Lecture 23 – MRI Instrumentation

This lecture will cover: *(CH5.14-5.16, 5.20)*

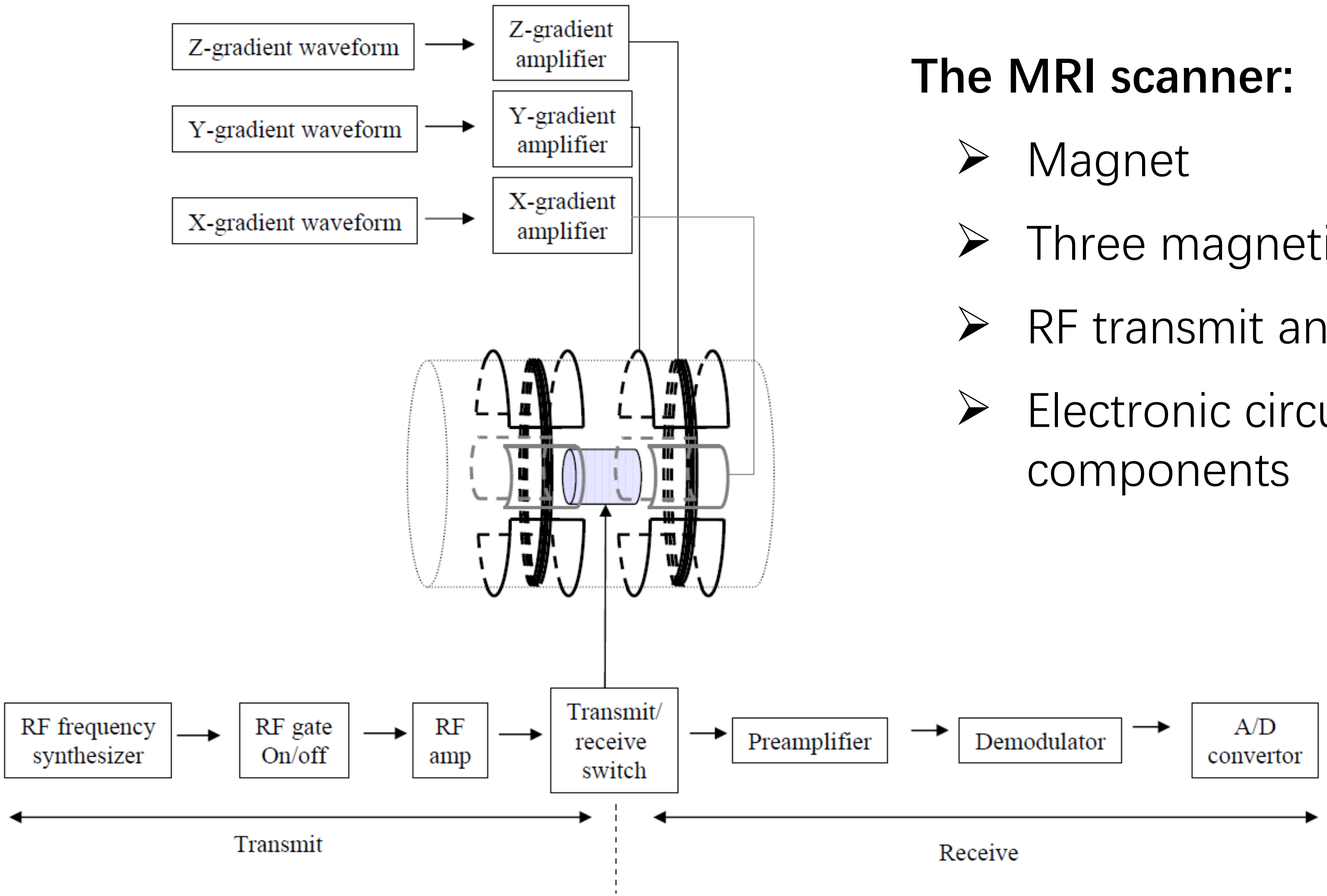
– MRI Instrumentation
- Superconducting magnet
- Magnetic field gradient coils
- Radiofrequency coils

– Fast MRI imaging
- Parallel imaging
- Echo planar imaging

– Imaging characteristics

(*Supplementary reading: The Essential Physics of Medical Imaging CH13.2*)

MRI instrumentation



The MRI scanner:

- Magnet
- Three magnetic field gradient coils
- RF transmit and receive coil
- Electronic circuits to control the 3 components

Fig. MRI system hardware components used to control the gradients and RF transmitter and receiver. Each gradient has a separate waveform and amplifier. The transmit and receive sides of the RF chain are separated by a transmit/receive switch.

Superconducting magnet design

- Aims of magnet design:
  - To produce the most homogeneous magnetic field over the sample;
  - To produce a stable magnetic field (1 ppb) over the course of an MRI scan.
- The magnetic field in a series of solenoid coil

$$B = \frac{\mu_0 n I}{\sqrt{L^2 + 4R^2}}$$

Where μ_0: the permeability of free space ($1.257 * 10^{-6}$ TmA^{-1})

n: the number of turns

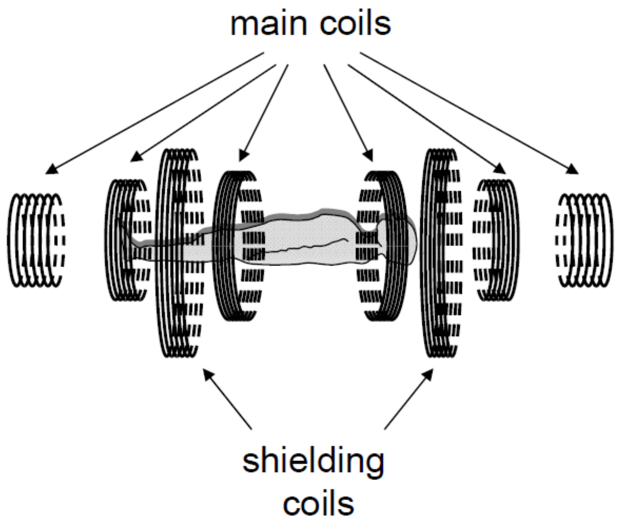
I: the current passing through the wires

L: the length of the solenoid

R: the radius of the solenoid

Superconducting magnet design

- The most commonly used superconducting material is the alloy of NbTi.
- The wires are wound in aluminum former and fixed with epoxy adhesive;
- The windings are house in the cryostat (低温恒温器) filled with liquid helium;
- The current can circulate indefinitely after "energizing" the magnet;



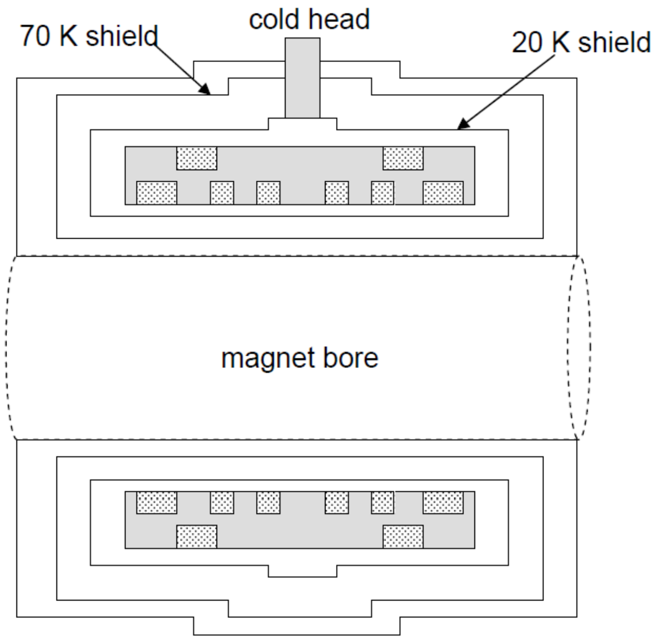


Fig. (left) The solenoidal coils used to produce a homogeneous static magnetic field. Six solenoids (main coils) are positioned along the z-axis. Two shielding coils are used to reduce the effect of the 'stray field' outside the magnet. (right) Cutaway of a superconducting magnet. The grey tinted areas represent those filled with liquid helium to make the wires superconducting. Two aluminium radiation shields are kept at 20 K and 70 K.

Magnet


ANN
NEWS
提供映像
MRI検査 ボンベ挟まれ死亡
知っておきたい! NEWS


中新视频
MRI 촬영실
Magnetic resonance imaging

Magnetic field gradient coils

- The gradient coils are constructed of copper, water-cooled and wound on cylindrical formers bolted to the inner bore of the magnet.
- The current in gradient coils can be switched on and off in less than 1ms;
- The noise is caused by the vibrating of the whole cylinder.
- The aim in gradient coil design are:
  - To produce the maximum gradient per unit current;
  - To minimize the “rise” time of the gradient;
  - To achieve the maximum volume of gradient linearity (95%);

Magnetic field gradient coils

- z gradient coil: Maxwell-pair coil
  - Tow loops in opposite direction
  - Spaced by $\sqrt{3}$ times of radius a
  - The gradient efficiency

$$\eta = \frac{8.058 \times 10^{-7} nI}{a^2} \ \mathrm{T/m}$$

- x- and y- gradient coil: saddle coil
  - 4 arcs with each subtending an angle of 120°;
  - Spaced in z-direction by 0.8 times the radius;
  - The length of arc is 2.57 times the radius;
  - The gradient efficiency

$$\eta = \frac{9.18 \times 10^{-7} nI}{a^2} \ \mathrm{T/m}$$

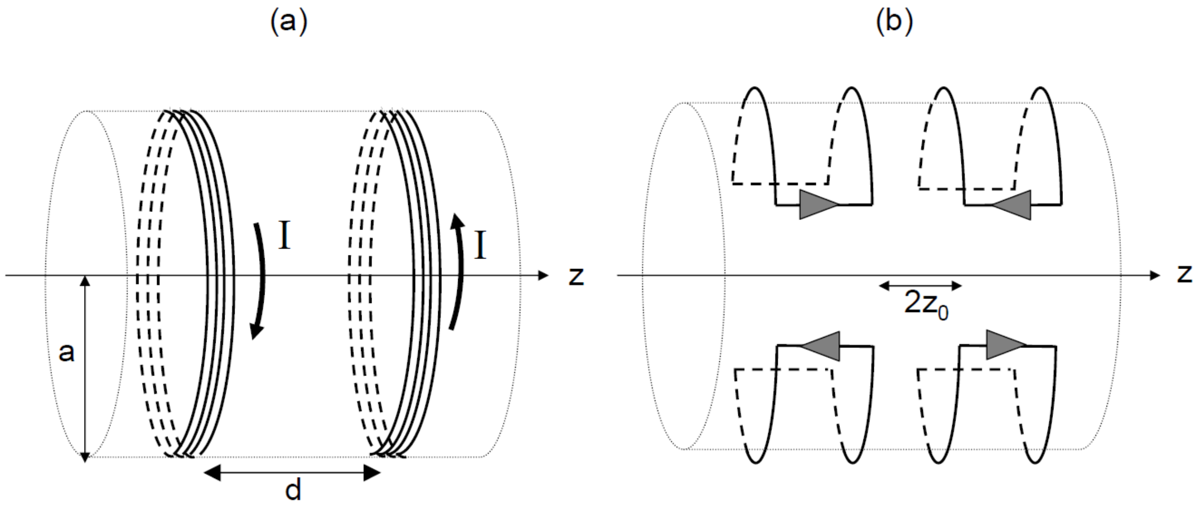


Fig. (a) Basic geometry of a Maxwell-pair gradient coil which produces a magnetic field gradient in the z-direction. The two halves consist of equal numbers of turns of wire, with equal currents in each half but flowing in opposite directions. The magnetic fields from the two halves cancel in the center of the gradient set. In the configuration shown, the left-hand half produces a magnetic field opposite to B_0 (using Fleming's left-hand rule) and therefore the resonant frequency increases in the positive z-direction. (b) Basic saddle-geometry for producing a transverse gradient, in this case in the y-direction. The direction of current in the four identical components of the gradient coil is indicated by the arrows. The design for producing an x-gradient is the same, but rotated by 90°.

Figure-of-merit

- The three criteria (homogeneity, switch speed and efficiency) for judging gradient performance can be combined into figure-of-merit defined as:

$$\beta = \frac{\eta^2}{L\sqrt{\frac{1}{V}\int\left(\frac{B(r)}{B_0(r)} - 1\right)^2 d^3r}}$$

where L: the inductance of the coil

$B_0(r)$: the desired magnetic field

$B(r)$: the actual magnetic field

V: the volume of interest over which the integral is evaluated

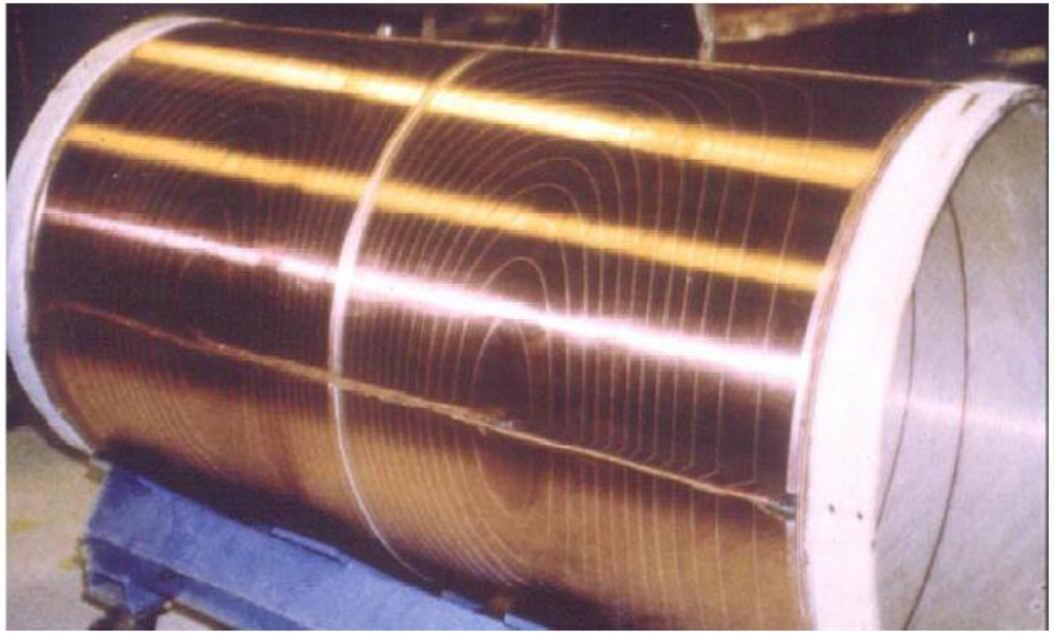


Fig. A fingerprint gradient set, used to produce a y-gradient. The design is usually milled from a solid piece of copper and mounted on to a fiber-glass cylindrical former, which is then securely bolted on to the inside of the magnet bore.

Radiofrequency coils

- The dual role of RF coil
  - To transmit RF pulse into the body
  - To detect the magnetization via induction
- The resonant frequency of LC circuit:

$$\omega_{\text{res}} \approx \frac{1}{\sqrt{LC}}$$

- The highest efficiency is achieved by matching the size of RF coil as closely as possible to the size of body part being imaged;

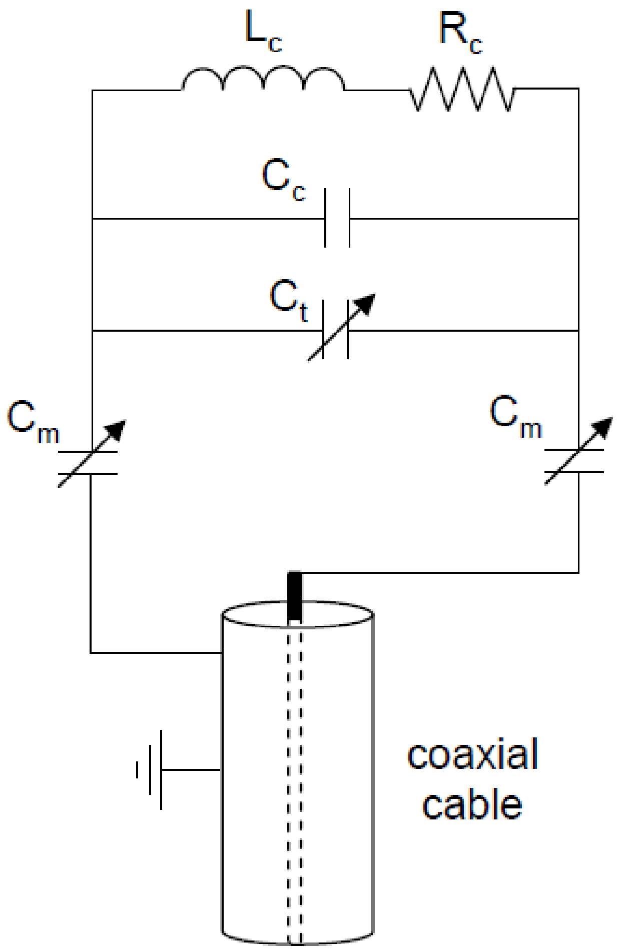
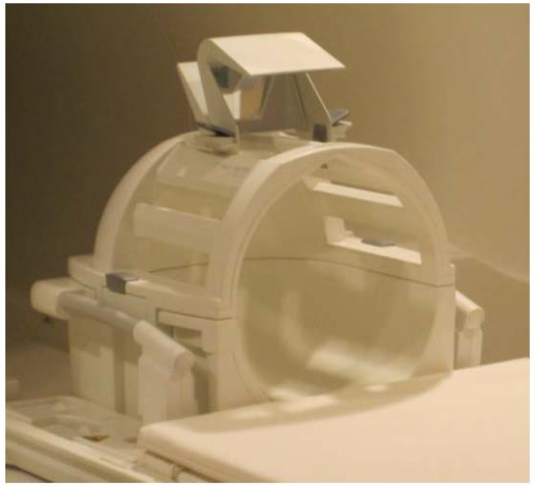


Fig. (a) Basic electrical circuit used to impedance match the RF coil to 50Ω at the Larmor frequency. The intrinsic inductance, capacitance and resistance of the RF coil are denoted by L_c, C_c and R_c, respectively. Three external capacitors (C_t and two equal value capacitors C_m) are connected to the RF coil to form the impedance matching network. The arrows denote that these can be variable capacitors whose value can be changed for each patient if required. The coil and impedance matching network are connected to the rest of the MRI system via a coaxial cable with 50Ω impedance.

RF coils for different body parts

(a)

(b)



(c)

Fig. (a) A commercial knee coil for a 3 Tesla magnet, formed from twelve rungs in a birdcage geometry. The structure has an upper and lower half to facilitate patient positioning. (b) Similar structure for a head coil, with a mirror attached so that the patient can see out of the end of the magnet, which is useful if visual stimuli are needed for functional MRI experiments. (c) Multi-element body array for 3 Tesla consisting of 24 separate elements. Each element is electrically decoupled from the other individual elements. The array is used to receive the MR signal, with a large body coil which is fixed inside the bore of the magnet used to transmit the RF pulses.

Lecture 23 – MRI Instrumentation

This lecture will cover:

- MRI Instrumentation
  - Superconducting magnet
  - Magnetic field gradient coils
  - Radiofrequency coils
- **Fast MRI imaging**
  - **Parallel imaging**
  - **Echo planar imaging**
- Imaging characteristics

Alias for fast imaging

- Aliasing due to the large phase encoding steps;

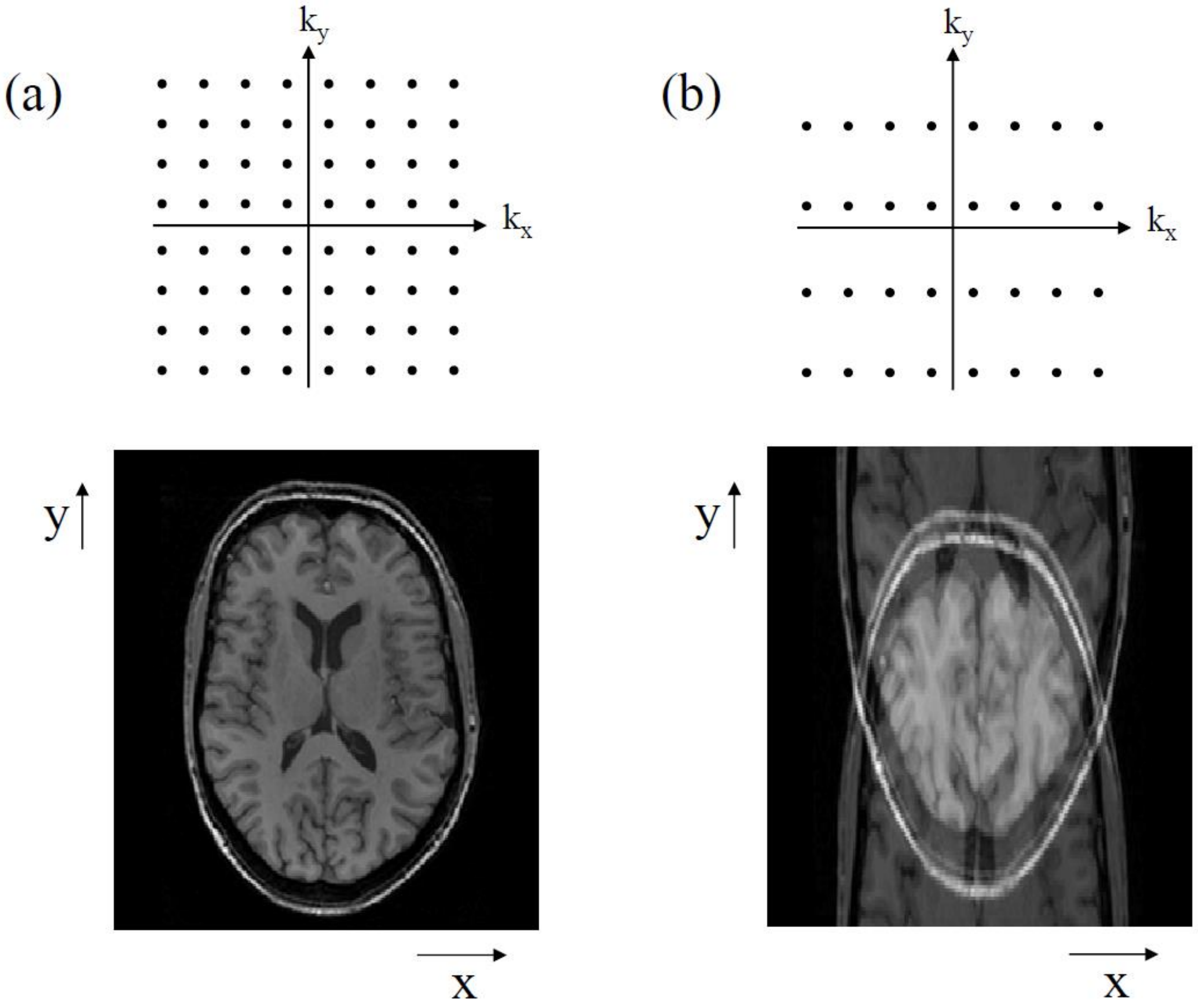


Fig. (a) A full k-space matrix (top) gives an unaliased image (bottom) provided that the Δk_x and Δk_y increments are correctly set. (b) If data acquisition speed is increased by a factor of 2 by acquiring only alternate k_y lines, then the Δk_y is doubled, the image field-of-view in the y-dimension is halved, and the image is aliased along the y-dimension.

Parallel imaging with coil arrays

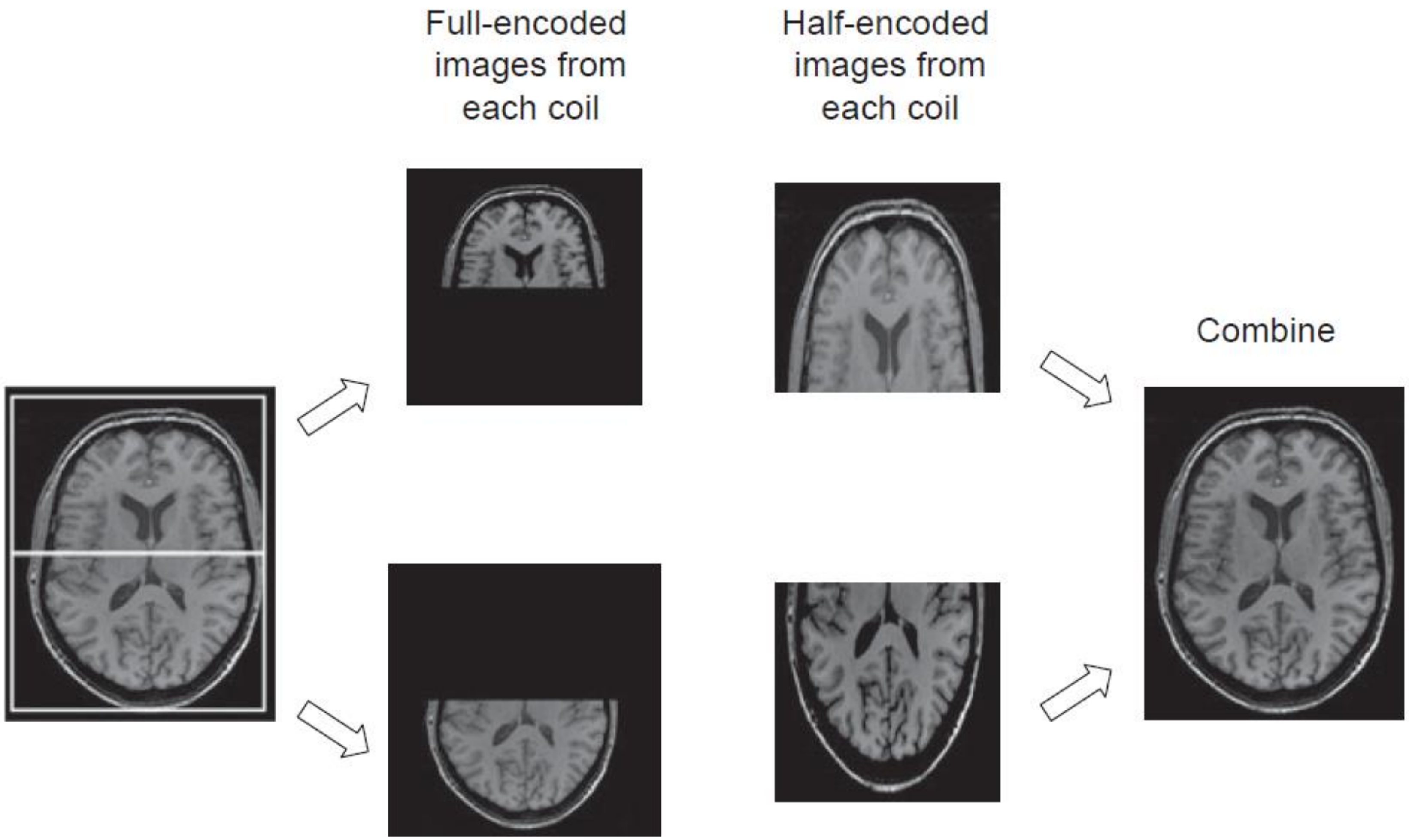


Fig. General principle of parallel imaging. Two RF coils, shown as white rectangles on the left, are used to acquire the MRI signal. If a full field-of-view image is acquired using full k-space encoding, then the two images from the two coils contain signal from either the top or bottom of the brain only. If only every other line of k-space is acquired, then the field of view in the y-dimension is halved, and so the images from each coil appear stretched, but are not aliased. Stitching together of the two images, and dimensional rescaling by a factor of two produces an unaliased image acquired in half the time of the fully encoded image.

Example of parallel imaging

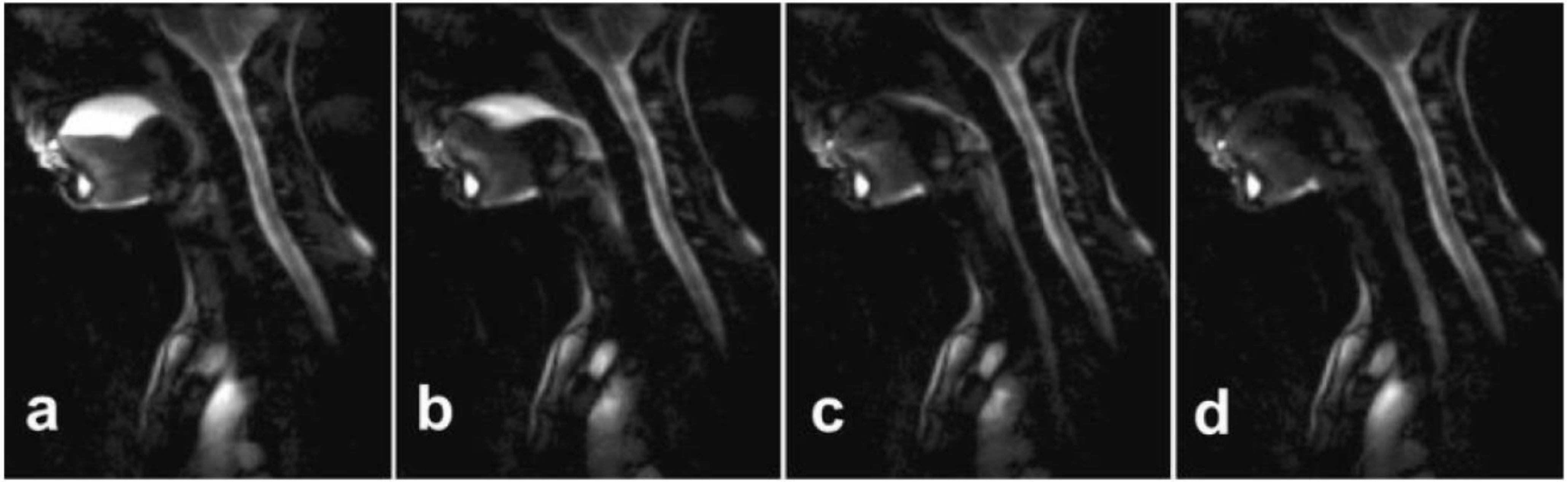


Fig. (a-d) Successive images from a subject who is swallowing water (bright signal). Data acquisition speed was increased four times over normal using parallel imaging.

Echo planar imaging

- Echo planar imaging (EPI, 平面回波序列);

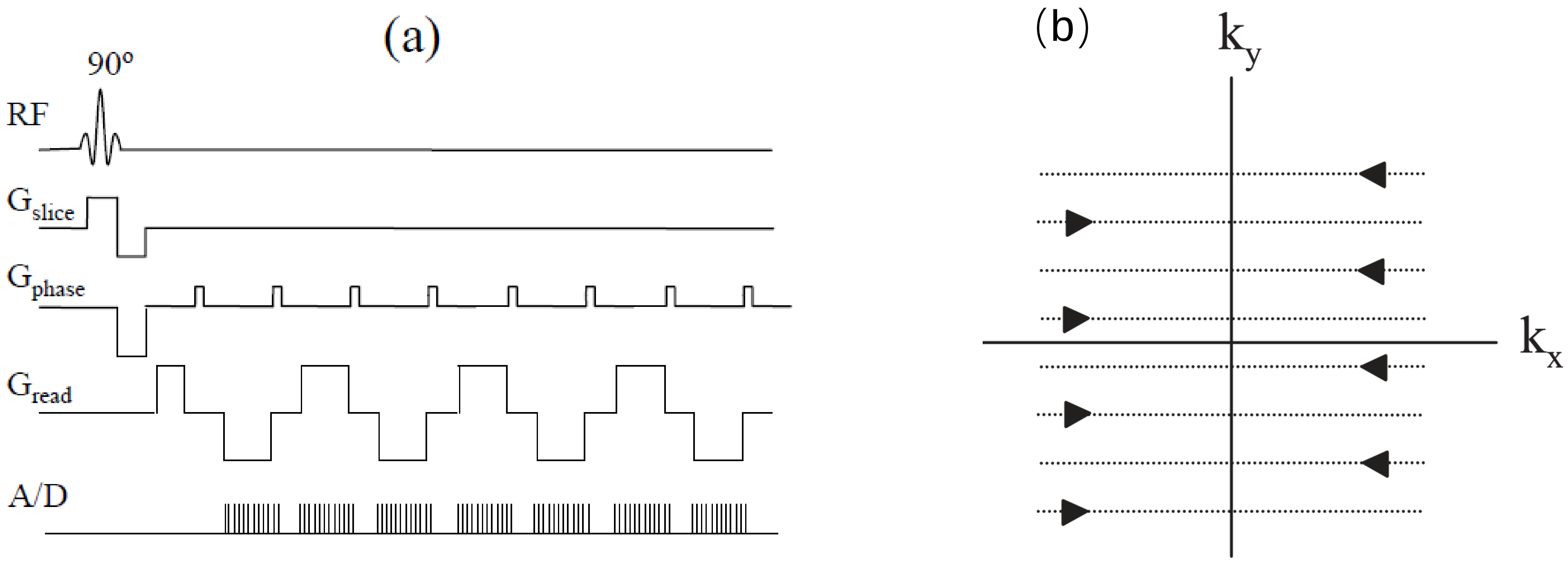


Fig. (a) Basic echo planar imaging sequence used for single-shot rapid MRI, and (b) Corresponding k-space coverage. The first data point acquired is at the maximum negative values of k_y and the maximum positive values k_x and (bottom left) and proceeds according to the arrows, ending up at the maximum positive values of k_x and k_y (top right) via a zig-zag trajectory.

Example of EPI

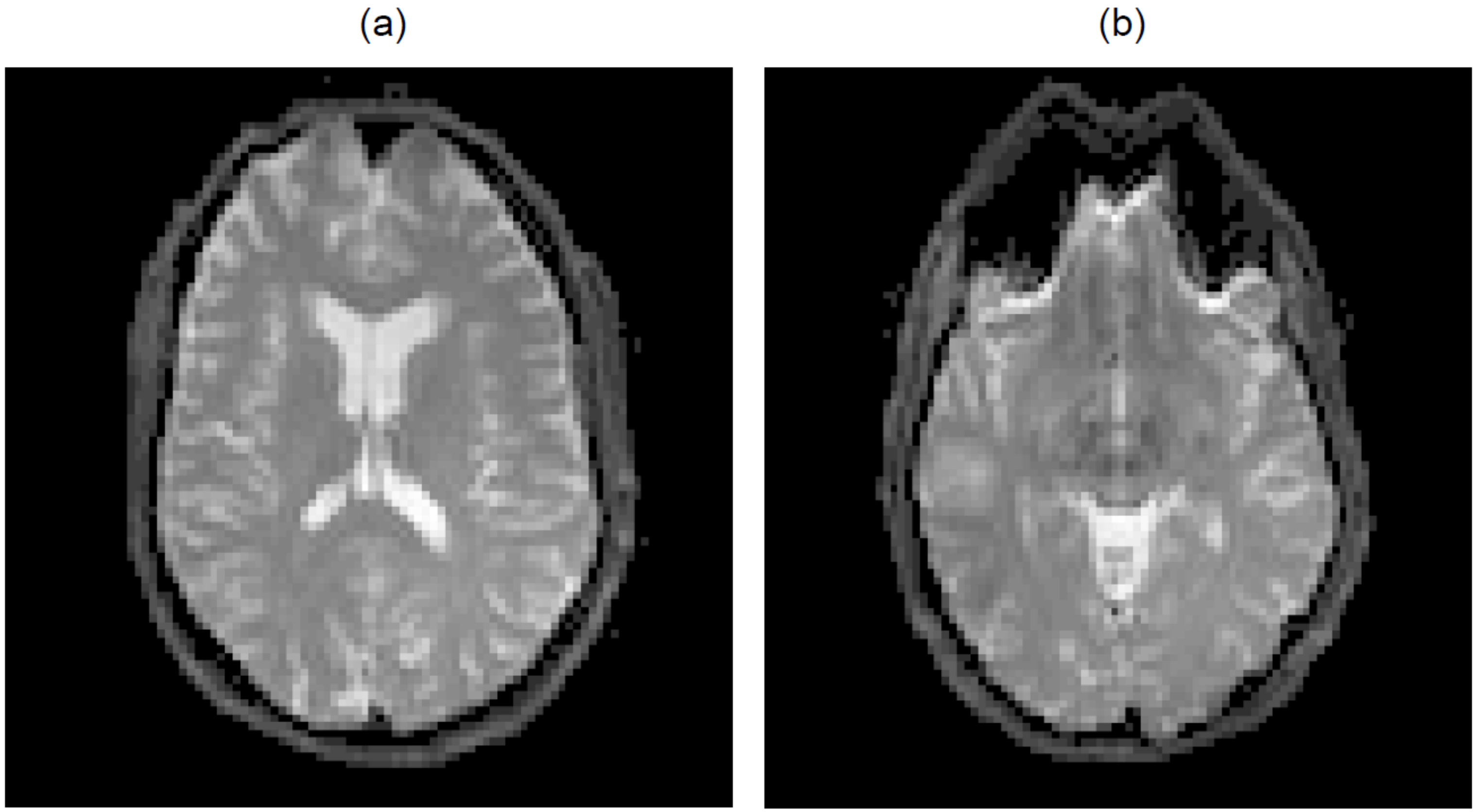


Fig. Two single-shot EPI images acquired in a patient's brain. Slice (a) was acquired in the middle of the brain, whereas slice (b) was acquired in an area close to the nasal cavities. The severe image distortions in (b) arise from the very short T_2* values in brain tissue close to the tissue/air interface.

Lecture 23 – MRI Instrumentation

This lecture will cover:

- MRI Instrumentation
  - Superconducting magnet
  - Magnetic field gradient coils
  - Radiofrequency coils
- Fast MRI imaging
  - Parallel imaging
  - Echo planar imaging
- **Imaging characteristics**

SNR

- B_0 field strength.
  - The signal is proportional to net magnetization M_0 and procession frequency ω_0 (B_0^2)
  - The noise arises from random voltage induced in the RF coils proportional to the $\sqrt{B_0}$
  - The higher B_0 ,the larger T_1, the shorter T_2 and the smaller image intensity for a given TR and TE;

- Image parameters
  - SNR decrease with short TR, high tip angle, long TE;
  - SNR decrease with increase of spatial resolution;
  - SNR is proportional to the slice thickness;
  - SNR increases by repeat image sequence. i.e. averaging ($\sqrt{N}$);

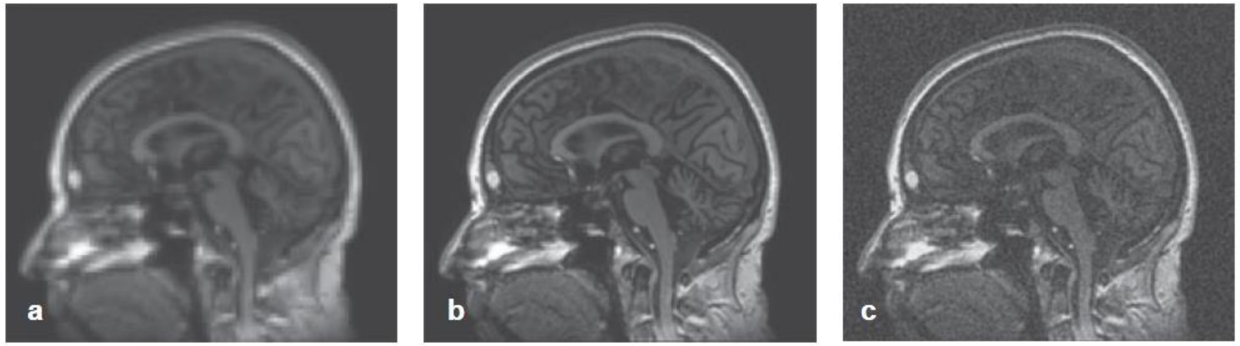


Fig. The trade-offs between SNR and spatial resolution. The three images are acquired with identical TR, TE and slice thickness. The image in (a) has a data matrix size of 64*64: the SNR is high, but the image appears 'blocky' with poor spatial resolution. The image in (b) has a matrix size of 128*128: the SNR is still relatively high. In image (c) with a data matrix of 256*256 the image is very 'grainy' due to the poor SNR.

Resolution and contrast

➢ Spatial resolution

- Slice thickness, the FOV in the phase-encoded dimension divided by the number of phase-encoding steps and the FOV in the frequency-encoded dimension divided by the number of data points
- Trade-off between spatial resolution, SNR and acquisition time

➢ Contrast to noise

- Based on difference in proton density, T_1, T_2 or T_2^*;
- Can be manipulated by choice of TR and TE;
- Contrast in T_1-weighted sequence decreases with reduced gradient field strength

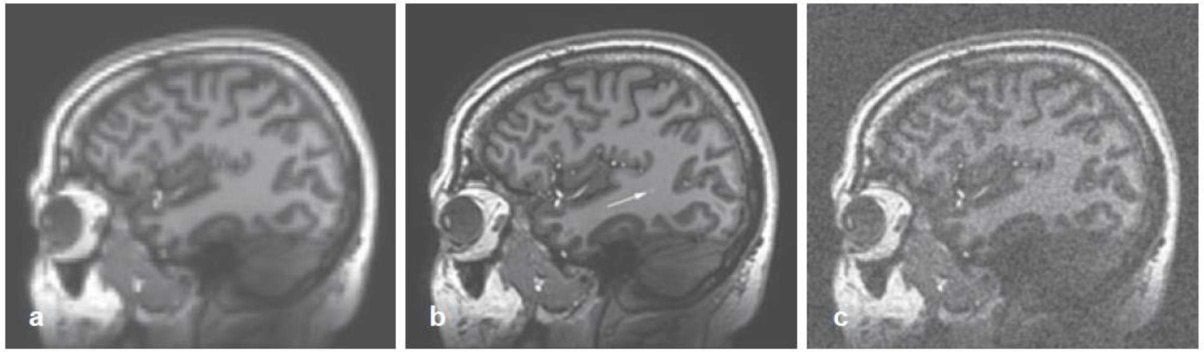


Fig. The trade-off between CNR, spatial resolution and SNR in MRI. A small hyperintense lesion is visible (white arrow) in image (b) which has a data matrix of 128*128. Using a lower data matrix of 64*64, image (a) produces a higher SNR, but the spatial resolution is not sufficient to see the lesion. Increasing the spatial resolution by acquiring a 256*256 data matrix, image (c), decreases the SNR so that the lesion is again not visible. The optimum spatial resolution for the best CNR in this case is given by image (b).