EL5823/BE6203 --- Medical Imaging - I

Ultrasound Imaging

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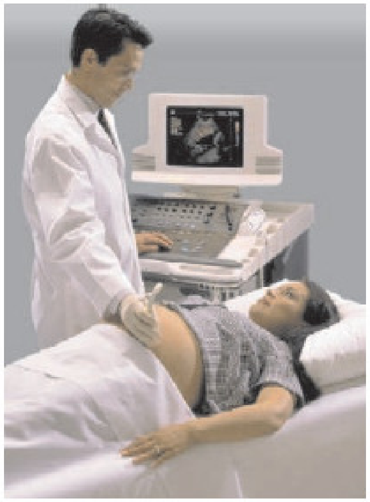
Based on J. L. Prince and J. M. Links, Medical Imaging Signals and Systems, and lecture notes by Prince. Figures are from the textbook except otherwise noted.

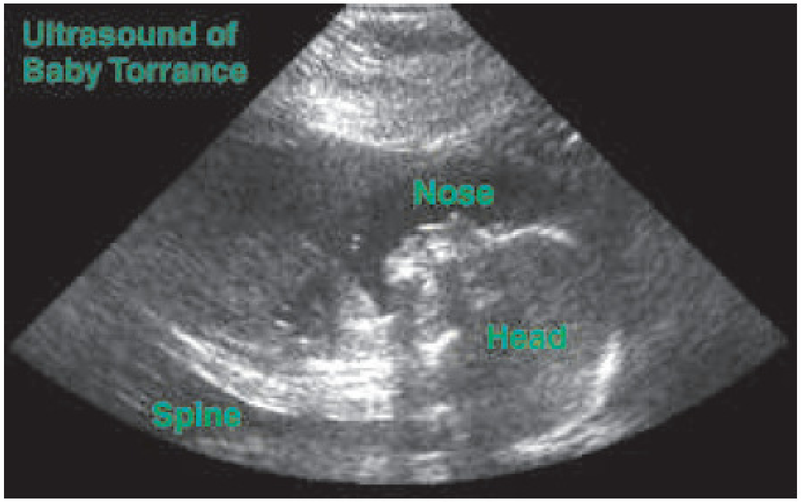
Lecture Outline

- Ultrasound imaging overview
- Ultrasound imaging system schematic
- Derivation of the pulse-echo equation
- Different ultrasound imaging modes
- Steering and focusing of phased arrays
- Doppler Imaging
- Clinical applications

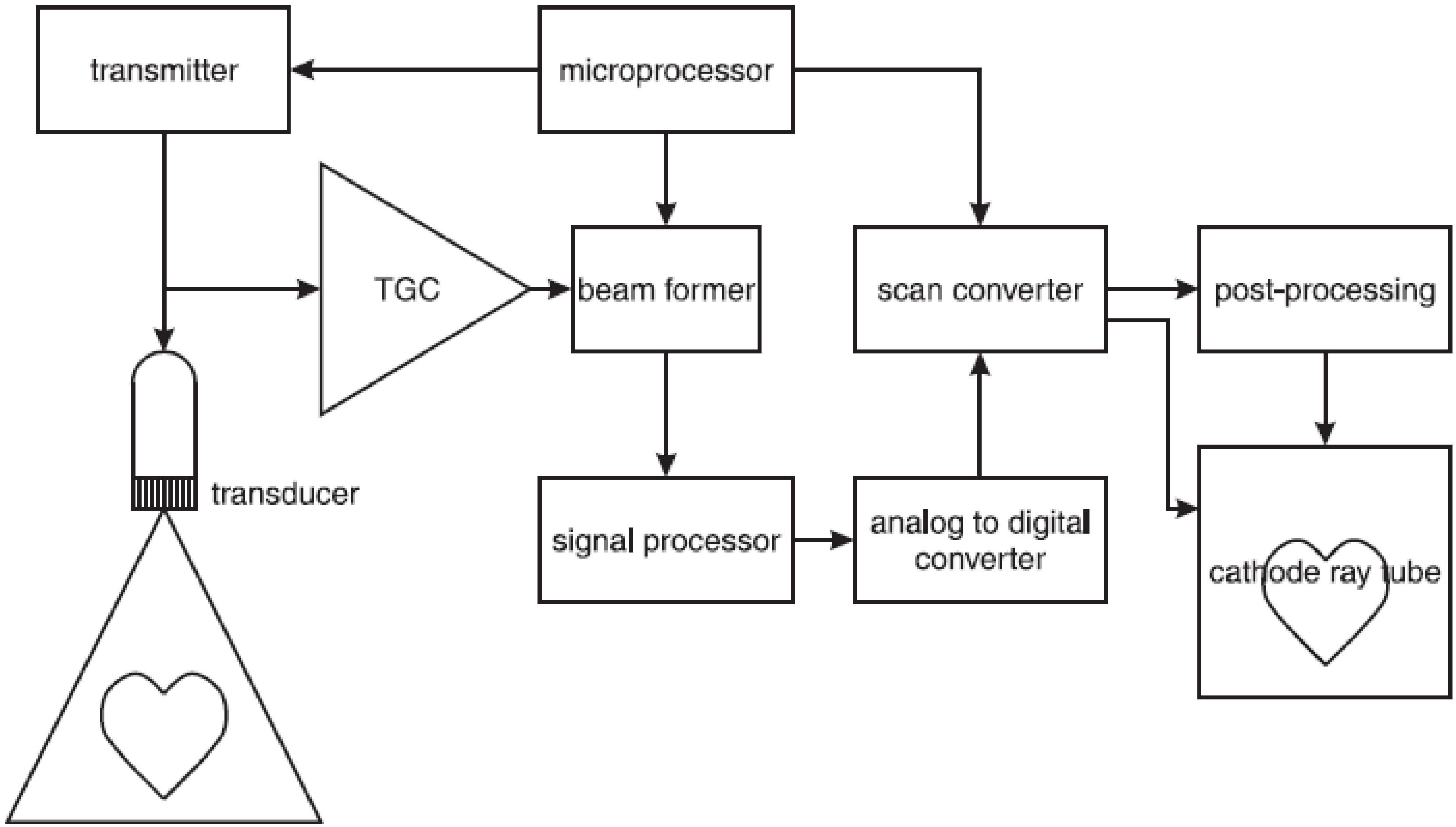
Ultrasound Imaging

- Measure the reflectivity of tissue to sound waves
- Can also measure velocity of moving objects, e.g. blood flow (Doppler imaging)
- No radiation exposure, completely non-invasive and safe
- Fast
- Inexpensive
- Low resolution
- Medical applications: imaging fetus, heart, and many others





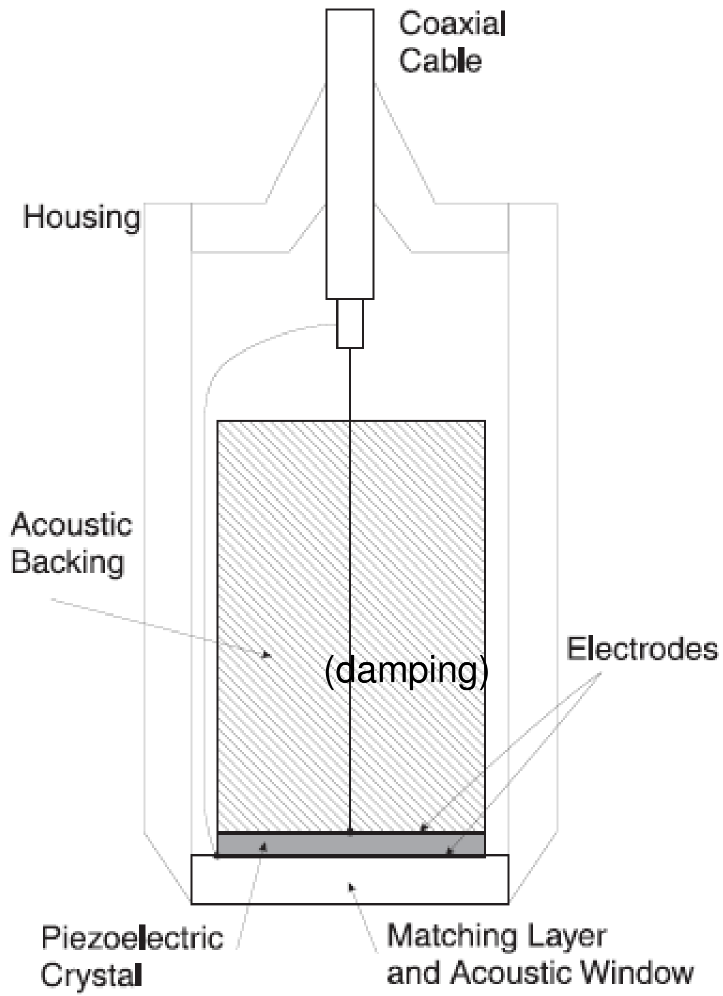
Schematic of an Ultrasound Imaging System

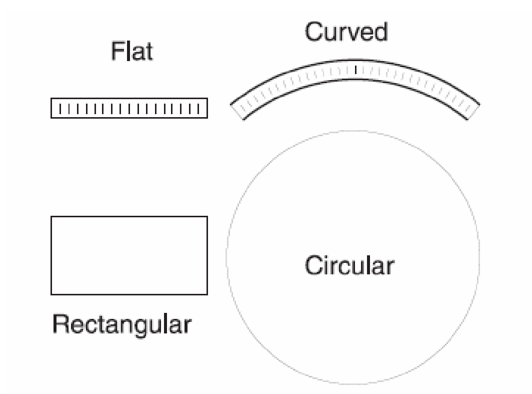


Functions of transducer

- Used both as Transmitter And Receiver
- Transmission mode: converts an oscillating voltage into mechanical vibrations, which causes a series of pressure waves into the body
- Receiving mode: converts backscattered pressure waves into electrical signals

Single Crystal Transducer (Probe)

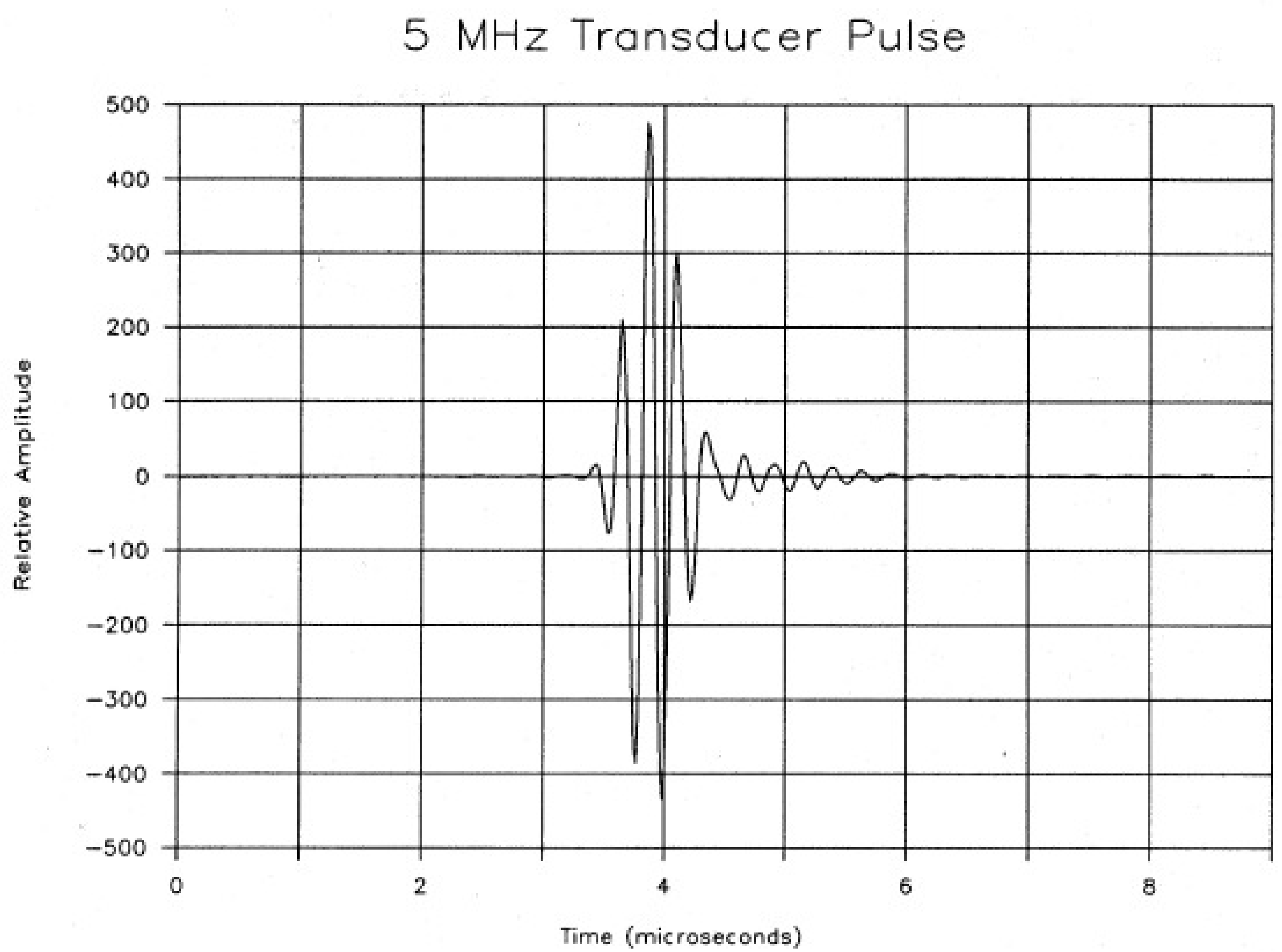




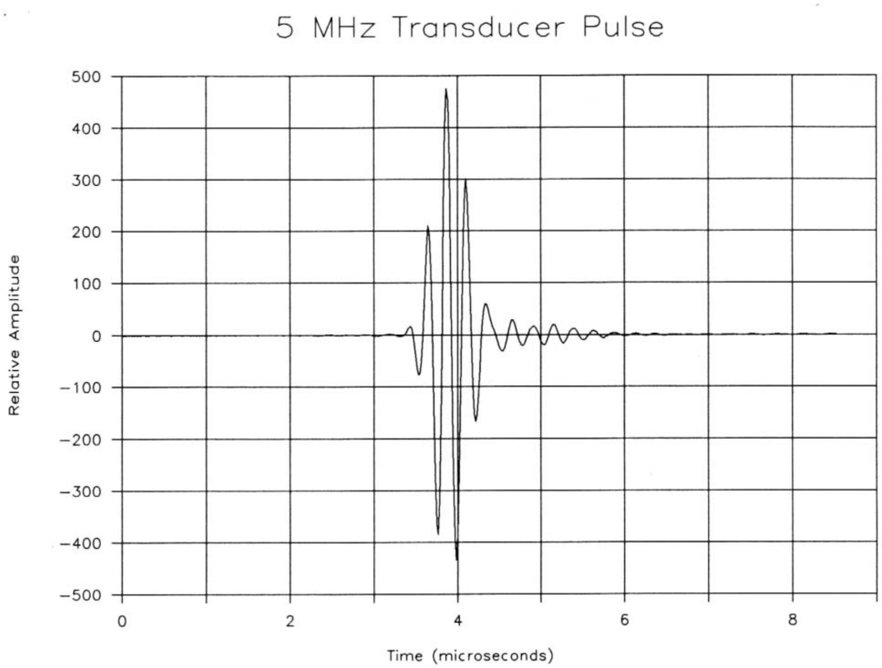
Pulse Echo Imaging

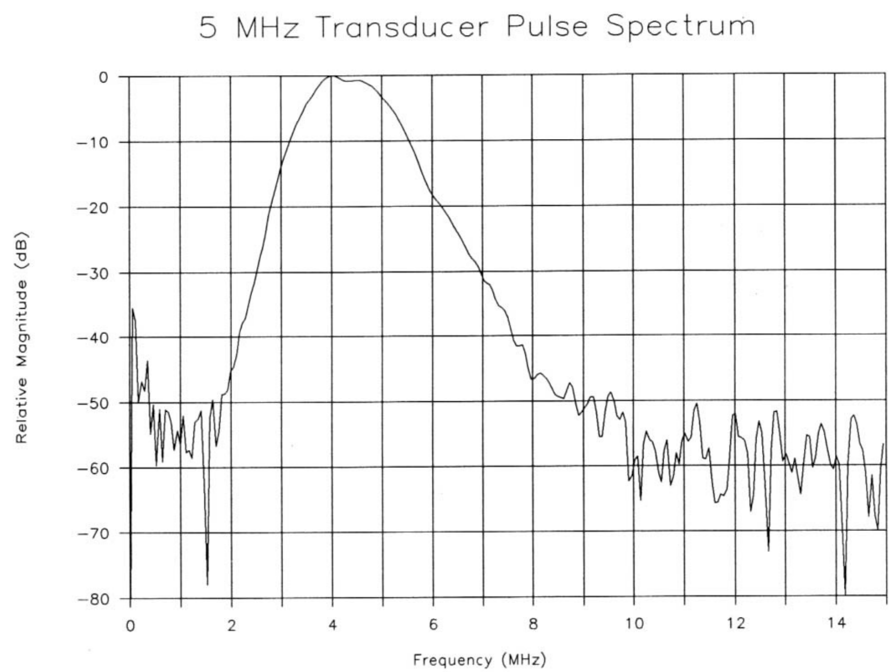
- Transducer is excited for a short period, generating a narrowband short pulse
- Detects backscattered wave (echo) generated by objects
- Repeat the above process, with the interval between two input pulses greater than the time for the receiver to receive the echo from the deepest object (2 d_max/c)

Typical Transmit Pulse



Spectrum of the Transmit Pulse





Product of a decaying envelop and a sinusoidal function

$$n_e(t)\cos(2\pi f_0 t)$$

$$n_e(t)\cos 2\pi f_0 t = \frac{1}{2} n_e(t)\left(e^{j2\pi f_0 t} + e^{-j2\pi f_0 t}\right)$$

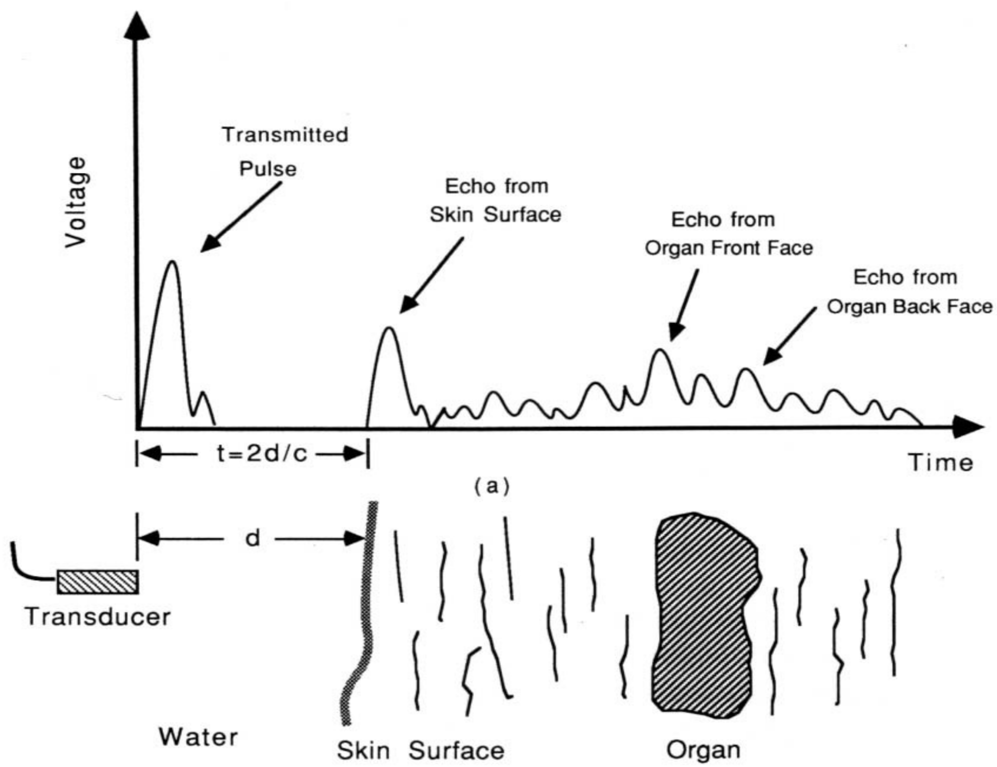
$$\Leftrightarrow \frac{1}{2} N_e(f) * [\delta(f - f_0) + \delta(f + f_0)]$$

$$= \frac{1}{2}[N_e(f - f_0) + N_e(f + f_0)]$$

Narrow band (around f0) pulse

What is the Received Signal?

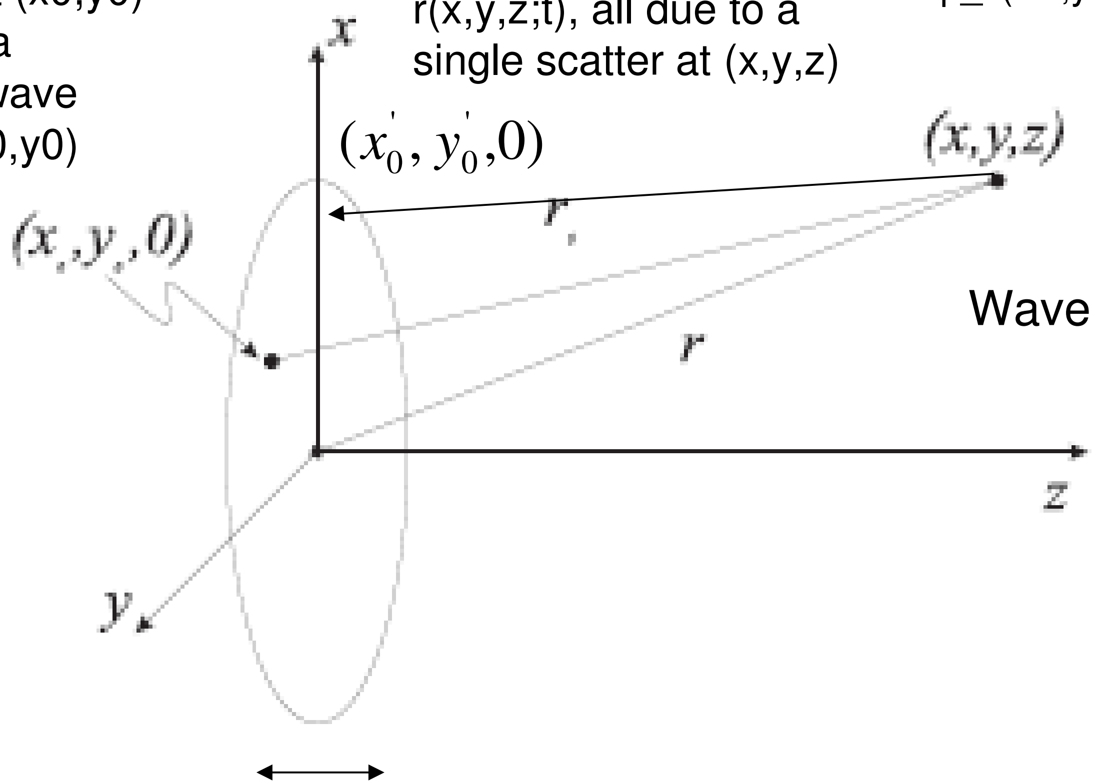
- What is the received signal (backscattered signal) at the transducer ?
- How is it related to the reflectivity in the probed medium?



The scattered signals over all (x0’,y0’) leads to a voltage signal r(x,y,z;t), all due to a single scatter at (x,y,z)

A scatter at (x,y,z) reflects p(x,y,z;t); generating p_s(x0’,y0’;t;x,y,z)

Each point (x0,y0) produces a pressure wave p(x,y,z,t;x0,y0)



Wave

p(x,y,z;t) is superposition of above waves over all points on the transducer face

Vibrating in z direction

Complex Signal Representation

- We will represent the input signal as the Real part of a complex signal to simplify derivation

- Complex signal:

$$\mathtt{n}(t) = n_e(t)e^{j\phi}e^{-j2\pi f_0 t}$$

- Complex envelope is $\tilde{n}(t) = n_e(t)e^{j\phi}$
- The pulse is

$$n(t) = \mathrm{Re}\{\mathtt{n}(t)\}$$

- The envelope is

$$n_e(t) = |\mathtt{n}(t)|$$

Derivation of the Pulse Echo Equation

- Pressure wave produced by point (x0,y0)

(each point acts as a dipole rather than a monopole, Signal is strongest in the direction orthogonal to the dipole)

$$p(x,y,z,t;x_0,y_0)=\frac{z}{r_0^2}n(t-c^{-1}r_0)$$

$$r_0^2=(x-x_0)^2+(y-y_0)^2+z^2$$

- Total pressure at (x,y,z) is superposition of above due to all x0,y0 in the transducer face

$$p(x,y,z,t)=\iint s(x_0,y_0)\frac{z}{r_0^2}n(t-c^{-1}r_0)dx_0dy_0$$

$$s(x,y)=1,\ \text{if } (x,y) \text{ in face; } =0, \text{otherwise}$$

- Reflected signal due to scatterer at (x,y,z) with reflectiveity R(x,y,z) is a spherical wave, with signal at transducer position x0',y0'

$$p_s(x_0^{'},y_0^{'},t)=R(x,y,z)\frac{1}{r_0^{'}}p(x,y,z;t-c^{-1}r_0^{'})$$

- The generated electrical signal (voltage) depends on reflected signal at all points in the transducer

$$r(x,y,z,t)=K\iint s(x_0',y_0')\frac{z}{r_0^{'}}p_s(x_0^{'},y_0^{'};t)dx_0^{'}dy_0^{'}$$

z/r0' due to dipole pattern

$$=KR(x,y,z)\iint s(x_0',y_0')\frac{z}{r_0'^2}\left\{\iint s(x_0,y_0)\frac{z}{r_0^2}n(t-c^{-1}r_0-c^{-1}r_0^{'})dx_0dy_0\right\}dx_0^{'}dy_0^{'}$$

- The total response for scatters at all possible (x,y,z) $r(t)=\iint r(x,y,z,y)dxdydz$

Plane Wave Approximation

- Excitation pulse envelope arrives at all points at a given range simultaneously.
- Mathematically,

$$\mathrm{n}(t - c^{-1}r_0 - c^{-1}r_0') \approx \mathrm{n}(t - 2c^{-1}z)e^{jk(r_0 - z)}e^{jk(r_0' - z)}$$

where wavenumber is

$$k = 2\pi f_0 c^{-1}$$

and range equation gives

$$ct = 2z$$

$$n(t) = n_e(t)e^{j\phi}e^{-j2\pi f_0 t}$$

$$n(t - c^{-1}r_0 - c^{-1}r'_0) = n_e(t - c^{-1}r_0 - c^{-1}r'_0)e^{j\phi}e^{-j2\pi f_0(t - c^{-1}r_0 - c^{-1}r'_0)}$$

Approximation : $r_0 \approx r'_0 \approx z$

$$n_e(t - c^{-1}r_0 - c^{-1}r'_0) = n_e(t - 2c^{-1}z)$$

Also using $t = 2z/c$ in the exponent

$$e^{-j2\pi f_0(t - c^{-1}r_0 - c^{-1}r'_0)} = e^{-j2\pi f_0 c^{-1}(2z - r_0 - r'_0)} = e^{-jk(z - r_0)}e^{-jk(z - r'_0)}$$

This approximation enables us to separate integration over x0,y0 and that over x0’,y0’

Field Pattern and Pulse-Echo Equation

- Define field pattern as (depends only on transducer face, not pulse)

$$q(x, y, z) = \iint s(x_0, y_0) \frac{z}{r_0^2} e^{jk(r_0 - z)} dx_0 dy_0$$

- Then received signal (from single scatterer) is

$$\mathrm{r}(x, y, z; t) = KR(x, y, z)\mathrm{n}(t - 2c^{-1}z)[q(x, y, z)]^2$$

Basic pulse-echo signal equation

- From all scatters, and considering attenuation in material with μ_a:

$$\mathrm{r}(t) = K \iiint R(x, y, z) \mathrm{n}(t - 2c^{-1}z) e^{-2\mu_a z} [q(x, y, z)]^2 dx dy dz$$

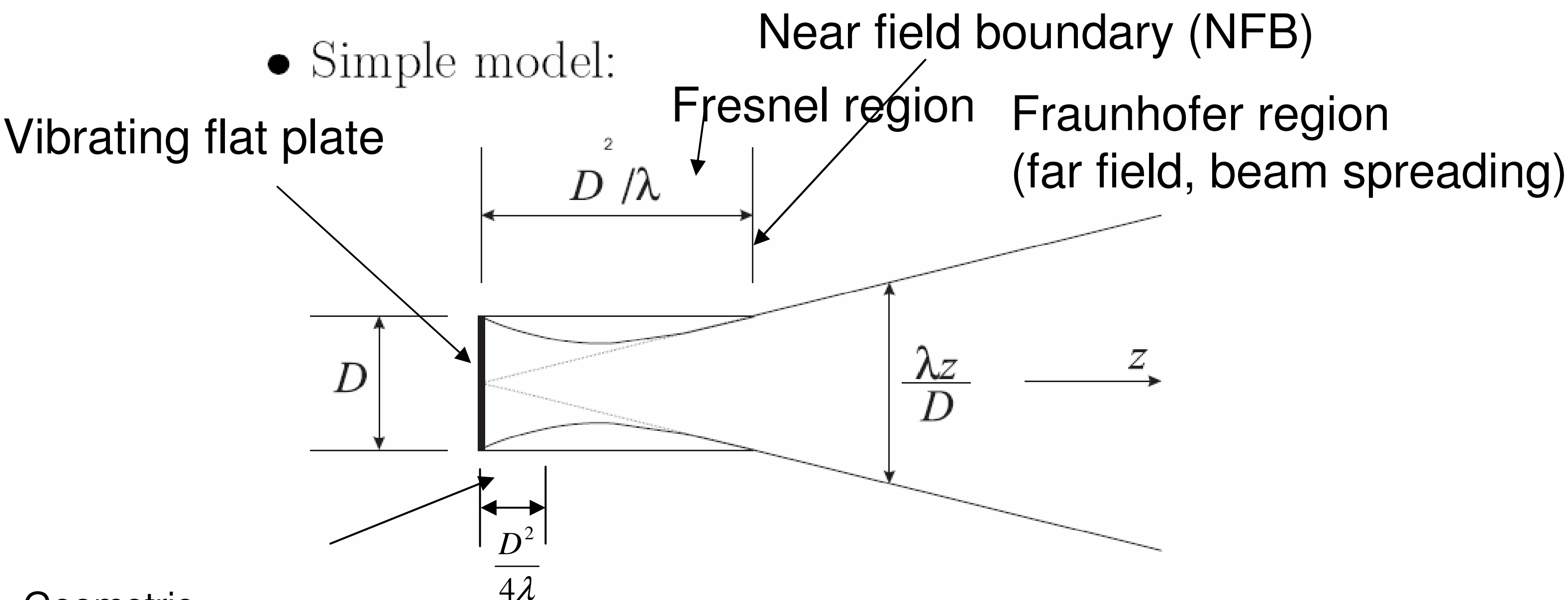
Paraxial Approximation

- Pattern is large near the transducer axis
- Then $r_0 \approx z$
- Field pattern becomes

$$q(x, y, z) \approx \frac{1}{z} \iint s(x_0, y_0) e^{jk(r_0 - z)} dx_0 dy_0$$

This is the same result that we would have got had we assumed that all points on the transducer act as spherical wave generators and receivers

Fresnel & Fraunhofer Approximation



Geometric approximation: Wave is confined in a cylinder

Fresnel and Fraunhofer Approximations

- Both involve phase approximations
- Fresnel field pattern (valid in Fresnel region)

$$q(x,y,z) \approx \frac{1}{z} s(x,y) ** e^{jk(x^2+y^2)/2z}$$

- Fraunhofer field pattern (further approximation, in far field)

$$q(x,y,z) \approx \frac{1}{z} e^{jk(x^2+y^2)/2z} S\left(\frac{x}{\lambda z}, \frac{y}{\lambda z}\right)$$

for $z \geq D^2/\lambda$.

S(u,v): FT of s(x,y)

General Pulse-Echo Equation

- Define

$$\tilde{q}(x, y, z) = z q(x, y, z)$$

- Fresnel or Fraunhofer satisfies

$$\mathtt{r}(t) = K \frac{e^{-\mu_a c t}}{(ct)^2} \iiint R(x, y, z) \mathtt{n}(t - 2c^{-1} z) \tilde{q}^2(x, y, z) dx dy dz$$

Time Gain Compensation

- Amplitude of $\mathtt{r}$ decays predictably
- Compensate with time-varying gain

$$\mathtt{r}_c(t) = g(t)\mathtt{r}(t)$$

$$\iiint R(x,y,z)\mathtt{n}(t - 2c^{-1}z)\tilde{q}^2(x,y,z)\,dx\,dy$$

$$g(t) = {(ct)^2}\Big/{e^{-\mu_a ct}}$$

Envelope Detection

- In A-mode of ultrasound imaging, the envelope of r_c(t) is detected and used as the output signal
- e_c(t)=envelope of r_c(t)

$$
\begin{aligned}
e_c(t) &= \Big| \iiint R(x,y,z) \\
&\quad \mathrm{n}(t - 2c^{-1}z)\tilde{q}^2(x,y,z)\,dx\,dy\,dz \Big| \\
&= \Big| \iiint R(x,y,z) \\
&\quad n_e(t - 2c^{-1}z)e^{j2kz}\tilde{q}^2(x,y,z)\,dx\,dy\,dz \Big|
\end{aligned}
$$

Since n(t) and q(x,y,z) depend only on the source, e_c(t) is affected by the reflectivity R(x,y,z) in the imaged body, or e_c(t) ~ R(0,0,z=ct/2)

Transducer Motion and Range Equation

Previous result assumes the transducer is located at (0,0,0). When the transducer is at arbitrary (x0,y0), q(x,y,z) is changed to q(x-x0,y-y0,z)

- Move transducer to (x_0, y_0); yields $e_c(t; x_0, y_0)$.
- Use range equation as $z_0 = ct/2$.
- Then $e_c(\cdot)$ estimates <u>reflectivity</u> at (x0,y0,ct/2)

$$
\begin{aligned}
\hat{R}(x_0, y_0, z_0) &= e_c(2z_0/c; x_0, y_0) \\
&= \left| \iiint R(x, y, z) e^{j2kz} n_e(2(z_0 - z)/c) \right. \\
&\quad \left. \cdot \tilde{q}^2(x - x_0, y - y_0, z) dx dy dz \right|
\end{aligned}
$$

Geometric Approximation (Fresnel Region)

With geometric approximation, we assume

$$\tilde{q}(x, y, z) = zq(x, y, z) \approx s(x, y)$$

Then

$$\hat{R}(x, y, z) = K\left|R(x, y, z)e^{j2kz} *** h(x, y, z)\right|$$

$$h(x, y, z) = \tilde{s}^2(x, y)n_e\left(\frac{z}{c/2}\right)$$

$$\tilde{s}(x, y) = s(-x, -y)$$

- If we ignore the e^{jkz} term, Received signal (envelope) at (x,y,t) is R(x,y,z) (z=tc/2) convolved with h
- h(x,y,z) can be thought of as a blurring function. Its support region defines the resolution cell of the imaging system
- With geometric approximation, the resolution cell has the same dimension as the transducer face in (x,y) plane, and the extend in z-direction = cT/2, if T is the length of the transmit pulse.

Fraunhofer Approximation

With Fraunhofer approximation and assuming $(x^2 + y^2)/2z \approx 0$ (correct when z >>x, y, i.e., near axis)

$$\tilde{q}(x, y, z) = zq(x, y, z) = S(\frac{x}{\lambda z}, \frac{y}{\lambda z})e^{jk(x^2+y^2)/2z} \approx S(\frac{x}{\lambda z}, \frac{y}{\lambda z})$$

for $z \geq D^2 / \lambda$

$$\hat{R}(x, y, z) = K\left|R(x, y, z)e^{j2kz} *** h(x, y, z)\right|$$

$$h(x, y, z) = \left[S\left(\frac{x}{\lambda z}, \frac{y}{\lambda z}\right)\right]^2 n_e\left(\frac{z}{c/2}\right)$$

$S(u,v)$: Fourier transform of $s(x, y)$

= sinc function for square plate

Note that the spatial extend of the resolution cell now increases with Z. That means the lateral resolution decreases with Z. The lateral extension is also not finite (sinc function)

Speckle Noise

- Recall with either geometric or Fraunhofer approximation

$$\hat{R}(x, y, z) = K\left|R(x, y, z)e^{j2kz} *** h(x, y, z)\right|$$

- e^{j2kz}*R(x,y,z) modifies R(x,y,z)
- Can be thought of as modulating R(x,y,z) in z with frequency k/\pi
- e^{j2kz} has period = \pi/k = \lambda/2, so each sound period (\lamda) includes 2 cycles of the modulating wave. An ultra sound pulse typically includes several \lambda
- So there are multiple cycles in z-direction in the resolution cell

- Term e^{j2kz} is "fast-changing" sinusoid in resolution cell
- Gives rise to essentially "random" constructive and destructive interference

Example

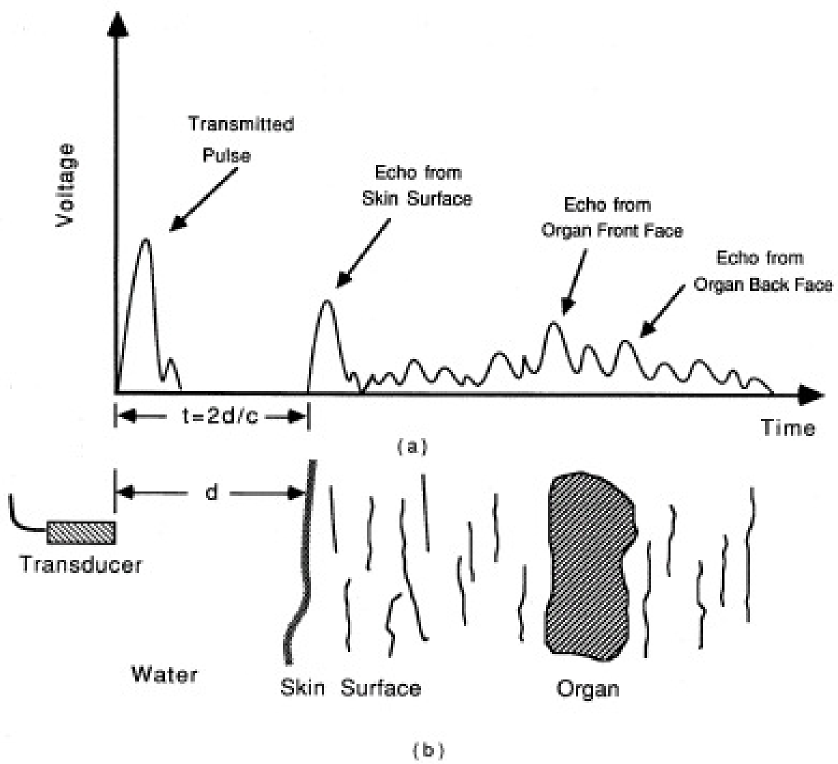
- Consider the case when a transducer is pointing down the z-axis, on which 2 scatters are located at z1 and z2, both in the Fraunhofer field. The envelope of the excitation pulse is rectangular with duration T. Sketch the envelope of the received signal.
- Go through in class
- Note: results depend on relative distances between z1 and z2

Ultrasound Imaging Modes

- A-mode
- M-mode
- B-mode

A-Mode Display

- Oldest, simplest type
- Display of the envelope of pulse-echoes vs. time, depth $d = ct/2$
  - Measure the reflectivity at different depth below the transducer position



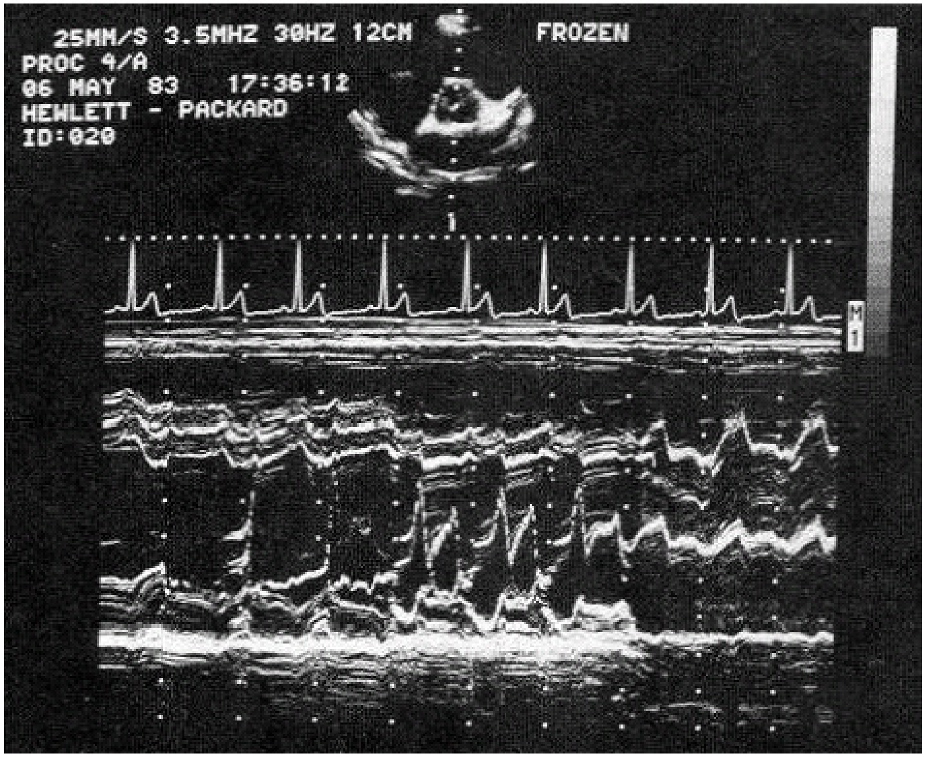
The horizontal axis can be interpreted as z, with $z=t\,c/2$

Application of A-Mode

- Applications: ophthalmology (eye length, tumors), localization of brain midline, liver cirrhosis, myocardium infarction
- Frequencies: 2-5 MHz for abdominal, cardiac, brain; 5-15 MHz for ophthalmology, pediatrics, peripheral blood vessels
- Used in ophthalmology to determine the relative distances between different regions of the eye and can be used to detect corneal detachment
  - High freq is used to produce very high axial resolution
  - Attenuation due to high freq is not a problem as the desired imaging depth is small

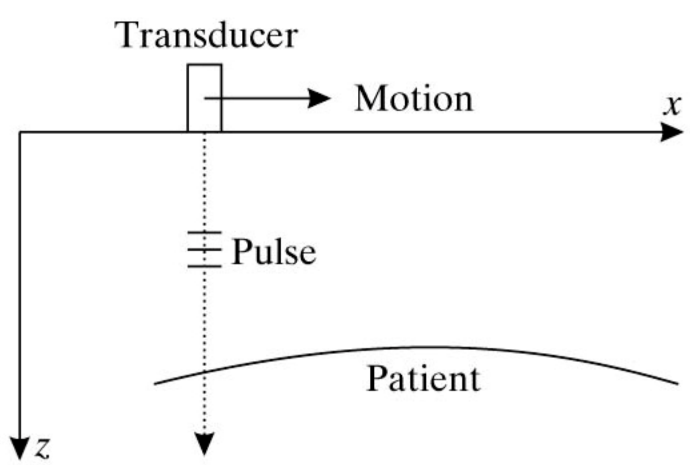
M-Mode Display

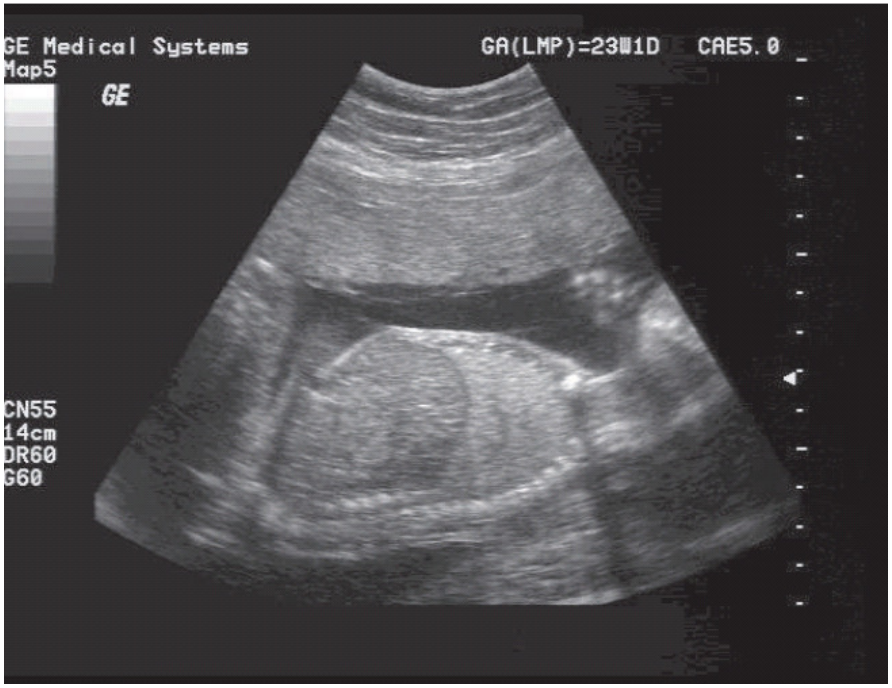
- Display the A-mode signal corresponding to repeated input pulses in separate column of a 2D image, for a fixed transducer position
  - Motion of an object point along the transducer axis (z) is revealed by a bright trace moving up and down across the image
  - Often used to image motion of the heart valves, in conjunction with the ECG



B-Mode Display

- Move the transducer in x-direction while its beam is aimed down the z-axis, firing a new pulse after each movement
- Received signal in each x is displayed in a column
- Unlike M-mode, different columns corresponding to different lateral position (x)
- Directly obtain reflectivity distribution of a slice! (blurred though!)





Application of B-Mode

- Can be used to study both stationary and moving structures
- High frame rate is needed to study motion (more later)
- Directly obtain reflectivity distribution of a slice! (blurred though!)
  - No tomographic measurement and reconstruction is necessary!

3-D Imaging

- By mechanically or manually scanning a phased array transducer in a direction perpendicular to the place of each B-mode scan
- Can also electronically steering the beams to image different slices

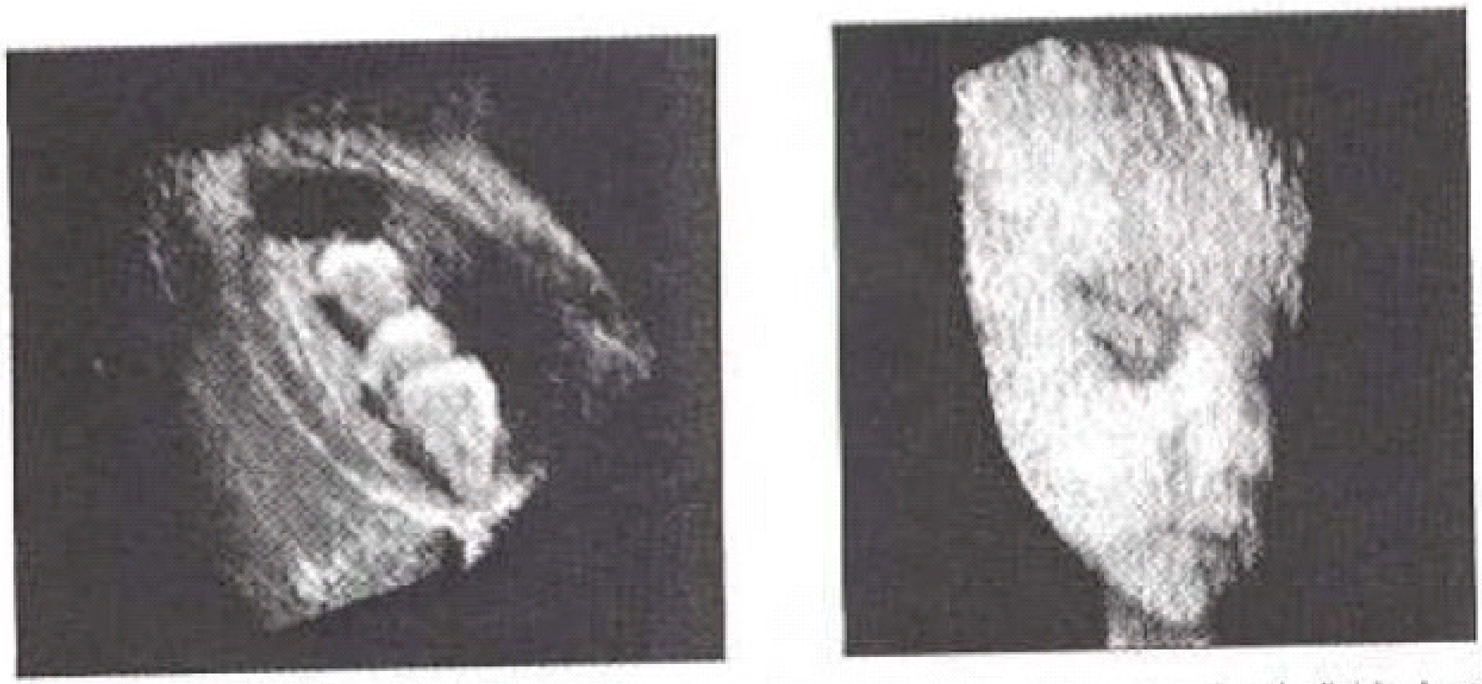


FIGURE 3.18. (Left) A three-dimensional abdominal ultrasonic scan showing individual gall stones. (Right) A three-dimensional image of a fetal head in utero (©2000 ATL Ultrasound).

From [Webb2003]

Depth of Penetration

- Recall that the amplitude of the wave attenuates in distance following

$$A(z) = A_0 e^{-\mu_a z} \qquad \mu_a = -\frac{1}{z}\ln\frac{A(z)}{A_0}$$

- Attenuation coefficient

$$\alpha = -20\log_{10} e \bullet \mu_a = -\frac{1}{z} 20\log_{10}\frac{A_z}{A_0}$$

  - \alpha is linearly increasing with f: alpha= a f

- System penetration is defined at the depth when 20 log 10 A0/Az=L (dB) (often 80dB)
- Combining above

$$af = \frac{1}{z} L \rightarrow z = \frac{L}{af} \qquad d_p = \frac{z}{2} = \frac{L}{2af}$$

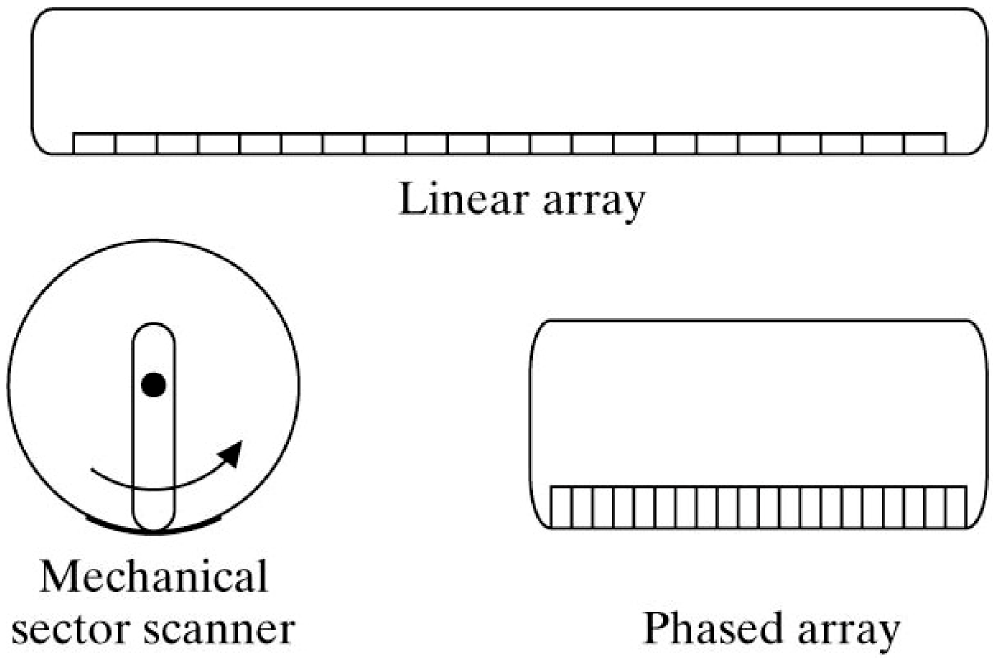
  - z is the round trip disiance, depth of penetration is half
  - Lower penetration at higher frequency
- Rule of thumb
  - L=80dB, a=1, d_p=40/f (Mhz) cm

Pulse Repetition Time

- The time needed to receive an echo at the maximum distance = $2d_p/c$
- Pulse repetition time $T_R >= 2d_p/c$
- Pulse repetition rate $f_R = 1/T_R$
- B-mode image frame rate
  - N pulses are required in each frame
  - Total time $= N T_R$
  - Frame rate $F = 1/NT_R = acf/NL$
- Typical numbers
  - 10-100 frames/sec
- Frame rate should be commensurate with object motion
- For faster moving objects,
  - reducing field of view and consequently N
  - Increasing frequency (at the cost of reduced depth of penetration)

B-mode Scanner Types

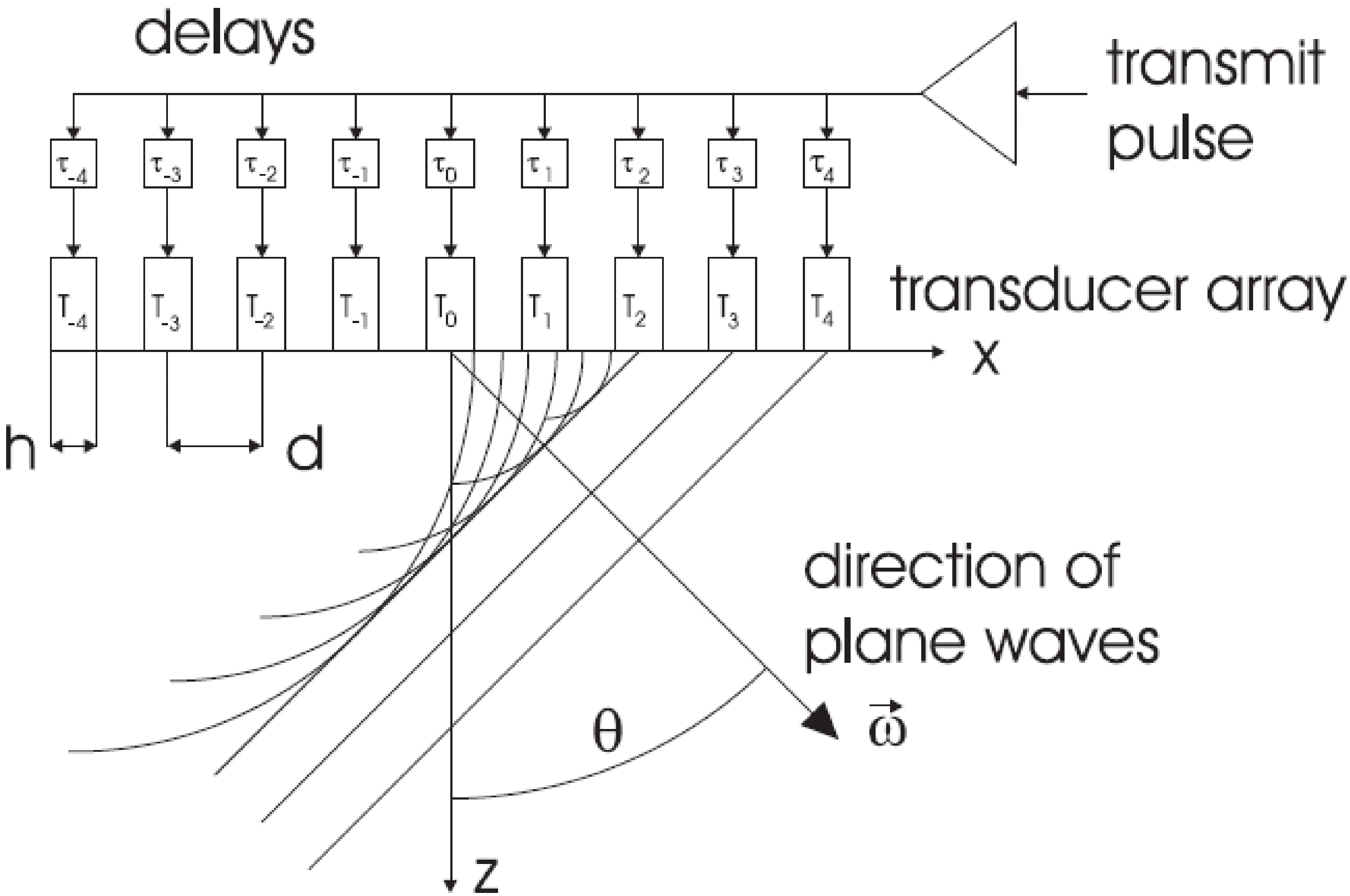
- To reduce scanning time for each frame, a B-mode scanner uses multiple transducers



Phased Arrays

- Phased array:
  - Much smaller transducer elements than in linear array
  - Use electronic steering/focusing to vary transmit and receive beam directions

Transmit Steering



Delays for Steering

- Extra distance that T0 travels than T1:

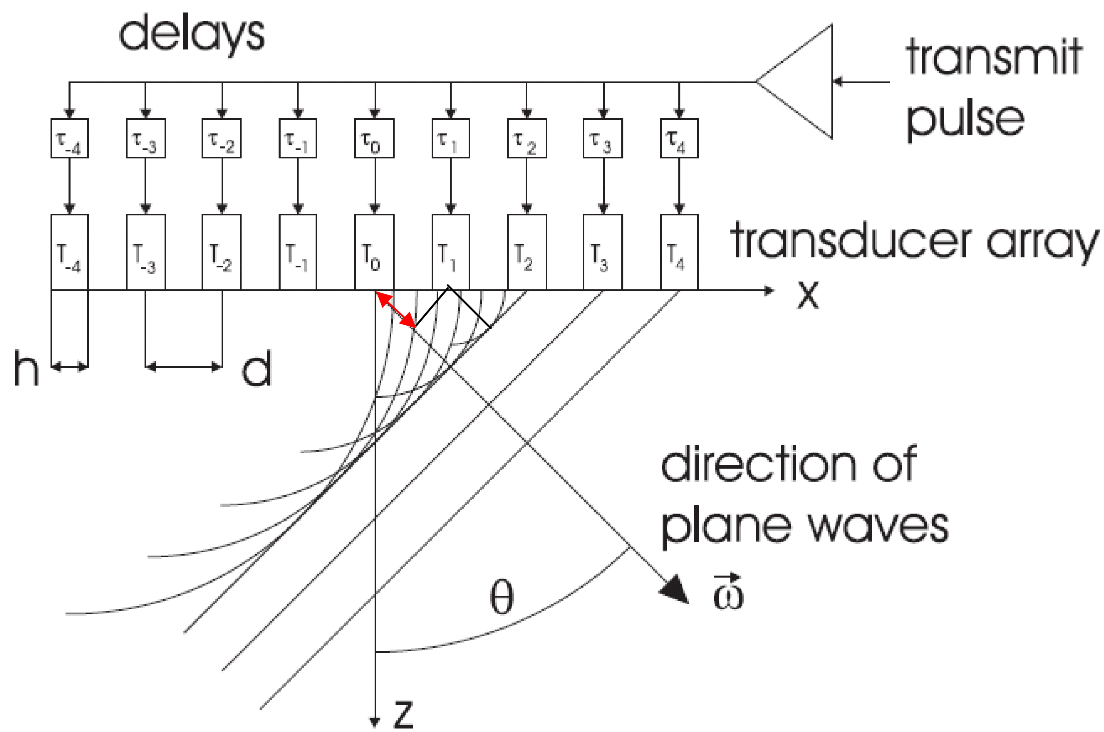
$$\Delta d = d \sin \theta$$

- For the wave from T1 to arrive at a point at the same time as T0, T1 should be delayed by

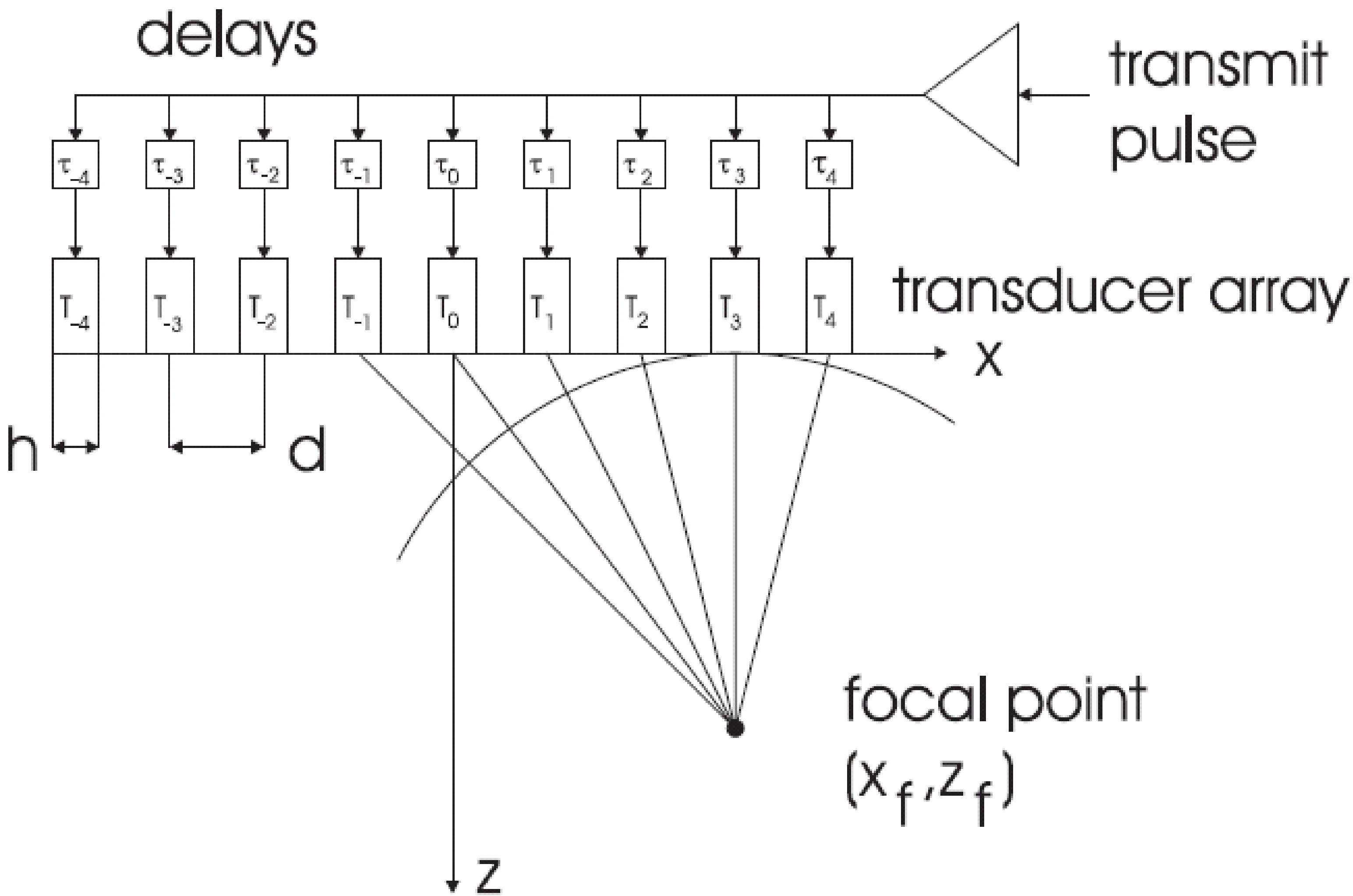
$$T = \Delta d / c = d \sin \theta / c$$

- If T0 fires at t=0, Ti fires at

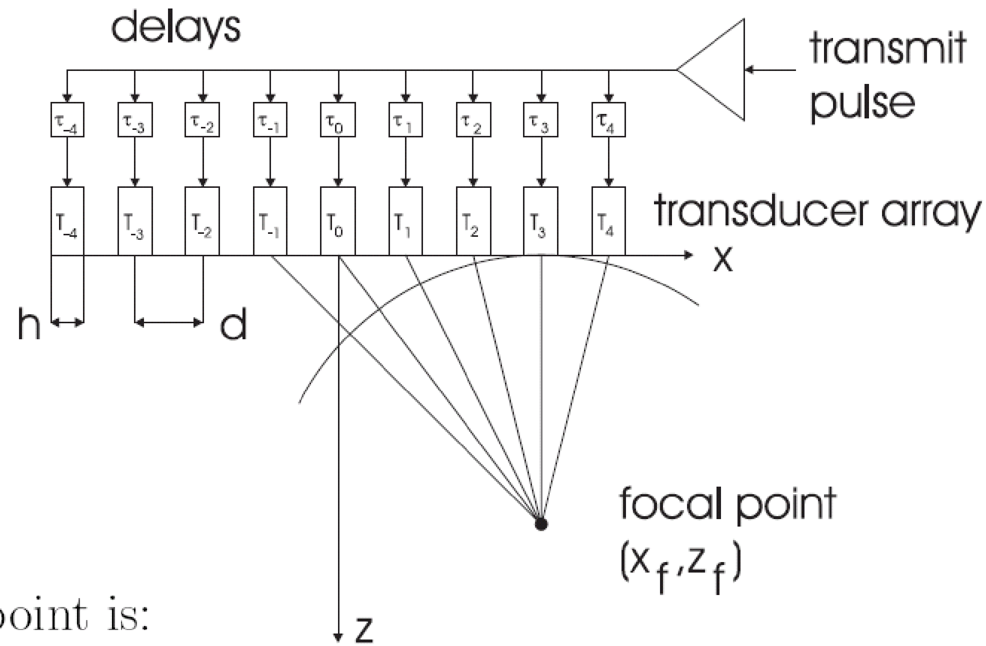
$$t_i = iT = id \sin \theta / c$$



Transmit Focusing



Delays for Focusing



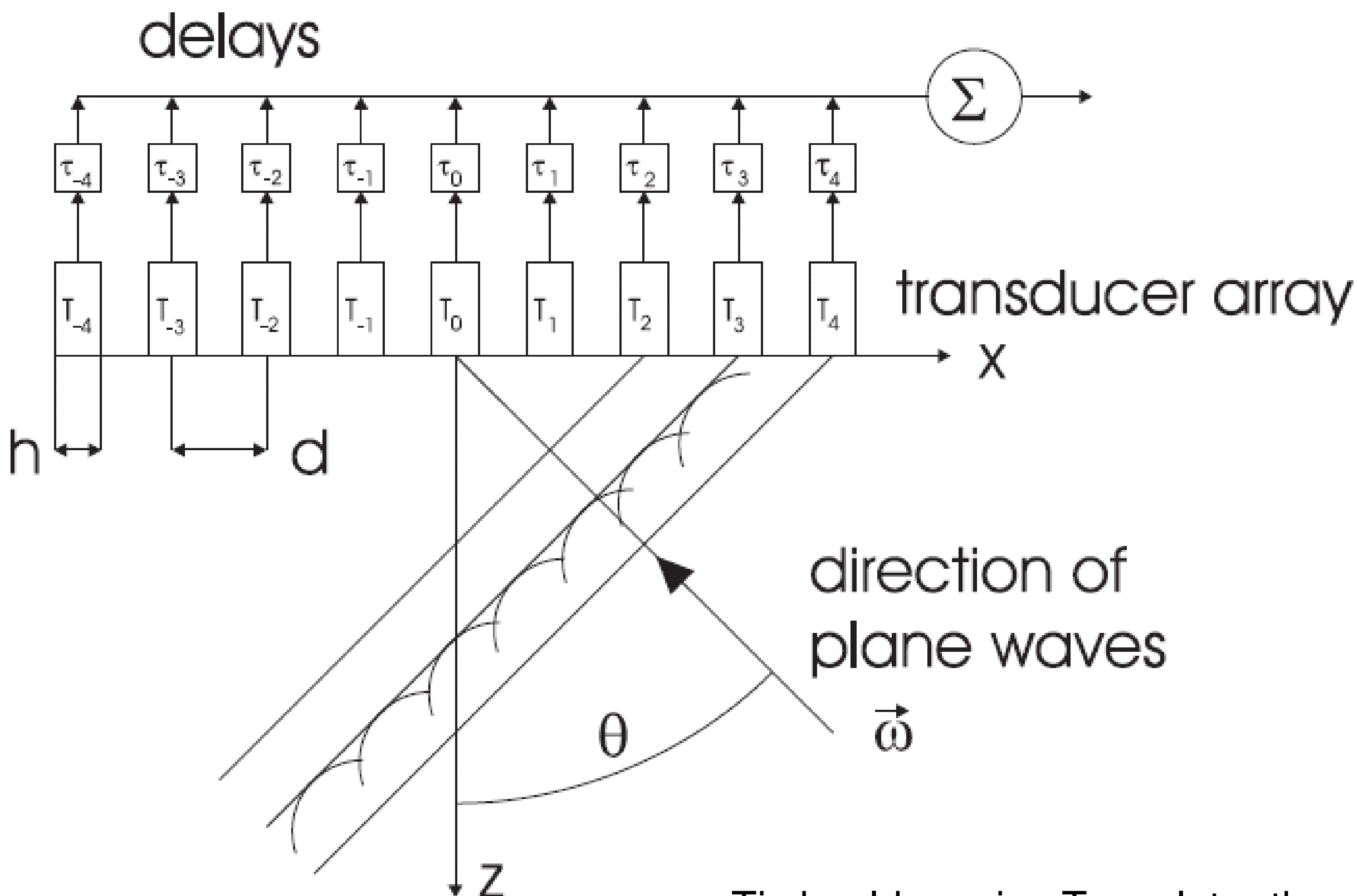
- Focal point at (x_f, z_f)
- T_i is at $(id, 0)$.
- Then range from T_i to focal point is:

$$r_i = \sqrt{(id - x_f)^2 + z_f^2}$$

- Assume T_0 fires at $t = 0$. Then T_i fires at

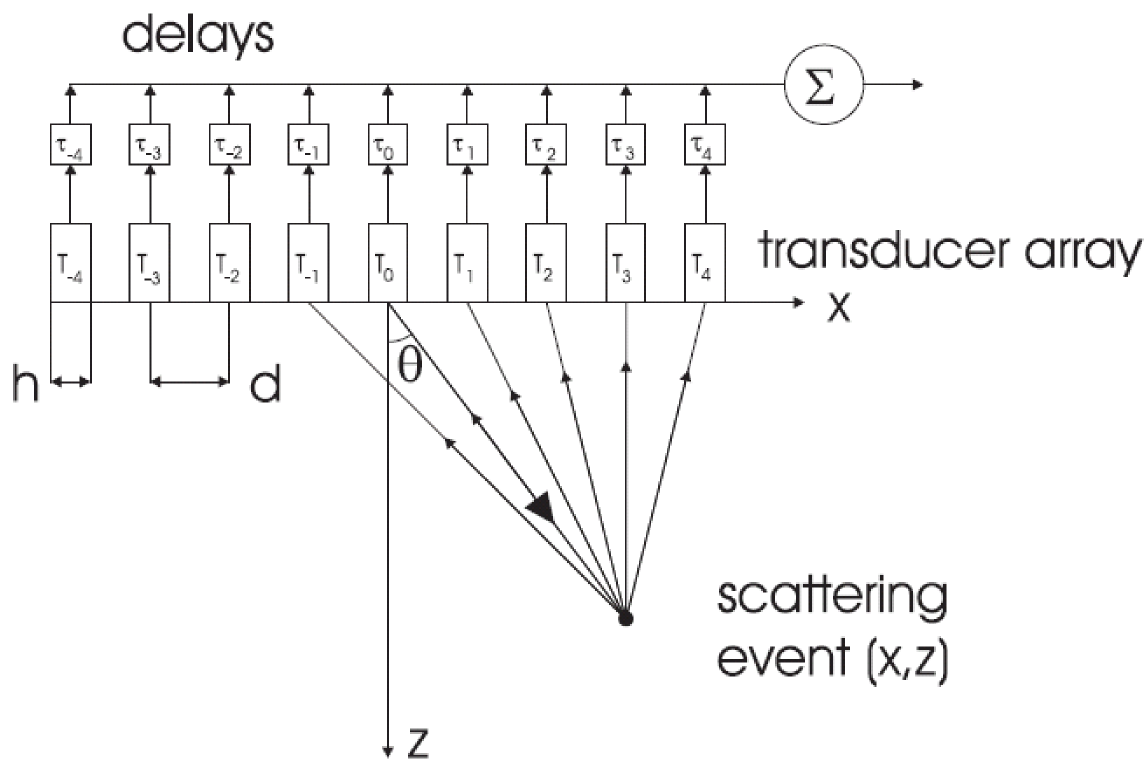
$$t_i = \frac{r_0 - r_i}{c} = \frac{\sqrt{x_f^2 + z_f^2} - \sqrt{(id - x_f)^2 + z_f^2}}{c}$$

Receive Beamforming



Ti should receive T sec later than T_i+1

Receive Dynamic Focusing



T0 fires at direction \theta, and all Ti's receive after a certain delay, so that they are all receiving signal from the same point at a particular time

Delay for Dynamic Focusing

- First consider a stationary scatterer at (x,z)
- Time for a wave to travel from T0 to the scatterer and then to Ti is

$$t_i = \left(\sqrt{x^2 + z^2} + \sqrt{(id - x)^2 + z^2} \right) / c$$

- Time difference between arrival time at T0 and at Ti

$$\Delta t_i = t_0 - t_i = \left(\sqrt{x^2 + z^2} - \sqrt{(id - x)^2 + z^2} \right) / c$$

- T0 fires at t=0, direction \theta. After time t,T0 reaches x(t)=ct \sin\theta, z(t)=ct \cos \theta (possible scatterer)
- Desired time delay is a function of t:

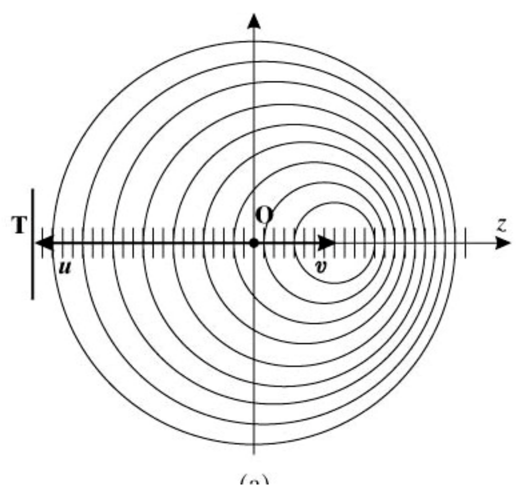
$$\tau_i(t) = t - \frac{\sqrt{(id)^2 + (ct)^2 - 2ctid \sin\theta}}{c} + \frac{Nd}{c}$$

Doppler Imaging

- Principles
  - Velocity vs. frequency shift (Doppler Freq)
- Used for blood flow velocity measurement
  - Stenosis or narrowing of the arteries causes blood flow velocity change
  - Children at risk of stroke have cerebral blood velocities 3-4 times of normal
- Imaging methods:
  - Continuous wave (CW) Doppler measurement
    - Demodulation
    - Time correlation
  - Pulse mode Doppler measurement

Doppler Effect: Moving source

- Doppler effect: change in frequency of sound due to the relative motion of the source and receiver
- Case 1: moving source (scatterer), stationary receiver (transducer)
  - Source moving away, wavelength longer, lower freq
  - Source moving closer, wavelength shorter, higher freq.



source (freq $= f_o$) moving with speed v opposite the wave direction:

one period $T = 1/f_o$

crest in wave moves a distance of $cT = c/f_o$ without source motion

source moves a distance of $vT = v/f_o$

With source motion, crest moves a distance of $cT + vT$.

The above can be thought of as the equivalent wavelength $\lambda_T = cT + vT$

Equivalent temporal frequency is

$$f_T = \frac{c}{\lambda_T} = \frac{c}{c+v} f_o$$

General case source moving in a direction θ:

(angle between source-> receiver vector and source motion vector)

$$f_T = \frac{c}{c - v\cos\theta} f_O$$

Doppler frequency:

$$f_D = f_T - f_O = \frac{v\cos\theta}{c - v\cos\theta} f_O \approx \frac{v\cos\theta}{c} f_O$$

$\theta > \pi/2$: source moving away from receiver, $f_D < 0$

$\theta <= \pi/2$: source moving towards receiver, $f_D > 0$

Doppler Effect: Moving Receiver

- Case 2: stationary source (transducer), moving receiver (target)
  - Transducer transmitting a wave at freq fs, wavelength =c/fs
  - Object is a moving receiver with speed v, with angle θ
  - Target moving away (θ>= π/2), sound moves slower
  - Target moving closer (θ<=π/2), sound moves faster

$$f_O = \frac{c + v\cos\theta}{c} f_S$$

Doppler frequency:

$$f_D = f_O - f_S = \frac{v\cos\theta}{c} f_S$$

$\theta >= \pi/2$: source moving away from receiver, $f_D < 0$

$\theta <= \pi/2$: source moving towards receiver, $f_D > 0$

Doppler Effect for Transducer

- Transducer:
  - Transmit wave at freq fs to object
  - Object moves with velocity v at angle θ
  - Object receives a wave with freq fo =(c+vcosq)/c fs
  - The object (scatterer) reflects this wave (acting as a moving source)
  - Transducer receives this wave with freq

$$f_T = \frac{c}{c - v\cos\theta} f_O = \frac{c + v\cos\theta}{c - v\cos\theta} f_S$$

  - Doppler freq

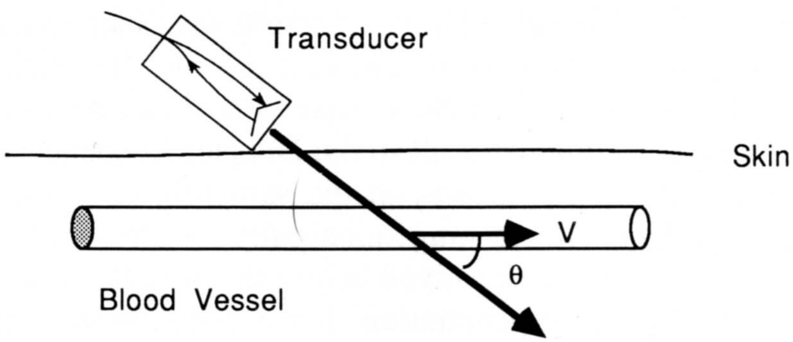
$$f_D = f_T - f_S = \frac{2v\cos\theta}{c - v\cos\theta} f_S \approx \frac{2v\cos\theta}{c} f_S$$

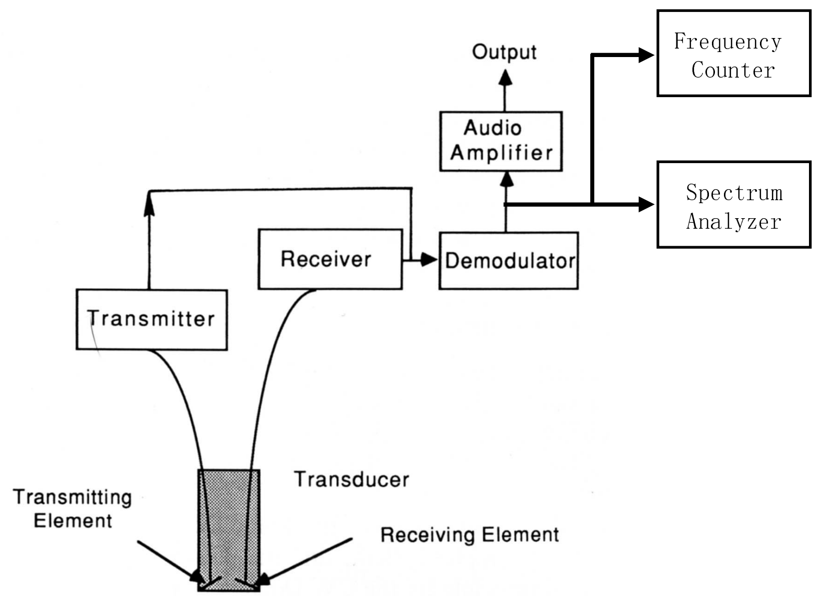
- Dopper-shift velocimeter: as long as θ not eq 90^0, can recover object speed from doppler freq.
- Doppler imaging: display f_D in space and time
  -

CW Doppler Measurement

- Separate transmitter and receiver
- Transducer at a fixed position, transmit and receive over a relatively long period of time
  - No depth information
  - Assumes a single moving object below the transducer
- Both transmitted and received signals can be considered as a sinusoidal signal with freq. f_s and f_T respectively
- Instrumentation:
  - Use AM modulation to deduce the frequency shift f_D
  - Use time-domain correlation

Instrumentation for demodulation based method





From Graber: lecture notes for BMI F05

Homodyne Demodulation

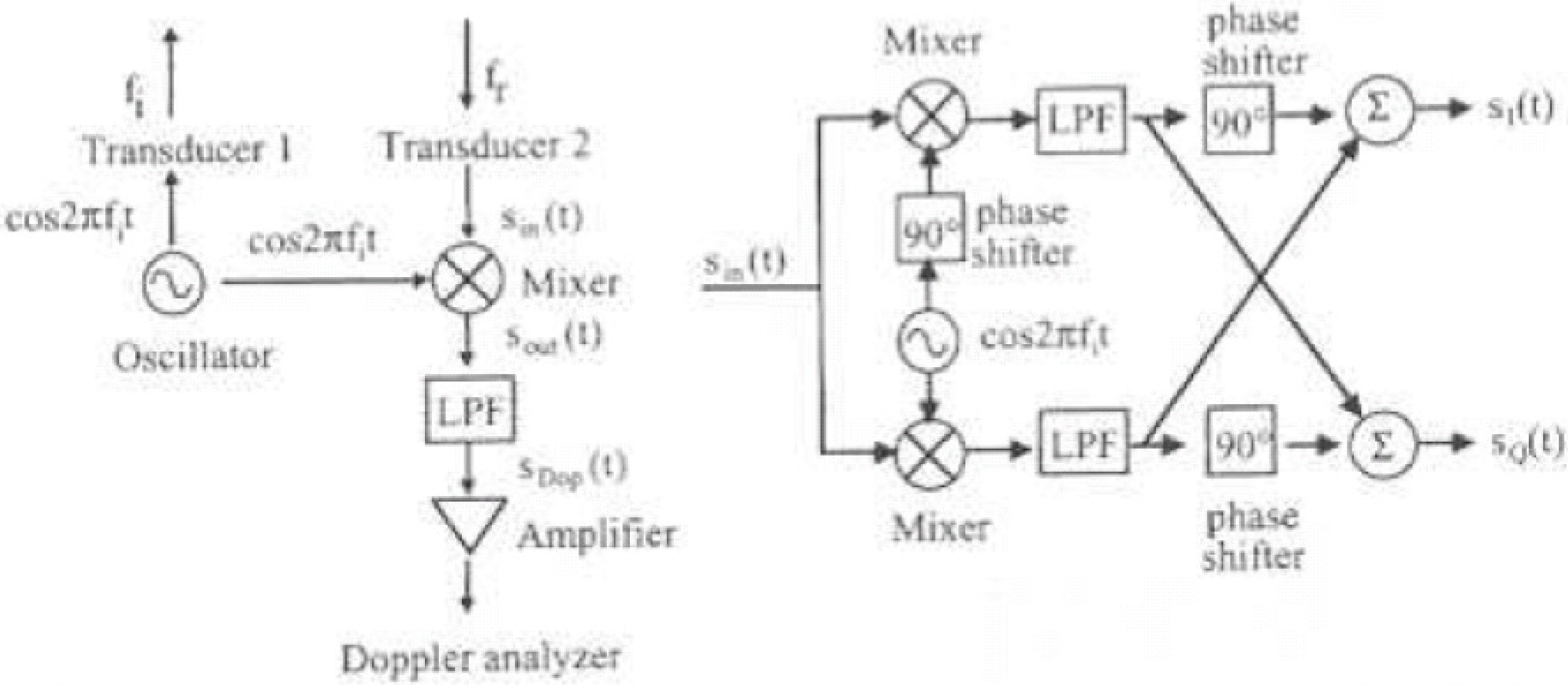
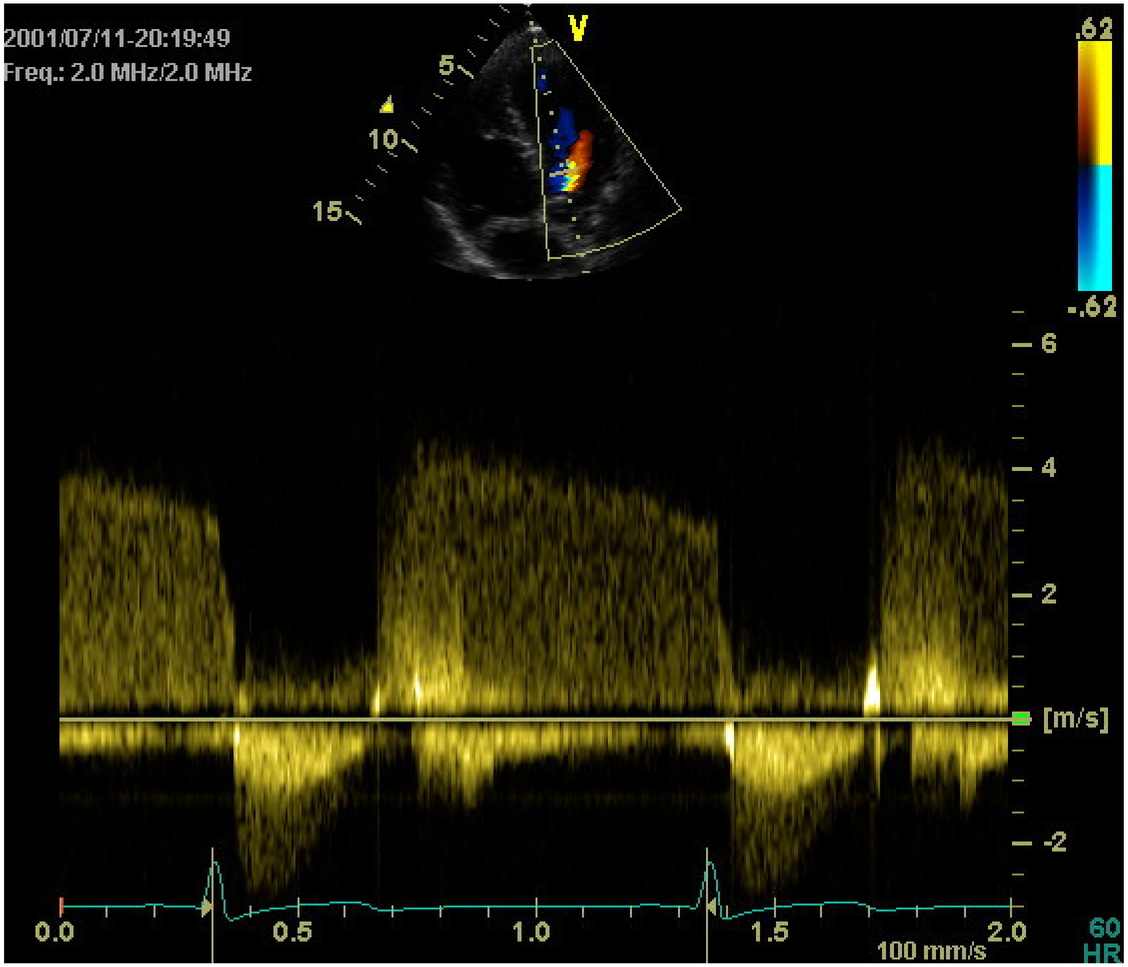


FIGURE 3.22. *(Left) A schematic of a homodyne demodulator used to extract the Doppler frequency in CW measurements. After mixing the received voltage $s_{in}(t)$ with the oscillator voltage, the output signal $s_{out}(t)$ is passed through a low-pass filter (LPF) and amplified. (Right) A schematic of a heterodyne demodulator which can be used to resolve directional ambiguity in the CW Doppler signal.*

From [Webb2003]

CW Doppler Measurement Example



From: Graber Lecture notes for BMS F05

Time Correlation

- See Figure 3.26 in [Webb]
- Performing correlation of two signals detected at two different times
- Deducing the time shift \tau (correspondingly distance traveled) that yields maximum correlation
- Determine the velocity

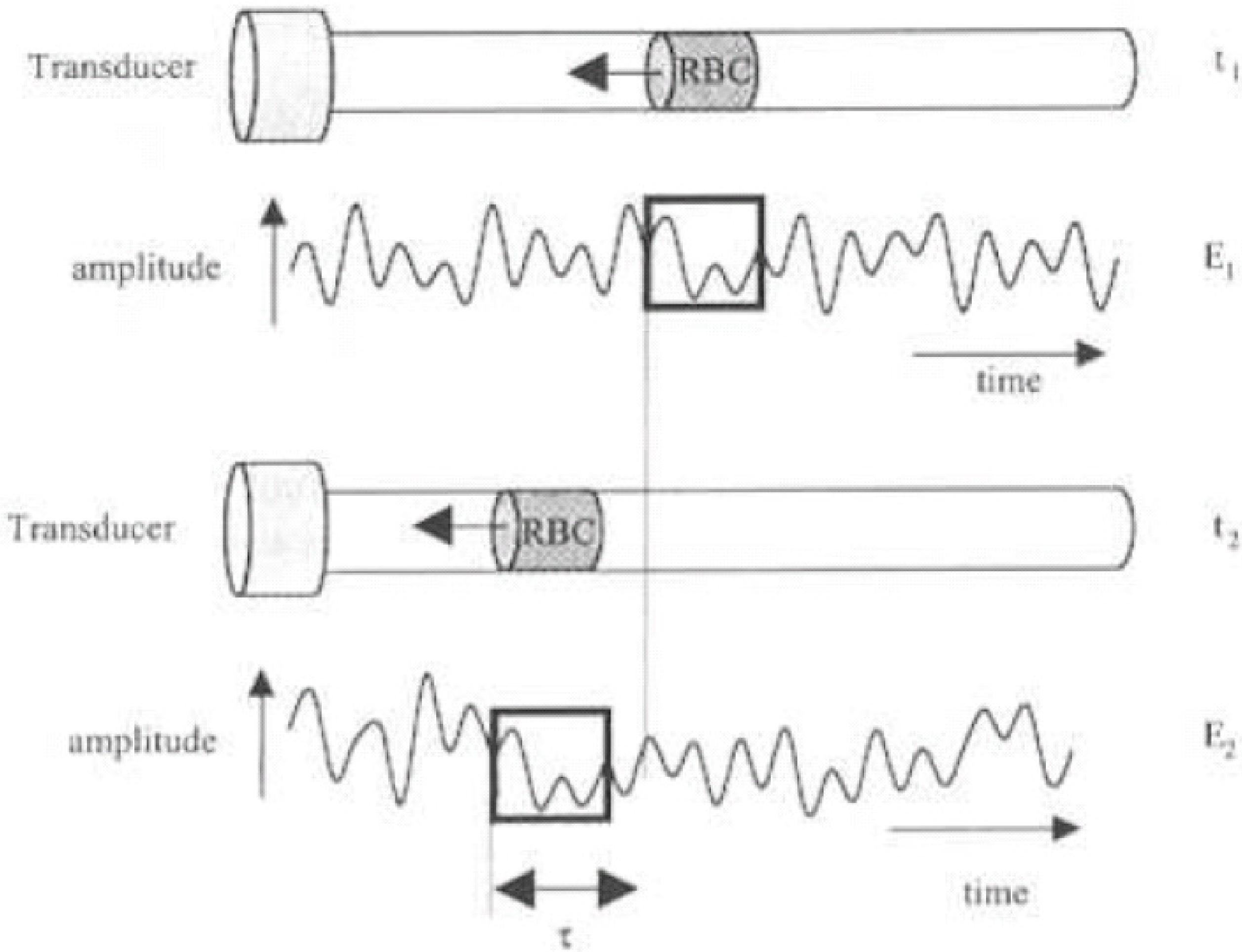


FIGURE 3.26. A diagram showing the basis of time-domain correlation methods for measuring blood velocity. A pulse of ultrasound is transmitted at time t_1 and the backscattered echo E_1 recorded. A second pulse is transmitted at time t_2 and the signature signal from the particular group of RBCs is time-shifted by an amount τ in the corresponding echo E_2. Correlation methods, as described in the text, are used to estimate the value of τ and hence the blood velocity.

Pulse Mode Doppler Measurement

- Use only one transducer
  - Transmits short pulses and receives backscattered signals a number of times
- Can measure Doppler shifts in a specific depth
- Schematic
  - See Figure 3.24 in [Webb]

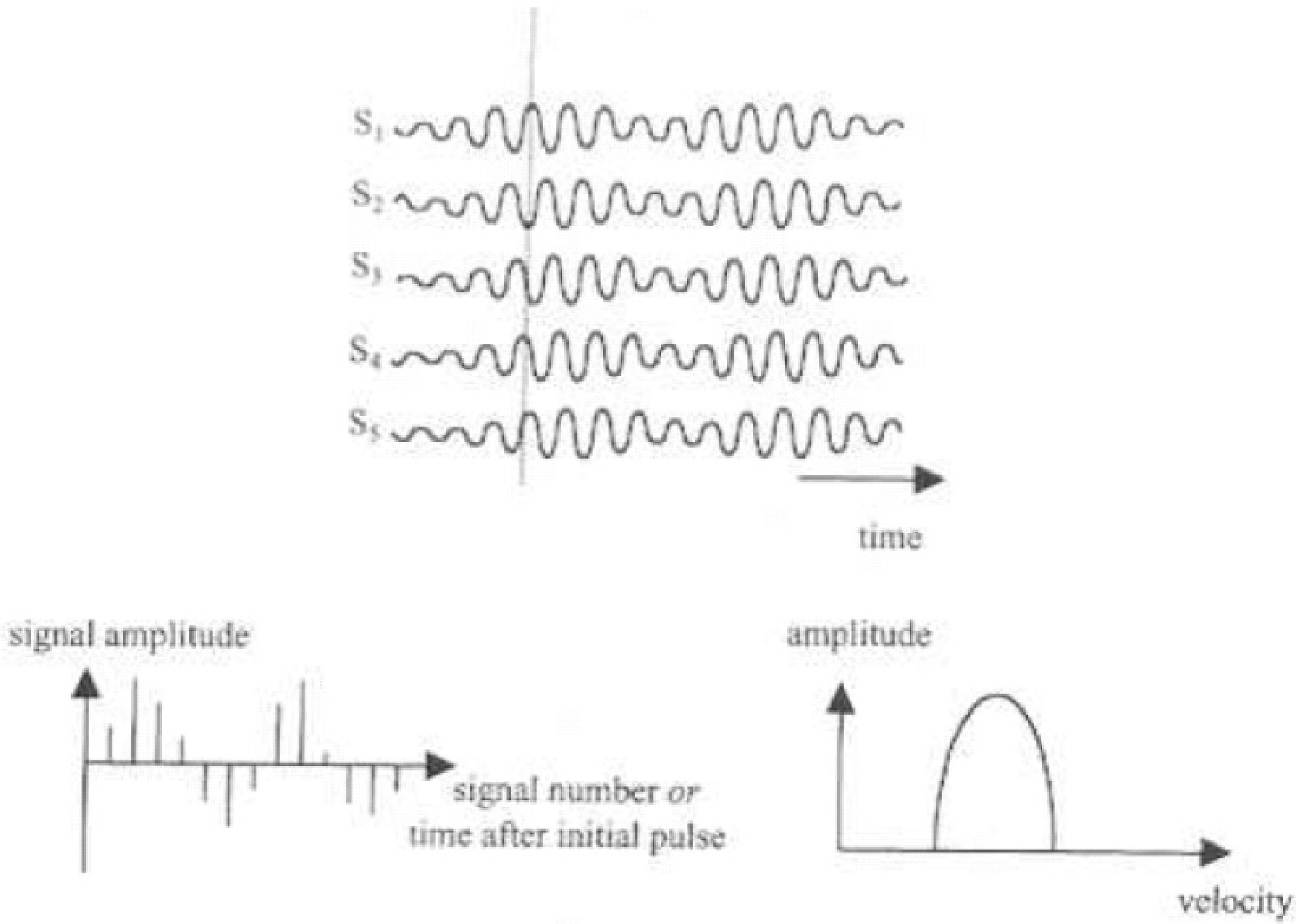


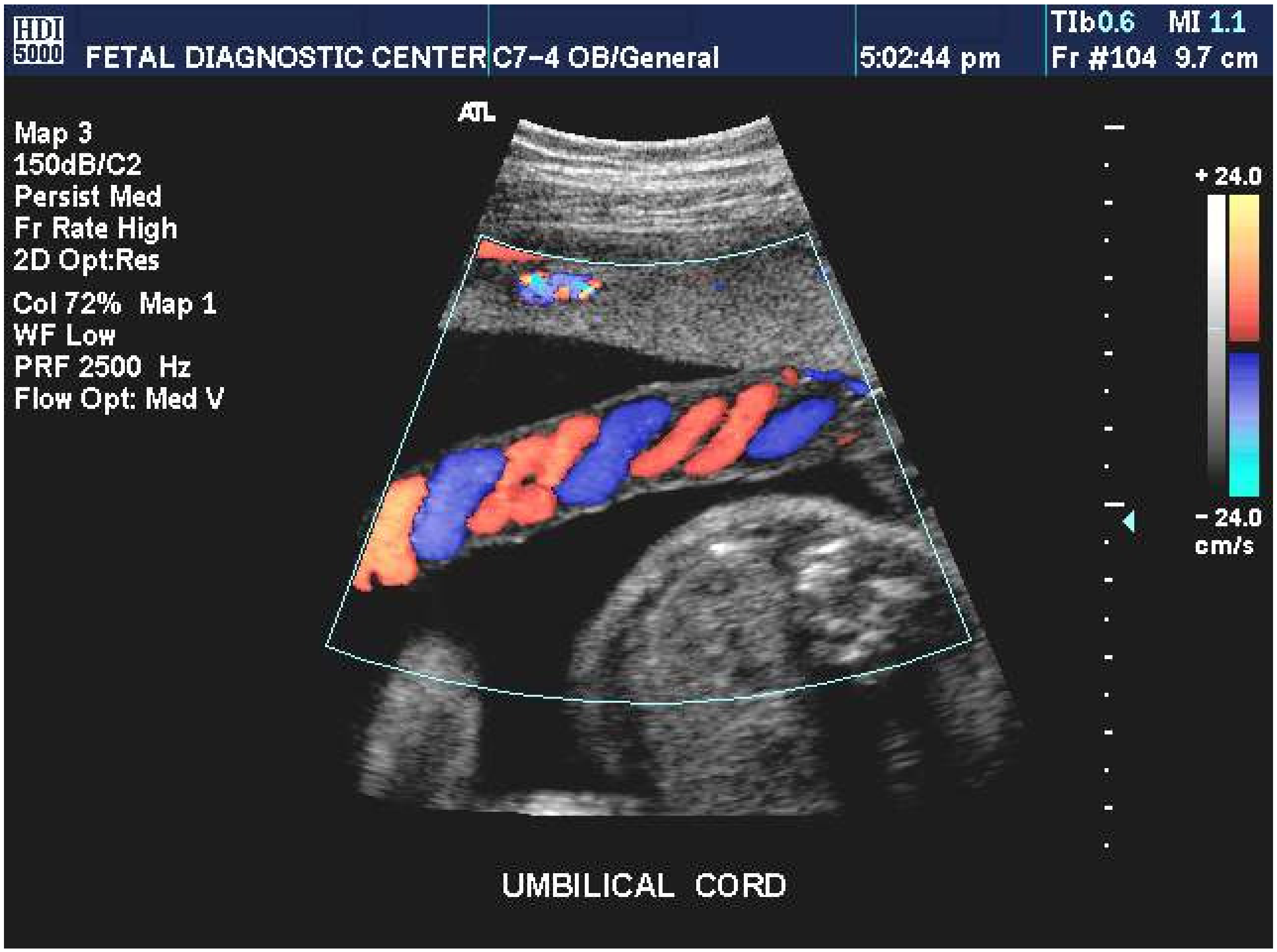
FIGURE 3.24. The basic processing steps in pulsed-mode Doppler ultrasound. (Top) After each pulse in a pulse train a backscattered signal (S_1, S_2, ..., S_5) is recorded. (Bottom left) The signal amplitude at a particular depth, corresponding to the dotted line in the top figure, is plotted as a function of the time after the initial pulse. (Bottom right) Fourier transformation of this plot results in the Doppler frequency, and hence blood velocity, distribution at the chosen location.

Duplex Imaging

- Combines real-time B-scan with US Doppler flowmetry
- B-Scan: linear or sector
- Doppler: C.W. or pulsed (f_c = 2-5 MHz)
- Duplex Mode:
  - Interlaced B-scan and color encoded Doppler images $\Rightarrow$ limits acquisition rate to 2 kHz (freezing of B-scan image possible)
  - Variation of depth window (delay) allows 2D mapping (4-18 pulses per volume)

From: Graber Lecture notes for BMS F05

Duplex Imaging Example



From: Graber Lecture notes for BMS F05

Clinical Applications

- Ultrasound is considered safe; instrument is less expensive and imaging is fast
- Clinical applications
  - Obstetrics and gynecology
    - Widely used for fetus monitoring
  - Breast imaging
  - Musculoskeletal structure
  - Cardiac diseases
- Contrast agents

See Handouts (Sec. 3.11, 3.12, 3.13 in [Webb])

Summary

- How transducer works
  - Properties of piezoelectric crystal, needs for damping and matching layer
- Pulse-Echo equation
  - Relation between excitation pulse n(t), transducer face s(x,y) and received signal r(t)
  - General equation and different approximation
  - Need for time gain compensation
  - Envelope of received signal after time gain compensation is an estimate of the reflectivity distribution in the z-direction below the transducer
    - Blurring function correspond to different approximation
    - Resolution cell depends on transducer face and excitation pulse
    - Lateral resolution depends on z with Fraunhofer approximation
      - Beam width increases with z
- Different US Scanning mode
  - A-mode (reflectivity in z for fixed (x,y) position), M-mode (motion trace in z for fixed (x,y)), B-mode (reflectivity in one cross section)
  - 3D imaging
  - Doppler imaging

Summary (cnt'd)

- Phased Array Steering and Focusing
  - Time delay for transmit element for steering and focusing
  - Time delay for receive focusing and dynamic focusing
- Velocity measurement using Doppler imaging
  - CW vs. time gated
- Clinical applications of different US imaging modes

Reference

- Prince and Links, Medical Imaging Signals and Systems, Chap 11 and Sec. 10.5
- A. Webb, Introduction to Biomedical Imaging, Chap. 3
- Handouts from Webb: Sec. 3.10: Blood velocity measurements using ultrasound; 3.13: clinical applications of ultrasound

Homework

- Reading:
  - Prince and Links, Medical Imaging Signals and Systems, Sec. 10.5, Ch.11
  - Handouts: Webb Sec 3.10-3.13
- Problems:
  - P11.4 (note T should be T=2 \lambda/c)
  - P11.6
  - P11.7
  - P11.8
  - P11.9
  - P11.14
  - P10.12 (note f_R should be (c+v) f_0/c)
  - P10.13