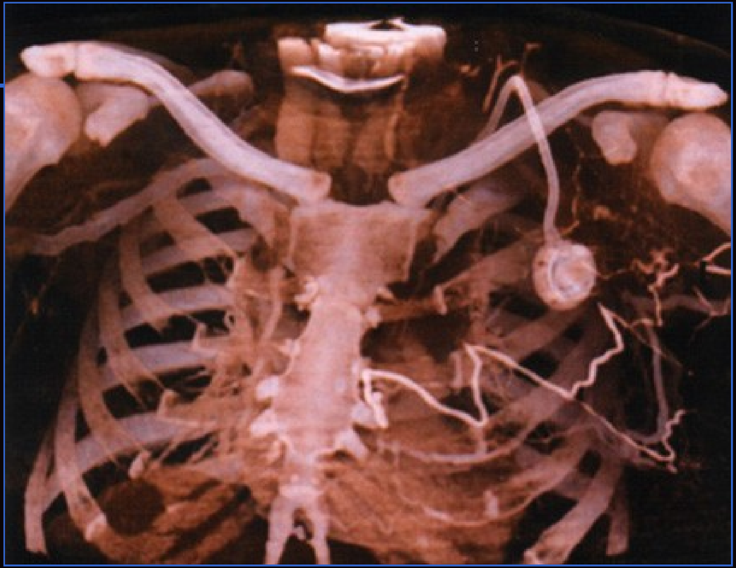
Principles of Computer Tomography (CT)



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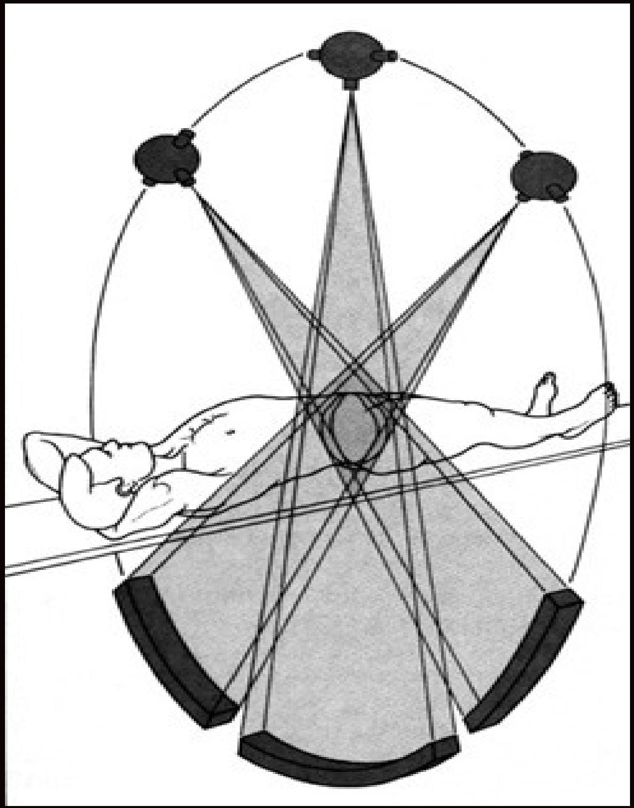
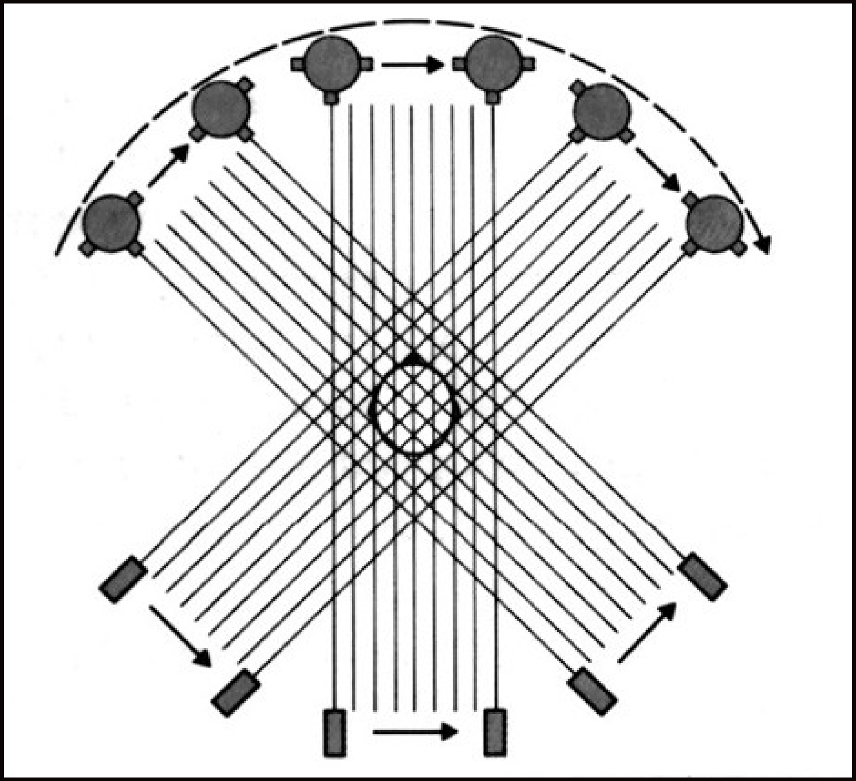
- CT images are acquired while the x-ray tube i rotating 360dg. around the patient
- The x-ray beam is collimated in axial orientation and divergent to encompass the patient‘s width in the other orientation.
- The intensity of attenuated x-rays emerging from the patient is measured with an array of minute detectors
- The detector array simultaneously rotates around the patient in unison with the x-ray tube (third generation of CT).



- multiple angular views of a 10mm thick slice or less are obtained, and from numerous attenuation values measured, a computer image is reconstructed using Fourier transform functions
- it takes 2-3 sec. to acquire an image – or „a slice“ and 10-15 to process the data for display.

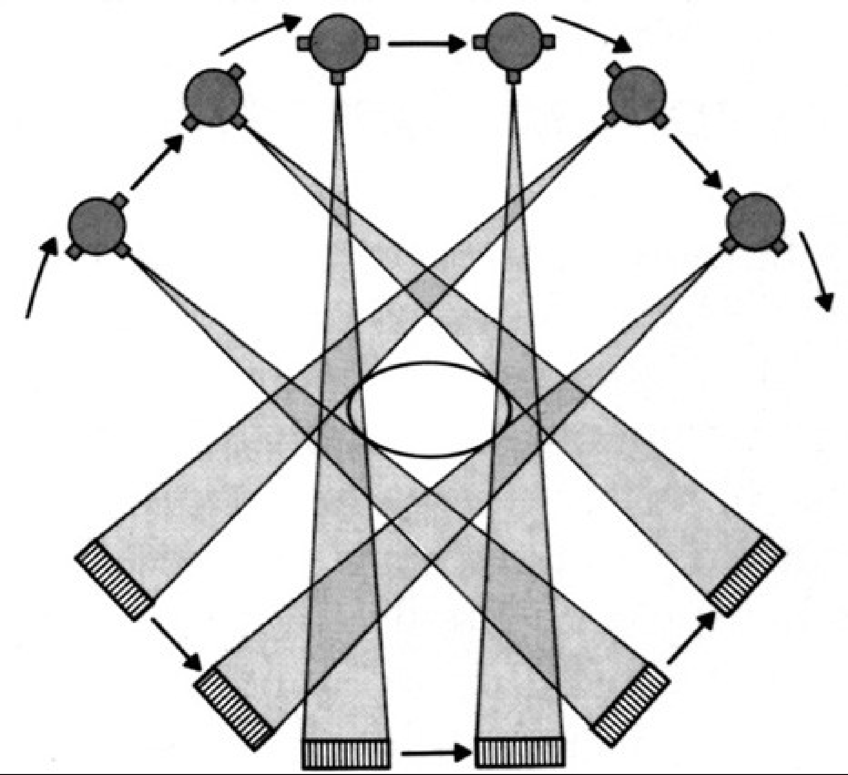
Generations of CT scanners – I.

- First generation used rotation and translation movement . There was one X-ray tube and one detector only.
- Parallel-beam
- It taked several minutes to acquire an image (one slice).



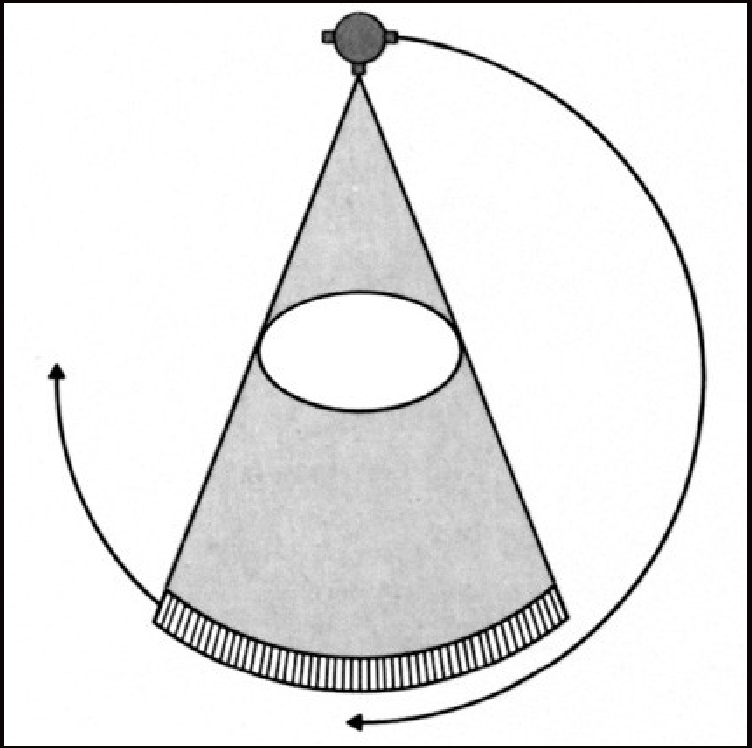
Generations of CT scanners – II.

- Second generation used also rotation and translation of X-ray tube and detectors – but there were several detectors (10-50).
- Small fan-beam
- It taked cca 10-20 seconds to acquire an image (one slice).



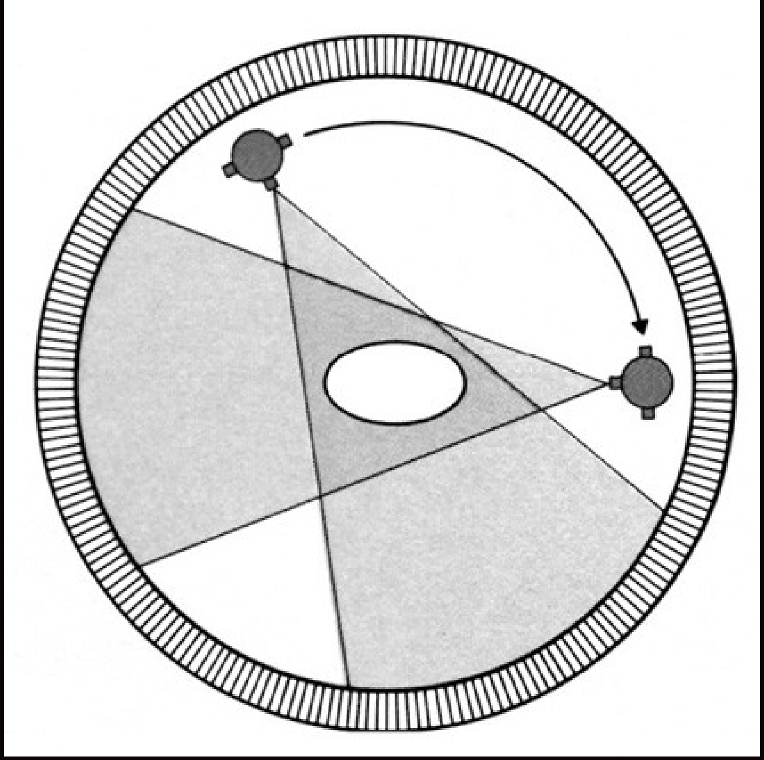
Generations of CT scanners – III.

- Third generation use wide row of detectors (300-600) - large fan-beam.
- The detector array simultaneously rotates around the patient in unison with the x-ray tube (without translation movement).
- The most used type at the moment.
- It takes cca 0,5-4 seconds to acquire an image (one slice).

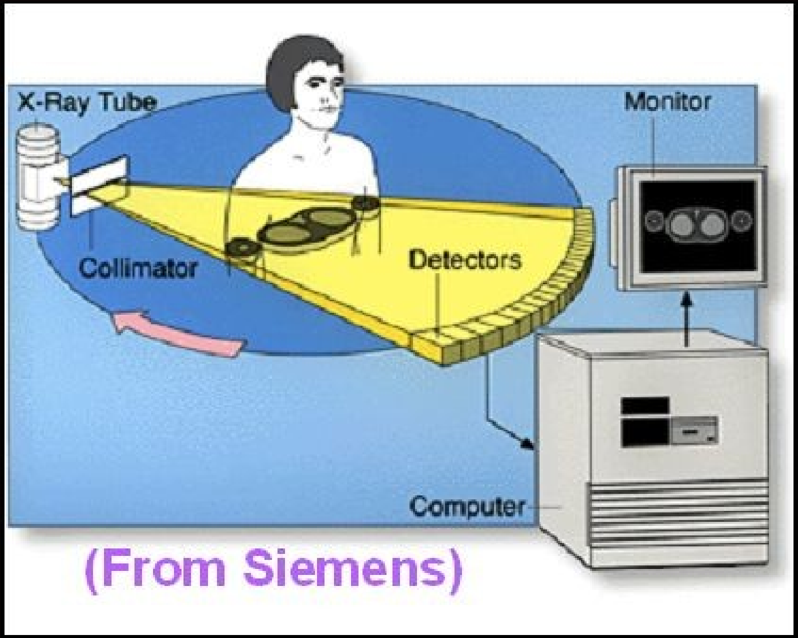


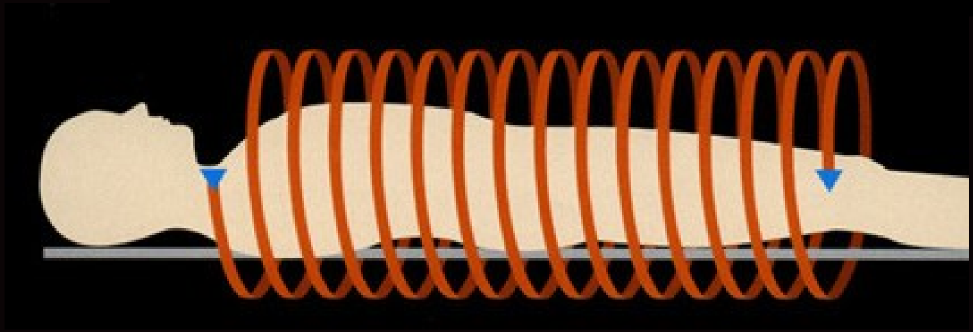
Generations of CT scanners – IV.

- Detector ring
- Source-rotation
- Large fan-beam
- Misrepresenting geometry
- Complicated balancing of rotor

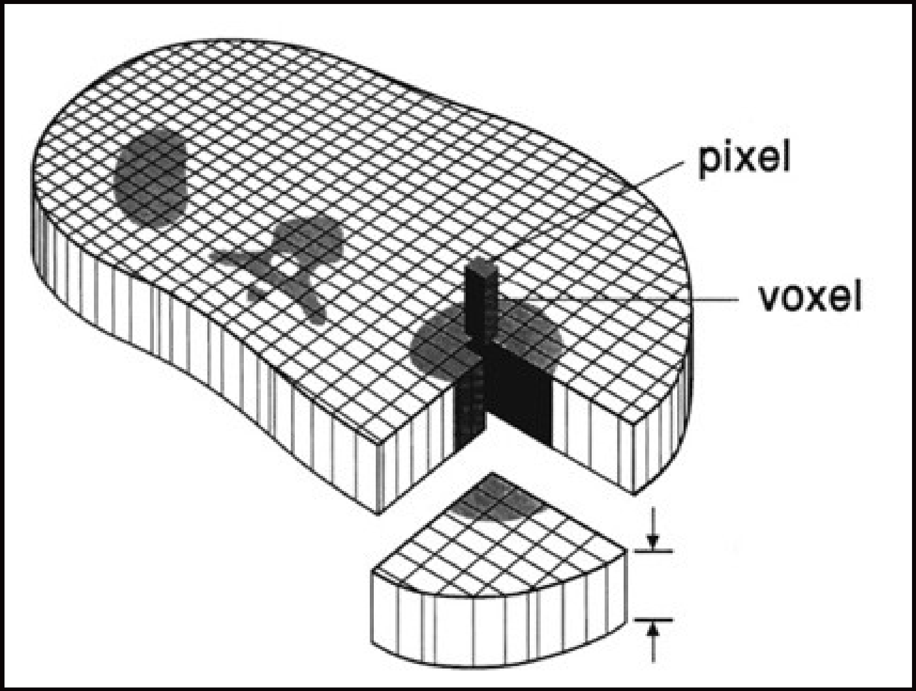


- Multiple images may also be obtained without ihnterrupting the x-ray beam between slices
- A continuous rotation of the tube and simultaneous movement of the tabletop through the gantry characterize the spiral or the helical CT
- The slice is controlled by collimating the beam in axial direction and may be as thin as 1mm (for high resolution imaging)

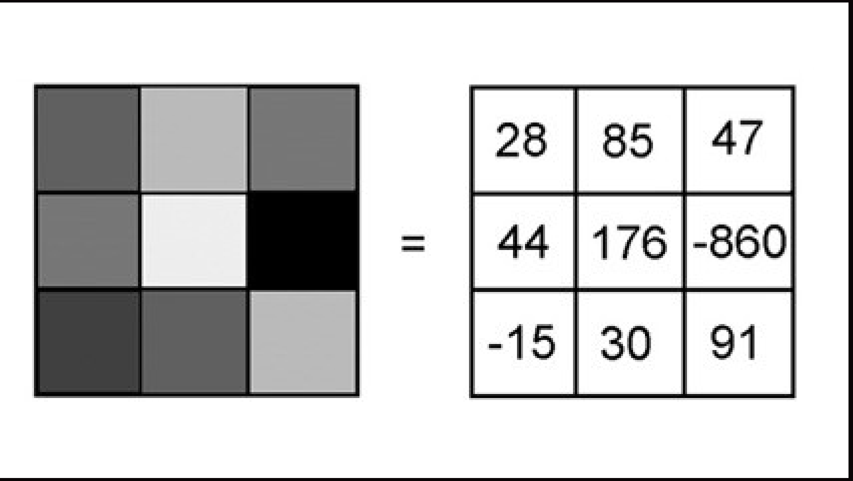




- CT images are displayed in a two-dimensional plane on a 512x512 matrix of image elements or pixels
- A pixel represents a tissue volume or voxel that usually measures 1x1x10mm
- The longest dimension of voxel is oriented parallel to the long axis of the patient and represents slice thickness



- Each voxel, respecitvely pixel in two–dimensional imaging, has its proper level of absorption coefficient – it is called Hounsfield unit (HU) (sir Godfrey Newbold Hounsfield constructed the first CT)
- Specific shade of gray colour is assigned each number of HU

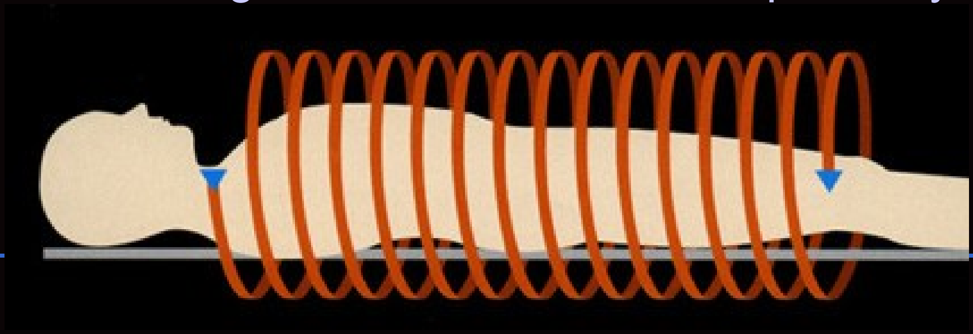


Conventional CT scanners

- Employ fan of x-ray beams and a large detector array
- Involves alternating patient translation and x-ray exposure
- Each rotation of x-ray tube generates data from which a corresponding axial image is reconstructed

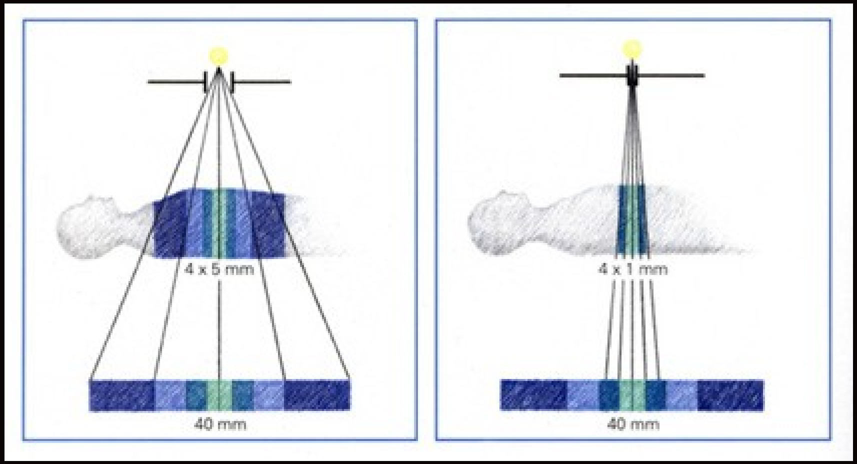
Helical (spiral) CT

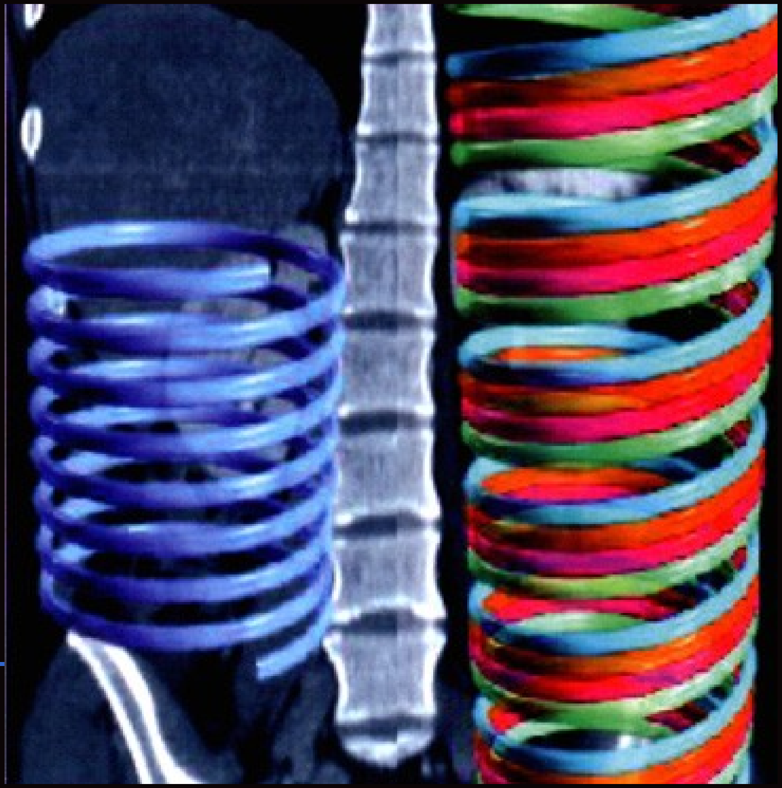
- Slip-ring technology (no electrical cables connecting gantry to ground) allows source detector assembly to rotate continuously. Previously, frequent, abrupt changes between scans were necessary to permit winding and unwinding of cables
- Simultaneous patient translation and x-ray scanning generates volume of data
- X-ray beam traces a helix of raw data from which axial images must be generated
- Each rotation generates data specific to an angled plane of section
- To create true axial image, data points above and below desired section must be interpolated to estimate value in axial plane
- Thus, interval between reconstructed images can be chosen retrospectively



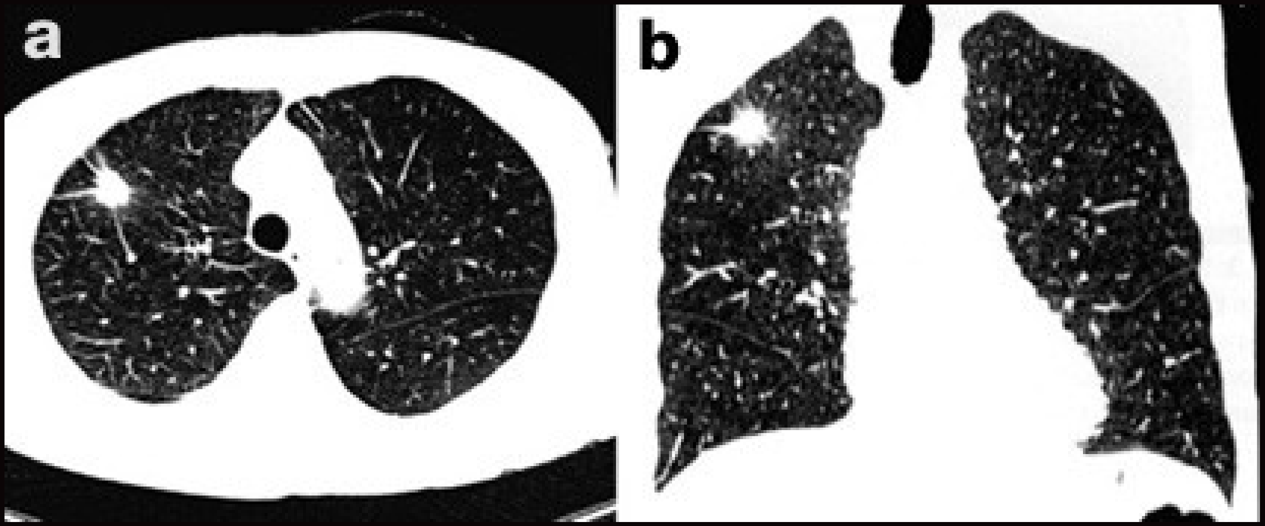
Multislice CT

- Several rows of detectors each above another
- MS- detector array segmented in z axis, a mosaic
- –Allows for simultaneous acquisition of multiple images in scan plane with ONE rotation.

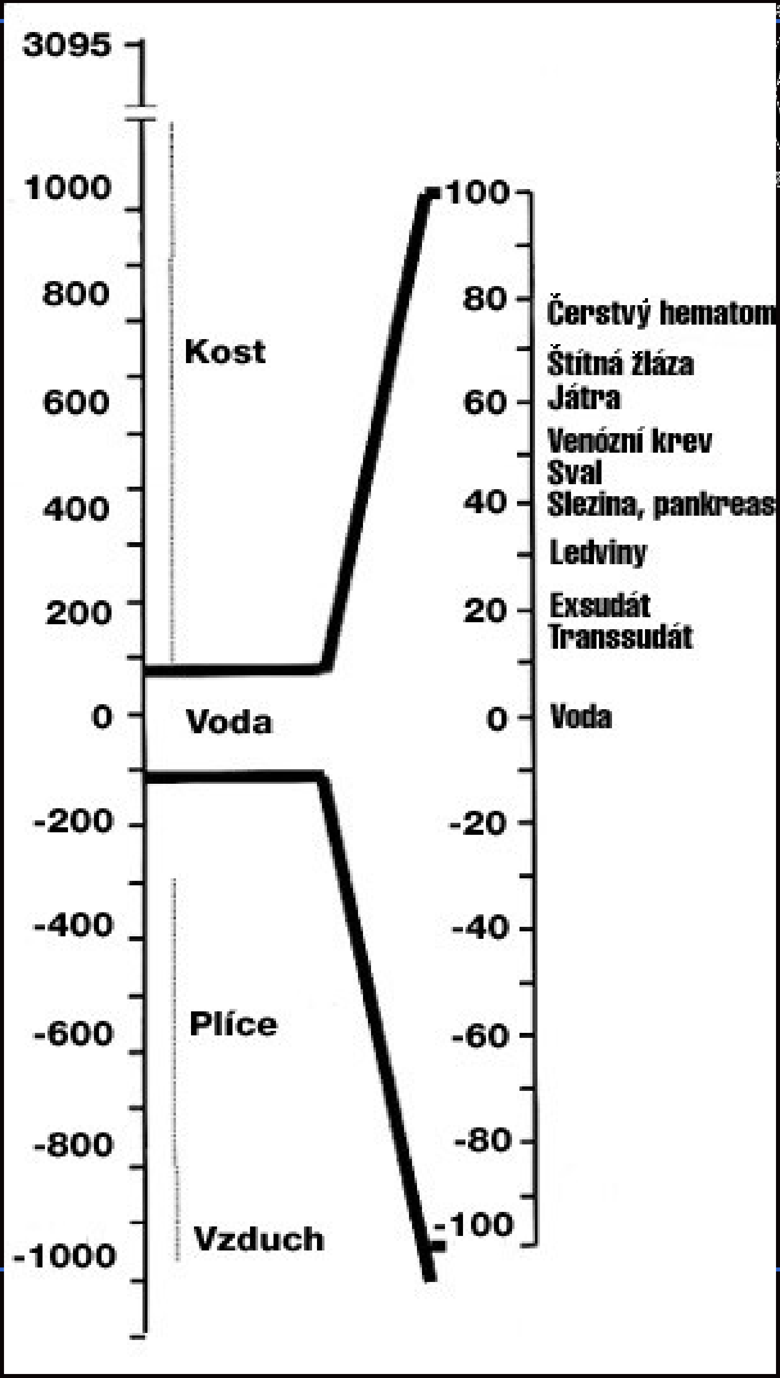




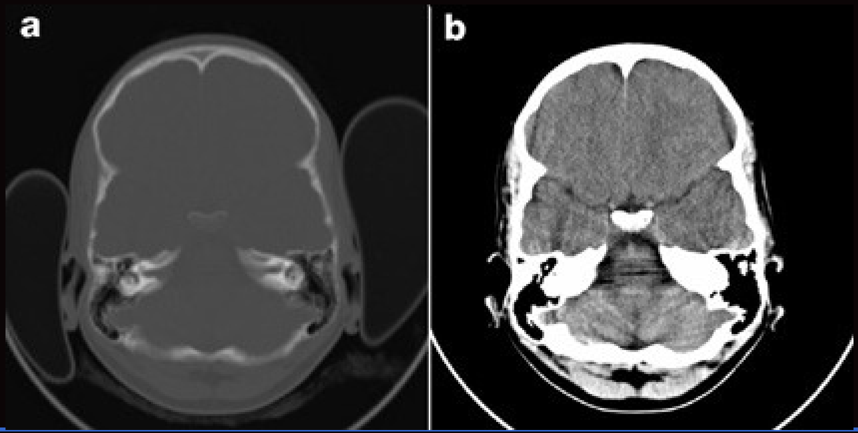
- Possibility of isotropic geometric resolution in coronal and sagital plain (excellent quality of multiplanar reconstructions).



- X-ray attenuation values are measured in Hounsfield units (HU) and are represented on the Hounsfield scale
- water is arbitrarily assigned a value of 0HU, air -1000HU and dense cortical bone +1000HU.



- The human eye is capable of distinguishing only 30-60 levels of gray – when presented with an image of 4000 levels of gray, the eye cannot perceive such small density differences.
- Two controls provide for selection of window width and window level. These allow high-contrast viewing around the area of interest.
- The window width selects only a segment of the Hounsfield scale and displays that „window“ at the full 256 gray levels. All values above the window width are displayed as white and all values below are diasplayed as black.



bone window- a,

brain window -b

Contrast media application

- intravascular – intravenous, intraarterial (iodine contrast media – ionic or non-ionic, mostly hyperosmolar; nefrotrophic)
- peroral (isodense - water, hypodense -air, hyperdense – iodine or barium enemas)
- intrathecal (liquor spaces, isoosmolar non-ionic high quality contrast media)
- intracavital

Contraindications

- Allergic patient
- Severe renal insufficiency
  - (risk of contrast-induced nefropathy)
- Acute stroke
  - (dissrupted blood brain barrier)
- Thyreotoxicosis
- Paraproteinemia
  - (Bence-Jonesovy protein can preccipitate in renal tubules and cause acute tubular necrosis)

Complications

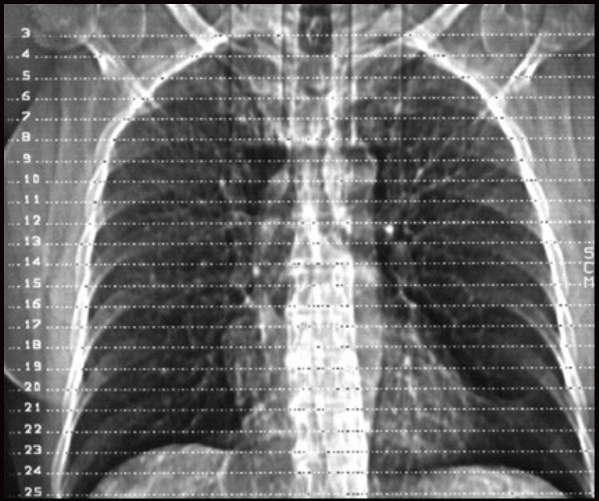
- Adverse reaction – due to chemotoxicity – heat feelings
- Allergic reaction
- Trophic changes in paravascular application

Algorithm of CT examination

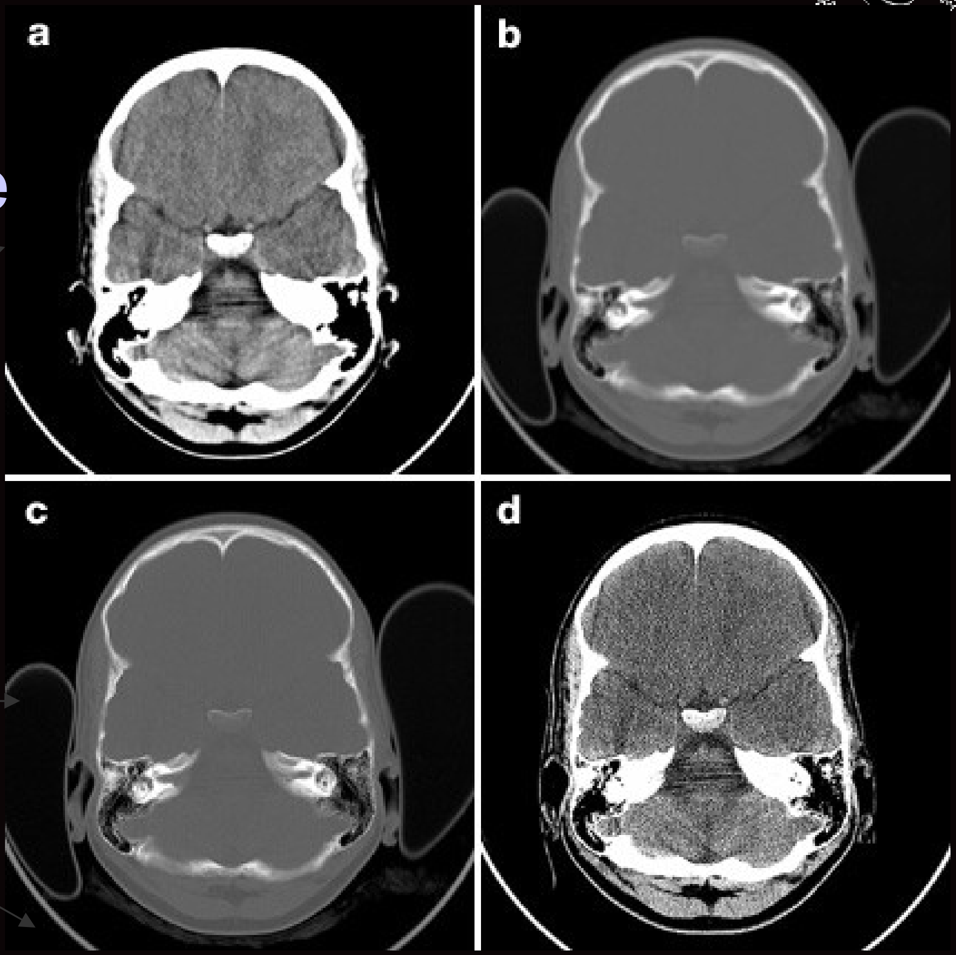
1. Topogram
2. Scanning parametres
3. Reconstruction parametres
4. Postprocessing, archiving of the data

Topogram

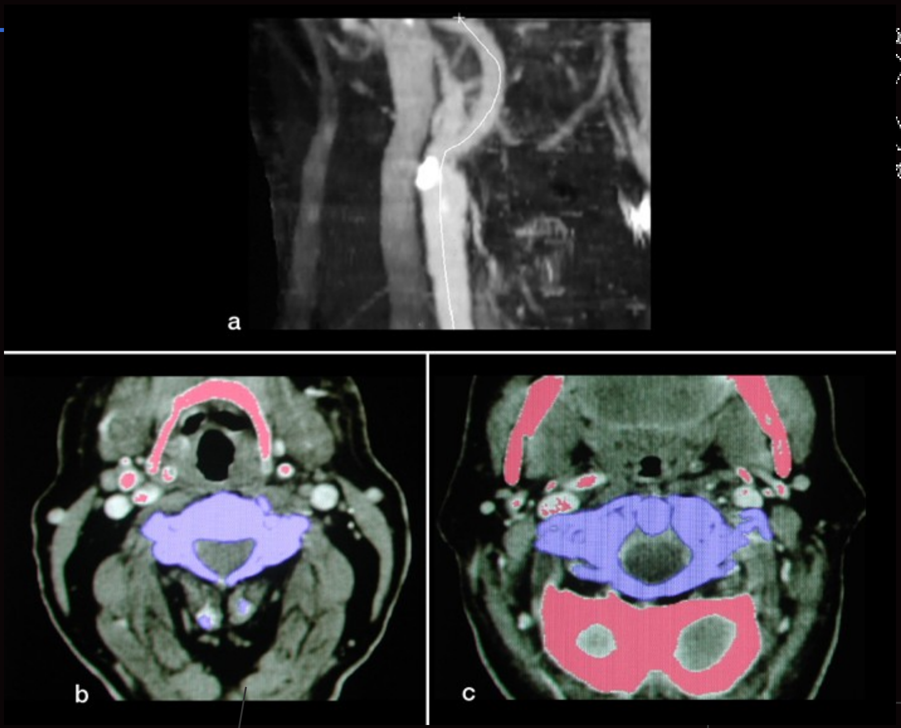
Range of examination
Slice thickness and number



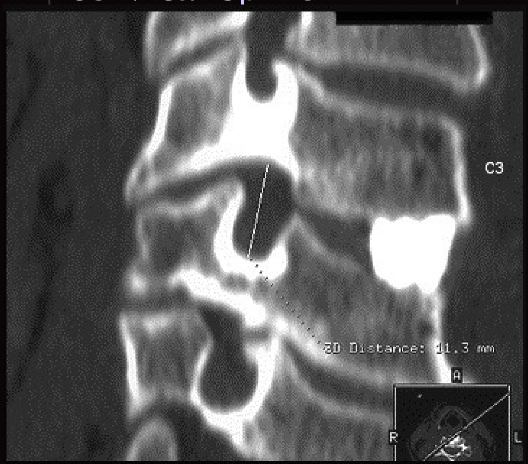
- Two main algorithms of reconstruction of the image:
- Soft tissue (higher contrast resolution) – a,b
- HR – high resolution (superior geometric resolution – c,d)



Postprocessing

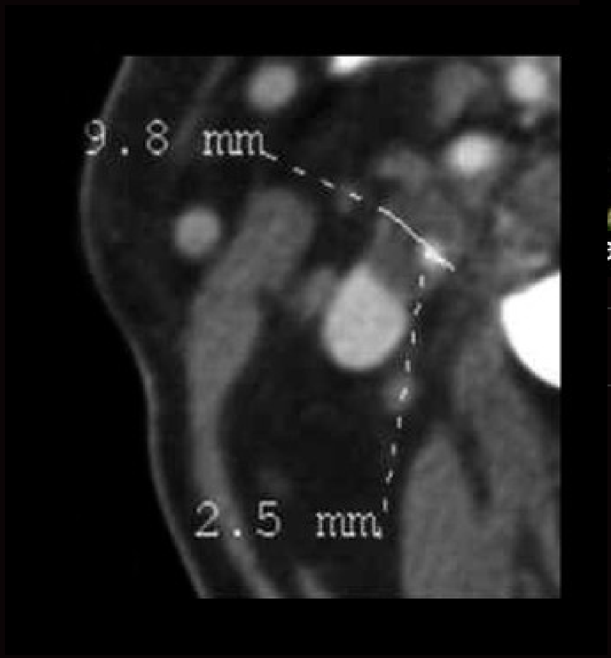


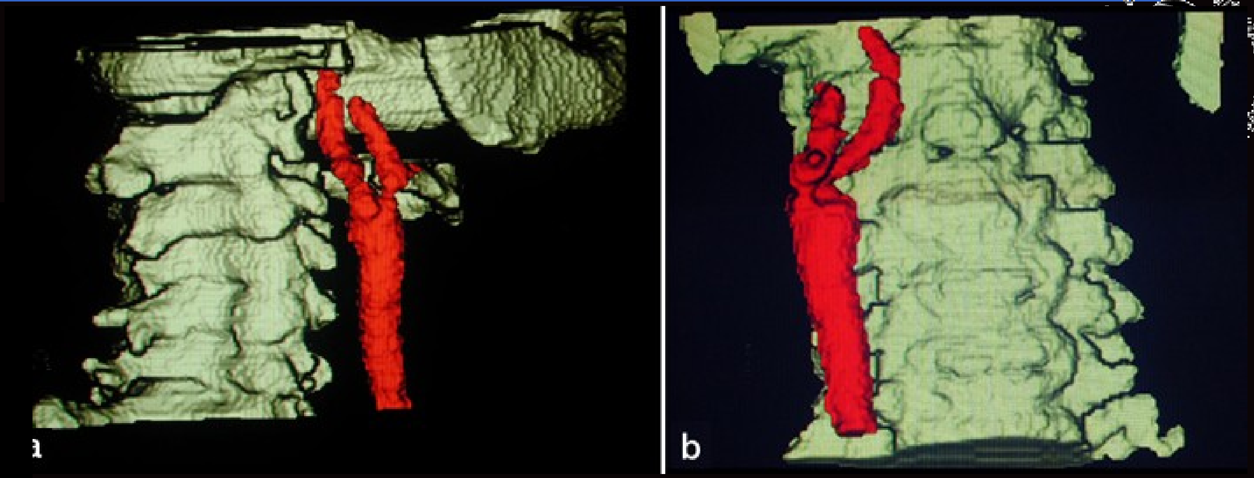
2D oblique view of cervical spine



CT angiography of carotid arteries – MIP (maximum intensity projection) reconstructions

CTA



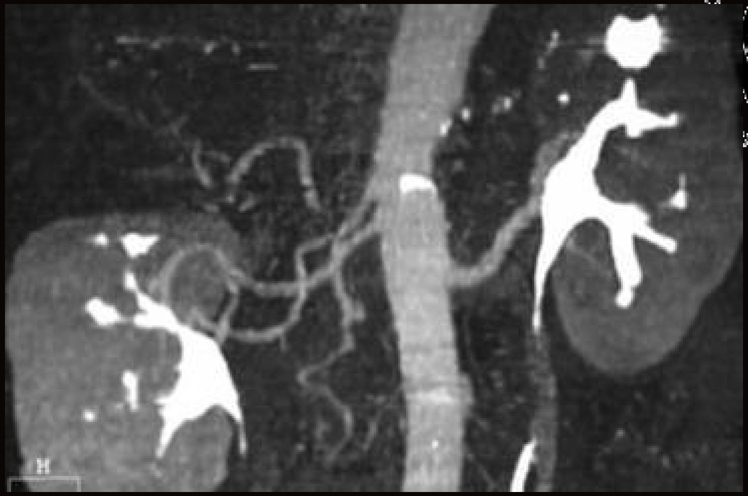


CTA – shadow surface display - SSD_

Stenosis qiantiffication

CT angiography of the aorta

V. cava superior sinistra.

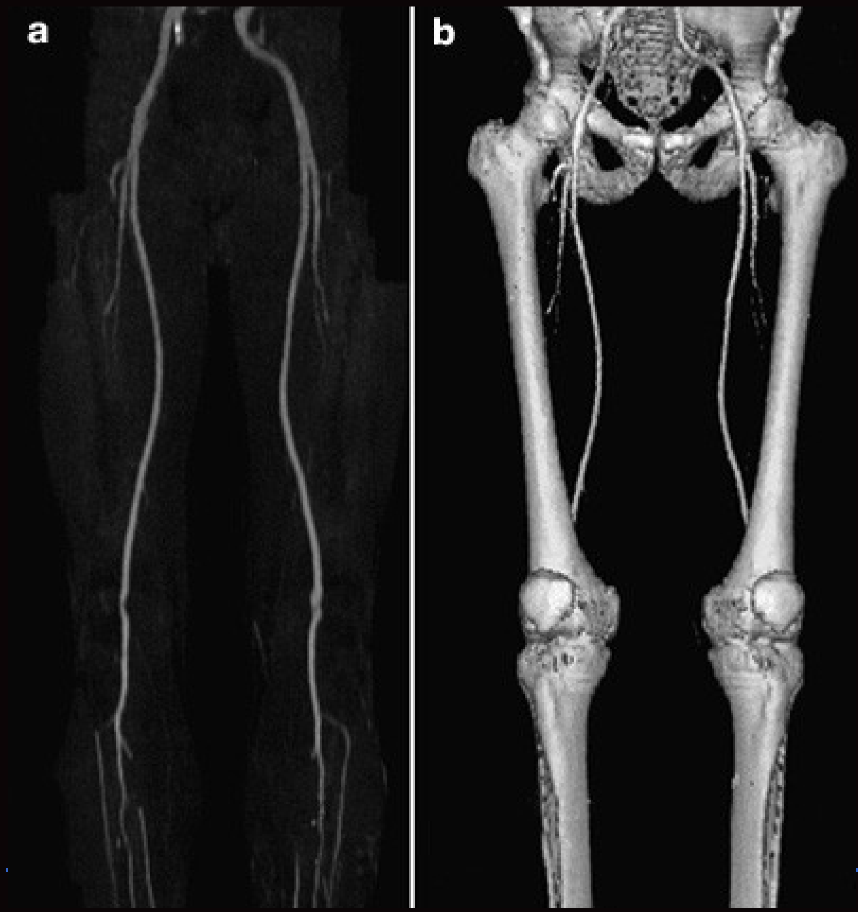


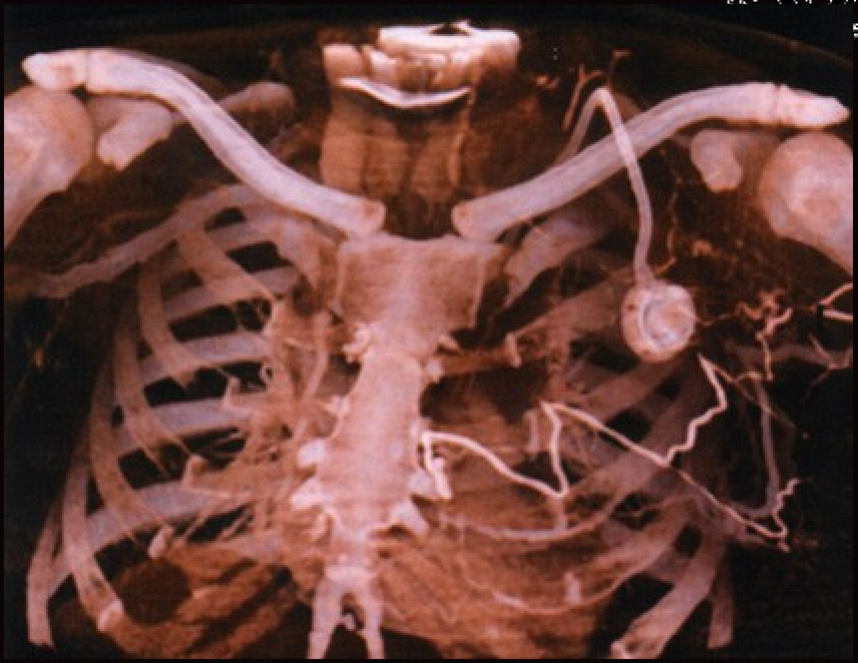
CTA - A. renalis duplex dx., truncus coeliacomesentericus

3D reconstruction of multiple facial bones fractures

3D – manubrium sterni fract.

Left – MIP, right – SSD





Volume rendering technique (VRT)

Principles of MR imaging

Bohatá Š.

Magnetic resonance

- Hydrogen nuclei (1H) have a magnetic behavior
- We use a strong magnetic field (0,2-2T)
- radiofrequency pulses - a resonance of frequencies - transfer of energy - 1H raising from lower to higher energy states
- signal from protons = from the pacient body - induction of a current in a receiver coil - computer reconstruction - resulting image

Disadvantages of MRI

- Strong magnetic field! (*whole patient is placed in)*
- duration of examination - *untill 60 min.*
- limited examination space
- the price = availability
- limited examination field *(brain + Csp., C+Th, Th+L)*

Benefits of MRI

- Noninvasive technique
- unprecedented soft tissue contrast
- arbitrary plane of cross-section
- MR angiography, ERCP, PMG *(without a contrast medium)*
- contrast media - Gd *(minimal risk of allergic reaction)*

Contraindications of MRI

ABSOLUTE

- presence of magnetic metallic implants
- pacemaker/ICD

RELATIVE

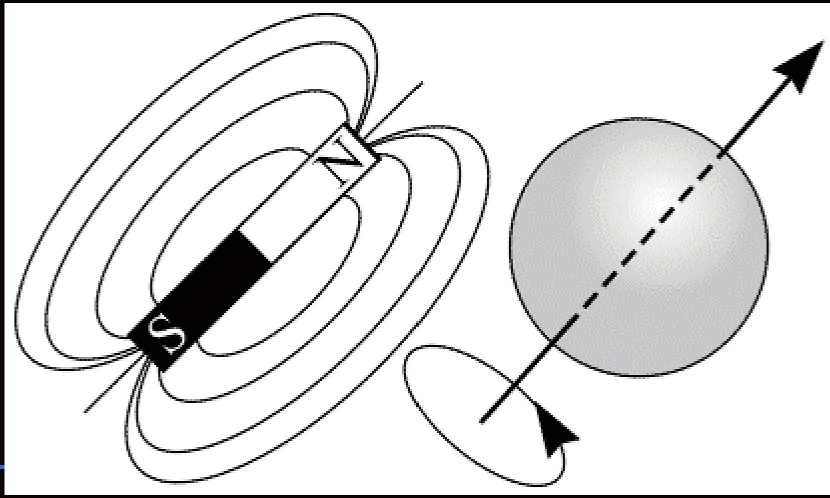
- claustrophobia
- non-cooperative patient
- I. trimestr of pregnancy

MR principles

- Nuclear magnetic resonance (NMR) is a non-invasive means of obtaining clinical images and of studying tissue metabolism in vivo. Bloch and Purcell independently discovered NMR in 1946. Six years later they were awarded the Nobel Prize for their achievements. Since then, the development of NMR spectrometers and NMR scanners has led to the opening up of whole new branches of physics, chemistry, biology and medicine.

Nuclei with an odd number of protons and neutrons possess a property called spin. In quantum mechanics spin is represented by a magnetic spin quantum number. Spin can be visualised as a rotating motion of the nucleus about its own axis. As atomic nuclei are charged, the spinning motion causes a magnetic moment in the direction of the spin axis. This phenomenon is shown in figure. The strength of the magnetic moment is a property of the type of nucleus. Hydrogen nuclei (1H), as well as possessing the strongest magnetic moment, are in high abundance in biological material. Consequently hydrogen imaging is the most widely used MRI procedure.

A charged, spinning nucleus creates a magnetic moment which acts like a bar magnet (dipole).



- The first step in creating a magnetic resonance image is placing the subject in a strong magnetic field.

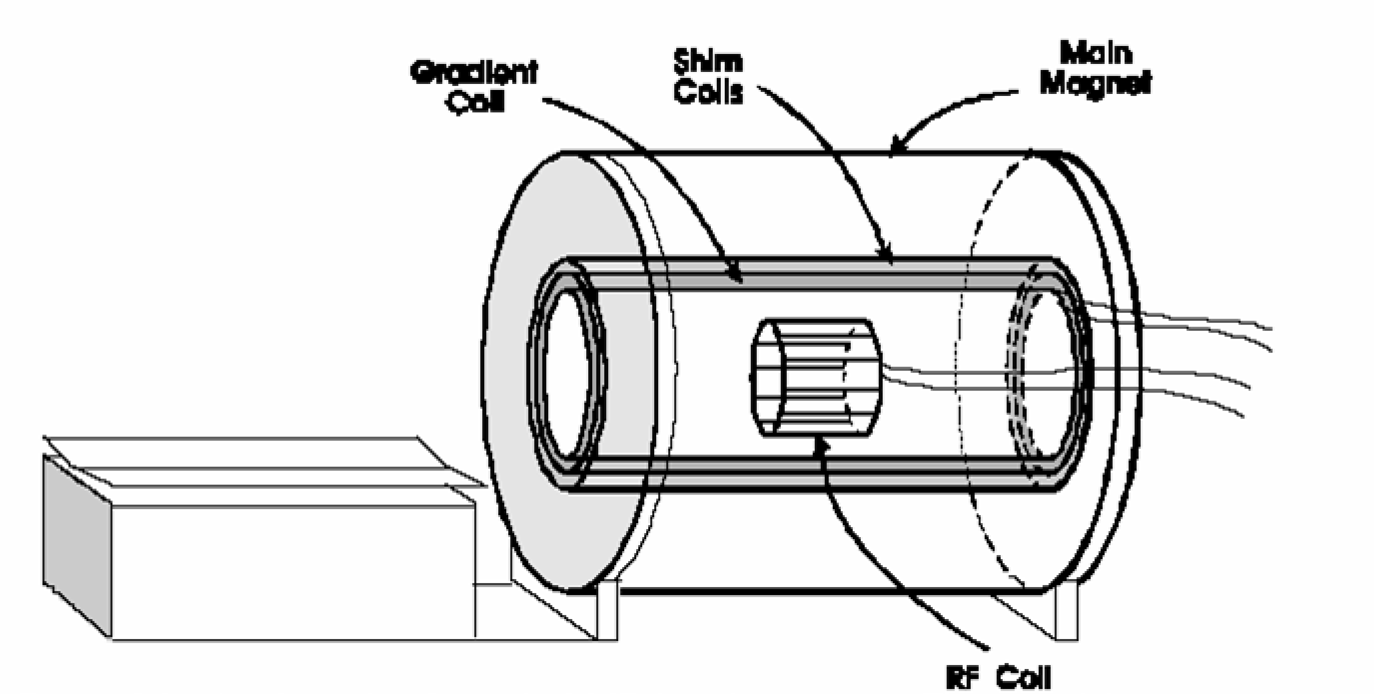
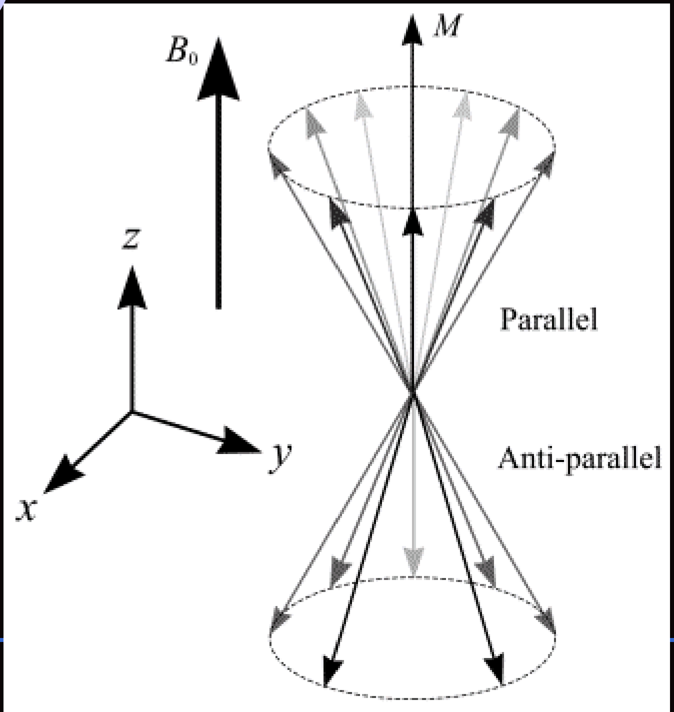
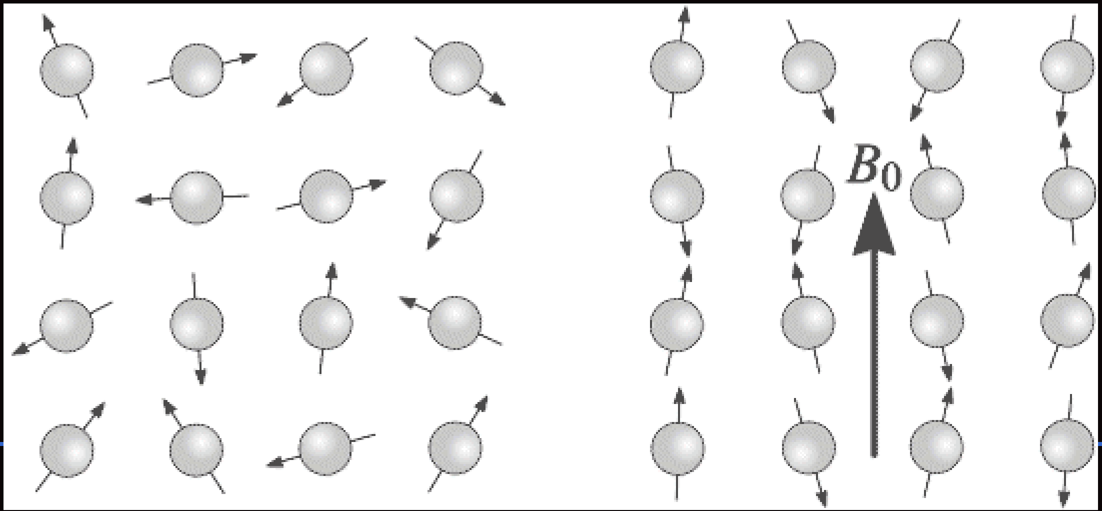


Figure 9.1: A schematic of a typical MR imaging system. The essential components include the magnet producing the main magnetic field, shim coils, a set of gradient coils, an RF coil, and amplifiers and computer systems (not shown) for control of the scanner

- Presence of a strong magnetic field causes the nuclear spins of certain atoms within the body, namely those atoms that have a nuclear spin dipole moment, to orient themselves with orientations either parallel or antiparallel to the main magnetic field (Bo).

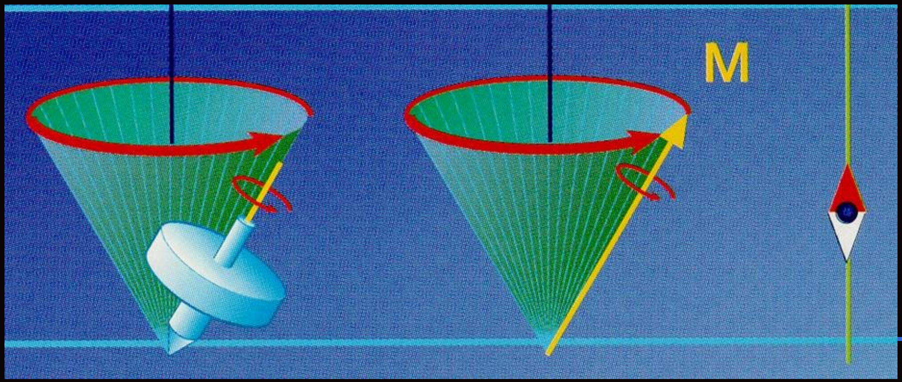


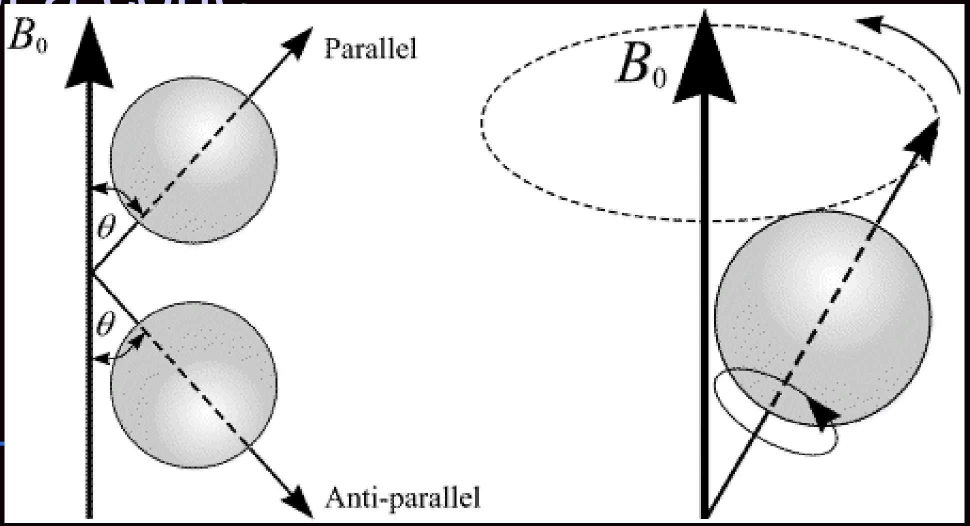
- A collection of 1H nuclei (spinning protons) in the absence of an externally applied magnetic field. The magnetic moments have random orientations. (b) An external magnetic field B0 is applied which causes the nuclei to align themselves in one of two orientations with respect to B0 (denoted parallel and anti-parallel).



The spin axes are not exactly aligned with B0, they **precess** around B0 with a characteristic frequency

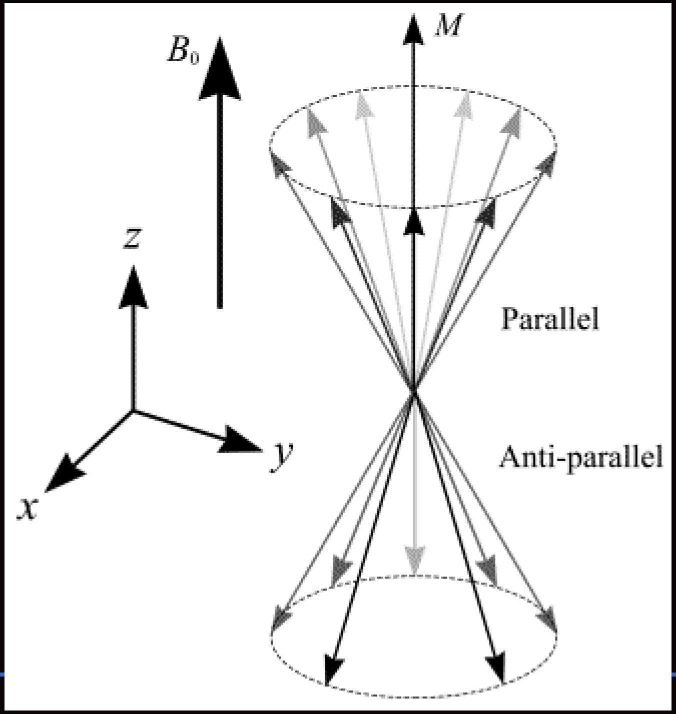
- (a) In the presence of an externally applied magnetic field, B0, nuclei are constrained to adopt one of two orientations with respect to B0. As the nuclei possess spin, these orientations are not exactly at 0 and 180 degrees to B0. (b) A magnetic moment precessing around B0. Its path describes the surface of a cone





- The nuclei precess about Bo with a frequency called the resonance or a Larmor frequency.
- Because the parallel state is the state of lower energy, slightly more spins reside in the parallel configuration, creating a net magnetization represented by a vector M.

A collection of spins at any given instant in an external magnetic field, B0. A small net magnetisation, M, is detectable in the direction of B0.



Magnetic resonance occurs when a radiofrequency pulse , applied at the Larmor frequency, excites the nuclear spins raising them from their lower to higher energy states. Classically this can be represented by a rotation of the net magnetization,

away from its rest of equilibrium state.

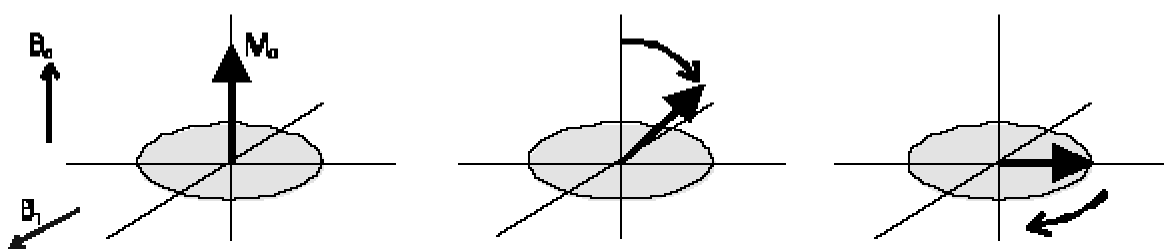
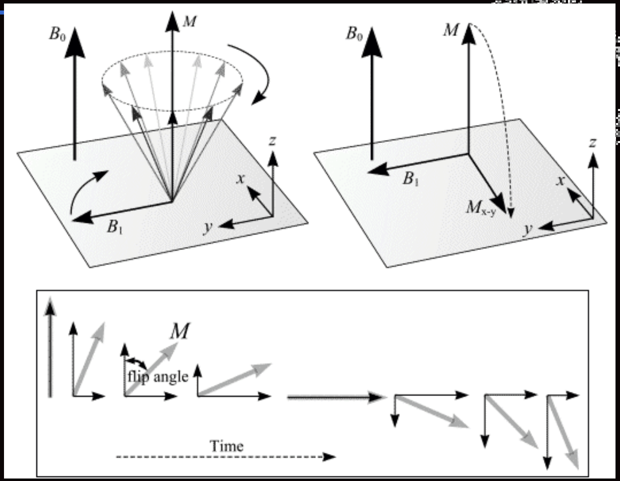
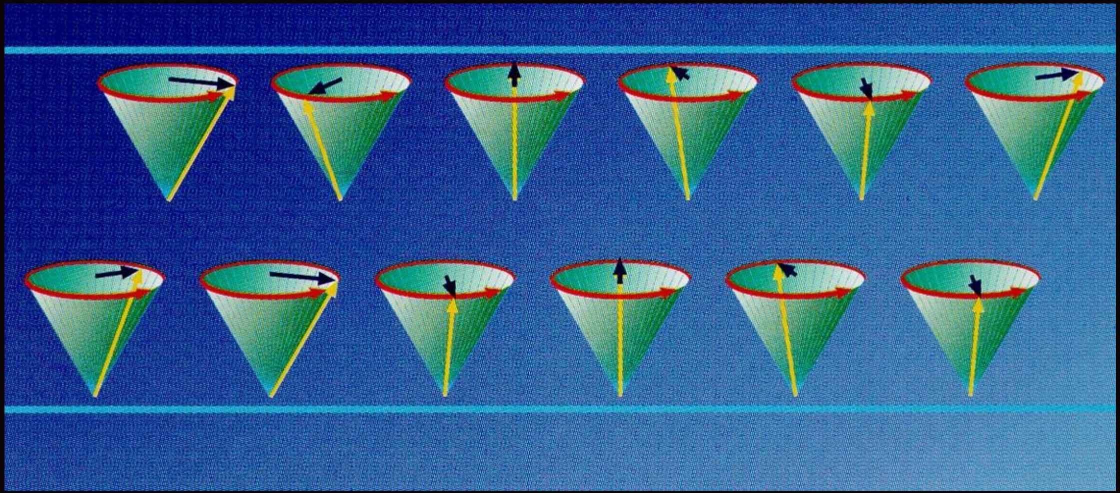
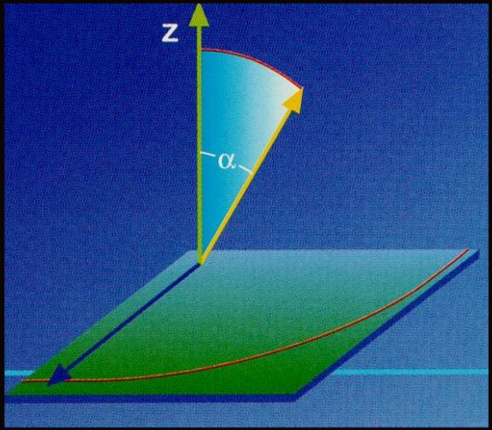


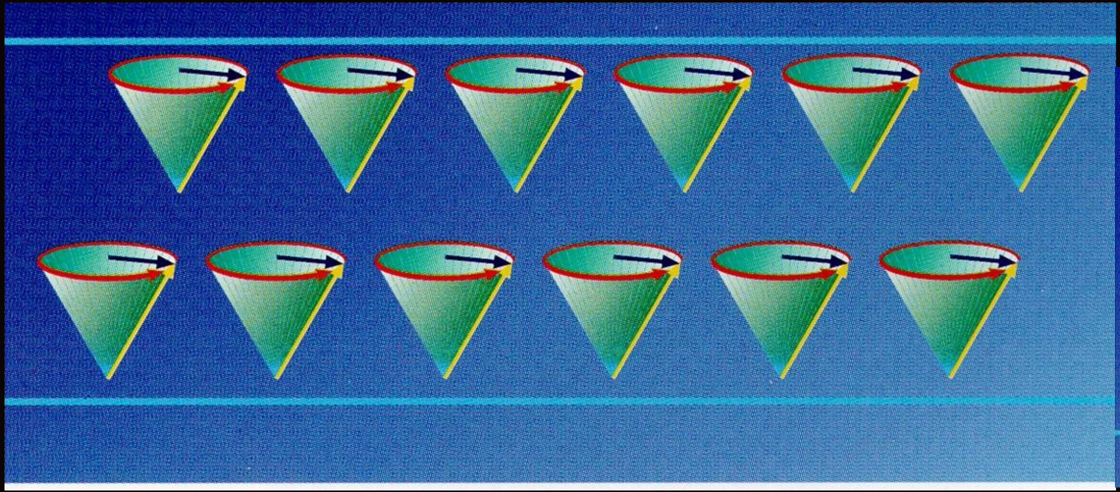
Figure 9.2: A series of vector diagrams illustrating the excitation of a collection of spins by applying an alternating magnetic field, in this case a 908 radio-frequency (RF) pulse (represented here as B_1). B_0 indicates the direction of the main magnetic field. The first 2 vector diagrams are in a frame of reference rotating with the radio-frequency pulse. As a result, the alternating magnetic field can be represented by a vector in a fixed direction. Application of the RF pulse flips the magnetization into the transverse plane, after which the magnetization continues to precess about the main magnetic field.

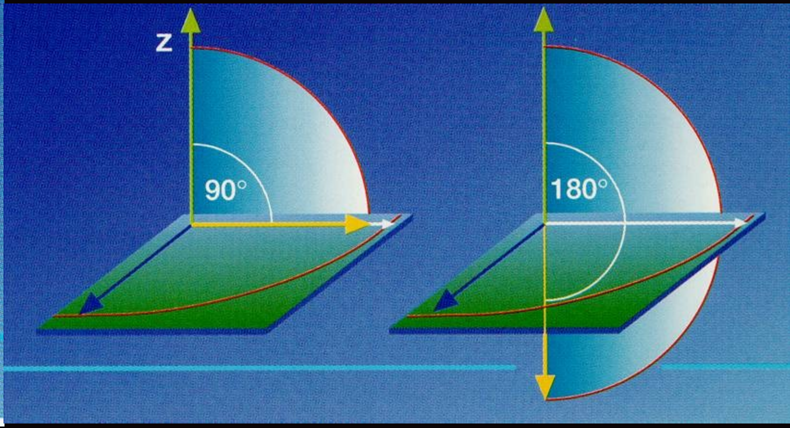
After a 90 degrees RF pulse, M lies in the x-y plane and rotates about the z-axis. The component of M in the x-y plane decays over time. An alternating current is induced in the receiver coil.











- Once the magnetization is deflected, the RF field is switched of and the magnetization once again freely precesses about the direction of Bo.
- This time dependent precession will induce a current in a receiver coil.
- The resultant exponentially decaying voltage, referred to as the free induction decay (FID), constitutes the MR signal.

After a 90 degrees RF pulse, M lies in the x-y plane and rotates about the z-axis. The component of M in the x-y plane decays over time. An alternating current, shown in Figure (b), is induced in the receiver coil.

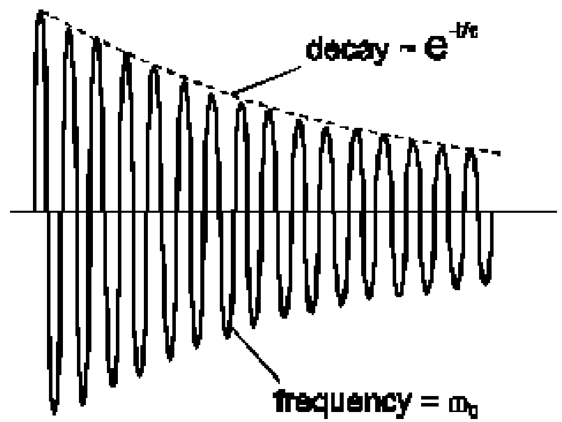


Figure 9.3: The signal acquired after excitation in the absence of applied magnetic field gradients is a decaying sinusoid, called the free induction decay (FID). This signal is characterized by two parameters – the amplitude and the frequency, which depend on the number and type of spins being studied and the magnetic environment that the spins are in.

- During the period of free precession the magnetization returns to its original equilibrium state by a process called ***relaxation*** which is characterized by two time constants, T1 and T2.

T1 and T2 depend on certain physical and chemical characteristics unique to tissue type, therefore contributing substantially to the capability of MRI to produce detailed images of the human body with unprecedented soft tissue contrast.

Reconstruction of MR image

- How do we make the voxels (volume elements)?
- How do we know, from which part of the body cames the signal, how we make the final image?
- Complex and complicated process, we need a physical and mathematical device

Slice Selection

- The first step is the selection of a slice, which is achieved by applying a magnetic field gradient along the z-axis (Gz) during a 90o rf pulse of a specific frequency bandwidth (period 1 of Figure 9.7). When the slice select gradient, Gz, is applied along the z-axis,the resonance frequencies of the protons become linearly related to position along the z-axis. Individual resonance frequencies correspond to individual planes of nuclei.

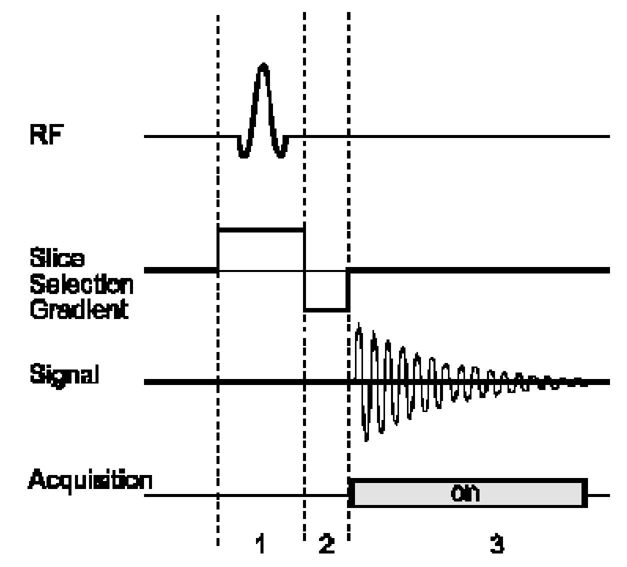
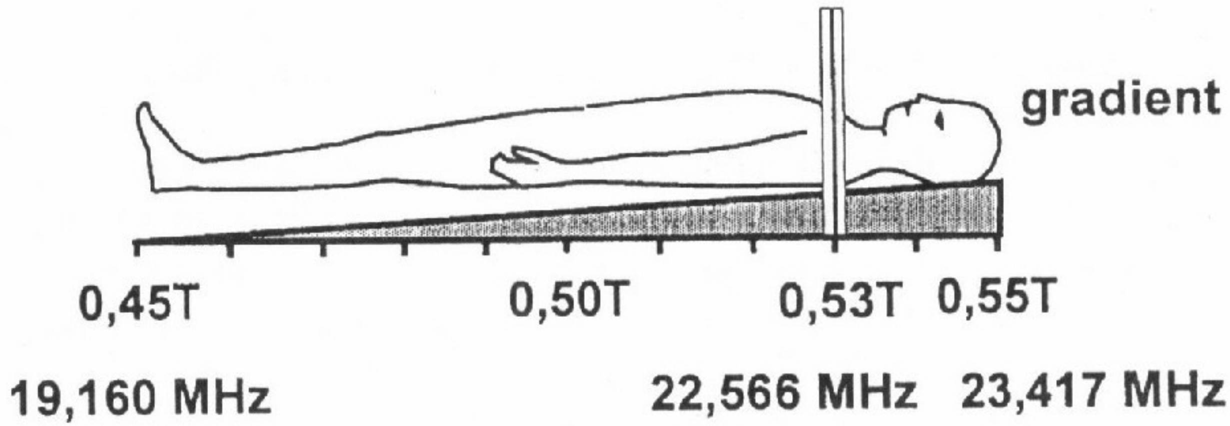
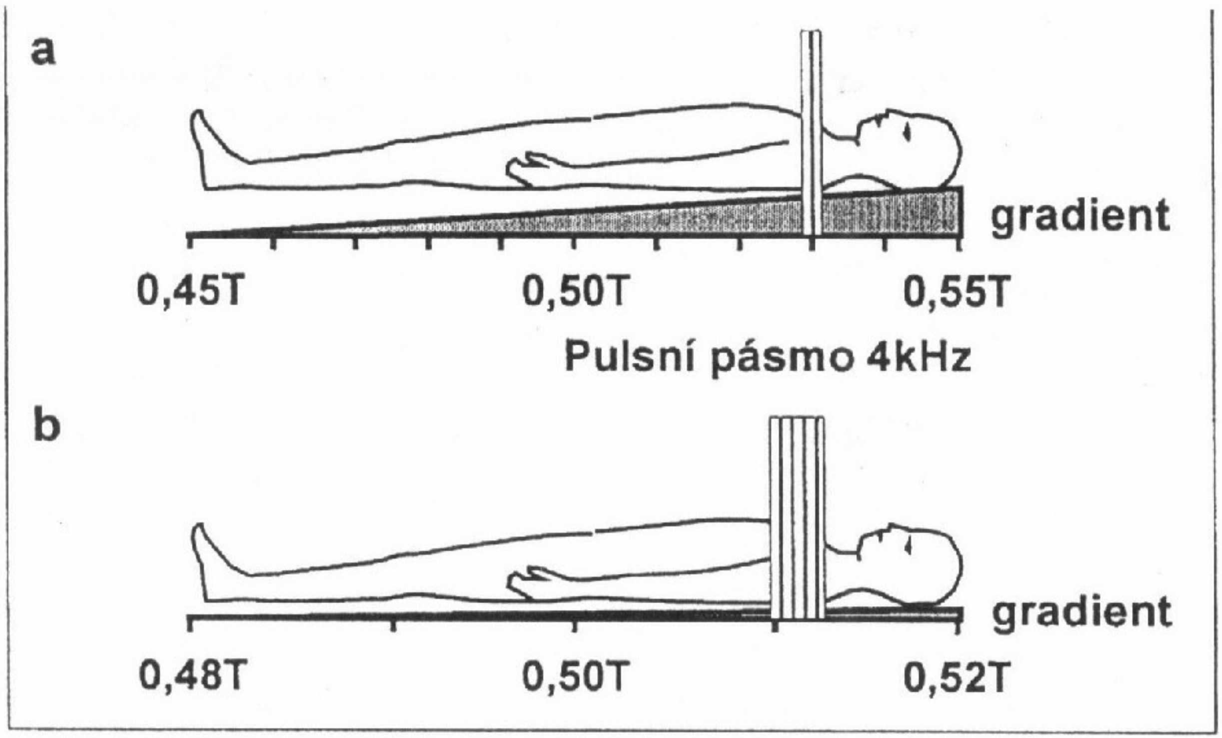


Figure 9.7: The sequence of RF power and gradient strength used for slice selection. To excite only one slice, a magnetic field gradient is applied during the excitation RF pulse.



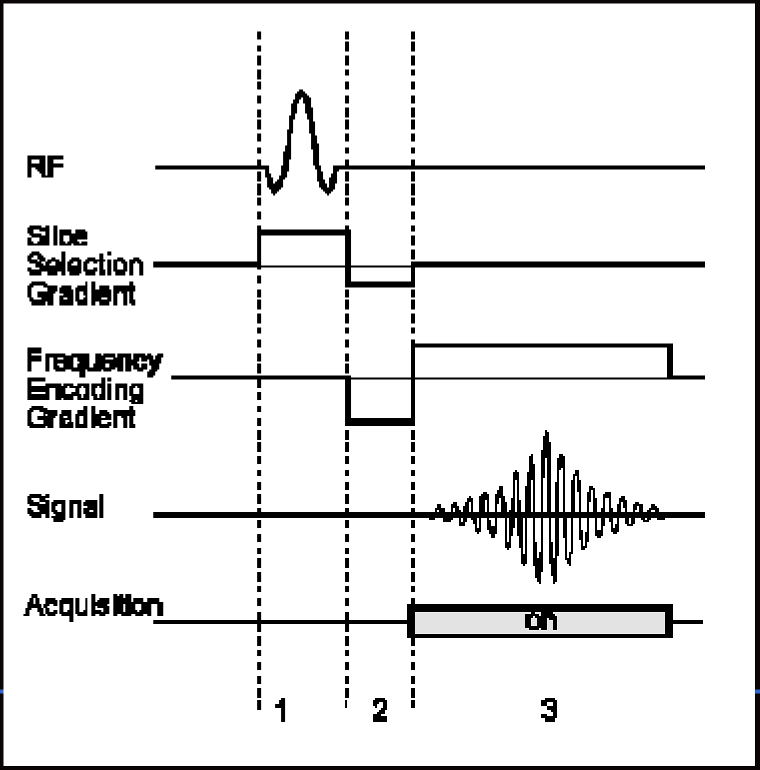


Dva možné postupy při volbě tloušťky tkáňového řezu

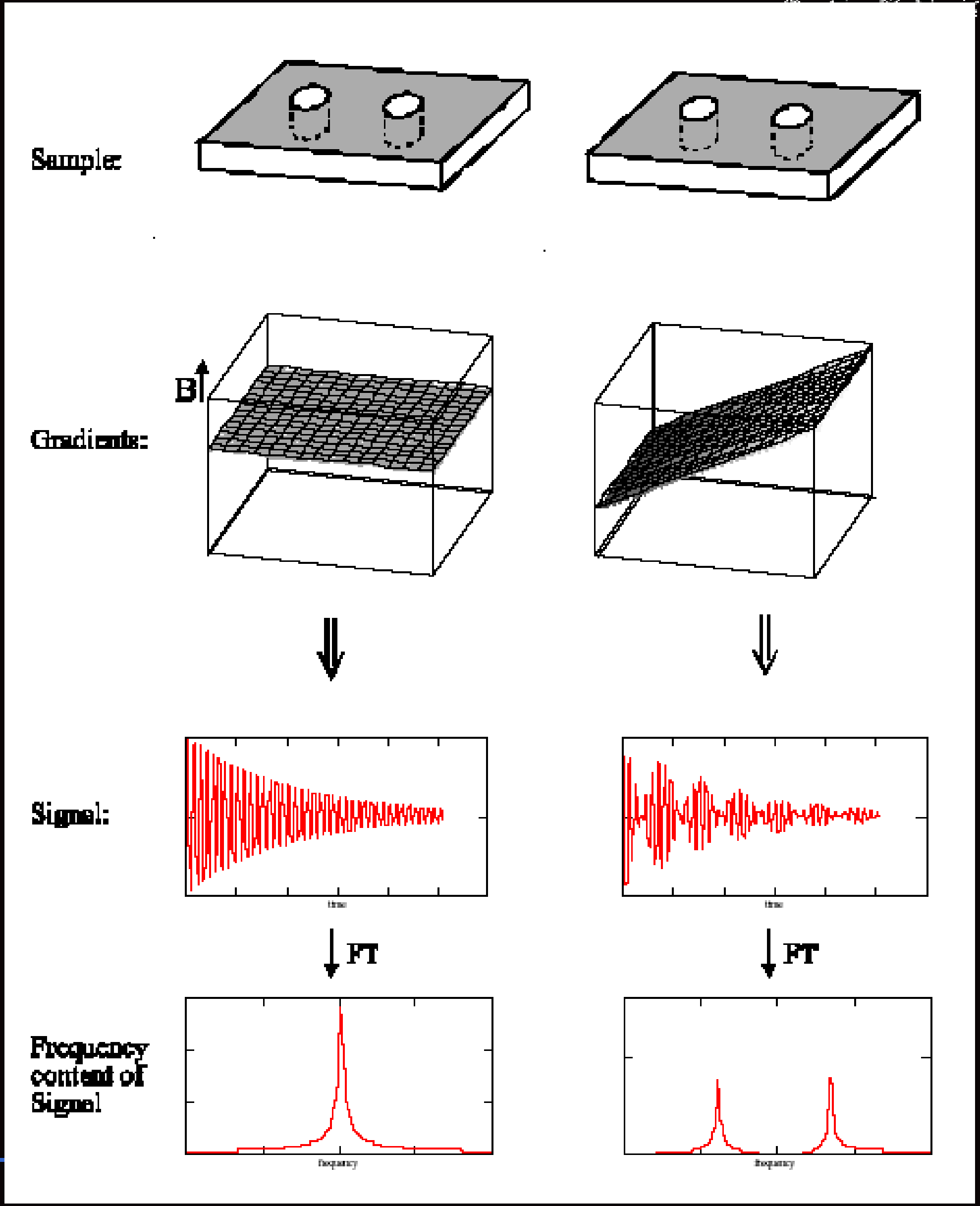
Frequency Encoding

- After slice selection, the next task is to distinguish signal from different spatial locations within this slice. This is accomplished in the x-direction by applying a gradient (Gx), the frequency-encoding gradient, during the acquisition of the signal. (time period 3 in Figure 9.11).

The sequence of RF power and gradient amplitudes used to excite one slice and encode the positions of the spins within that slice into the signal. In this "frequency encoding," the positions of the spins encoded by applying a magnetic field gradient in one of the directions in the excited slice during the acquisition.



A series of steps illustrating the concept of frequency encoding to distinguish the signal coming from two point sources of magnetization, e.g. small vials of water, in an object.*(left)* When no gradient is applied, both sources of magnetization resonate at the same frequency and the signal is a simple decaying sinusoid. When this signal is Fourier transformed, the signal is shown to contain only one frequency.*(right)*
When a gradient is applied, one of the sources of magnetization precesses at a higher frequency than the other. The resulting signal is an interference pattern of the two frequencies and is shown to contain by Fourier transformation to contain two distinct frequencies. Notice that the Fourier transformed signal is the projection of the amount of magnetization along the axis along which the gradient was applied.
That is, in this one dimensional case, the frequency content of the signal *is* the image.

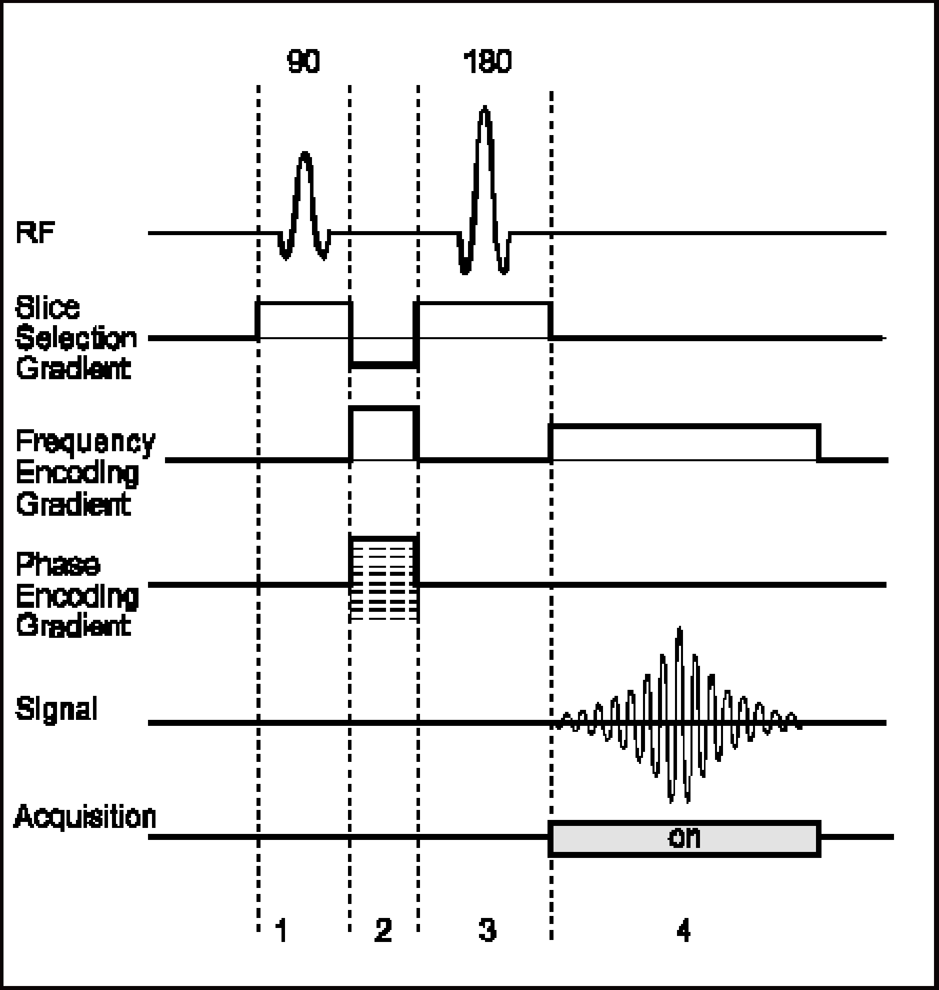


Phase Encoding

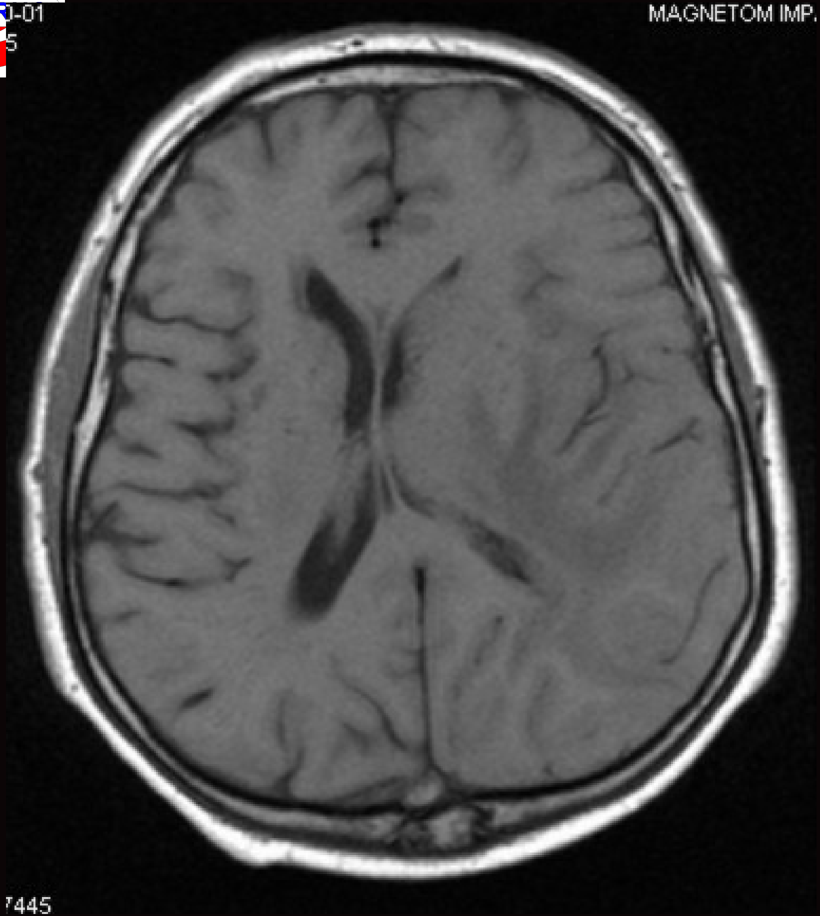
- Suppose a slice through a homogeneous sample has been selected and excited as described in Slice Selection section, and then frequency encoded according to the previous section. After a short time, the phase of the spins at one end of the gradient leads those at the other end because they are precessing faster. If the frequency encoding gradient is switched off, spins precess (once more) at the same angular velocity but with a retained phase difference. This phenomenon is known as phase memory.
- A phase encoding gradient is applied orthogonally to the other two gradients after slice selection and excitation, but before frequency encoding. The phase encoding gradient does not change the frequency of the received signal because it is not on during signal acquisition. It serves as a phase memory, remembering relative phase throughout the slice.

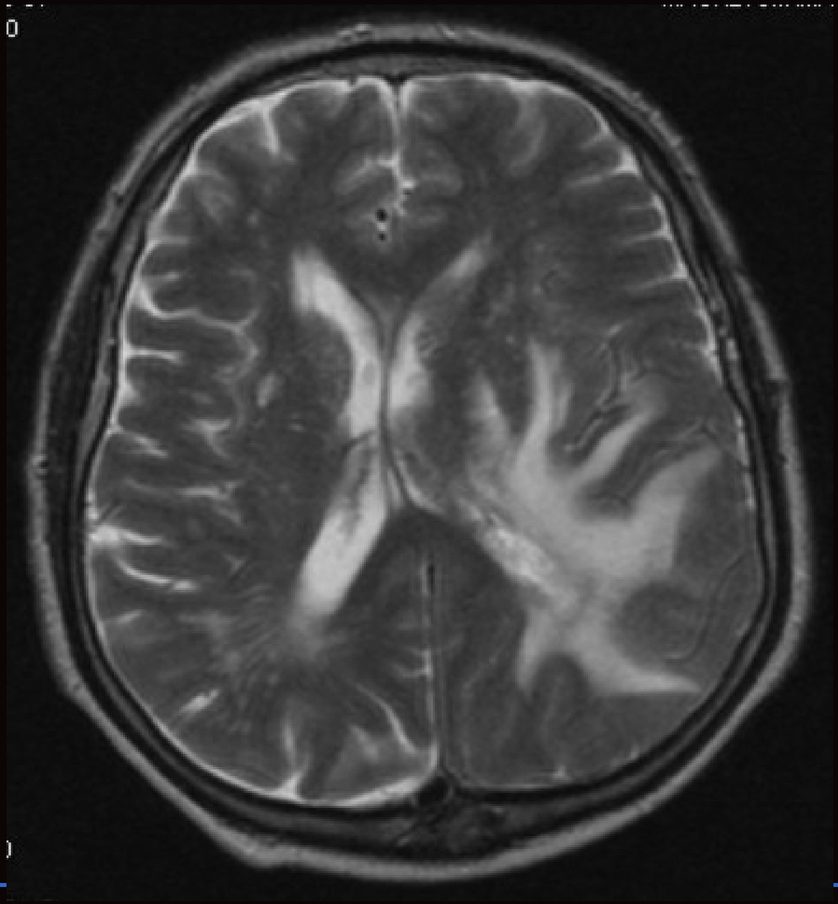
A complete pulse sequence diagram for the spin echo sequence.

Spatial locations of the spins are encoded into the signal by applying three orthogonal gradients, techniques that are called slice selection, frequency encoding, and phase encoding. In period 1, a 90 degree pulse and a slice selection gradient excite one slice. In period 2, the initial frequency encoding gradient and the phase encoding gradient are applied. In period 3, a 180 degree pulse is applied, along with a slice selection pulse (such that only the spins in the same excited slice are “flipped,”) and in period 4 the frequency encoding gradient is applied and the signal is acquired. The sequence shown here is repeated numerous times (128, 256, 512, etc. depending on the desired resolution) each time with a different strength of the phase encoding gradient.

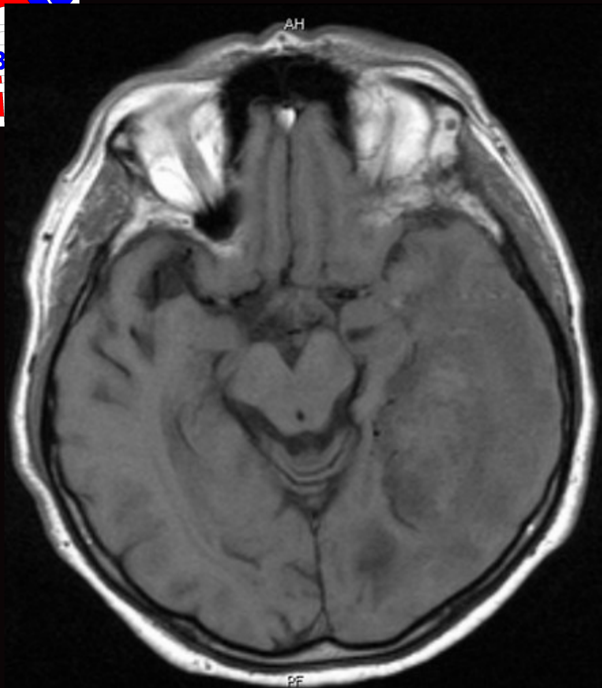


- Although the signal obtained from one acquisition (slice selection, phase encoding and frequency encoding) contains information from all voxels in the imaging slice, the information gathered from one iteration of this sequence is not sufficient to reconstruct an image. Consequently, the sequence has to be repeated with different settings of the phase-encoding gradient Gy.
- To construct a 256 x 256 pixels image a pulse sequence is repeated 256 times with only the phase encoding gradient changing.
- A Fourier transformation allows phase information to be extracted so that a pixel (x, y) in the slice can be assigned the intensity of signal which has the correct phase and frequency corresponding to the appropriate volume element. The signal intensity is then converted to a grey scale to form an image.

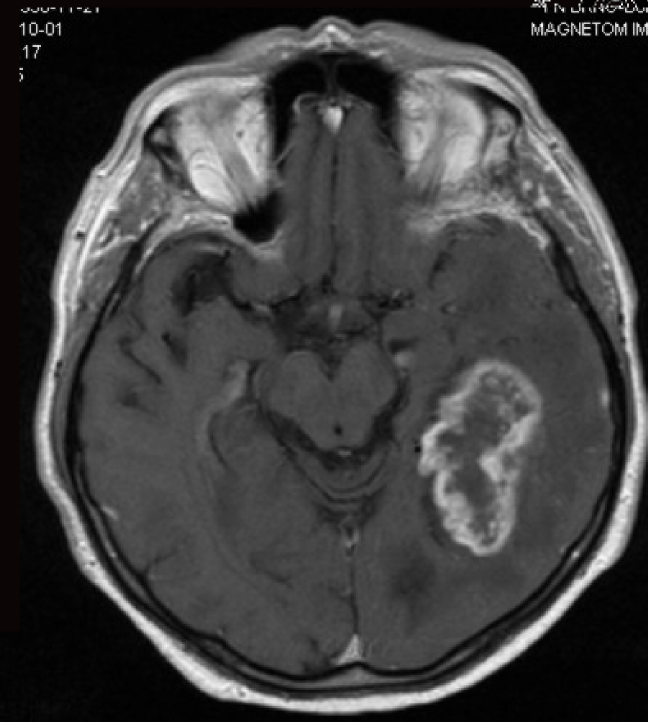


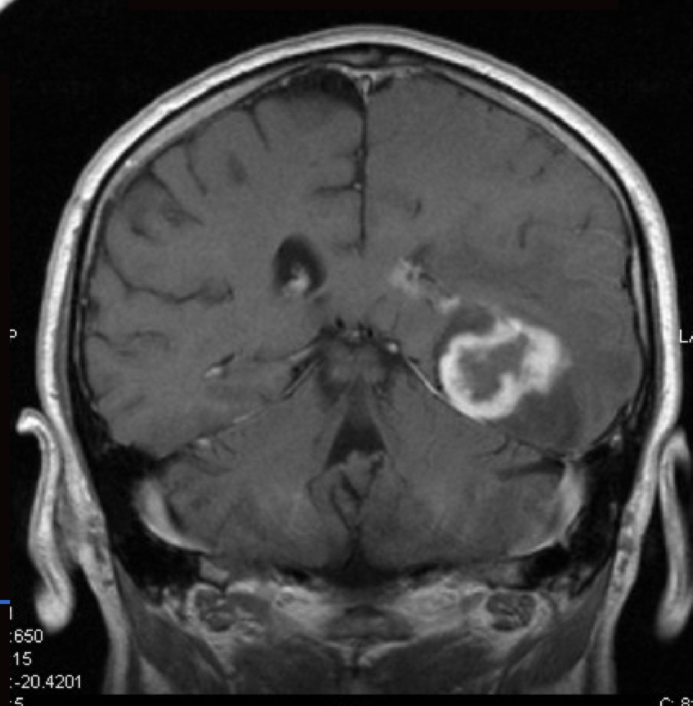


T1 weighted image(left) and
T2 weighted image (right),
axial plane – left brain
hemisphere edema



Precontrast T1 WI





Postcontrast T1 weighted images – cerebral tumor (glioblastoma) showing ring enhancement