Lecture 18 – Ultrasound Application

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

– 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)

Doppler effect

- The change in frequency or wavelength of a wave in relation to an observer who is moving relative to the wave source:
- If the source and receiver motion are aligned

$$f' = \frac{c \pm v}{c \mp u} f$$

Where f: the frequency of source

c: velocity of wave

u: the velocity of source

v: the velocity of receiver

- If the source and receiver motion are angled

$$f' = \frac{c \pm v\cos\beta}{c \mp u\cos\theta} f$$

Where β: the angle between the receiver motion and source-receiver line

θ: the angle between the source motion and source-receiver line

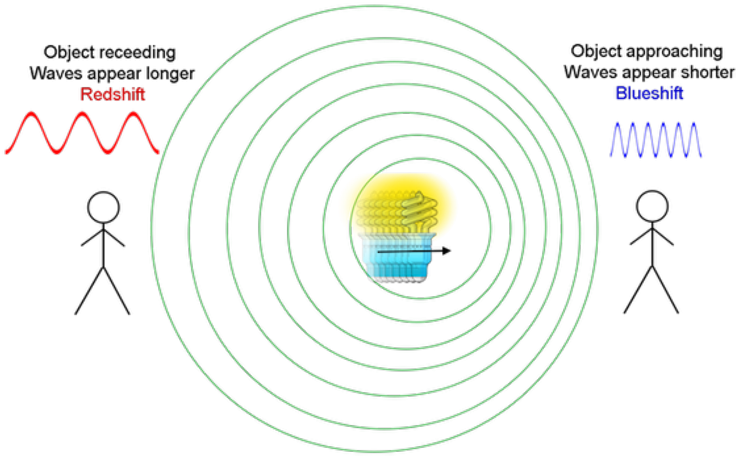


Fig. Doppler effect .

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

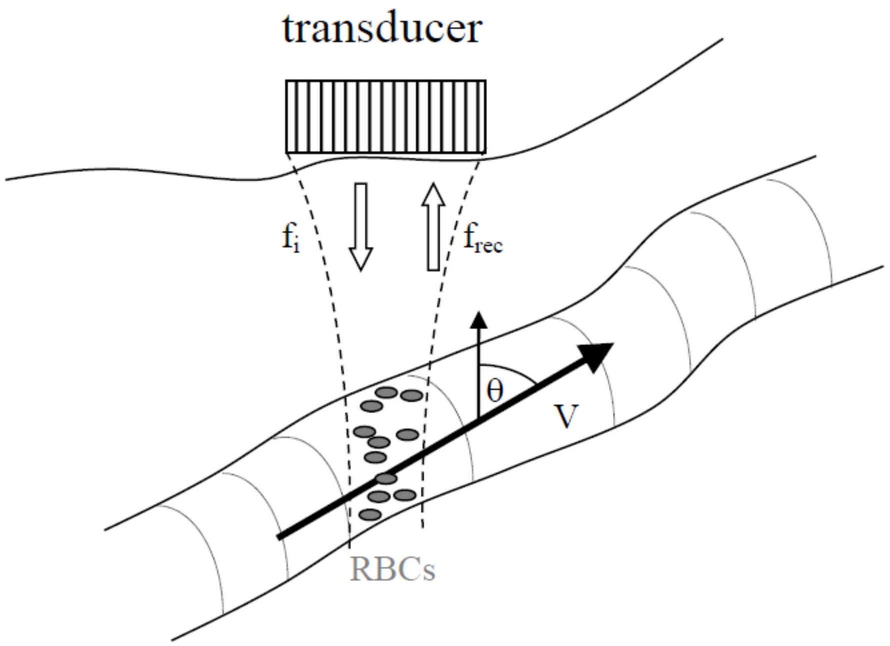


Fig. Showing the origin of the Doppler shift in ultrasound imaging of blood flow. The ultrasound beam is scattered from the RBCs in a vessel. The backscattered ultrasound beam is detected by the transducer at a slightly different frequency (f_{rec}) from that transmitted (f_i) into the body.

Pulse wave (PW) Doppler

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

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

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

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

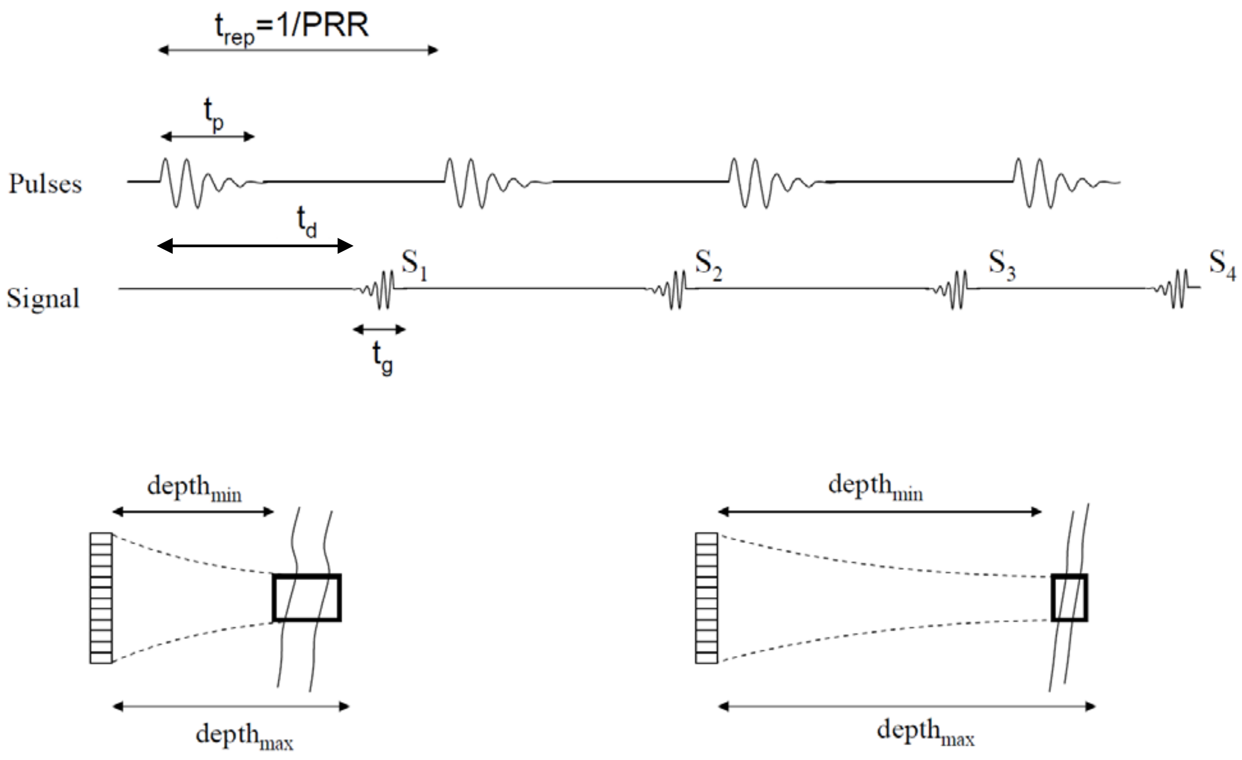


Fig. (top) General mode of operation of pulsed mode Doppler imaging. A series of signals S_1, S_2,. . . . .S_n are acquired to estimate the blood flow. (bottom) The parameters t_p, t_g and t_d are chosen to localize the received signal to the desired ROI, defined by the focal point of the phased array 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

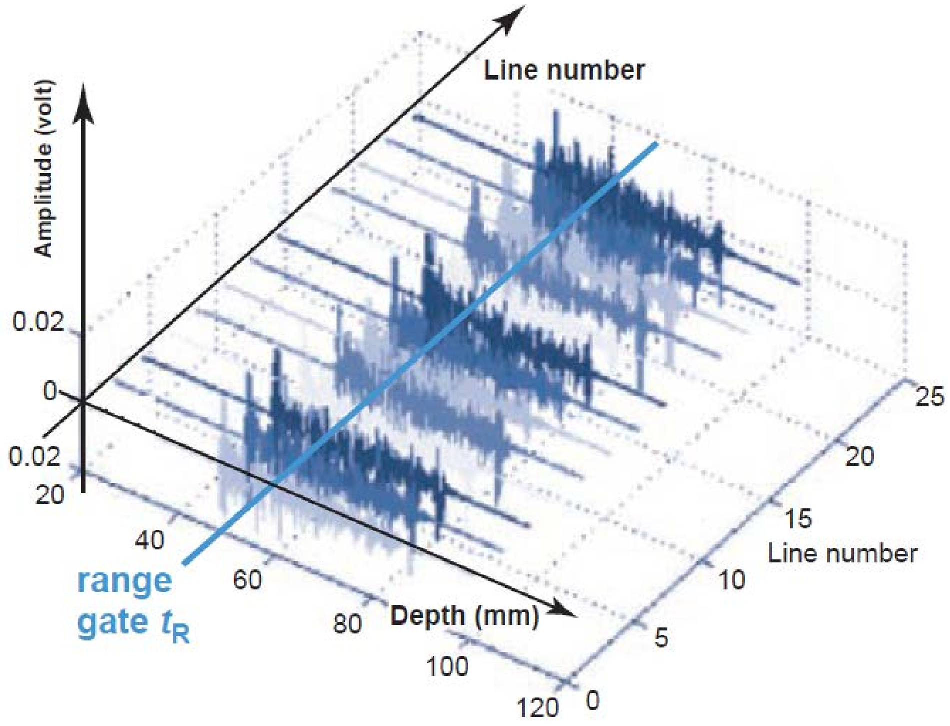


Fig. Pulsed wave Doppler uses the M-mode acquisition scheme and samples the subsequent reflected pulses at a fixed range gate t_R to calculate the Doppler frequency f_D.

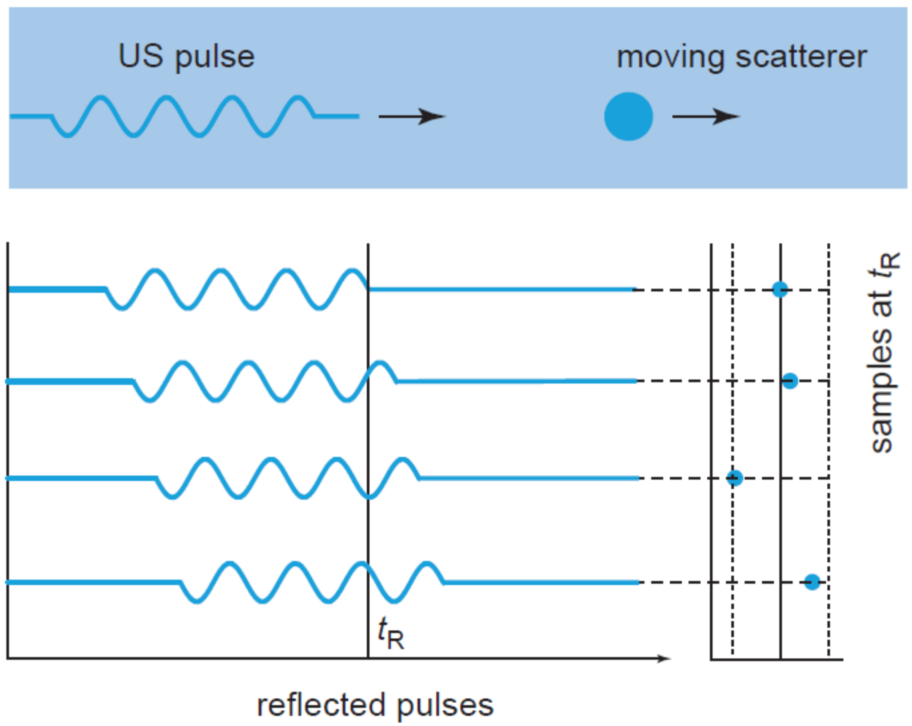


Fig. Schematic representation of the PW Doppler principle. Pulses are transmitted at a fixed pulse repetition frequency (PRF). They are reflected at a scatterer in motion. Because of this motion, the reflected pulses are dephased. Measuring each reflected pulse at the range gate t_R yields a sampled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

Phase shift

➢ The phase shift between two subsequent received pulses:

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

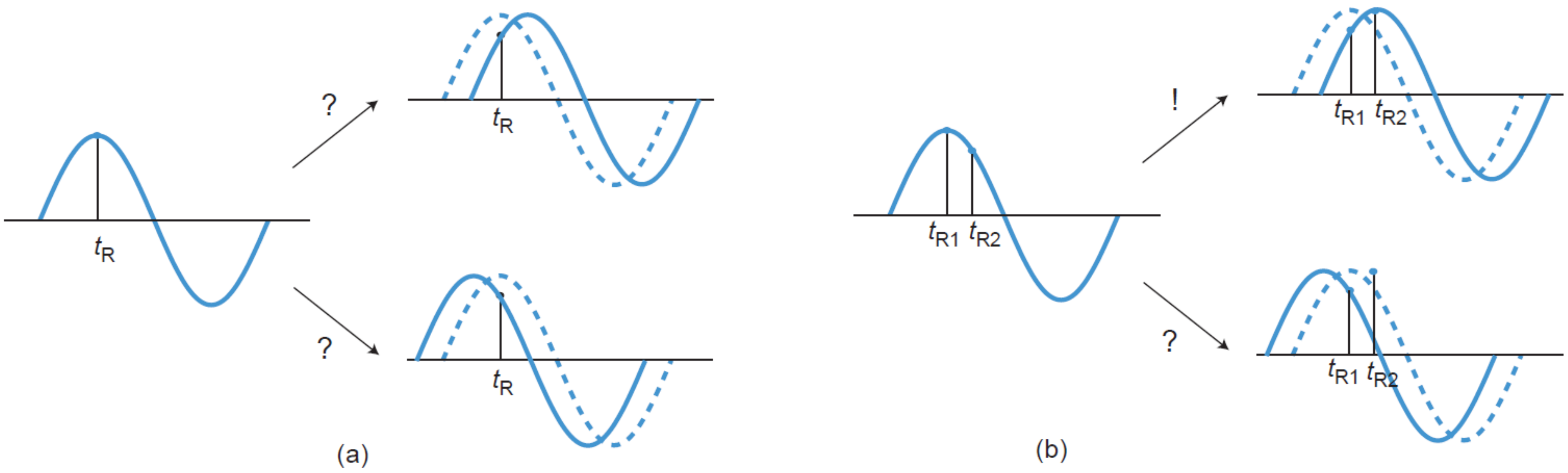
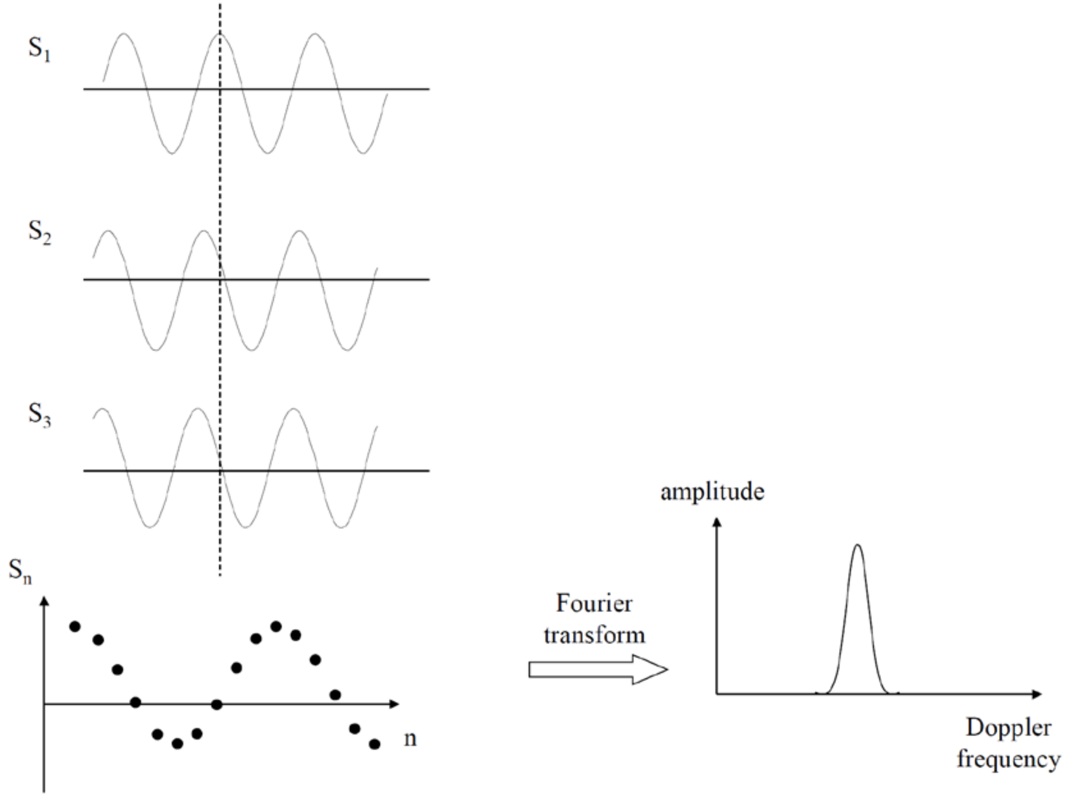


Fig. (a) If a single sample is acquired at the range gate, no directional information is obtained. (b) However, if a second sample is acquired slightly after the first one, the direction of motion is uniquely determined since a unique couple of samples within the cycle is obtained.

Spectral Doppler



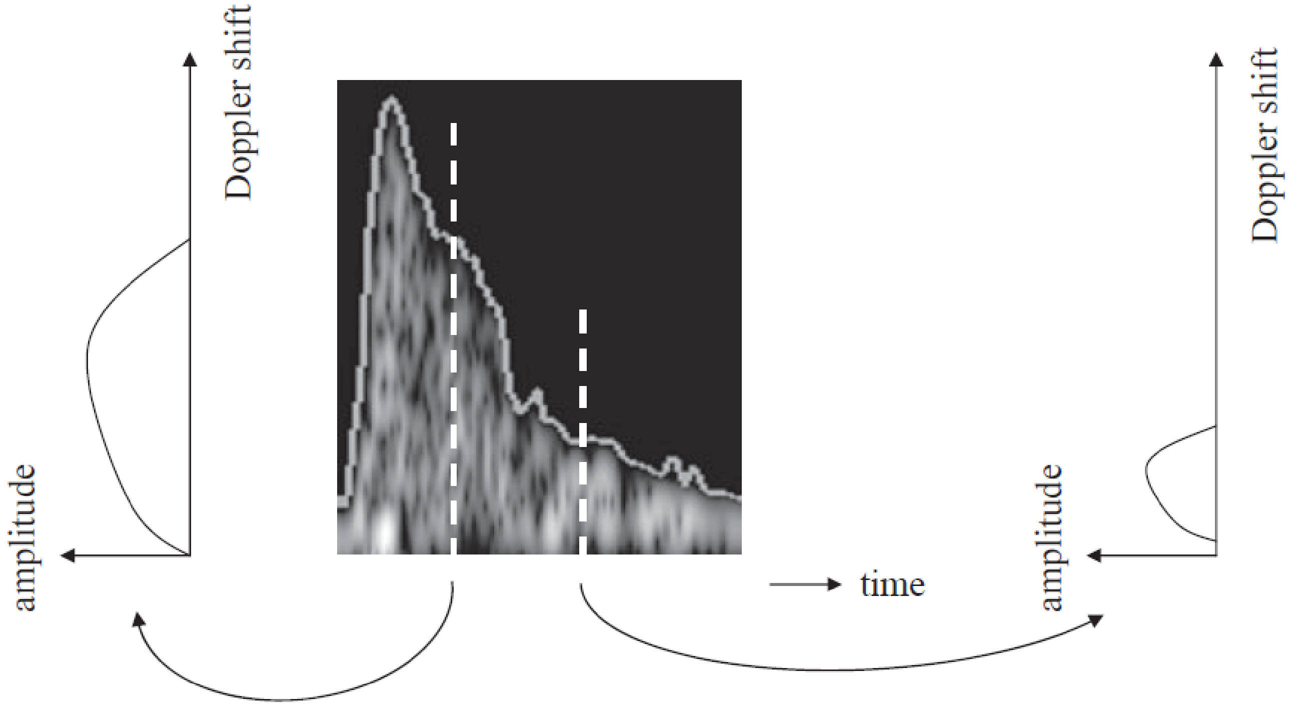
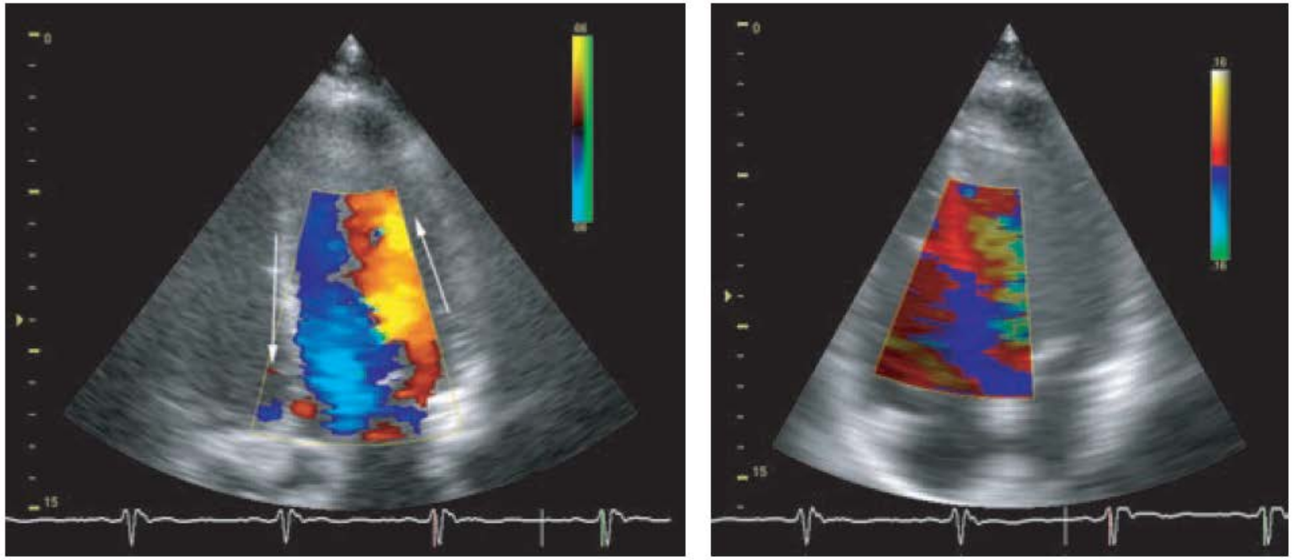


Fig. Steps in the production of a Doppler frequency distribution from one particular axial position within the ROI, determined by the particular time-point in signals S1. . .Sn that is analyzed. In this example a timepoint is chosen at the dashed line. A plot of the signals (Sn), taken at the dashed 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.

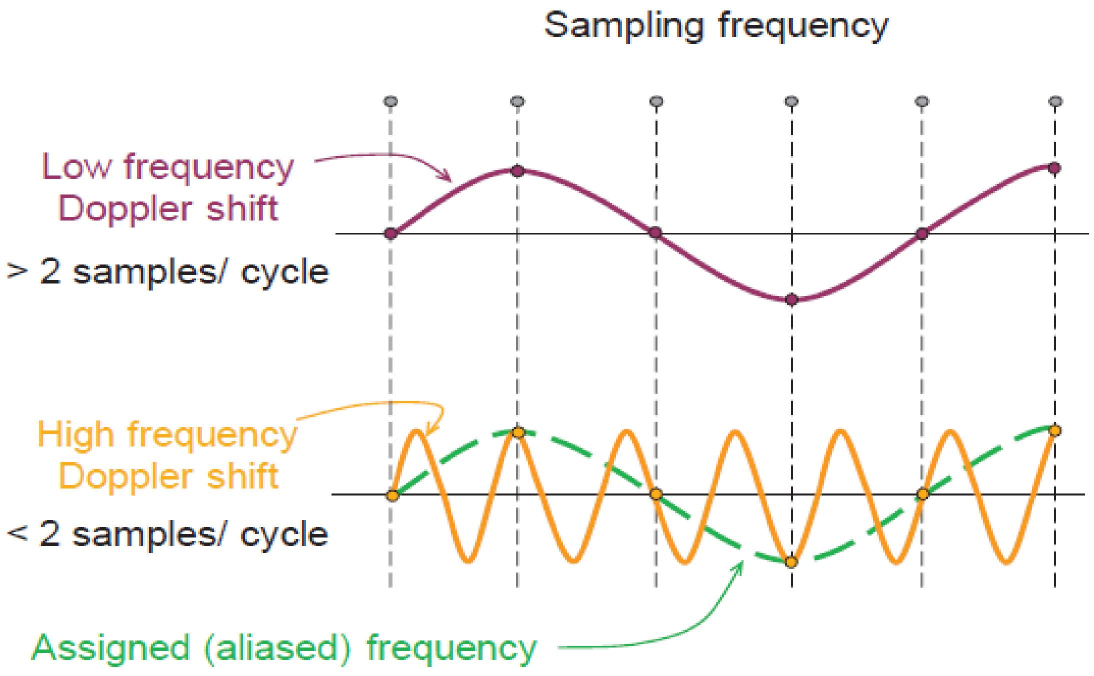


Fig. Aliasing occurs when the frequencies in the sampled signal are greater than ½ the PRF (sampling frequency). In this example, a signal of twice the frequency is analyzed as if it were the lower frequency and thus mimics (aliases) the lower frequency.

Lecture 18 – Ultrasound Application

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– Doppler Ultrasound

– **Ultrasound contrast agent**

– **Harmonic and pulse inversion imaging**

– Application

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;

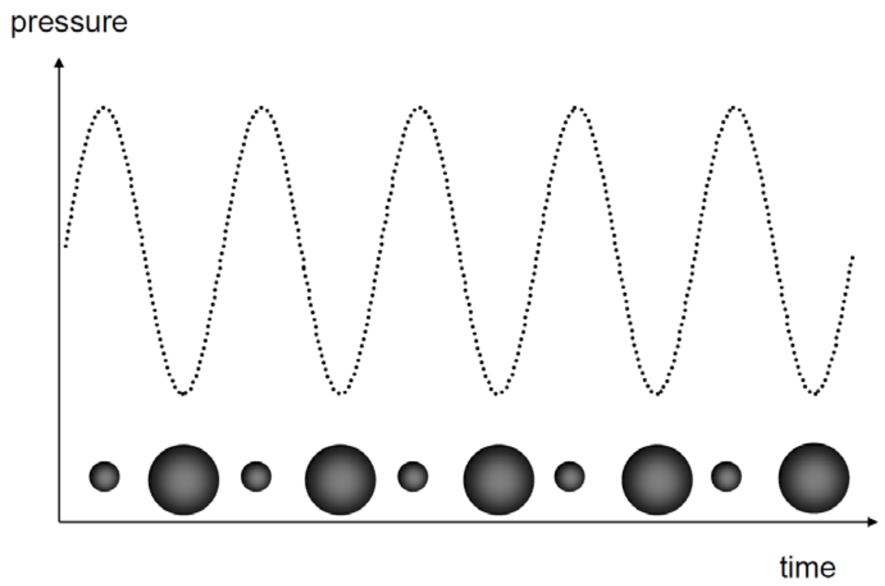

Fig. Change in the shape of a microbubble as an ultrasound pressure wave passes through the tissue in which the microbubble is located.

Harmonic and pulse inversion imaging

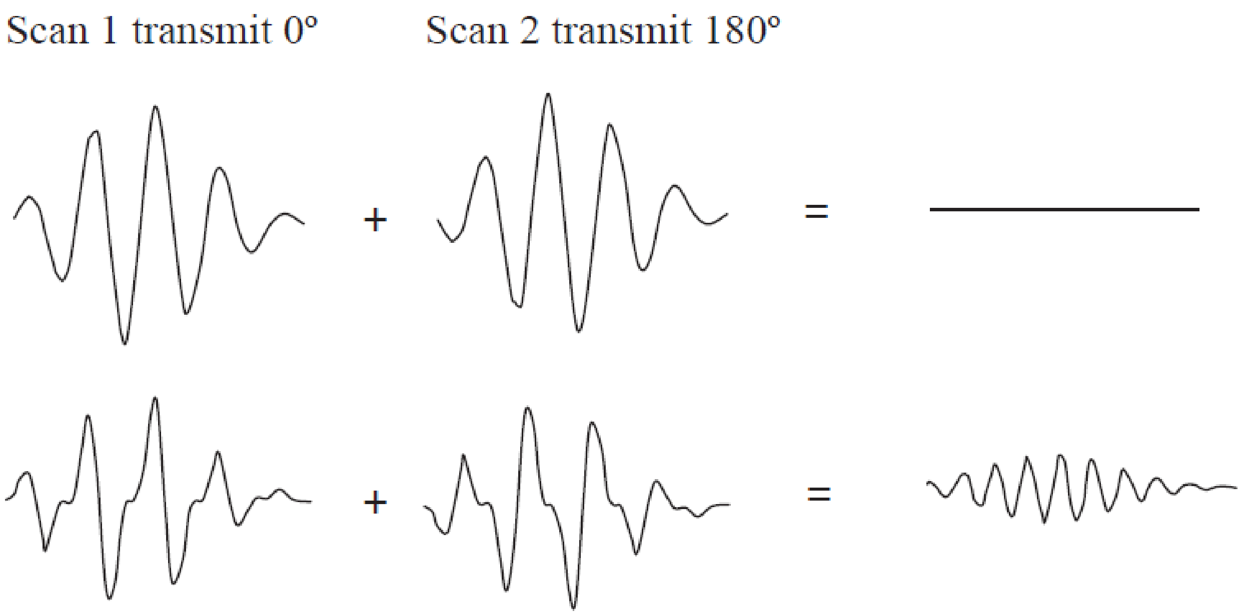

Fig. The principle of pulse inversion imaging. At the top, any signal which contains only components at f_0 is cancelled out by addition of the two scans. At the bottom, the signal contains components at both f_0 and $2\ f_0$: in the summed signal, the component at $2f_0$ remains.

- 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.

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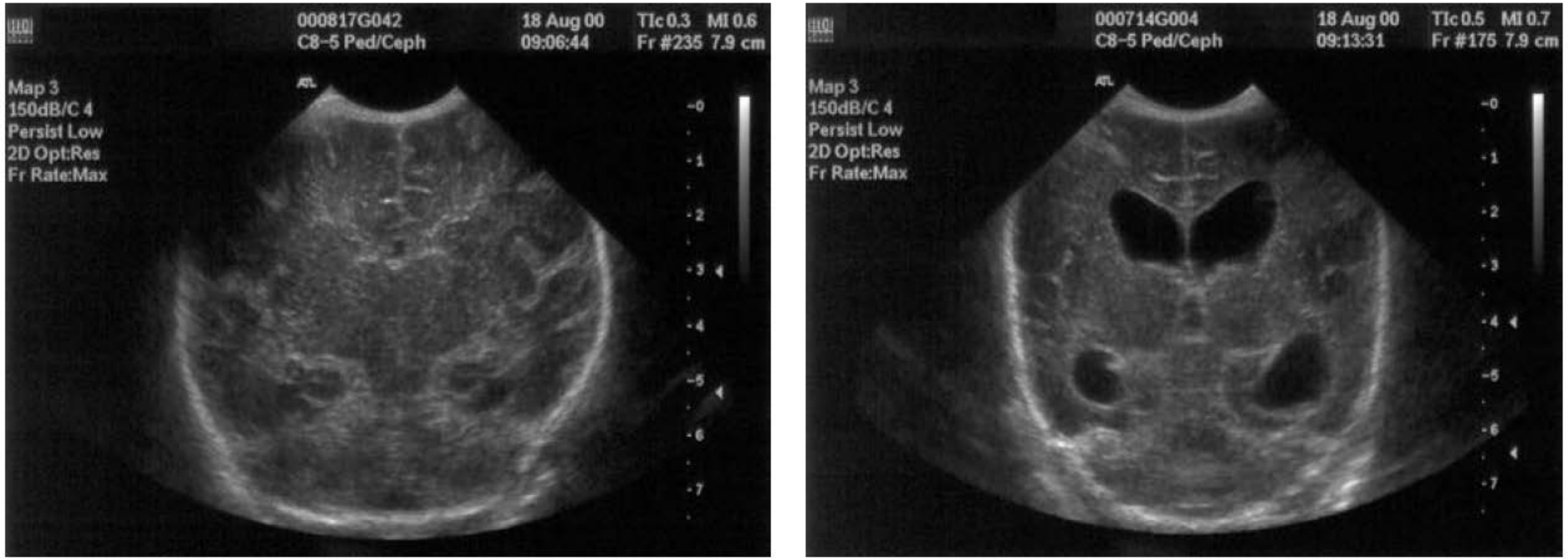
Structural Imaging

➢ Head



(a) (b)

Fig. (a) Normal cranial ultrasound. (b) Fluid-filled cerebral cavities on both sides as a result of an intraventricular hemorrhage.

Structural Imaging

- Neck

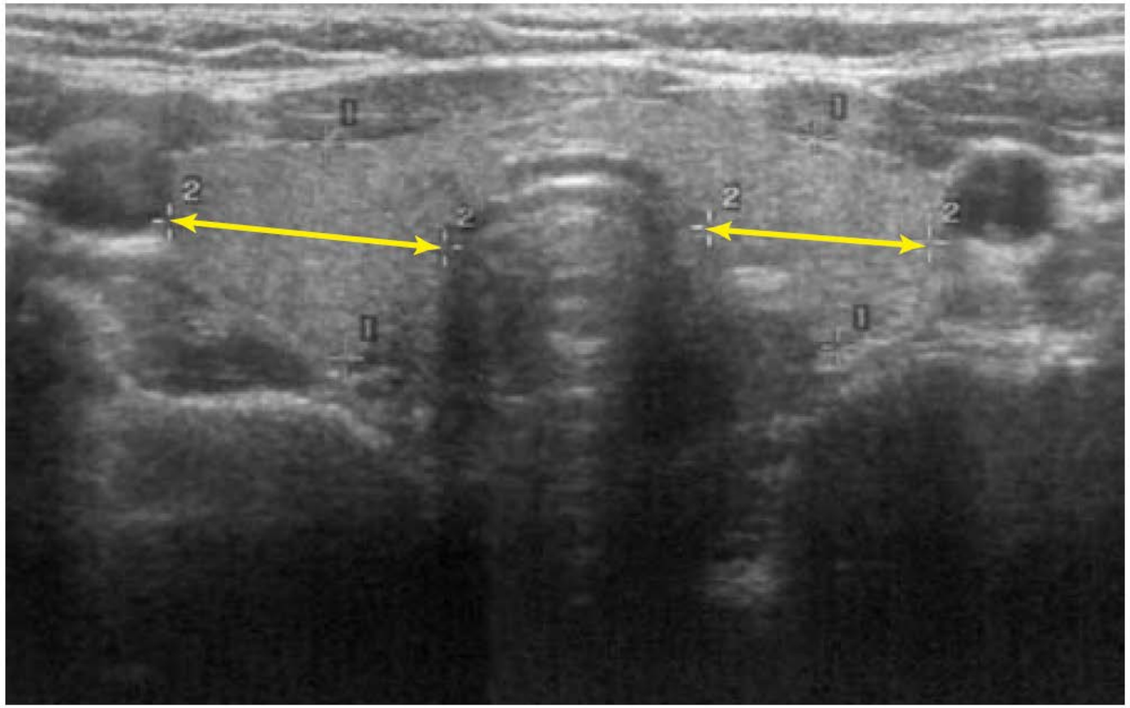


Fig. Ultrasound image of the thyroid showing a mild bilateral enlargement (arrows) suggesting an inflammatory disease or hormonal inbalance.

Structural Imaging

➢ Abdomen

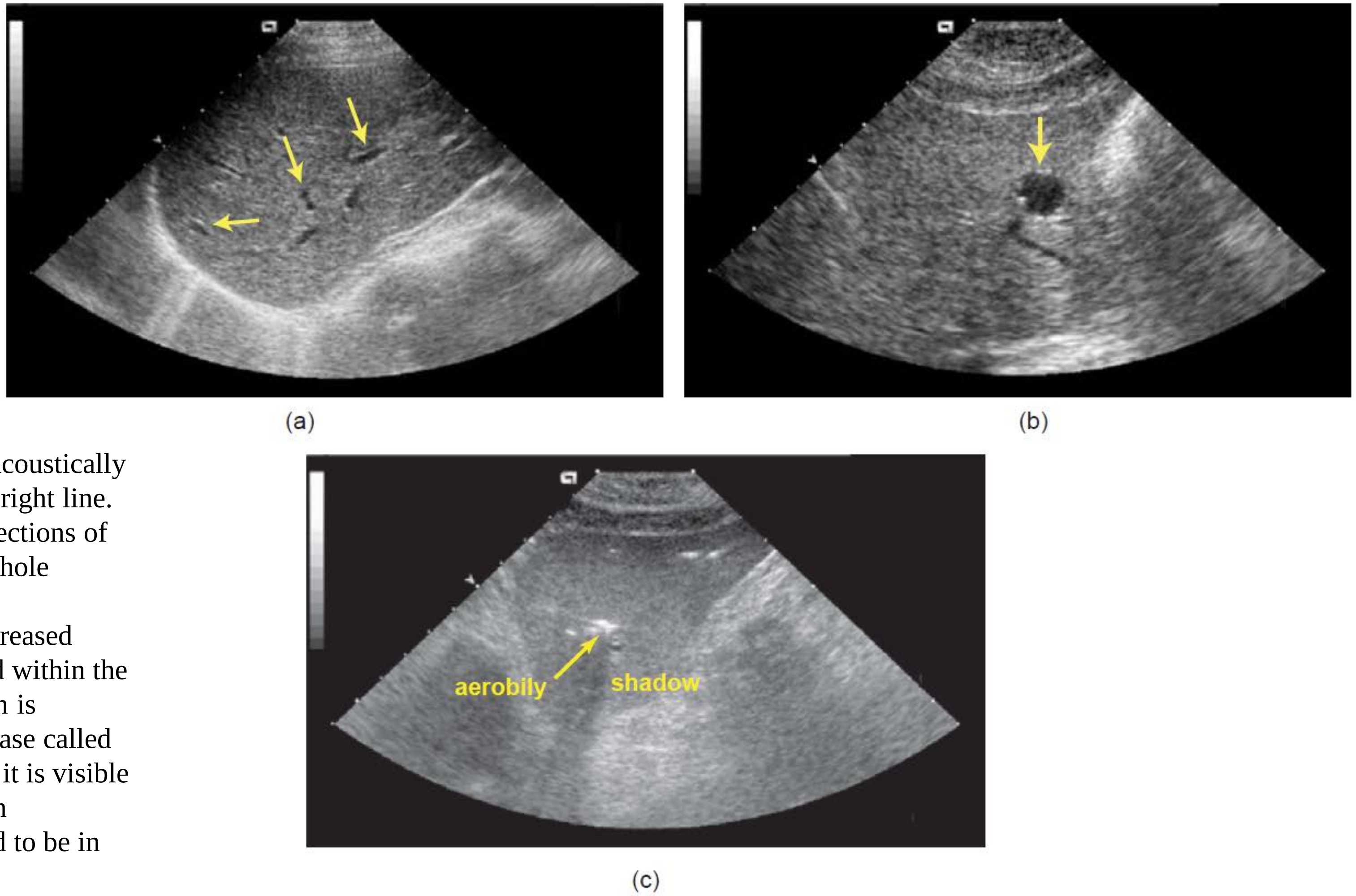


Fig. (a) B-mode image of a normal liver. The liver is an acoustically homogeneous, large organ. The boundary is visible as a bright line. The little black holes (arrows) inside the liver are cross-sections of blood vessels. (b) Liver with cyst visible as a large black hole (arrow). Note the so-called acoustic retroamplification, a hyperechoic region behind the cyst. The origin of this increased reflectivity is the reduced acoustic attenuation in the fluid within the cyst. (c) The opposite effect of acoustic retroamplification is observed when air is present in the bile ducts, as in a disease called aerobily. Because air is a perfect reflector for ultrasound, it is visible as a very bright reflection. Because of this extremely high attenuation, deeper regions cannot be imaged and are said to be in the ''acoustic shadow'' of the air.

Contrast echography

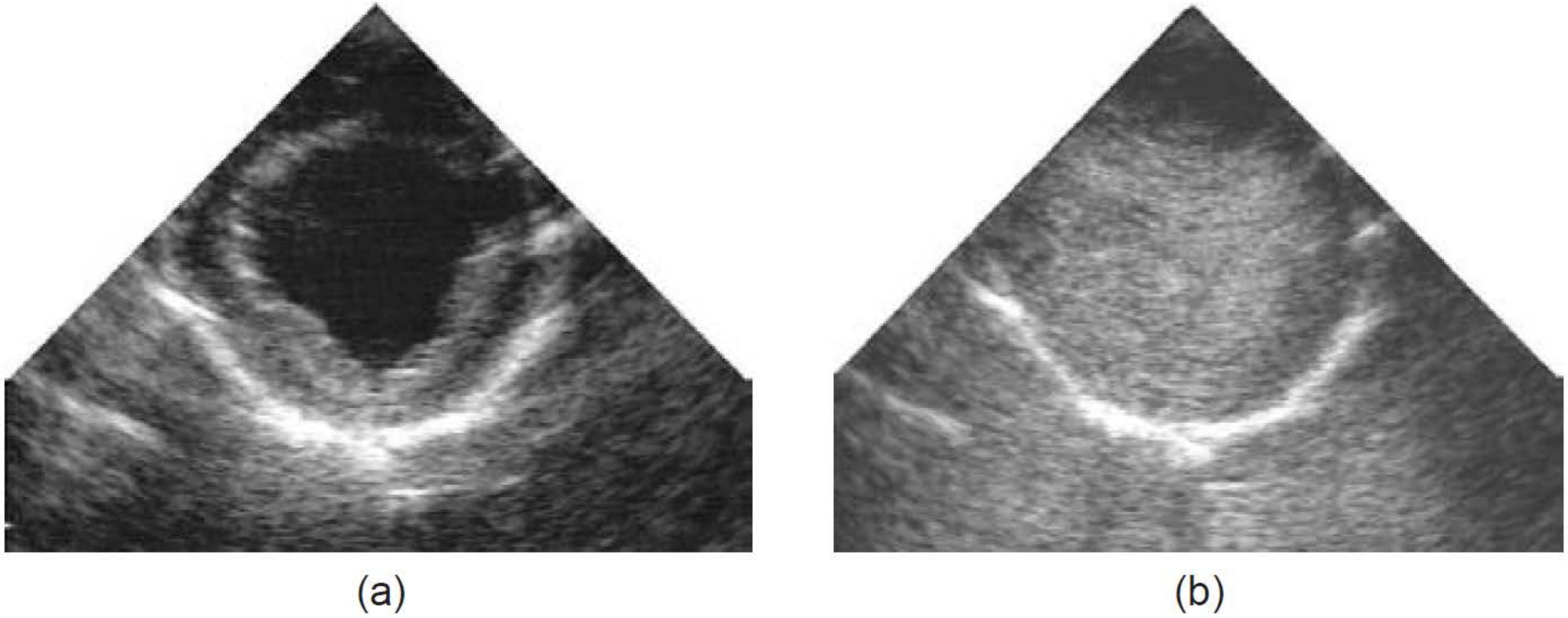


Fig. B-mode gray scale image of the left ventricle of the heart in a short-axis view before (a) and during (b) pervenous injection of a contrast agent.

Thermoacoustic imaging

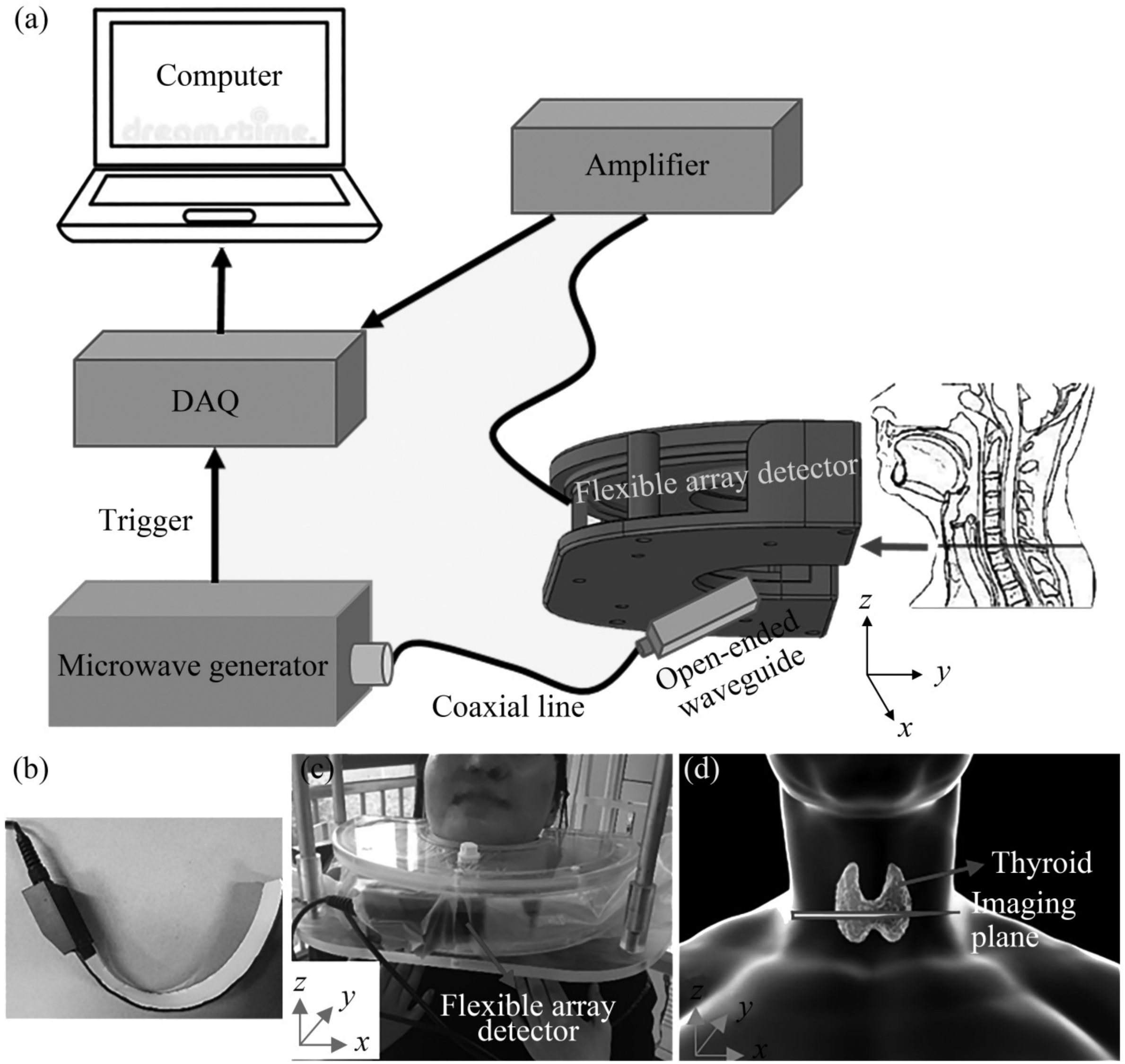
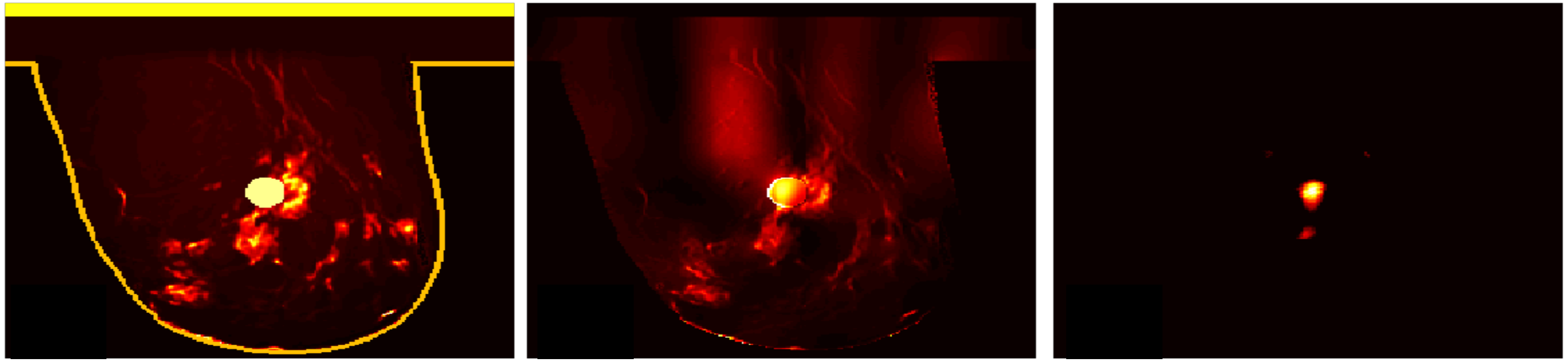
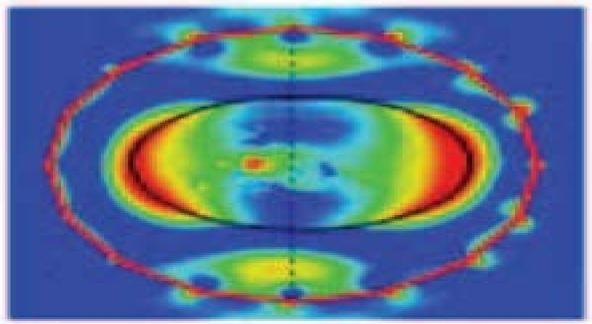


Fig. Schematic diagrams of microwave-induced thermoacoustic imaging system for human thyroid: (a) Schematic diagram of the microwave-induced thermoacoustic imaging experimental system for human thyroid.(b)Photograph of the flexible array detector.(c)Photograph of the customized thyroid stent experimental site.(d)Microwave-induced thermoacoustic ultrasound imaging plane schematic.

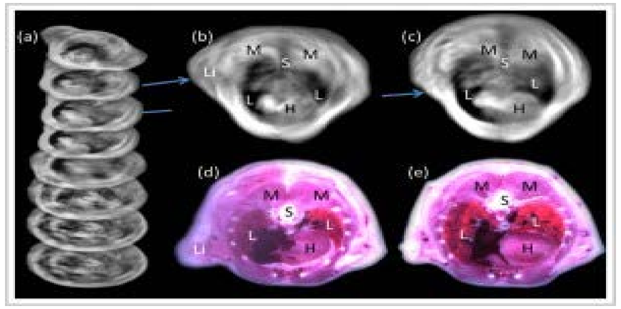
Thermoacoustic imaging



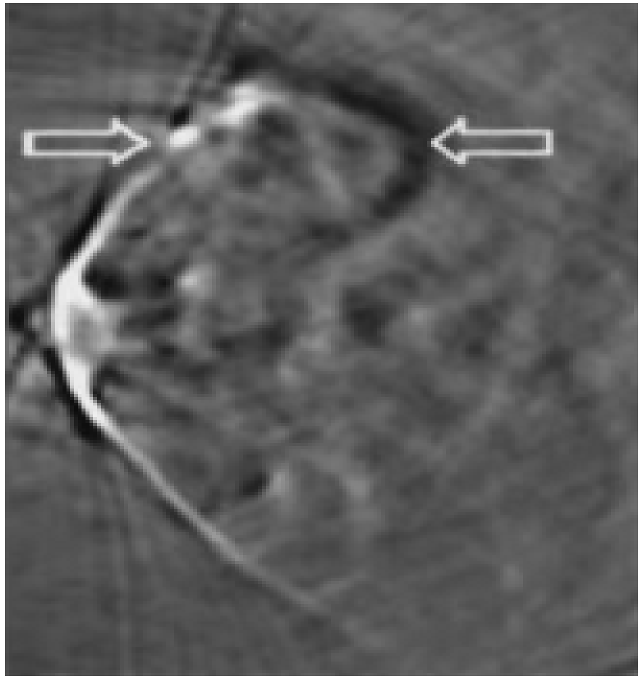
Focused Microwave Breast Hyperthermia



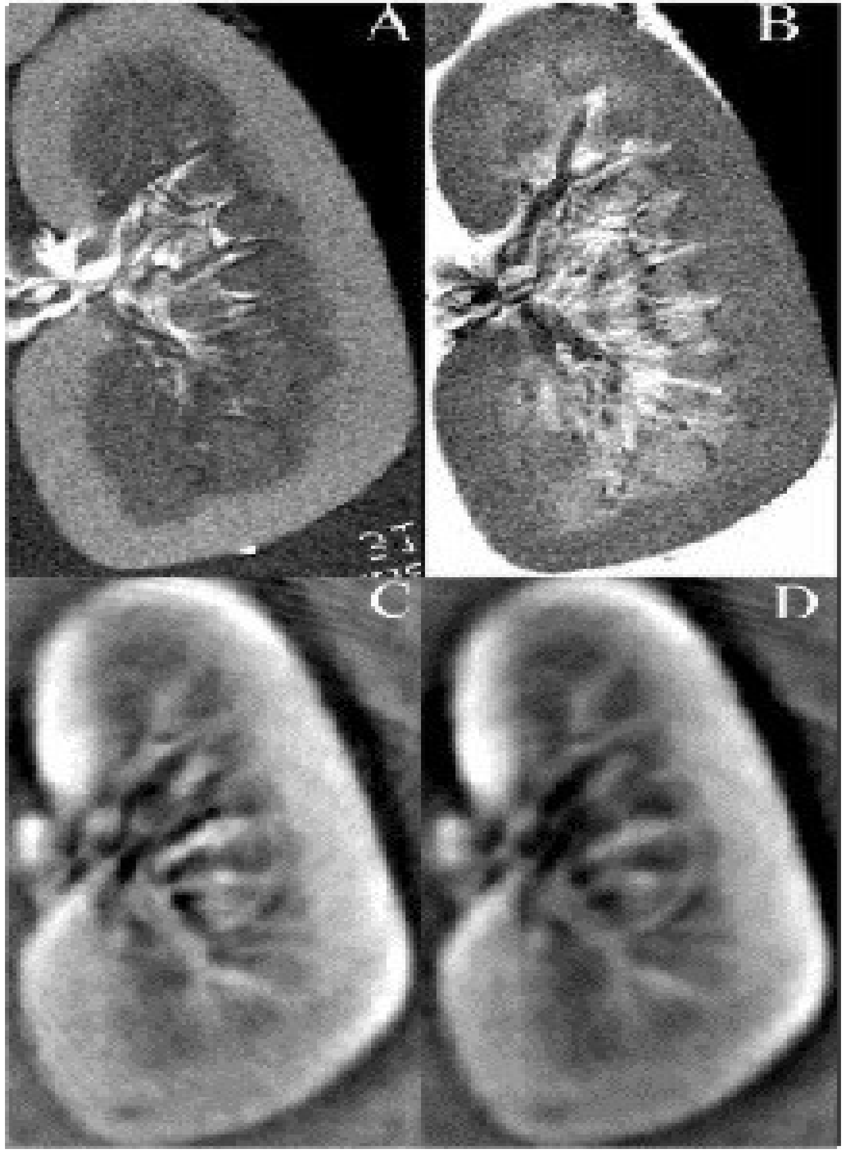
Temperature Monitoring



Skin Cancer

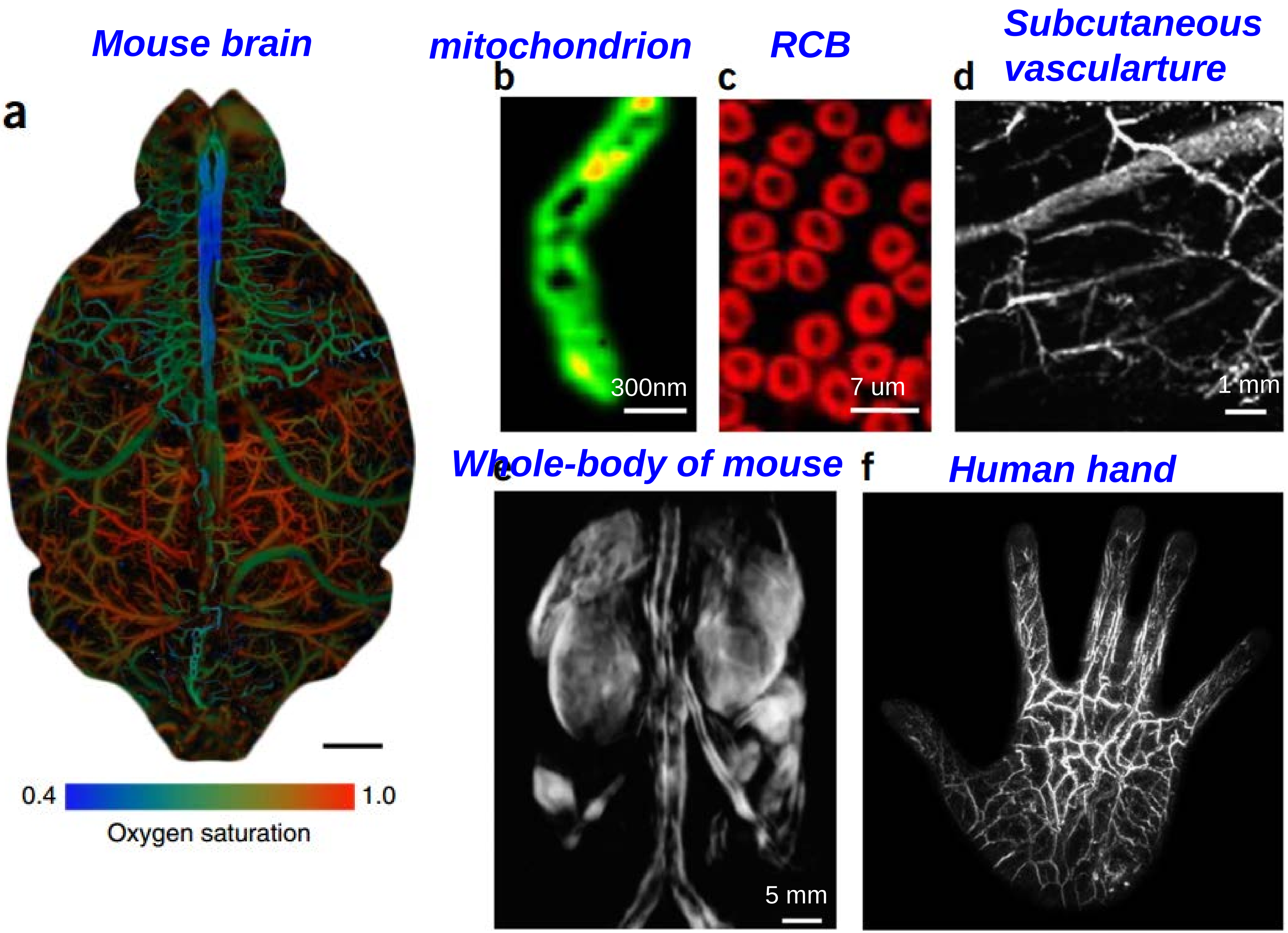


Human Breast



Lamb Kidney

PAI - Small-animal imaging

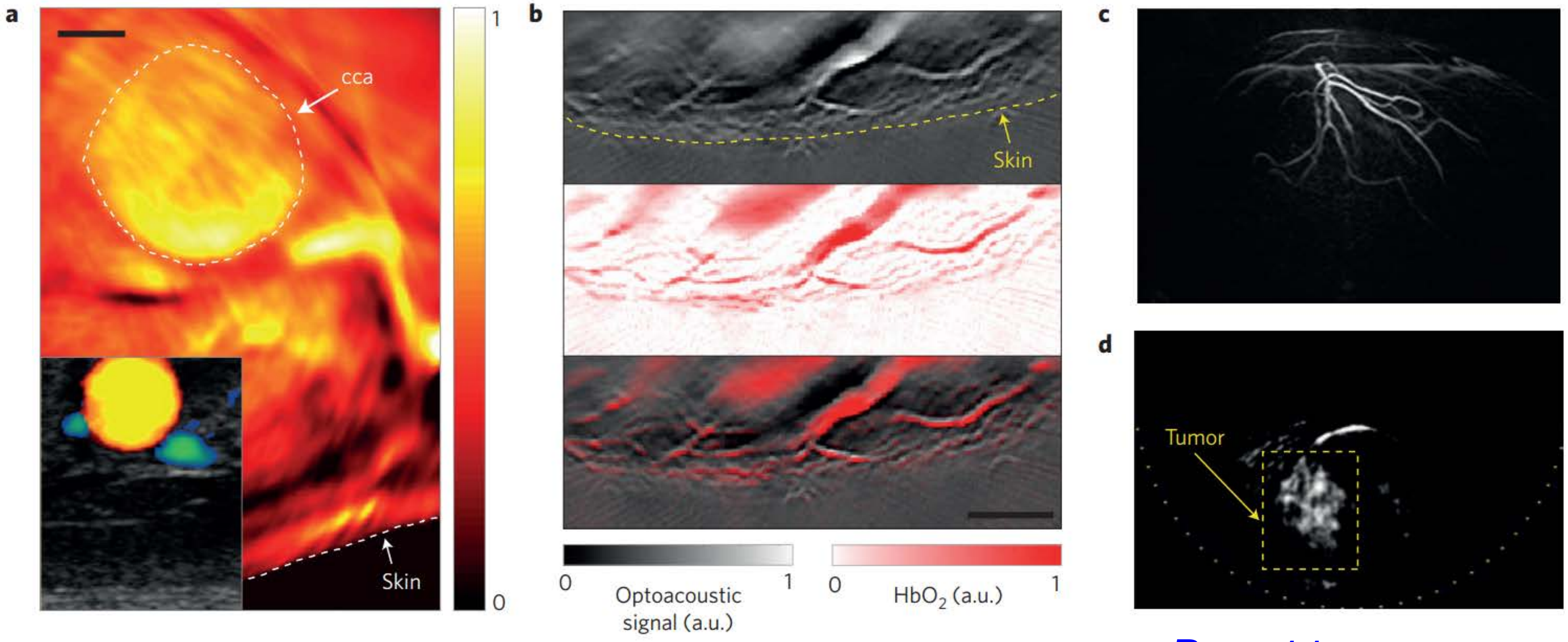


PAI - Human trials

Carotid artery of human

oxyhemoglobin in the human hallux

Blood vessels in the breast of a healthy volunteer



Breast tumour (ductal carcinoma)