Lecture 23 – Ultrasound Application

This lecture will cover: *(CH4.8-4.13)*

- Clinical diagnostic scanning modes
- Imaging characteristics
- Doppler Ultrasound
- Ultrasound contrast agent
- Harmonic and pulse inversion imaging
- Application

(*Supplementary reading: The Essential Physics of Medical Imaging CH14.5-14.9, FMI 6.5*)

Clinical diagnostic scanning modes

- **A (amplitude) mode :**

  a one-dimensional "line image" which is a plot of amplitude vs time

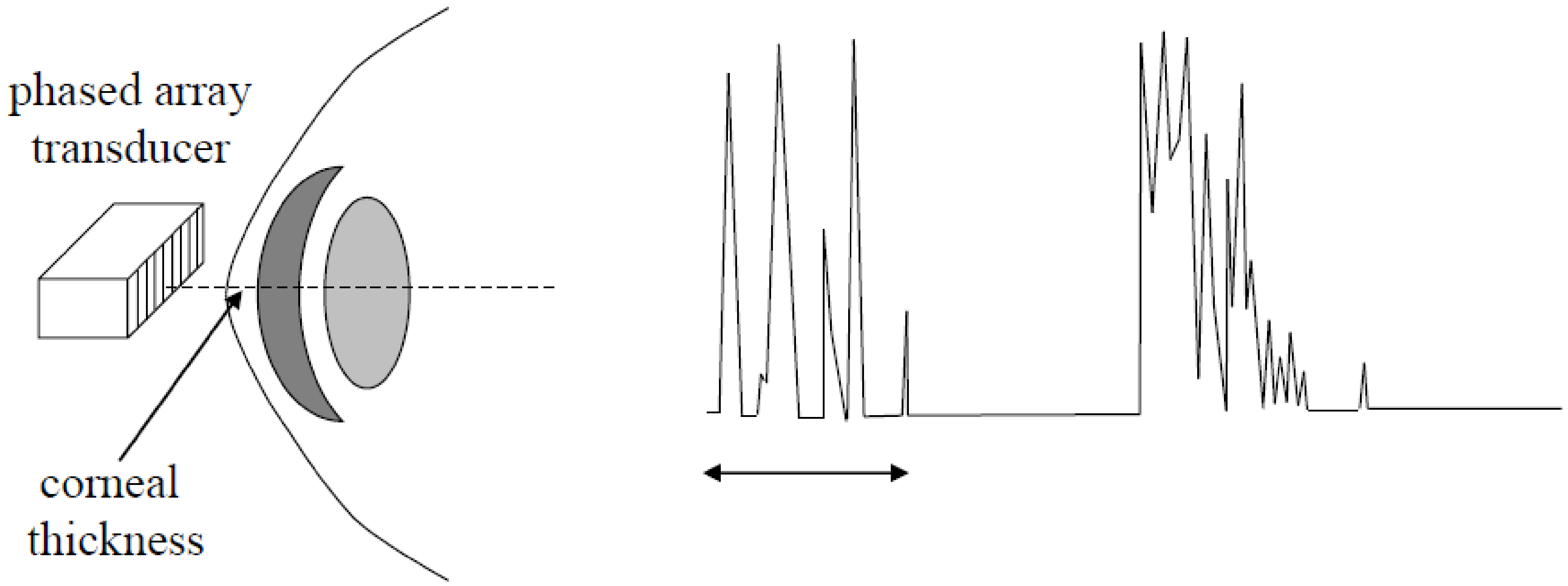

Fig. Use of A-mode ultrasound scanning to measure the corneal thickness of the eye. A single line of high frequency ultrasound is used, and the one-dimensional signal plot is shown on the right. The double headed arrow represents the thickness of the cornea..

Clinical diagnostic scanning modes

- **M (motion) mode:**

  a continuous series of A-mode lines and display them as a function of time.

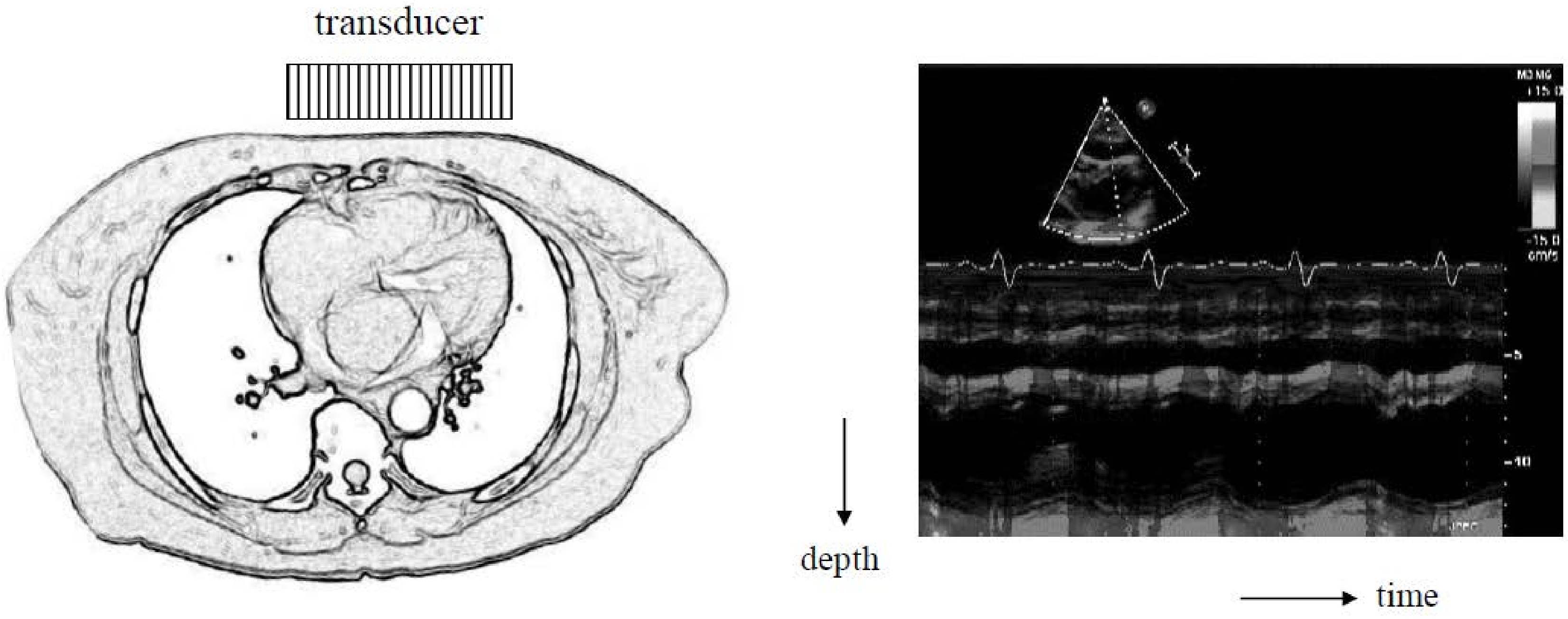

Fig. M-mode data acquisition. The transducer is placed above the heart and sends out a single line of ultrasound. An A-mode scan is recorded, and as soon as the last echo has been acquired, the A mode scan is repeated. The horizontal time-axis increments for each scan, and therefore a time-series of one-dimensional scans is built up. A straight line represents a structure that is stationary, whereas the front of the heart shows large changes in position.

Clinical diagnostic scanning modes

- **B (Brightness) mode:**
  - most commonly used in clinical diagnosis
  - a 2D image through a cross-section of tissue;
  - 3D imaging can be performed by multi-dimensional arrays

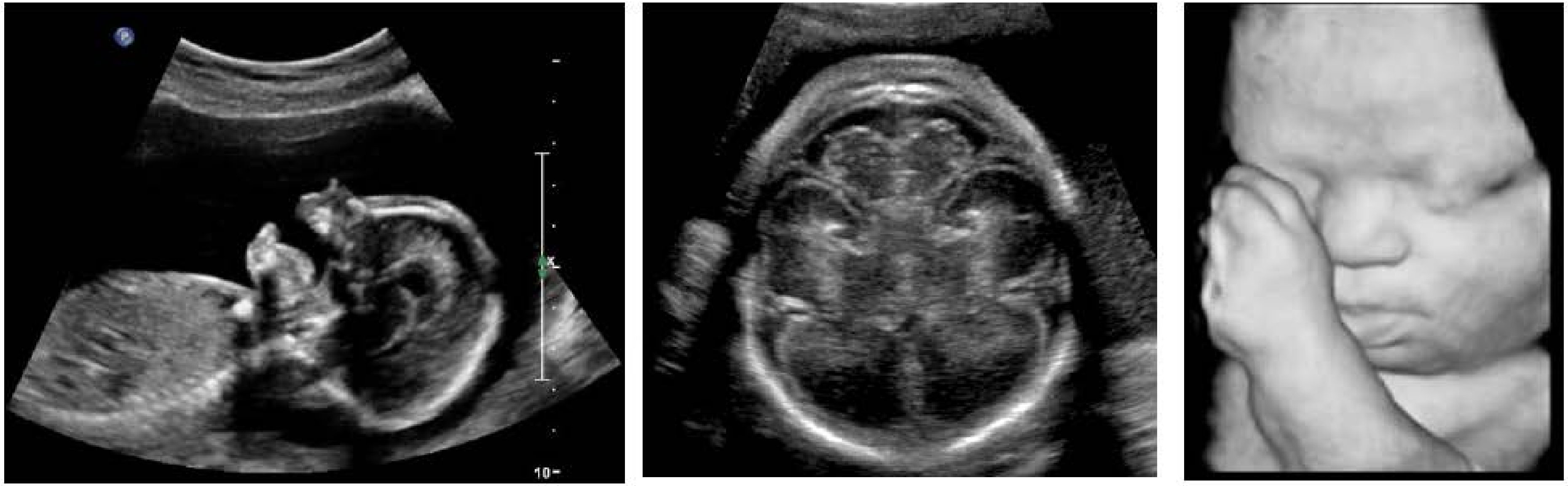

Fig. Two-dimensional B-mode scans of (left) 19 week fetus in the womb, and (centre) foetal brain. (right) Three dimensional foetal image using a two-dimensional array and mechanical steering.

Clinical diagnostic scanning modes

Compound scanning

- Acquire an ultrasound image from multiple angles and combine the images together;
- Reduce the speckles caused by scattering;
- Present the irregular curvatures without influence of structures parallel to the beam;
- Reduce other artifacts such as acoustic enhancement and shadowing;

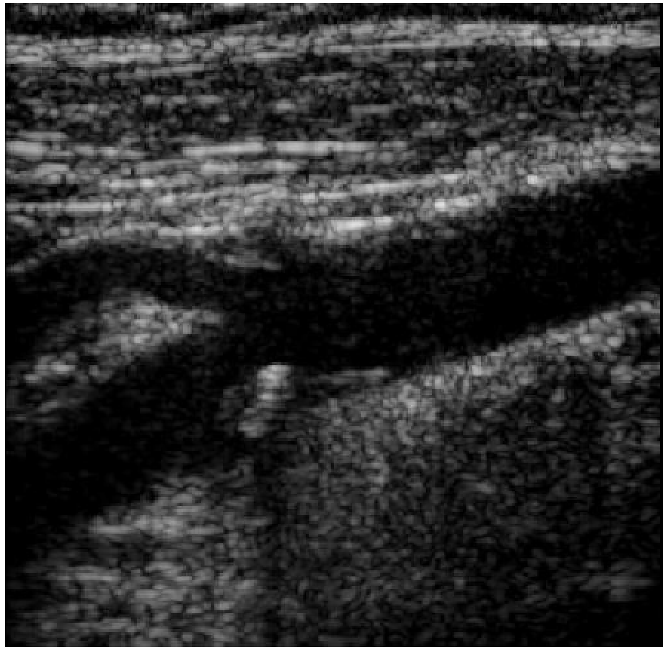

Fig. Comparison of a carotid artery bifurcation acquired using a conventional B-mode scan on the left, and a compound scan with nine different orientations on the right..

Image characteristics

- **Signal-to-noise**
  - The intensity of transmitted pulse;
  - The operating frequency of transducer: higher frequency, lower SNR;
  - The type of focusing: the higher focusing, the higher SNR at focal area, but lower SNR outside of depth-of-focus;
  - Noise sources: speckles from scattering and clutters from side lobes, grating lobes, multi-path reverberation and tissue motion;
- **Spatial resolution**
  - Lateral resolution: focusing and frequency
  - Axial resolution: ½ wavelength of ultrasound pulse, therefore higher damping, frequency provide better resolution
- **Contrast-to-noise:** similar to SNR

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Doppler Ultrasound

- Doppler effect occurs due to scattering from red blood cells (RBC) with A diameter of 7-10 μm;
- B-mode images acquired by phased array transducers to
  - localize specified vessels or their region
  - estimate the vessel size
  - measure blood velocities converted into blood flow values;
- Blood flow velocity

$$f_{\mathrm{r}} = f_{\mathrm{i}}\left[\frac{(c + v\cos\theta)}{c}\right]^2 = f_{\mathrm{i}} + \frac{2f_{\mathrm{i}}v\cos\theta}{c} + f_{\mathrm{i}}\left[\frac{v\cos\theta}{c}\right]^2$$

$$f_{\mathrm{D}} = |f_{\mathrm{i}} - f_{\mathrm{r}}| \approx \frac{2f_{\mathrm{i}}v\cos\theta}{c} \quad \Rightarrow \quad v = \frac{cf_{\mathrm{D}}}{2f_{\mathrm{i}}\cos\theta}$$

Where f_{D} is the Doppler shift

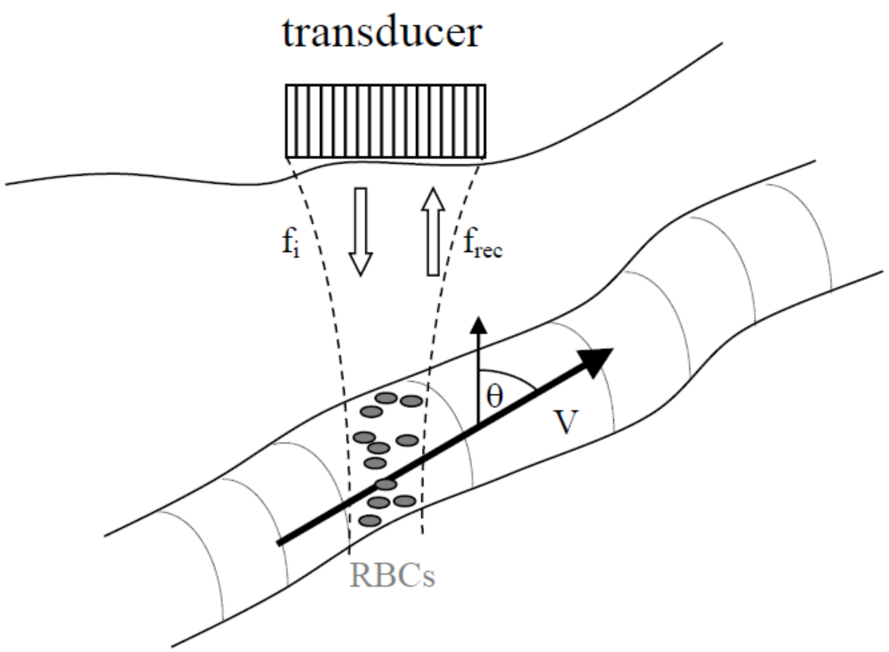


Fig. Showing the origin of the Doppler shift in ultrasound imaging of blood flow. The ultrasound beam is scattered from the RBCs in a vessel. The backscattered ultrasound beam is detected by the transducer at a slightly different frequency (f_{rec}) from that transmitted (f_i) into the body.

Pulse wave (PW) Doppler

- The phased array transducer used to:
  - Pulse transmission and receiving
  - Locate the region-of-interest (ROI)

$$\mathrm{depth}_{\mathrm{min}} = \frac{ct_d}{2}$$

$$\mathrm{depth}_{\mathrm{max}} = \frac{c(t_d + t_g)}{2}$$

Where by using a gate on receiving signals, the signals from tissue outside of the ROI are not recorded.

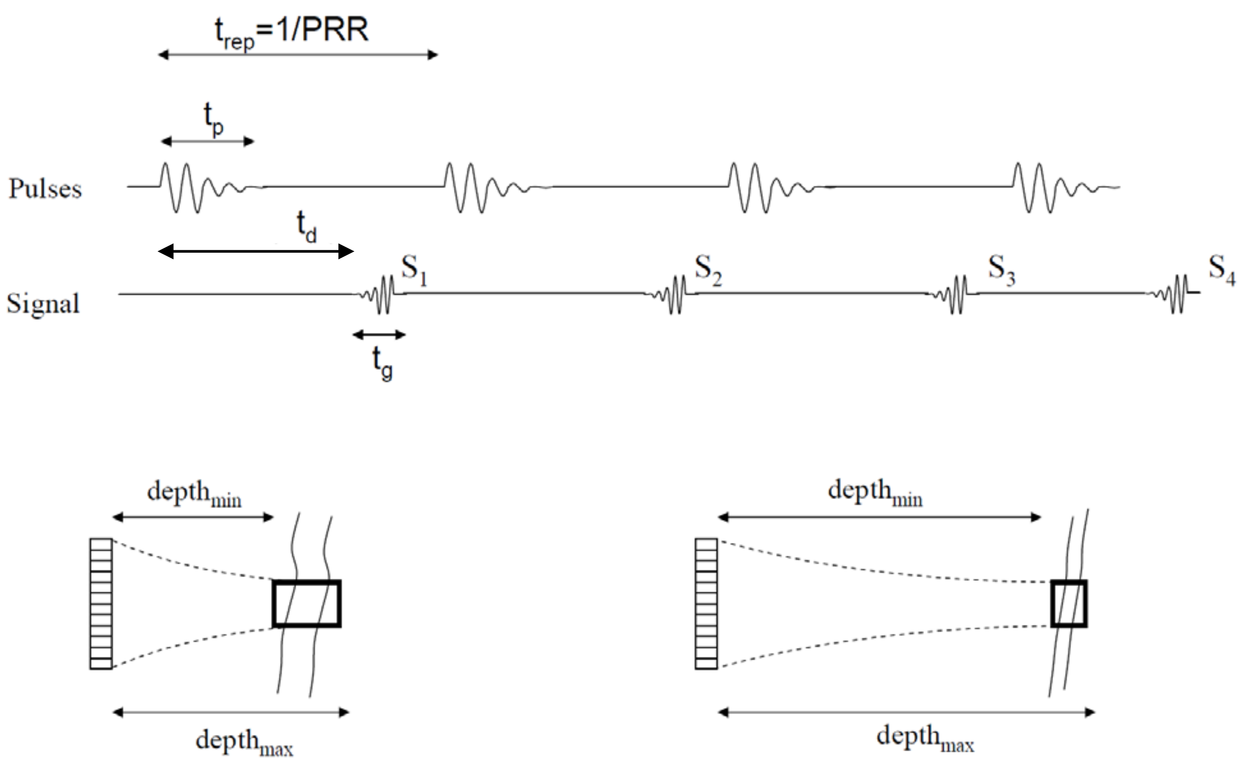


Fig. (top) General mode of operation of pulsed mode Doppler imaging. A series of signals S_1, S_2,. . . . .S_n are acquired to estimate the blood flow. (bottom) The parameters t_p, t_g and t_d are chosen to localize the received signal to the desired ROI, defined by the focal point of the phased array transducer and the minimum and maximum required depths: shown are examples of obtaining information from a vessel close to the surface (left) and deeper within the body (right).

Pulse wave (PW) Doppler

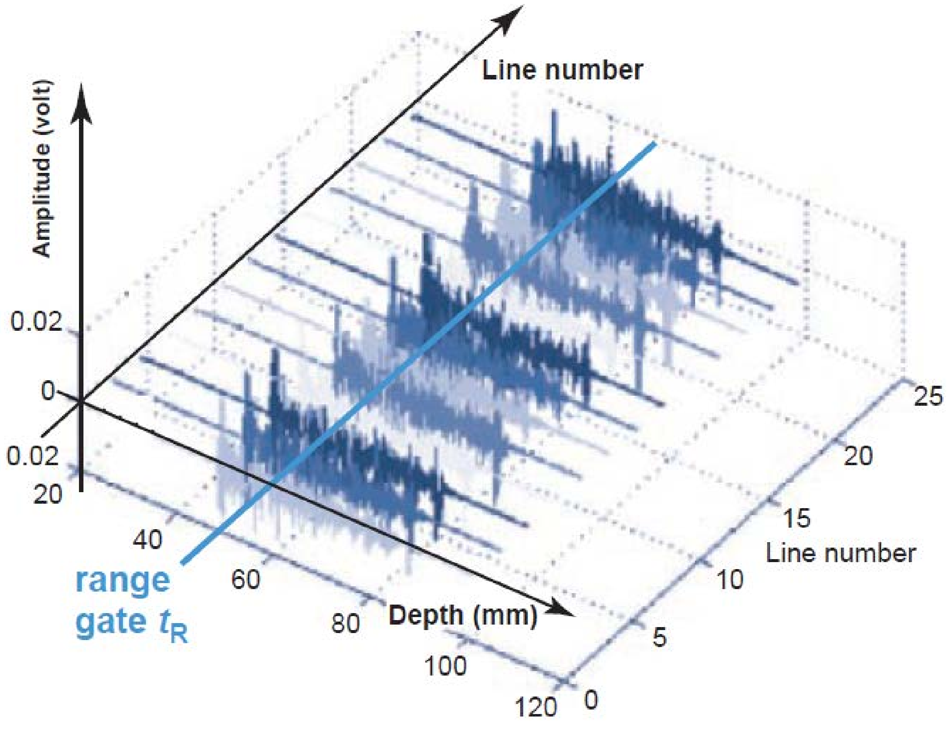


Fig. Pulsed wave Doppler uses the M-mode acquisition scheme and samples the subsequent reflected pulses at a fixed range gate t_R to calculate the Doppler frequency f_D.

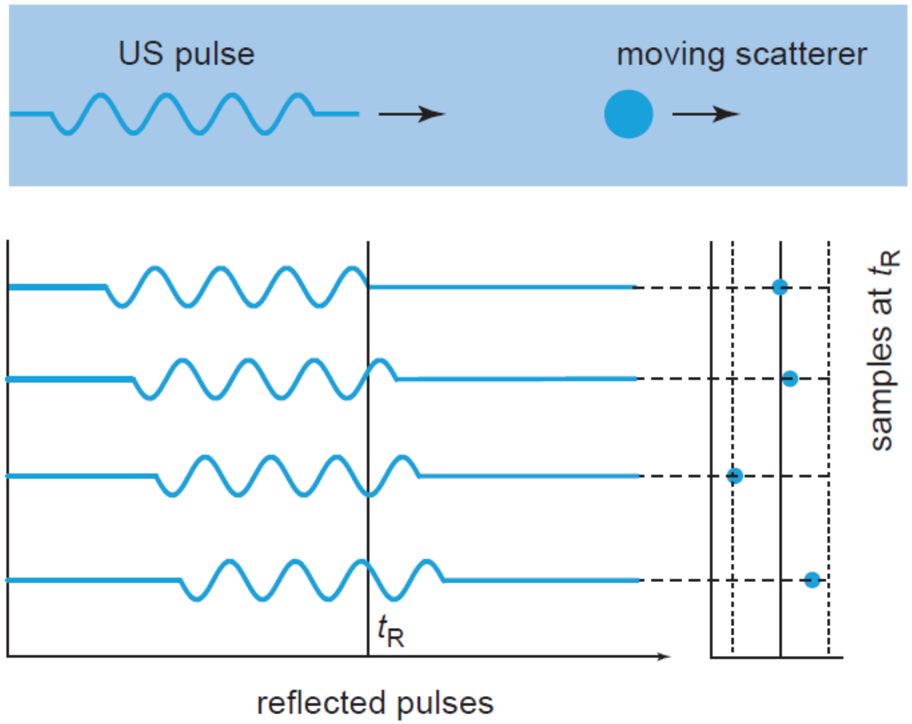


Fig. Schematic representation of the PW Doppler principle. Pulses are transmitted at a fixed pulse repetition frequency (PRF). They are reflected at a scatterer in motion. Because of this motion, the reflected pulses are dephased. Measuring each reflected pulse at the range gate t_R yields a sampled sinusoidal signal with frequency f_D.

Phase shift

➢ The phase shift between two subsequent received pulses:

$$\Delta\phi = 2\pi f_T \frac{2v_a T_{\mathrm{PRF}}}{c}$$

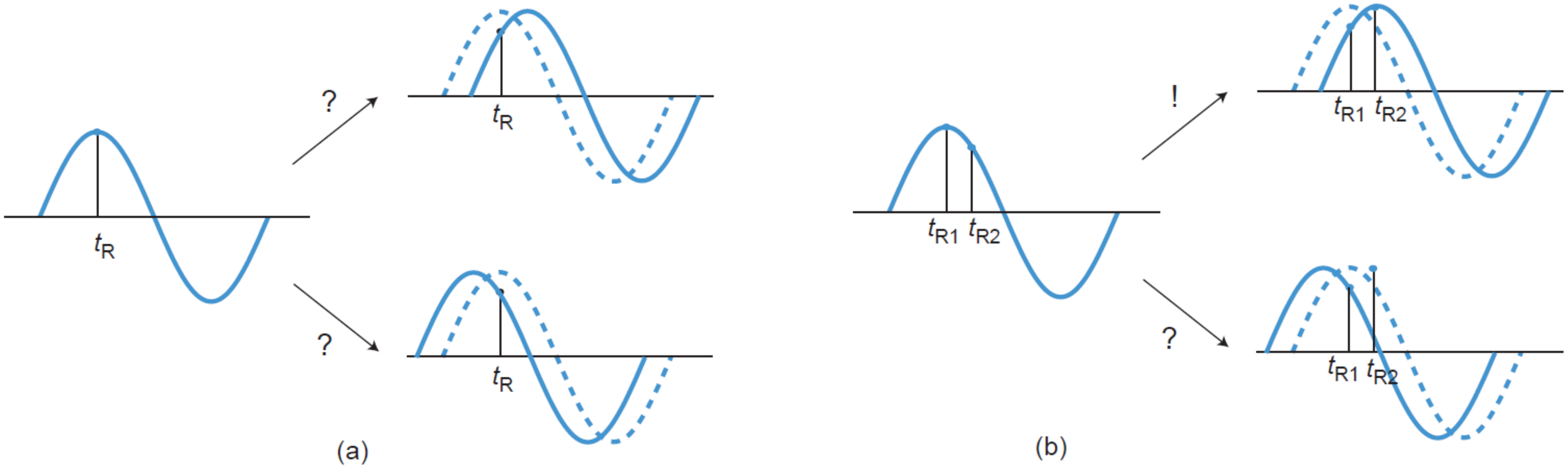


Fig. (a) If a single sample is acquired at the range gate, no directional information is obtained. (b) However, if a second sample is acquired slightly after the first one, the direction of motion is uniquely determined since a unique couple of samples within the cycle is obtained.

Spectral Doppler

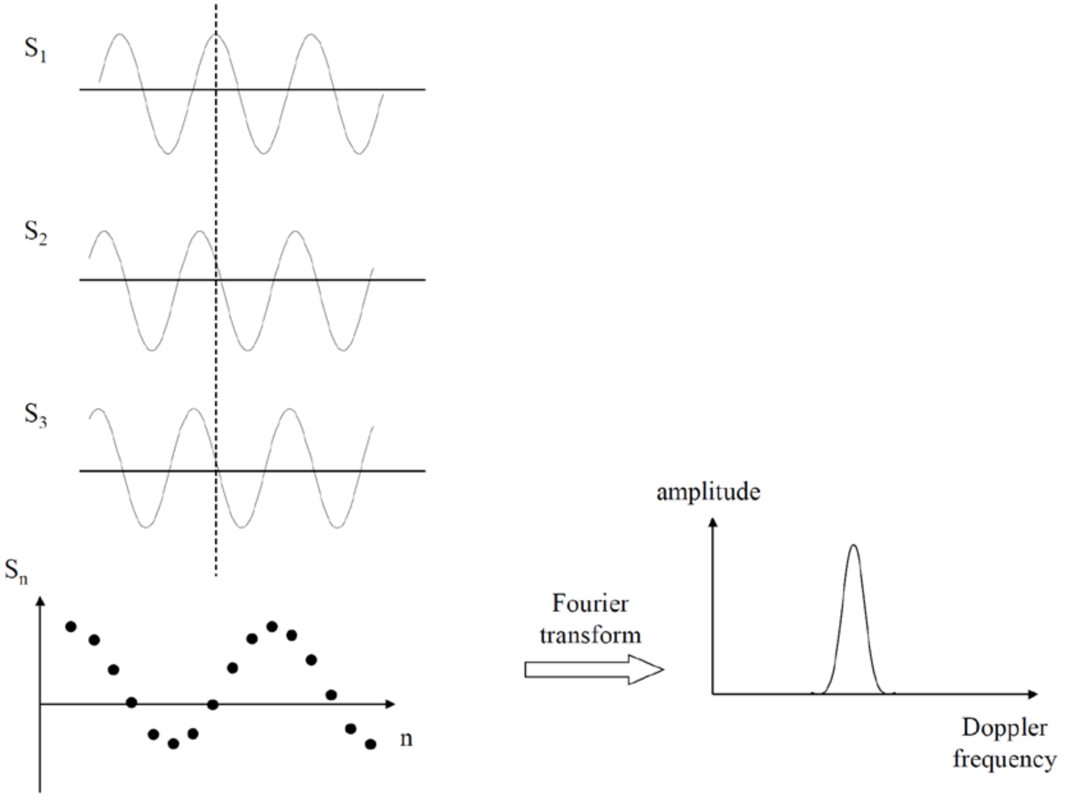


Fig. Steps in the production of a Doppler frequency distribution from one particular axial position within the ROI, determined by the particular time-point in signals S1. . .Sn that is analyzed. In this example a timepoint is chosen at the dashed line. A plot of the signals (Sn), taken at the dashed line, as a function of n is shown at the bottom left. Since the value of n is directly related to the time after the initial RF pulse is applied, a Fourier transform of the Sn vs. n plot gives the Doppler frequency spectrum shown on the right.

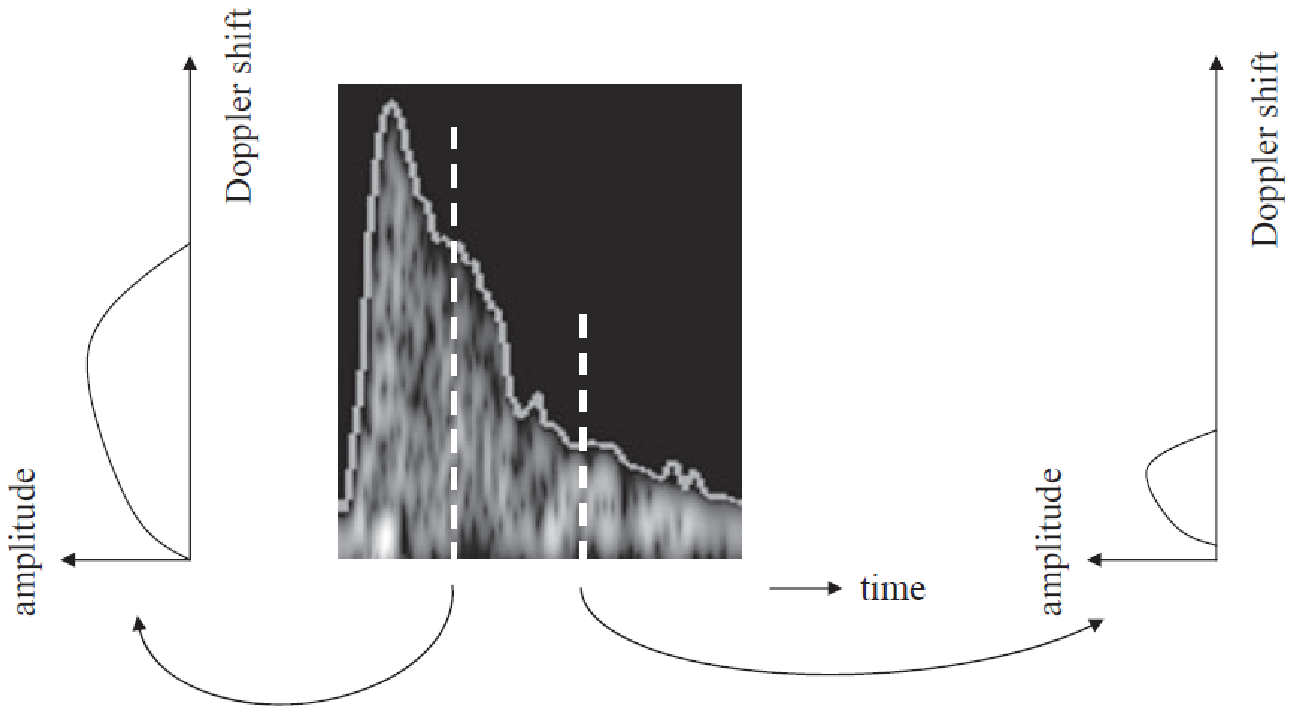
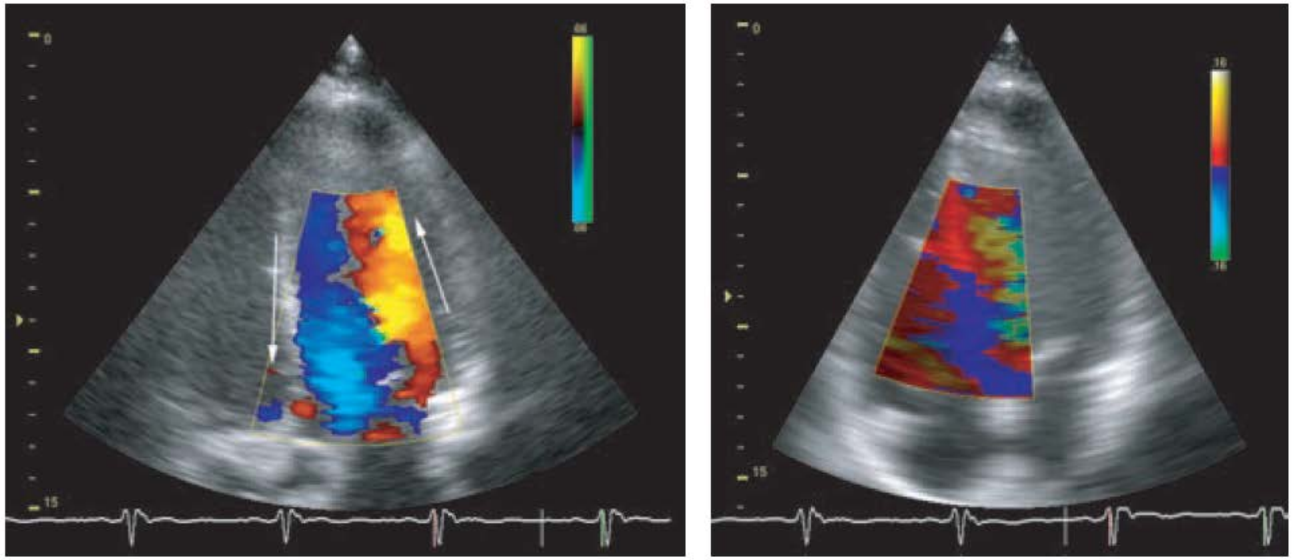


Fig. A spectral Doppler plot, with the amplitude of each frequency component of the Doppler shift being reflected in the intensity of the plot (white is the highest amplitude). The horizontal axis represents time. The left-hand plot shows high Doppler shift frequencies corresponding to high blood velocities just after the heart has reached full contraction and pumped blood into the arteries. The right-hand plot, which occurs at a later time, shows much lower Doppler shift frequencies, and coincides with the heart expanding to fill with blood ready for the next contraction.

Color Doppler

- Display flow information combined with a B-mode scan:
  - Blue/red represents flow towards/away from the transducer,
  - Intensity of color shows velocity;



(a) (b)

Fig. (a) Using color Doppler techniques, blood flow within the ventricles can be visualized. This image shows the flow in a normal left ventricle at the beginning of diastole. Red colors represent flow toward the transducer, coming from the left atrium through the mitral valve and into the left ventricle. Blue colors show the blood within the left ventricle flowing away from the transducer toward the aorta. (b) Doppler techniques can be used to acquire the slower, regional velocities of the heart muscle itself. Local velocities in the direction of the transducer are represented in red, and velocities away from the transducer are in blue.

Aliasing

- From the Nyquist theory, the maximum Doppler frequency shift that can be measured:

$$f_{\max} = \frac{PRF}{2}$$

- The maximum blood velocity

$$v_{\max} = \frac{cf_{\max}}{2f_i} = \frac{cPRF}{4f_i}$$

- The maximum depth

$$d_{\max} = \frac{c}{2PRF} = \frac{c^2}{8f_i v_{\max}}$$

- Continuous wave Doppler: no limit to depth and velocity measurement.

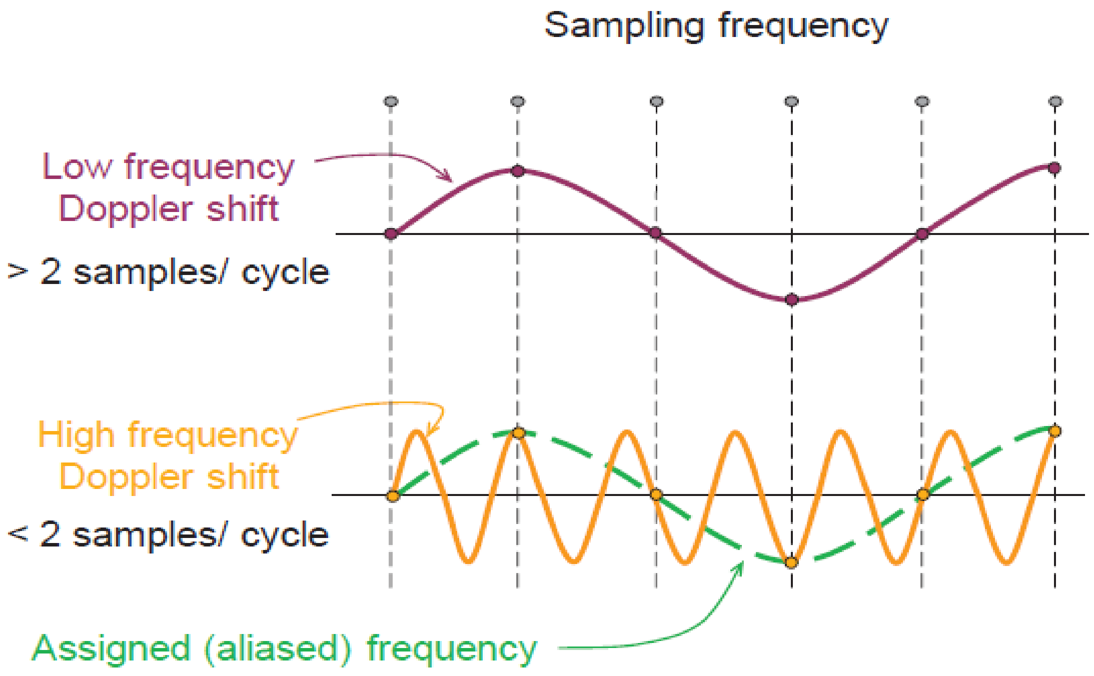


Fig. Aliasing occurs when the frequencies in the sampled signal are greater than ½ the PRF (sampling frequency). In this example, a signal of twice the frequency is analyzed as if it were the lower frequency and thus mimics (aliases) the lower frequency.

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Ultrasound contrast agent

- Goals:
  - To enhance the signal intensity of Doppler ultrasound;
  - To measure the blood perfusion in the heart and other organs;
- Microbubbles:
  - A diameter of 2-10μm with shells of a few tens of nm thick;
  - A resonance condition which corresponds to the ultrasound frequency at which the degree of expansion and contraction of the bubble is greatest;

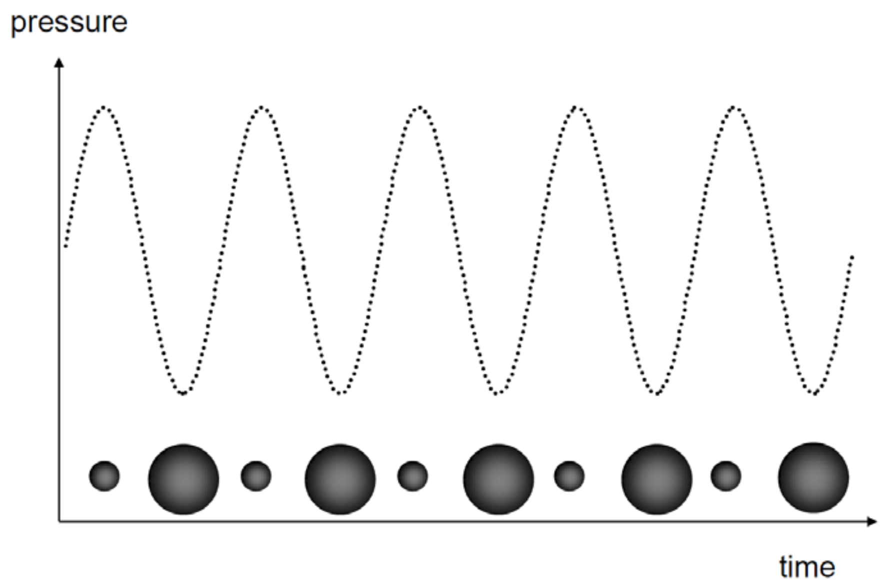

Fig. Change in the shape of a microbubble as an ultrasound pressure wave passes through the tissue in which the microbubble is located.

Harmonic and pulse inversion imaging

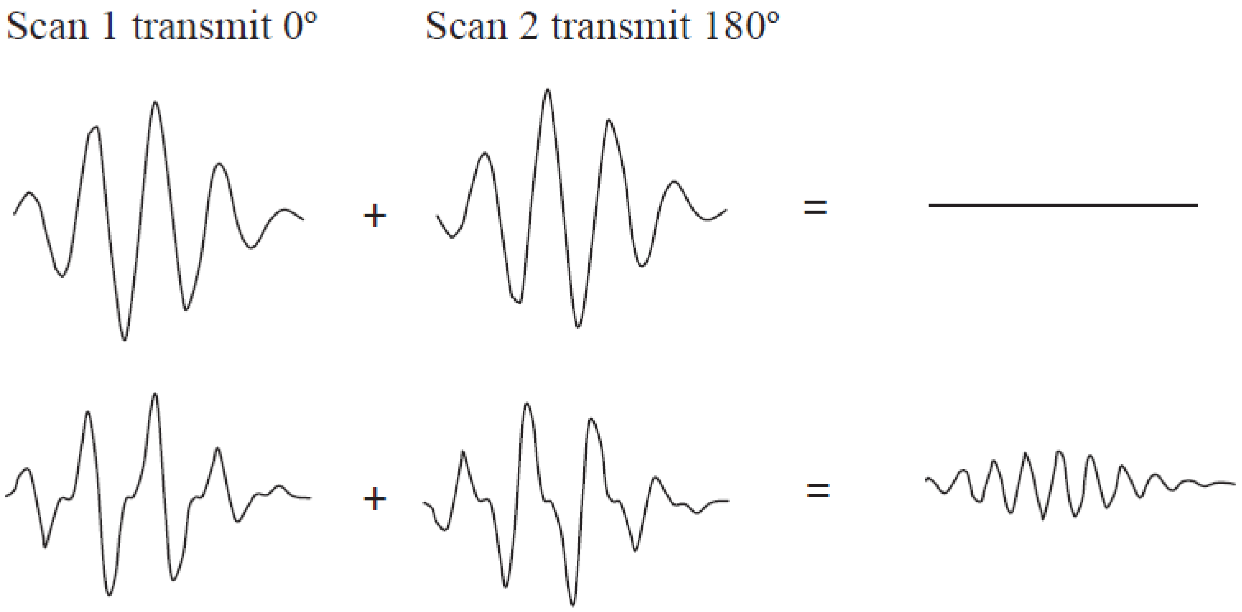

Fig. The principle of pulse inversion imaging. At the top, any signal which contains only components at f_0 is cancelled out by addition of the two scans. At the bottom, the signal contains components at both f_0 and 2 f_0: in the summed signal, the component at $2f_0$ remains.

- Harmonic frequency
  - Amplitude peaks at $2f_0$, $3f_0$, $4f_0$ ⋯ for Doppler ultrasound;
  - Nonlinear phenomenon;
  - High intensity ultrasound pressure on microbubbles;
- Pulse inversion
  - Signal cancellation between two transmitted pulses with 180° phase difference;
  - Nonlinear scattering signal do not completely cancel out.