → aliases earlier



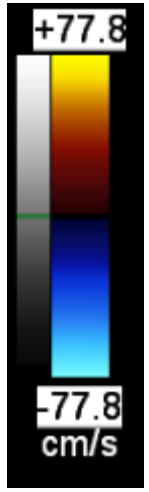


Colour Doppler

- Superimpose colour-coded velocity information on to conventional grey-scale image
- Uses same transducer for imaging and Doppler shift information
- Produces fast mean frequency estimate in a group of pixels; can also give an index of turbulence
- For each image pulse, require 3-15 pulses to obtain Doppler information

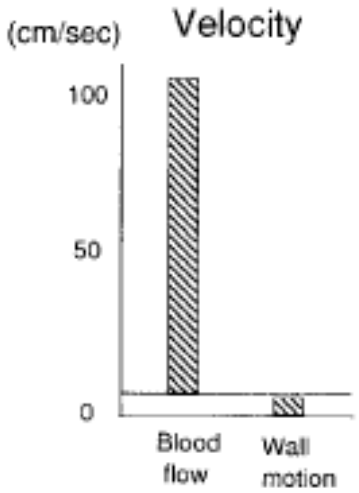
Colour and TDI

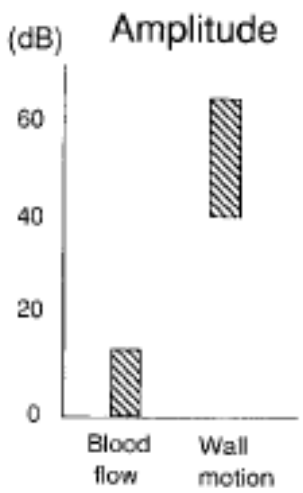
Filters are used to discriminate between myocardium and tissue in colour imaging:

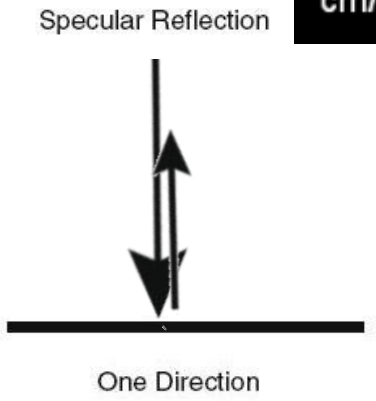


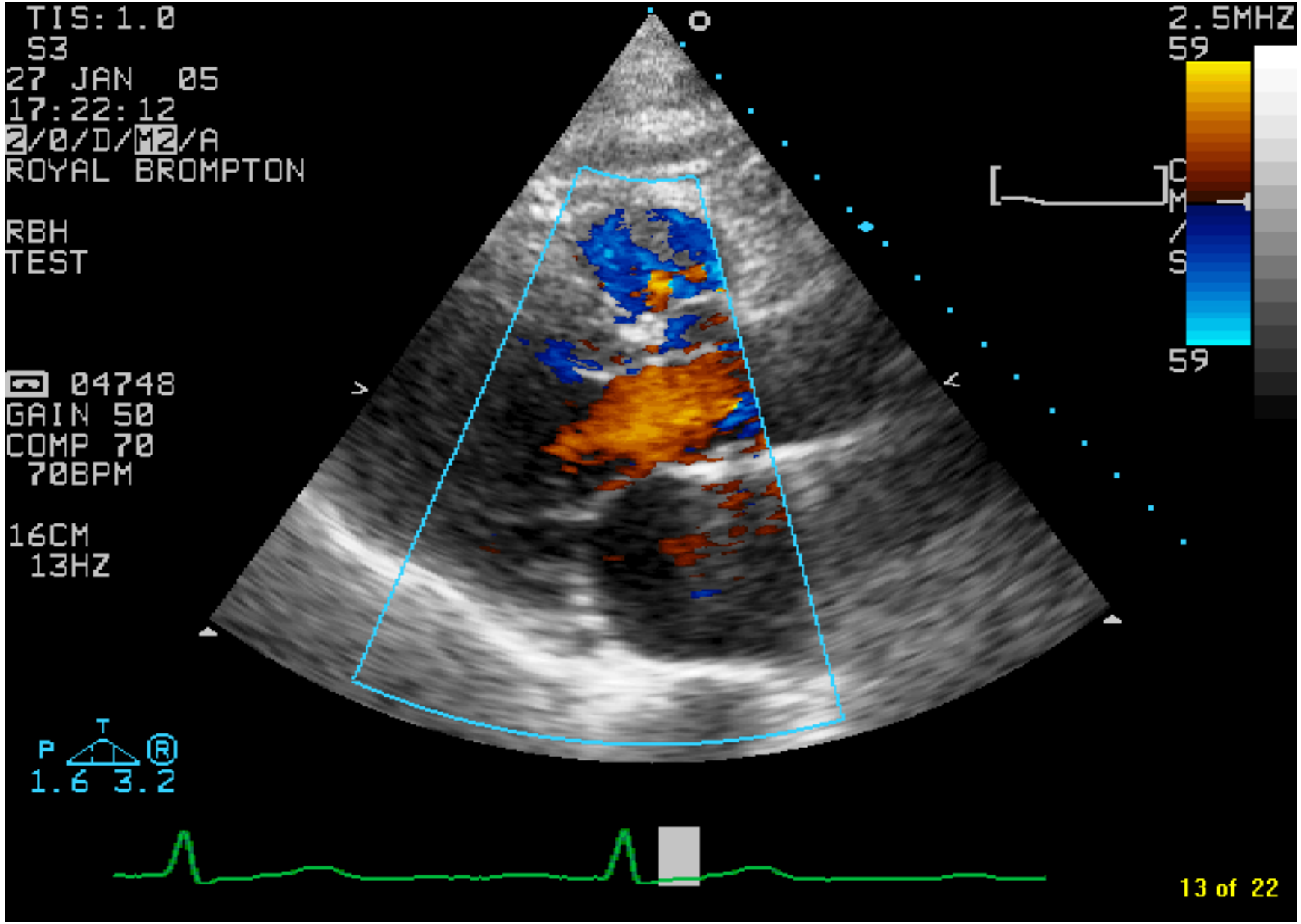
Blood is a low amplitude scatterer (recall Rayleigh scattering) with relatively quick velocities.

Myocardium is a high amplitude spectral reflector with relatively slow velocities.

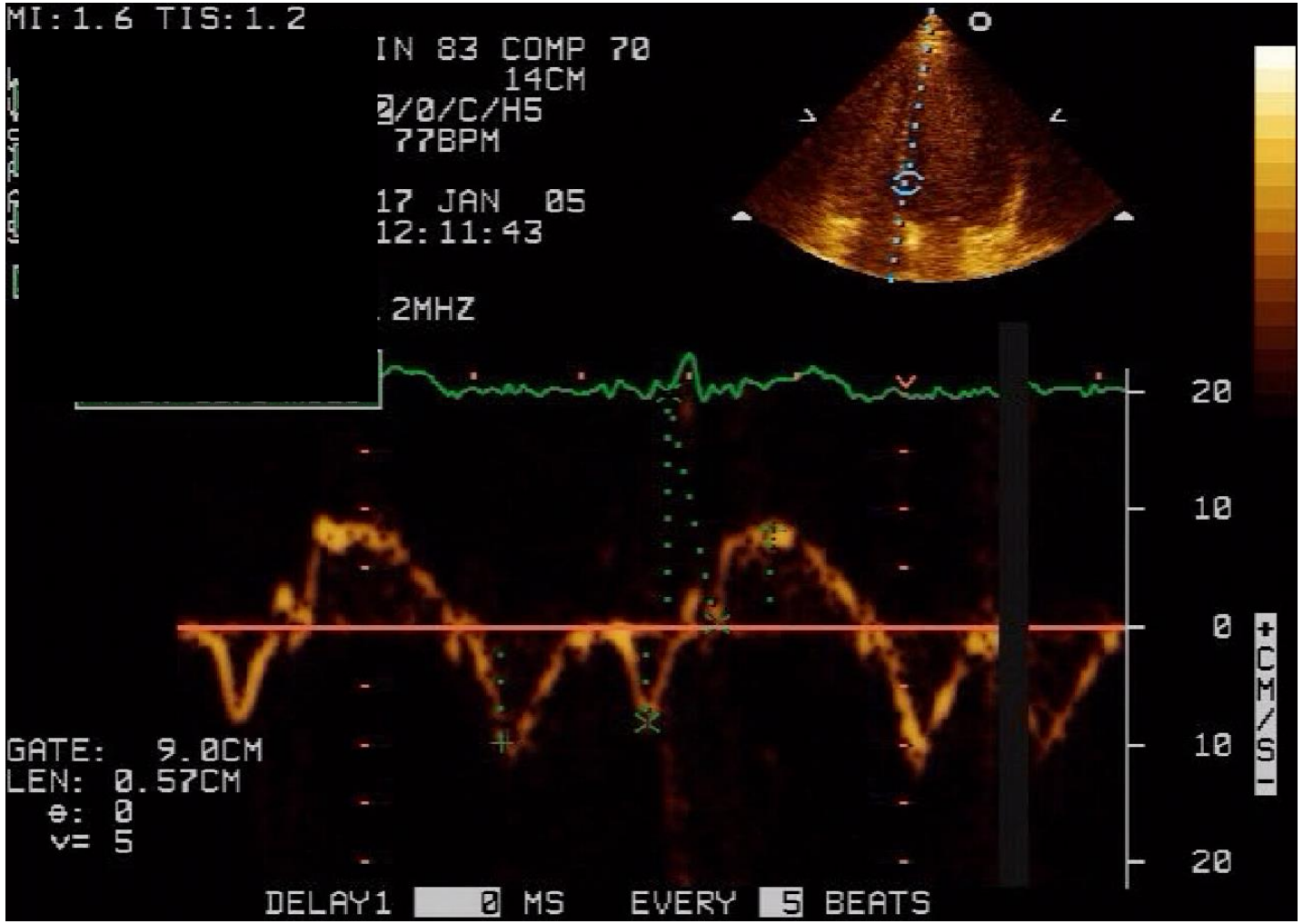




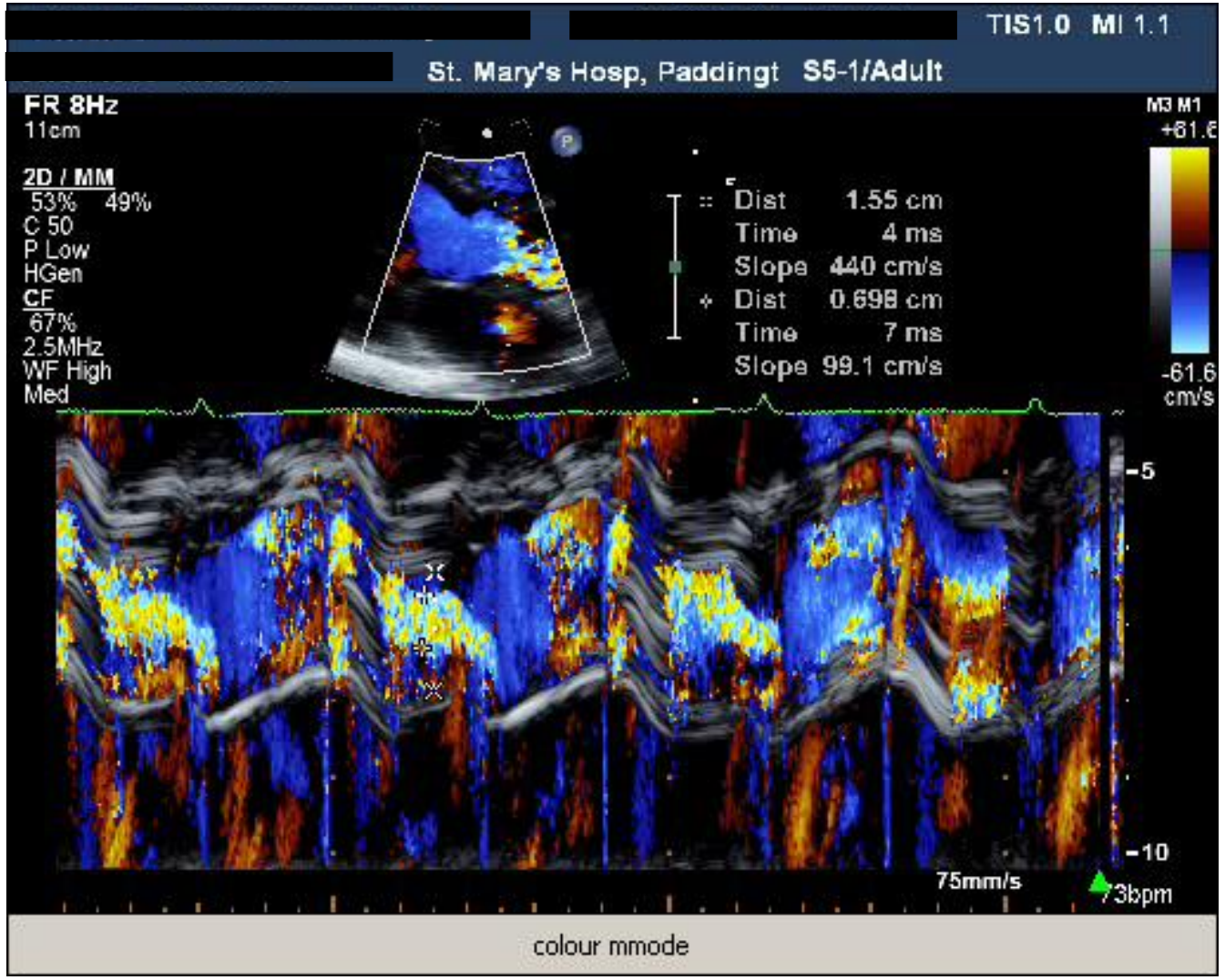


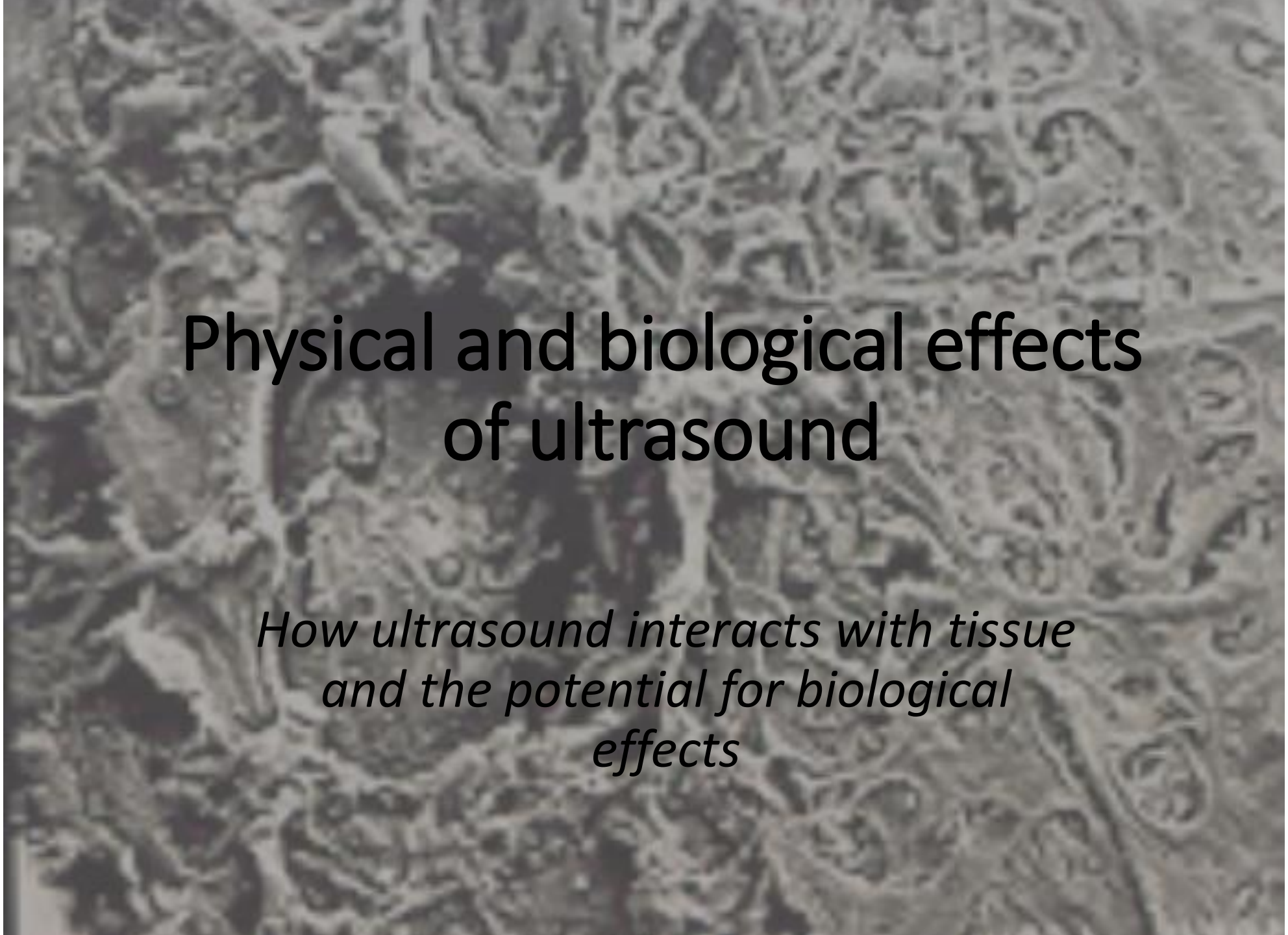

TIS:1.0
S3
27 JAN 05
17:22:12
2/0/D/M2/A
ROYAL BROMPTON
RBH
TEST
04748
GAIN 50
COMP 70
70BPM
16CM
13HZ
P T R
1.6 3.2
2.5MHZ
59
CM/S
59
13 of 22

Tissue Doppler (TDI)



Colour M mode





Physical and biological effects of ultrasound

How ultrasound interacts with tissue and the potential for biological effects

Safety

- Potential hazardous biological effects pul
  - Heating and cavitation effects
- Measurement of beam intensity
  - (Spatial Peak Temporal Average, [SPTA])
- Practical precautions:
  - power levels,
  - Peak intensities in different modes
  - On-screen indices.

Effects in tissue

Important to distinguish between

- physical effects
- biological effects
- clinical risk
- clinical outcomes
- relative importance of potential hazards in *diagnosis* and *screening*

Power and intensity

- Total power emitted by TDCR may be relatively low i.e. mW
- Local intensities, however may be relatively high i.e hundreds of mW

Tissue heating (unperfused tissues in vitro)

Tissue	Intensity mW cm-2	Time for 2 deg C temp.rise
Liver	100	3.7 mins
	1000	22 secs
Bone	100	2.9 secs
	1000	0.3 secs
Eye lens	100	33 secs
	1000	3.3 secs

Cooling

By conduction into surrounding tissue

By local blood perfusion

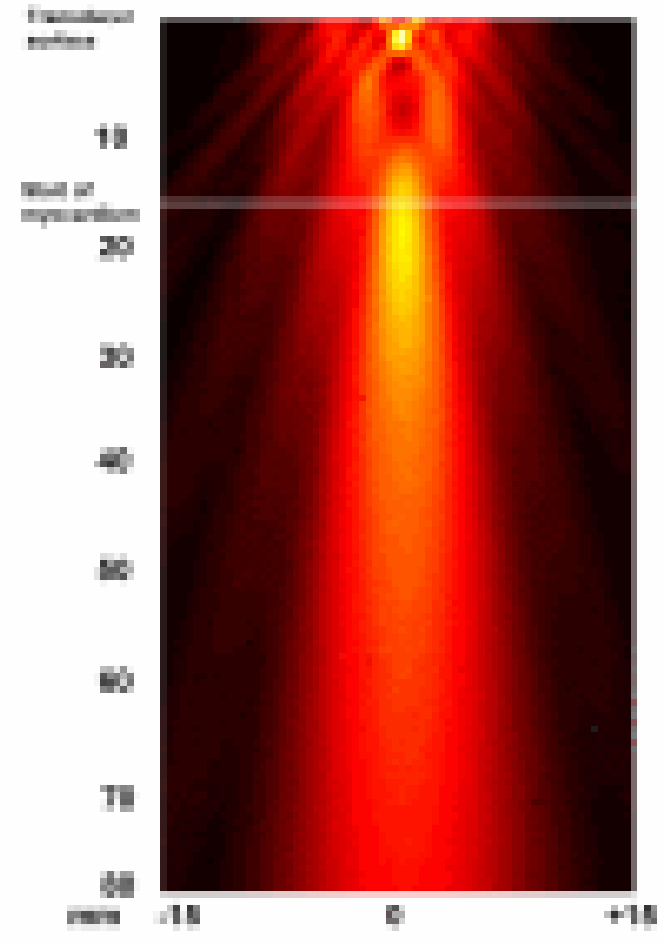
By thermoregulation

In-vivo

Risk in thermally unregulated tissues (e.g. foetus)

Risk in unperfused tissues (e.g. lens of eye)

Intensity in a beam



I_{SPTA} **Spatial peak, temporal average**

- Lowest for linear and convex array B-mode images.
- Higher in scan modes where beam is stationary
- Pulsed Doppler uses longer pulses and stationary beam, can produce highest values
- Continuous wave and colour Doppler also higher than B mode
- High at foci
- In sector scans can be high close to the transducer, where all beams in the scan pass through during image formation.

Pulse pressure

- Pulse pressure – peak rarefaction pressure (P-)

- Cavitation – the catastrophic collapse of oscillating bubbles, or tissue voids
- Cavitation can be destructive: ships' propellers, aircraft fuel lines
- Causes DNA damage
- Possible effects at air interfaces in e.g. lung – impedance mismatch causes large negative pressures

On screen indicators

- Thermal index (TI)– proportion of possible 1 deg C temperature rise under a range of scanning conditions. There are different TIs for different tissues
- Mechanical index (MI) – frequency corrected proportion of theoretical threshold of cavitation risk

Output Power

- Keep output power as low as practicable – use increased gain instead wherever possible

Goodbye!