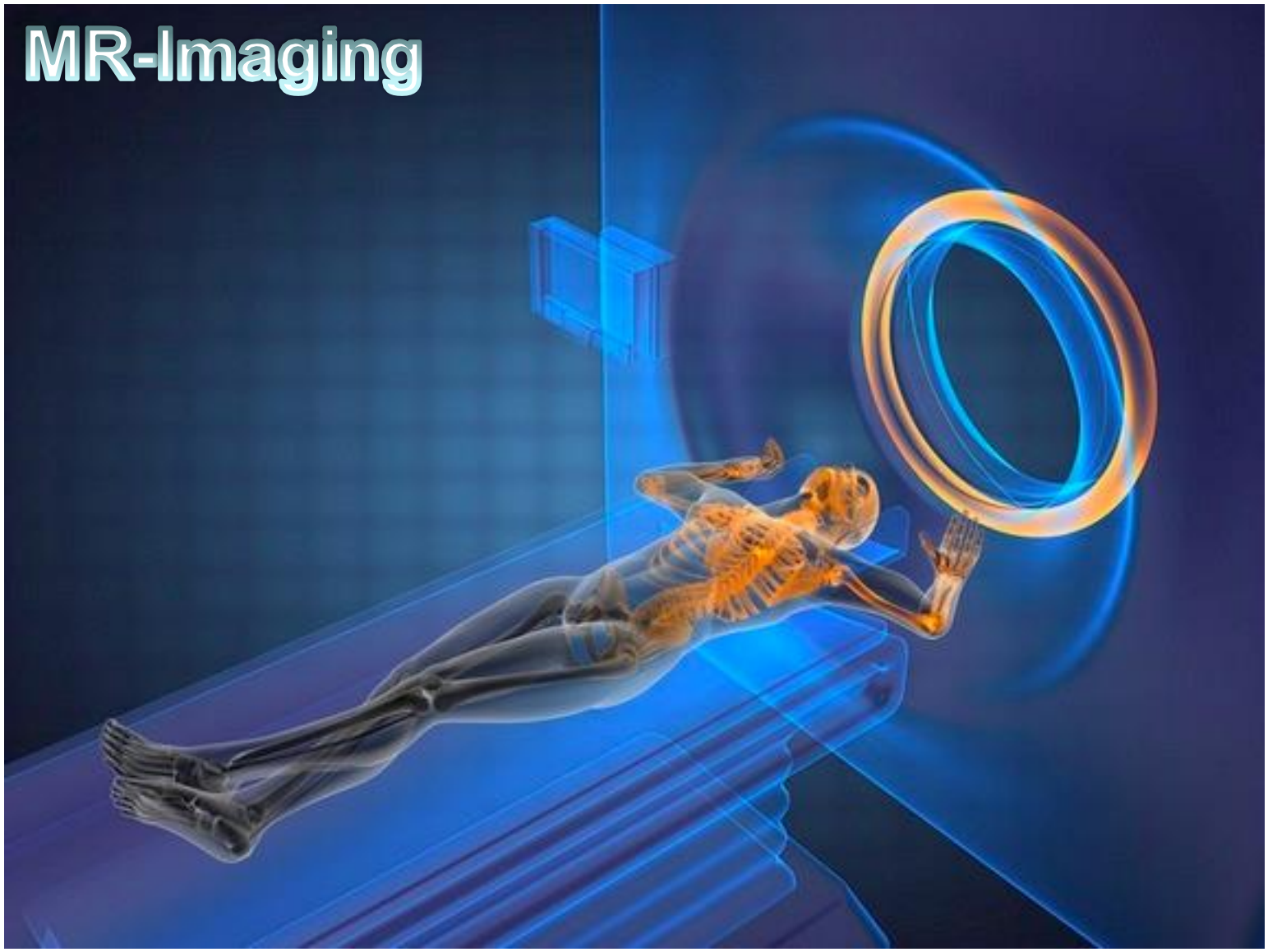


UiO : Department of Physics
University of Oslo

FYS 4250

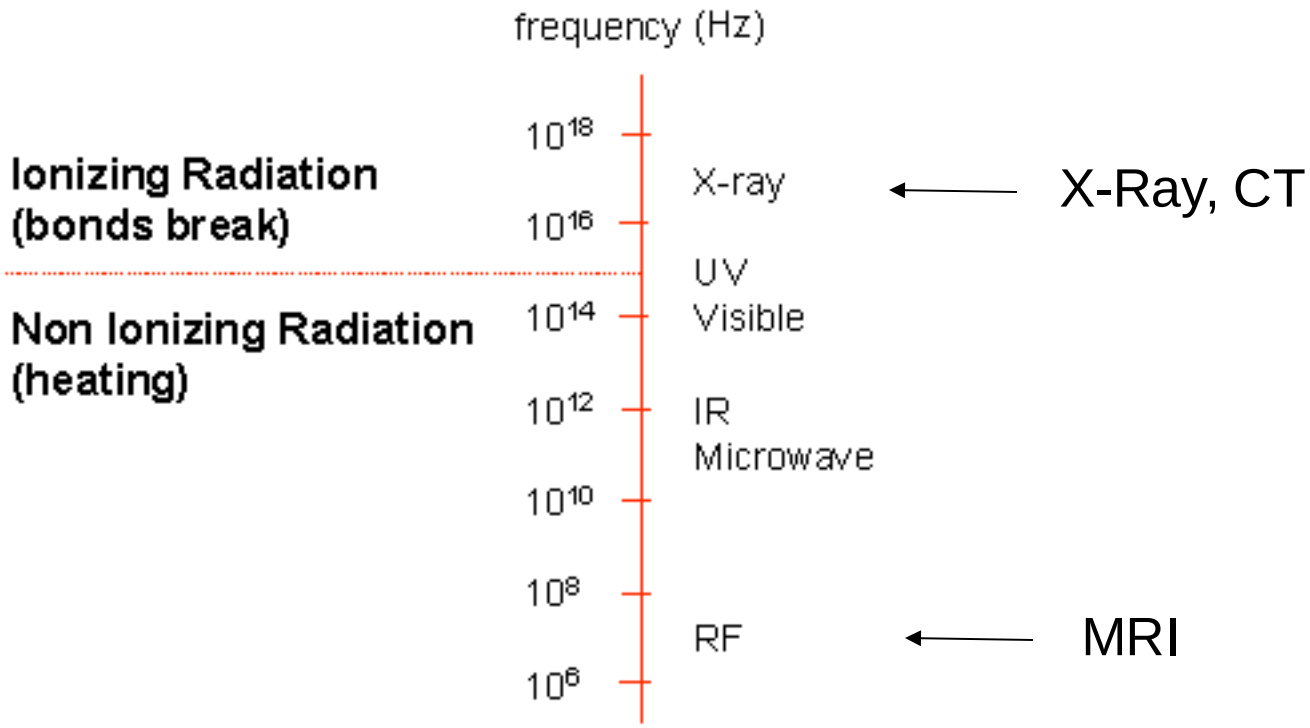
Lecture 14

Everything we see is a perspective, not the truth



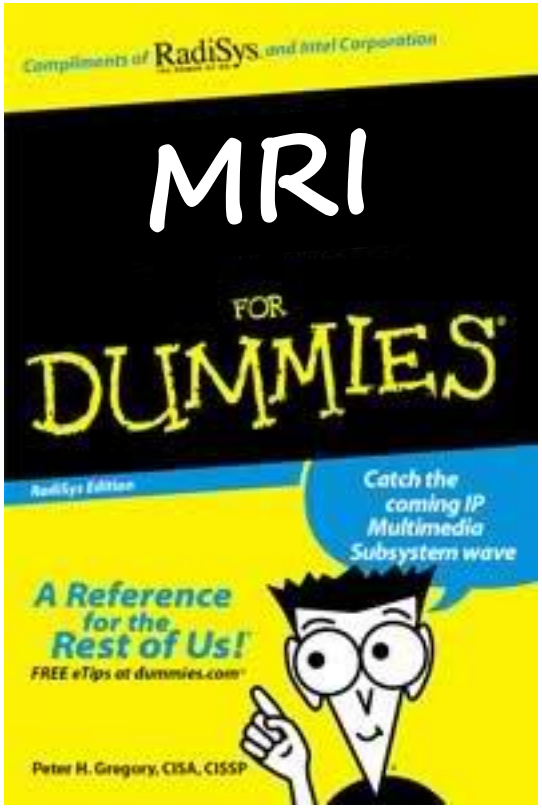
http://i.livescience.com/images/i/000/055/842/i02/MRI-scan-130415.jpg?1376591243

Electromagnetic Radiation Energy

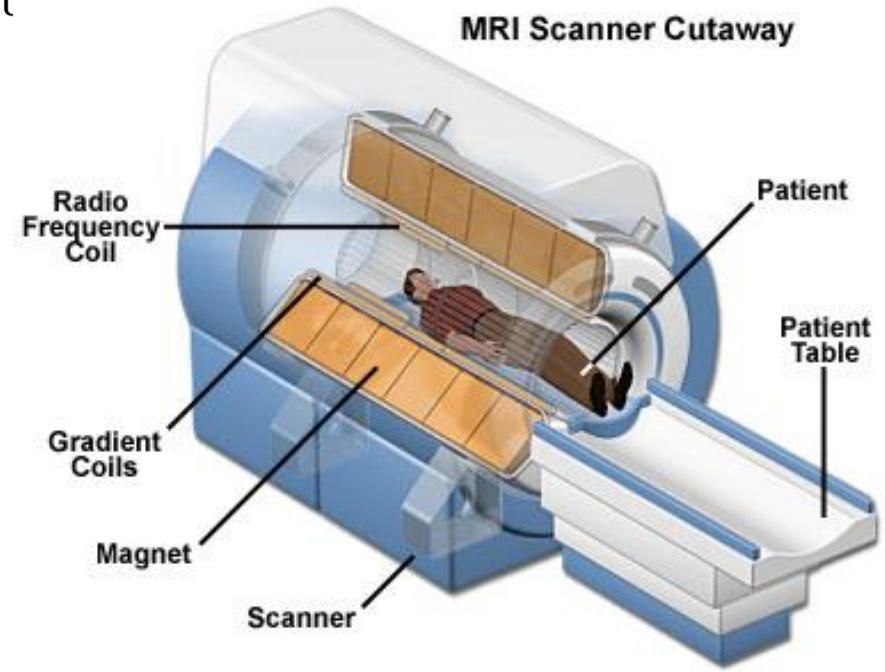


A brief recipe of MRI

1. Put the subject into a strong magnetic field
2. Pass radiowaves through the subject
3. Turn of the radiowaves
4. Recieve radiowaves coming back from the subject
5. Convert the measured RF-data to an image



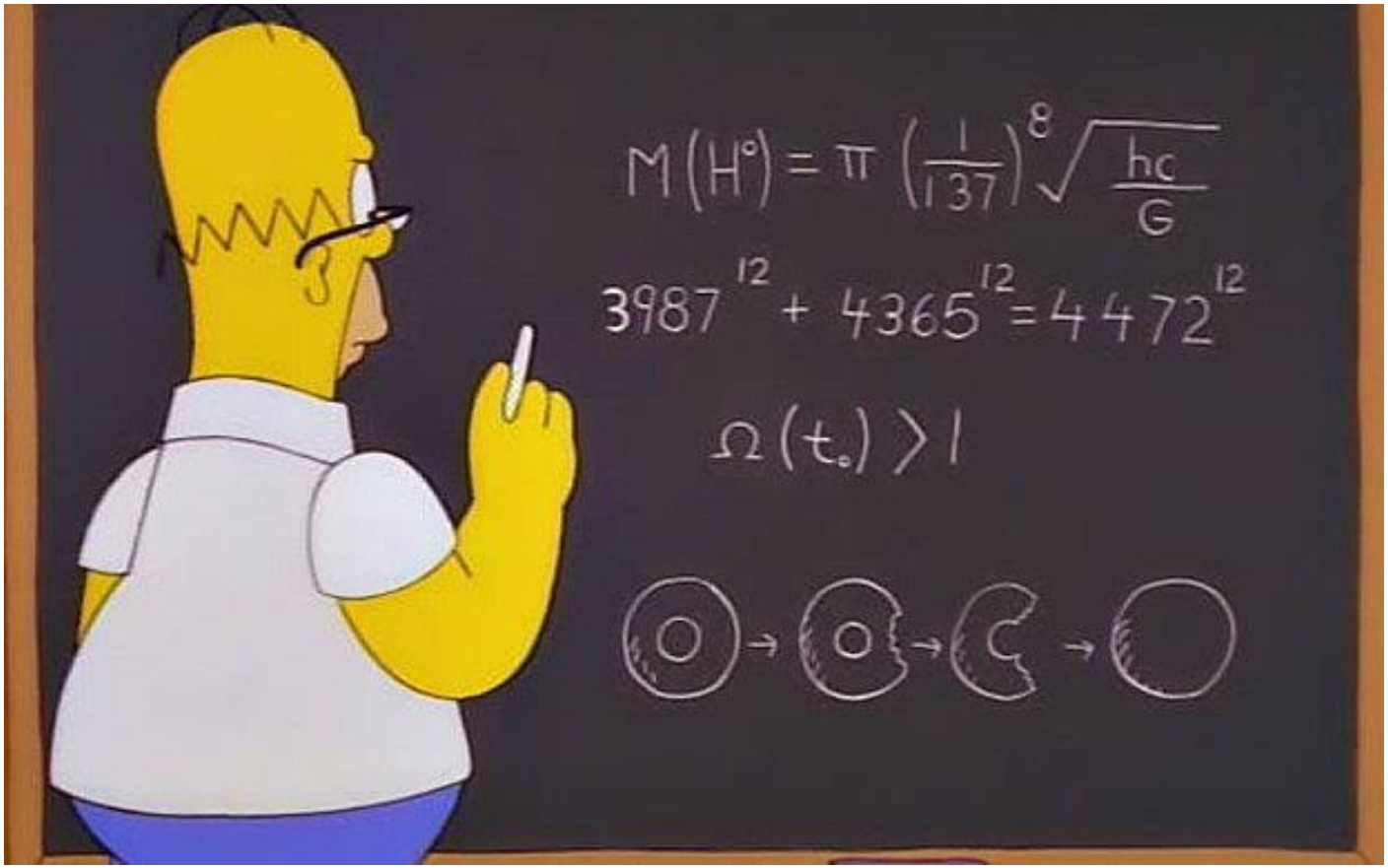
MRI





Source: Biomed resources

The basic physics of MRI



https://blog.nycdatascience.com/student-works/higgs-boson-machine-learning-challenge-3/

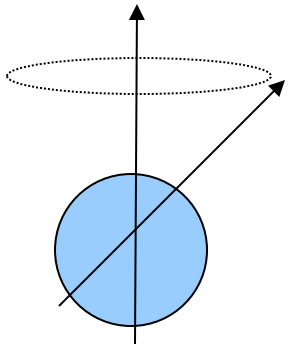
Prerequisites for depicted nucleus

- A nucleus that is to be pictured must have both:
  - Spin
  - Charge

Note! Nucleus with even protonnumbers cannot be used because the spin will cancel each other

Single-proton

- A single proton has a charge on the surface which is sufficient to form a small current-loop and generates a magnetic momentum μ

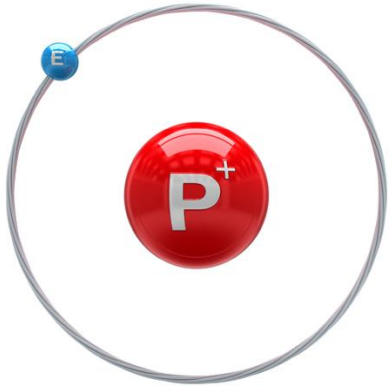


- The proton has also a mass that creates an angle-moment J due to the spin

Hydrogenatoms

- The hydrogenatom is the only large element in the body able to be depicted with MRI. (C, O and N have all even numbers in the proton number).
- Hydrogen is everywhere in the body, primarily combined to water

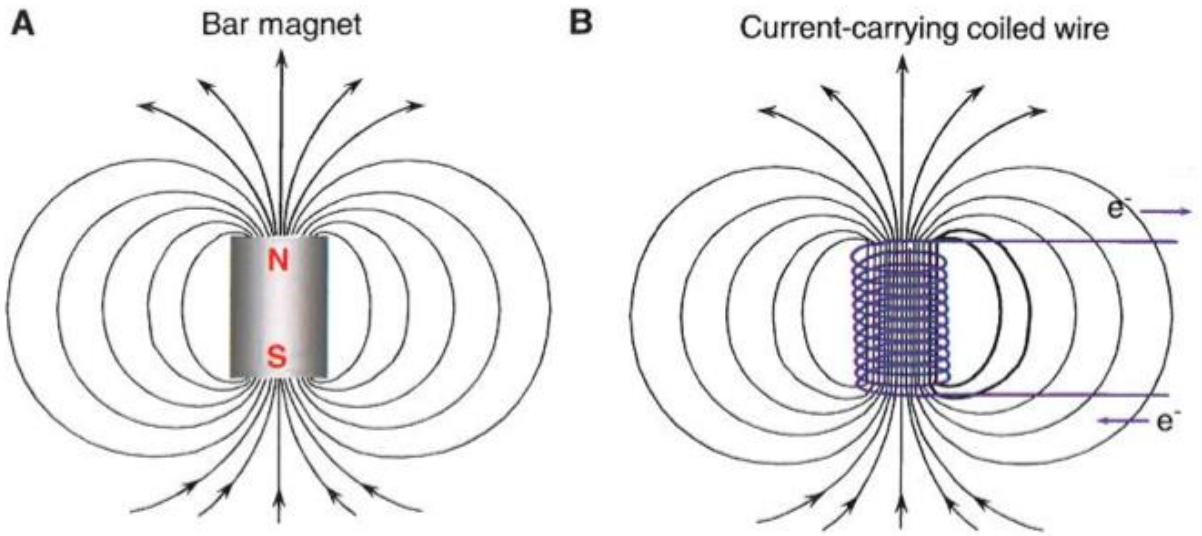
= All MRI-images are in fact a picture of hydrogen



Source: http://theenergycollective.com/sites/theenergycollective.com/files/imagepicker/488516/hydrogen%20economy.jpg

Magnetism is a fundamental property, generated by moving charges. Magnetic field is induced by a moving charge in a cable, use right hand rule to find the direction of the magnetic field

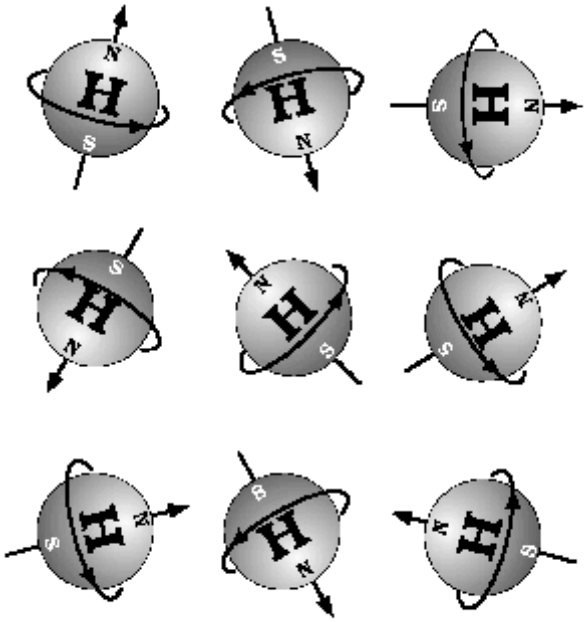
Magnetic fields



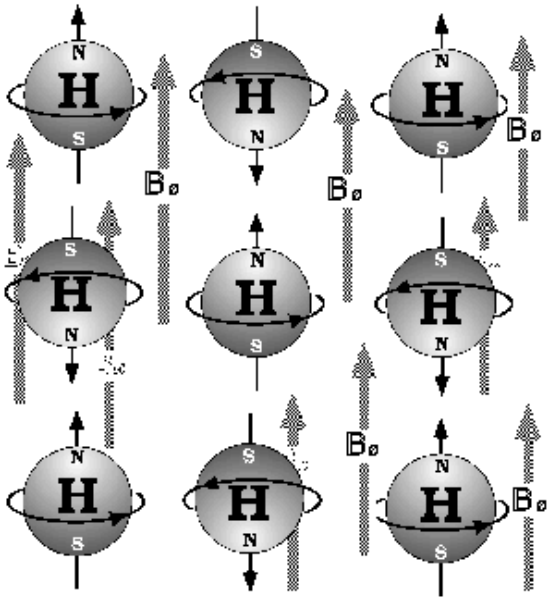
Dipole magnetic field

FIGURE 12-1 **A.** The magnetic field has two poles with magnetic field lines emerging from the north pole (*N*), and returning to the south pole (*S*), as illustrated by a simple bar magnet. **B.** A coiled wire carrying an electric current produces a magnetic field with characteristics similar to a bar magnet. Magnetic field strength and field density are dependent on the amplitude of the current and the number of coil turns.

Spinning Protons Act Like Little Magnets

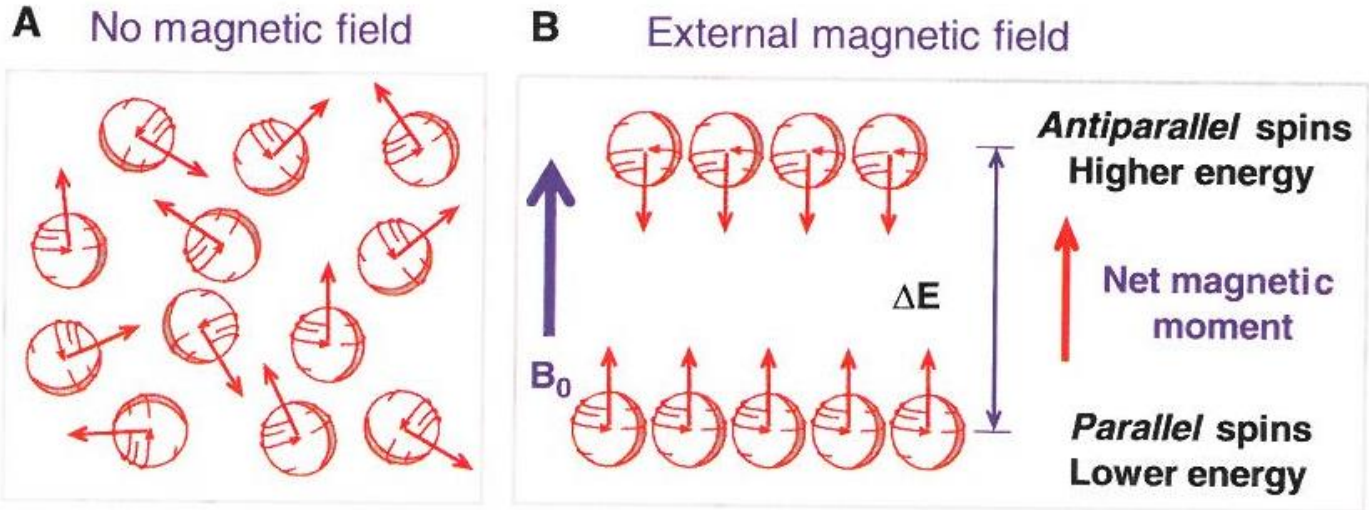


They Align With An External Magnetic Field (Bø)



- Ref:www.simplyphysics.com

Energy levels of free protons

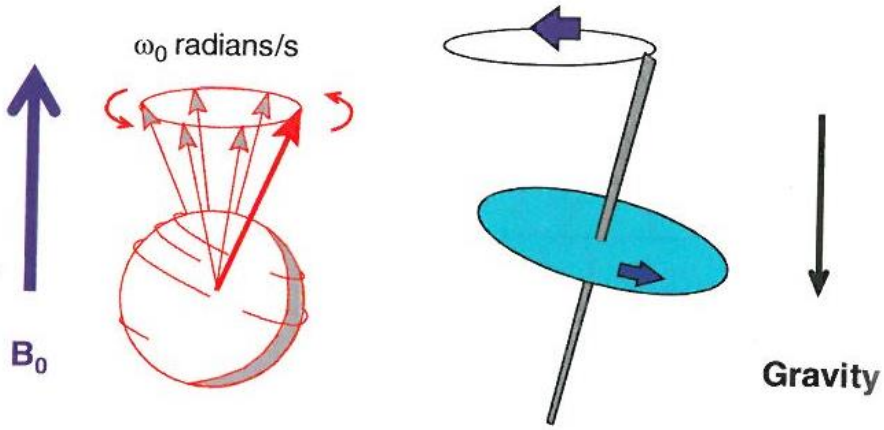


■ **FIGURE 12-7** Simplified distributions of "free" protons without and with an external magnetic field are shown. **A.** Without an external magnetic field, a group of protons assumes a random orientation of magnetic moments, producing an overall magnetic moment of zero. **B.** Under the influence of an applied external magnetic field, B_0, the protons assume a nonrandom alignment in two possible orientations: parallel and antiparallel to the applied magnetic field. A slightly greater number of protons exist in the parallel direction, resulting in a measurable *net magnetic moment* in the direction of B_0.

Magnetic characteristics of nucleus

$W_0 = \gamma B_0$

■ **FIGURE 12-8** A single proton *precesses* about its axis with an angular frequency, ω, proportional to the externally applied magnetic field strength, according to the *Larmor* equation. A well-known example of precession is the motion a spinning top makes as it interacts with the force of gravity as it slows.

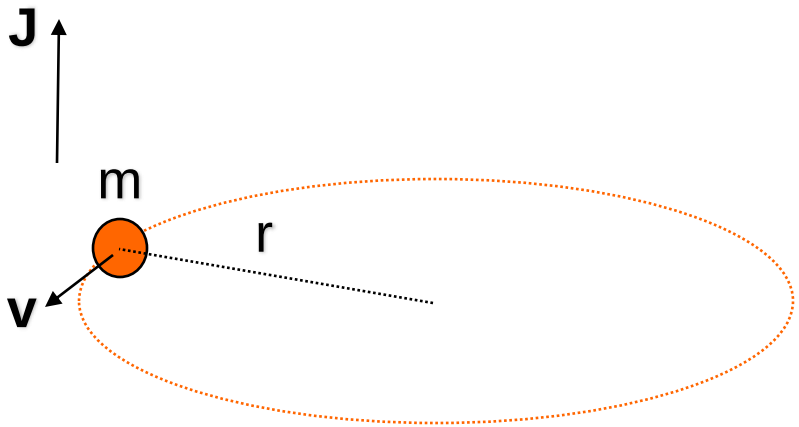


In addition to spin, there is a torque in a perpendicular direction from the applied magnetic field that causes precession. The Larmor equation describes the dependence between the magnetic field B0 and the angular precessional frequency w0.

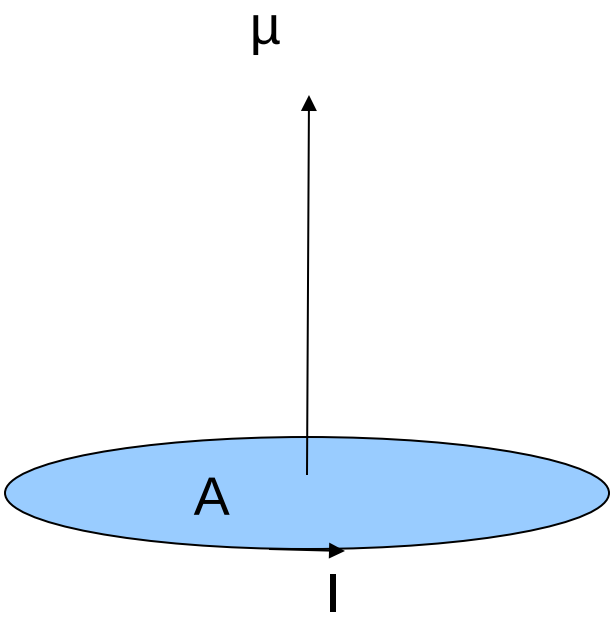
$$f_0 = \frac{\gamma}{2\pi} B_0$$

Angle momentum

$$\mathbf{J} = m\omega = m\mathbf{v}r$$

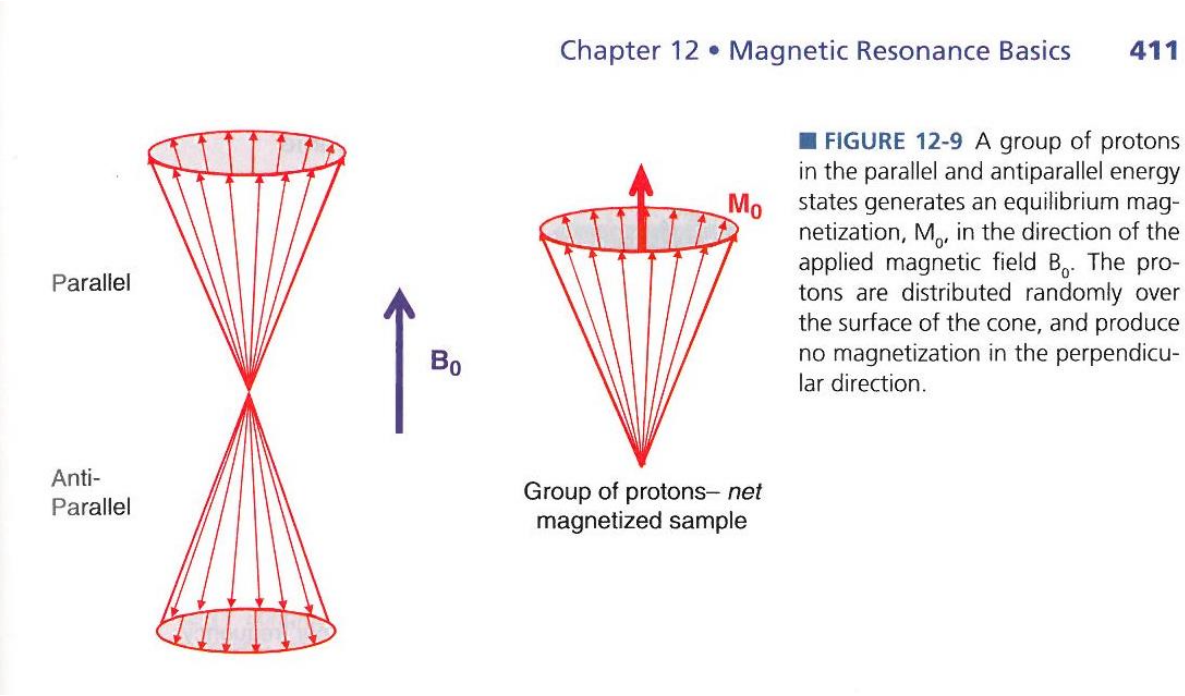


Magnetic momentum



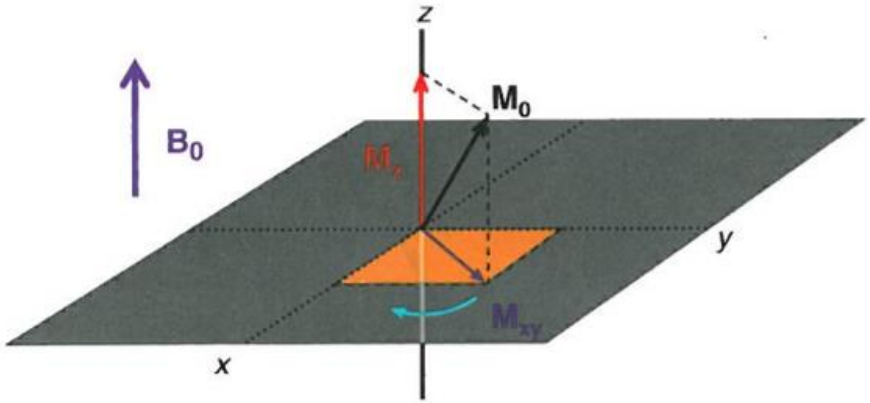
The magnetic momentum vector μ=IA

Equilibrium magnetization



The net magnetic moment at equilibrium is equal to the vector sum of the individual magnetic moments in the direction of B0. Perpendicular contributions are randomly distributed and sum to zero

Magnetization in three planes



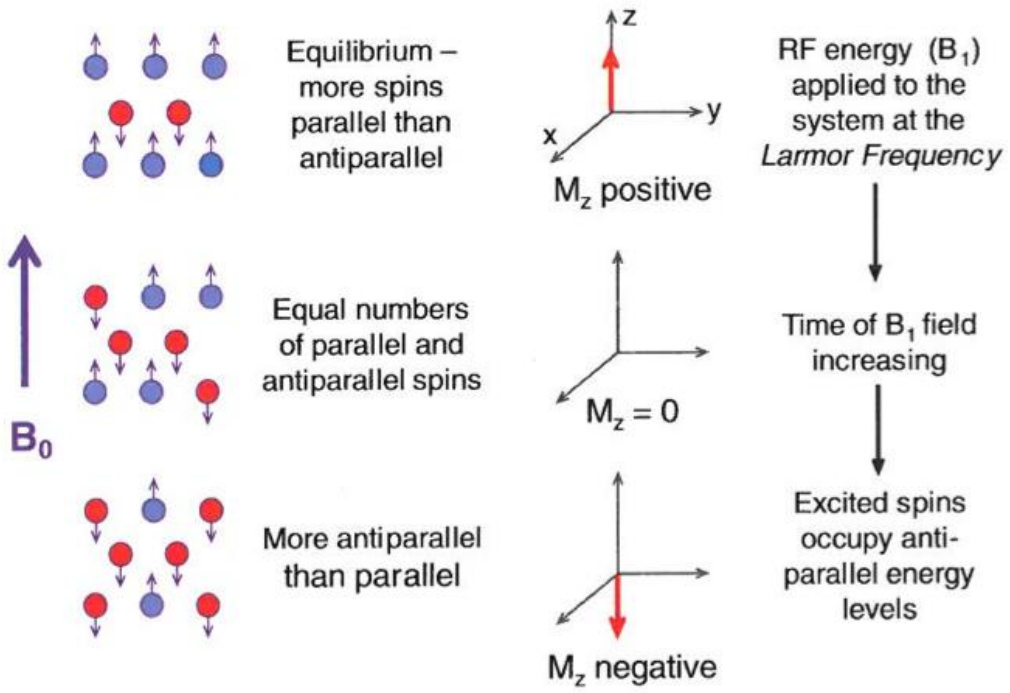
M_z: Longitudinal Magnetization: in z-axis direction

M_{xy}: Transverse Magnetization: in x-y plane

M_0: Equilibrium Magnetization: maximum magnetization

FIGURE 12-11 *Longitudinal magnetization,* M_z, is the vector component of the magnetic moment in the z direction. *Transverse magnetization,* M_{xy}, is the vector component of the magnetic moment in the *x-y* plane. *Equilibrium magnetization,* M_0, is the maximal longitudinal magnetization of the sample, and is shown displaced from the z-axis in this illustration.

Resonance and exitation



■ FIGURE 12-12 A simple quantum mechanics process depicts the discrete energy absorption and the time change of the longitudinal magnetization vector as RF energy equal to the energy difference of the parallel and antiparallel spins is applied to the sample (at the Larmor frequency). Discrete quanta absorption changes the proton energy from parallel to antiparallel. With continued application of the RF energy at the Larmor frequency, M_z is displaced from equilibrium, through zero, to the opposite direction (high energy state).

Larmour frequency

The energy difference between the two alignment states depends on the nucleus

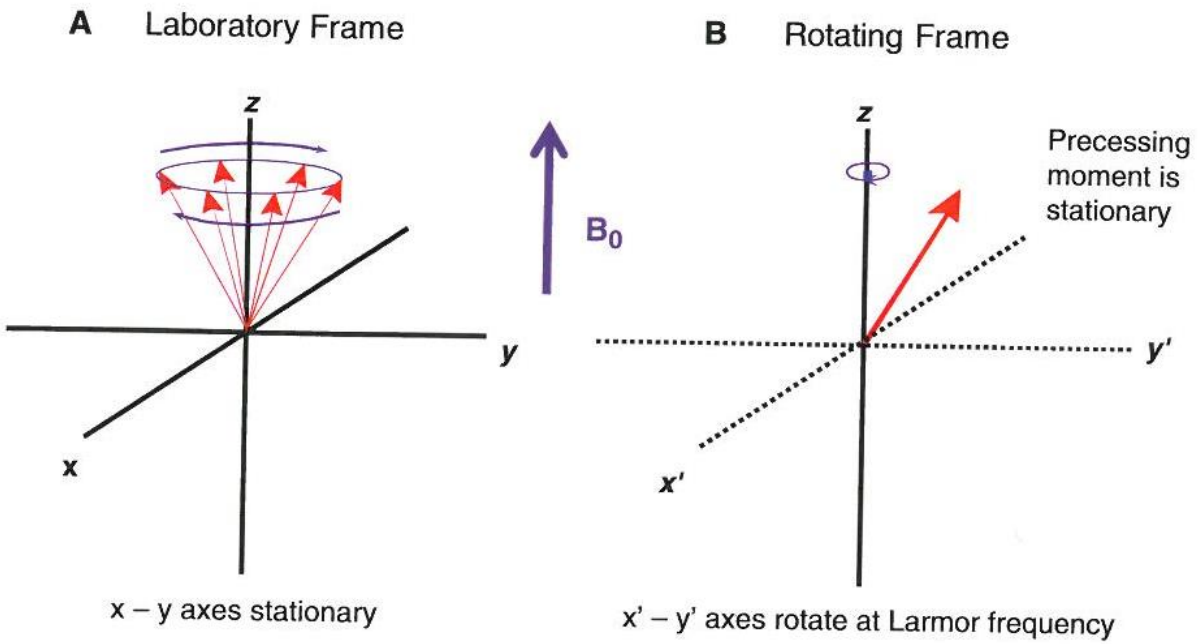
$$\Delta E = 2\,\mu_z\,B_o$$
$$\Delta E = h\,\nu$$

$$\nu = (\gamma/2\pi)\;B_o$$

known as Larmor frequency

$\gamma/2\pi$ **= 42.57 MHz / Tesla for proton**

Rotating frame

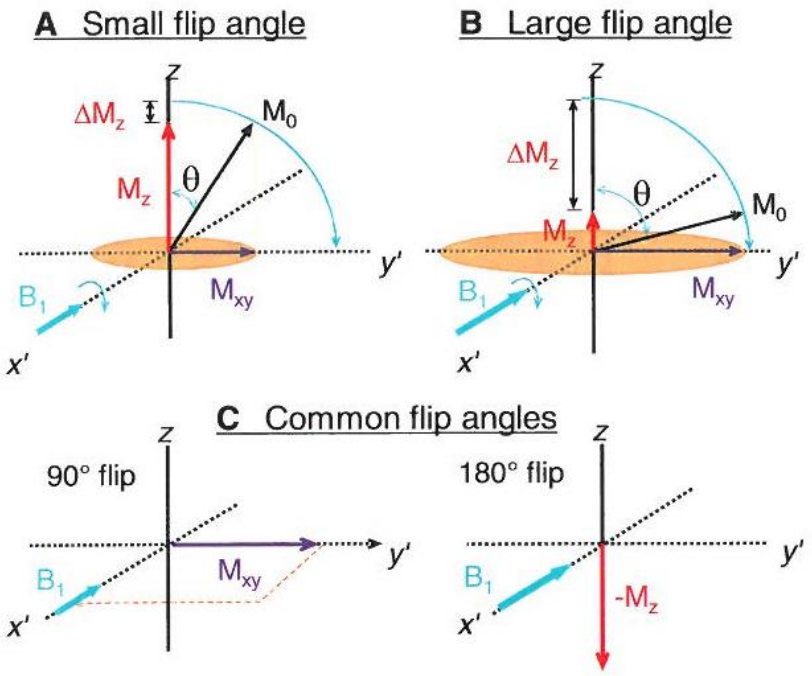


■ **FIGURE 12-10** **A.** The *laboratory frame of reference* uses stationary three-dimensional Cartesian coordinates: x, y, z. The magnetic moment precesses around the z-axis at the Larmor frequency as the illustration attempts to convey. **B.** The *rotating frame of reference* uses rotating Cartesian coordinate axes that *rotate about the z-axis* at the Larmor precessional frequency, and the other axes are denoted: x' and y'. When precessing at the Larmor frequency, the sample magnetic moment is stationary.

Bushberg et al, The essential physics of medical imaging

Flip angles is the degree of M_z rotation by the field B1 as it is applied along the x-axis or the y-axis perpendicular to Mz

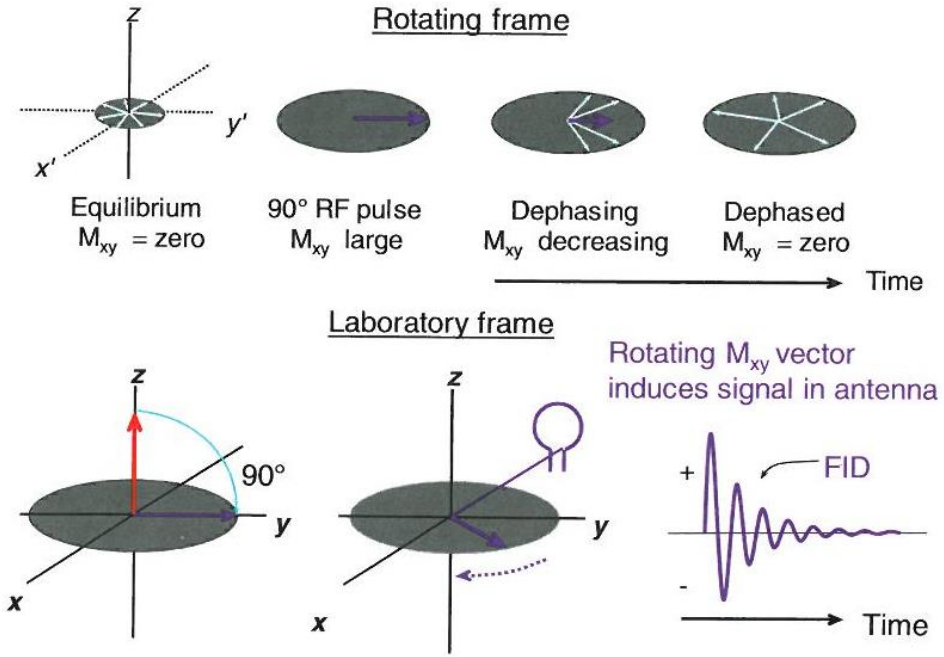
Flip angles



■ **FIGURE 12-14** Flip angles are the result of the angular displacement of the longitudinal magnetization vector from the equilibrium position. The rotation angle of the magnetic moment vector is dependent on the duration and amplitude of the B_1 field at the Larmor frequency. Flip angles describe the rotation of M_z away from the z-axis. **A.** Small flip angles (less than 45 degrees) and **B.** Large flip angles (75 to 90 degrees) produce small and large transverse magnetization, respectively.

Free induction decay = a damped sinusoidal electronic signal

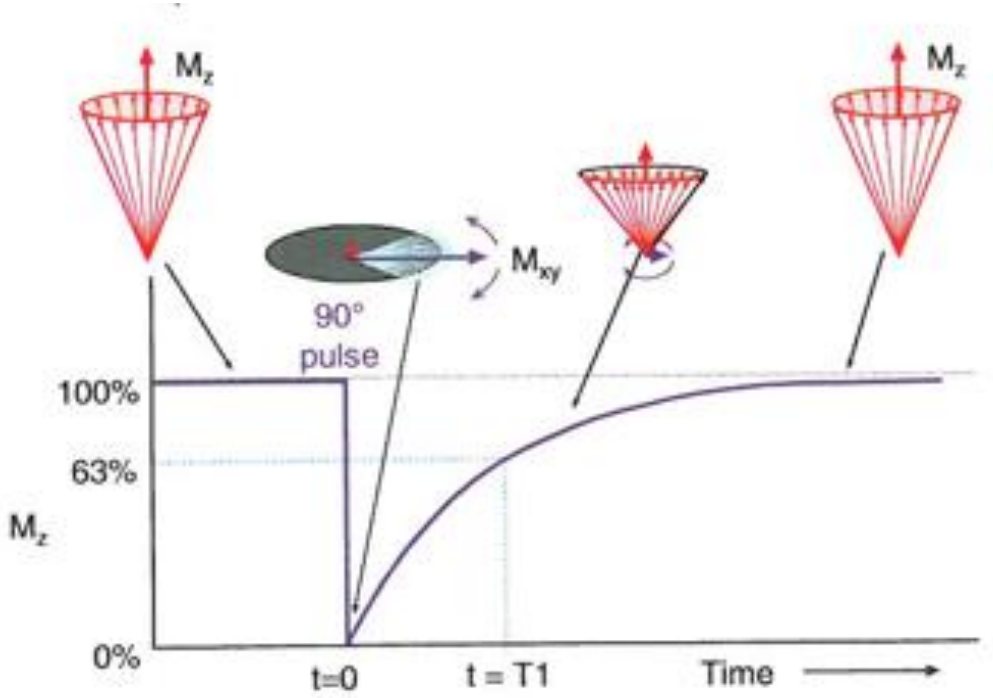
Dephasing of magnetic vector



■ **FIGURE 12-15** **Top.** Conversion of longitudinal magnetization, M_z, into transverse magnetization, M_{xy}, results in an initial phase coherence of the individual spins of the sample. The magnetic moment vector precesses at the Larmor frequency (stationary in the rotating frame), and dephases with time. **Bottom.** In the laboratory frame, M_{xy} precesses and induces a signal in an antenna receiver sensitive to transverse magnetization. A FID signal is produced with positive and negative variations oscillating at the Larmor frequency, and decaying with time due to the loss of phase coherence.

Bushberg et al, The essential physics of medical imaging

Return to equilibrium, T1 relaxation

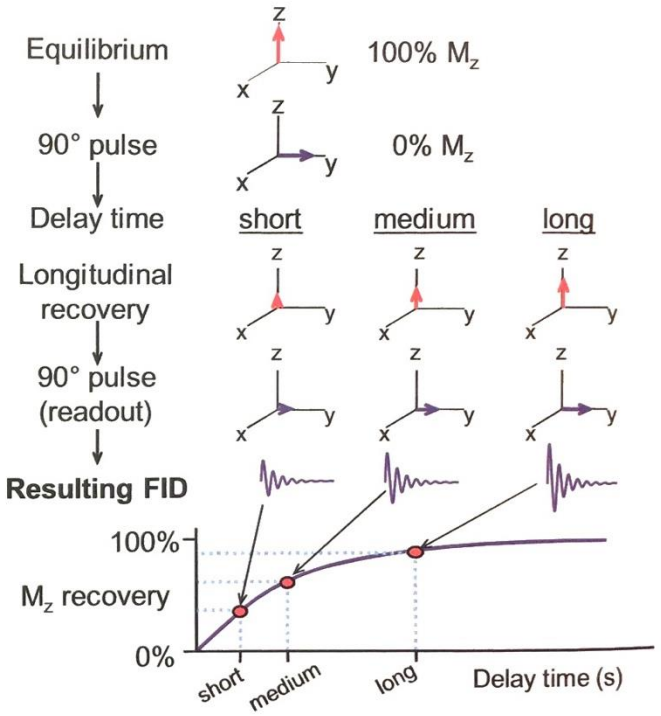


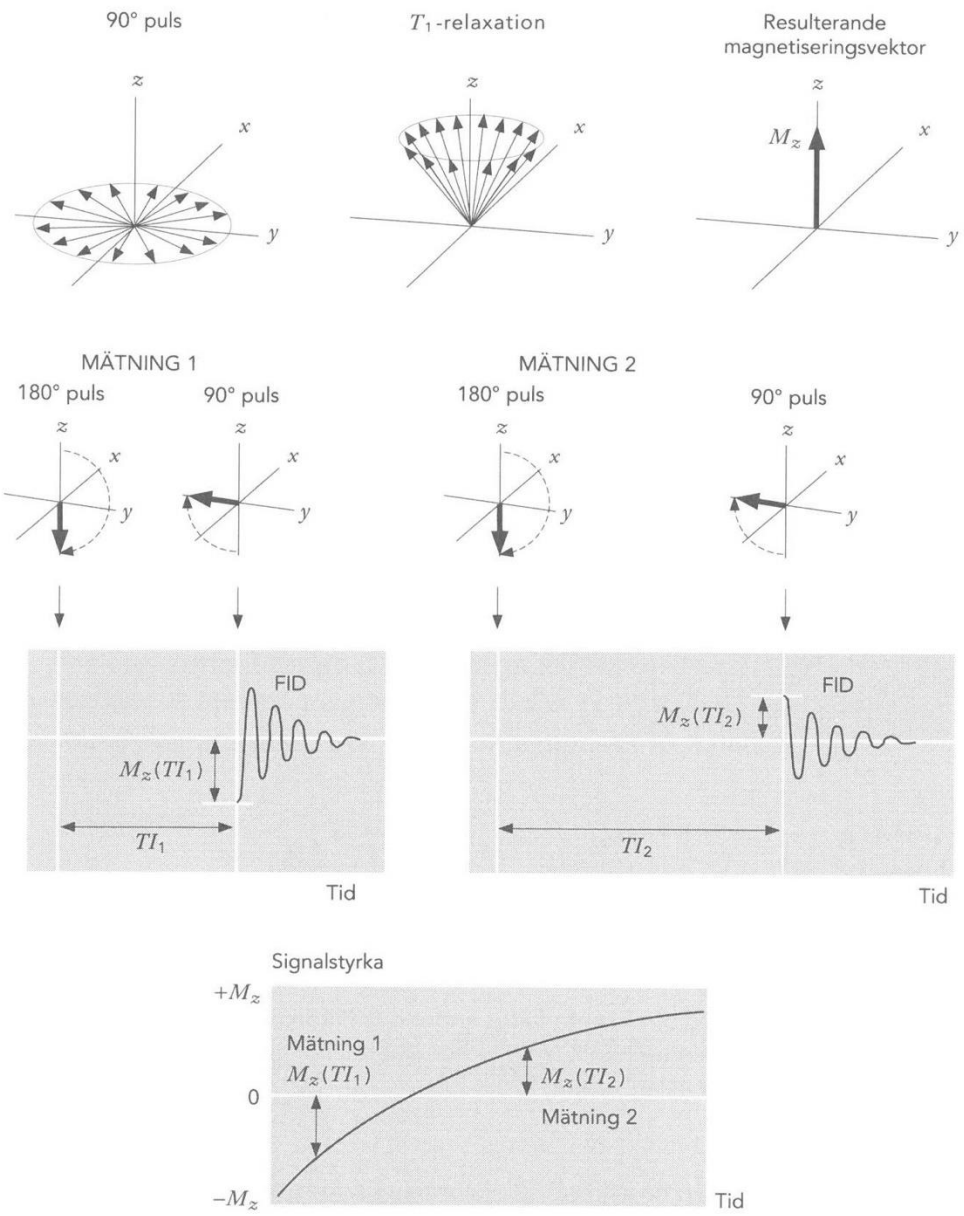
■ FIGURE 12-17 After a 90-degree pulse, M_z is converted from a maximum value at equilibrium to $M_z = 0$. Return of M_z to equilibrium occurs exponentially and is characterized by the spin-lattice T1 relaxation constant. After an elapsed time equal to T1, 63% of the longitudinal magnetization is recovered. Spin-lattice recovery takes longer than spin-spin decay (T2).

Bushberg et al, The essential physics of medical imaging

Return to equilibrium, T1 relaxation

■ **FIGURE 12-18** Spin-lattice relaxation for a sample can be measured by using various delay times between two 90-degree RF pulses. After an initial 90-degree pulse, $M_z = 0$, another 90-degree pulse separated by a known delay is applied, and the longitudinal magnetization that has recovered during the delay is converted to transverse magnetization. The maximum amplitude of the resultant FID is recorded as a function of delay times between initial pulse and readout (three different delay time experiments are shown in this example), and the points are fit to an exponential recovery function to determine T1.

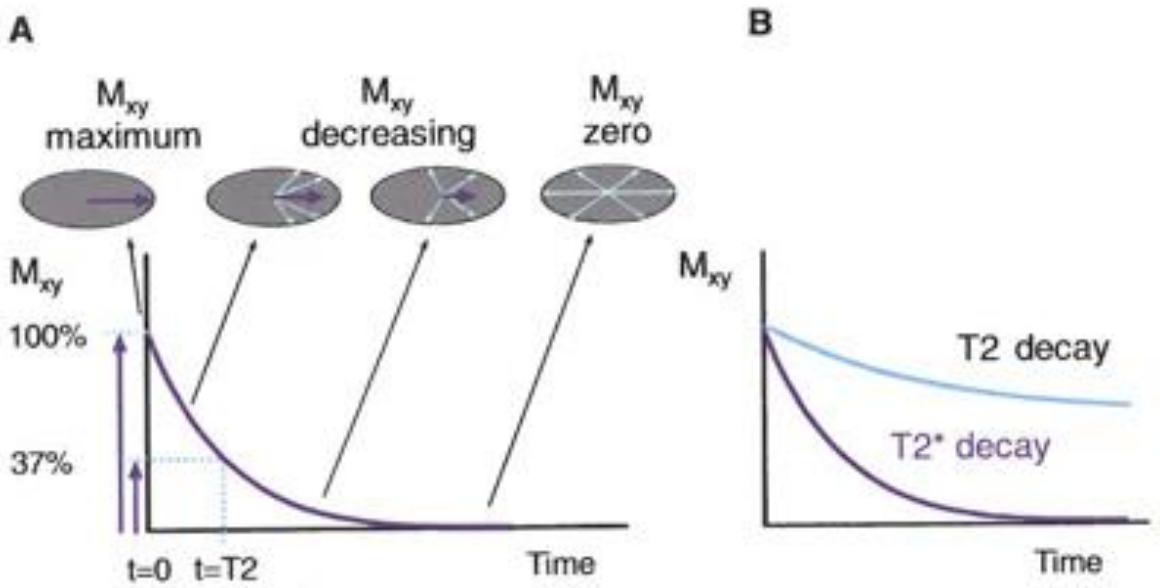




Figur 8:83. Överst visas hur T_1-relaxationen sker längs z-axeln (i urfasat tillstånd). Därunder visas hur T_1 mäts enligt metoden »inversion recovery». Först ges en 180° puls, som vrider spinnvektorerna i z-axelns negativa riktning. Efter tiden TI_1 ges en ny puls, som vrider spinnvektorerna 90° upp i xy-planet. Minst två mätningar erfordras för bestämning av relaxationskurvan. Full relaxation måste ske mellan mätningarna.

T_1 recording

Free induction decay: T2 Relaxation

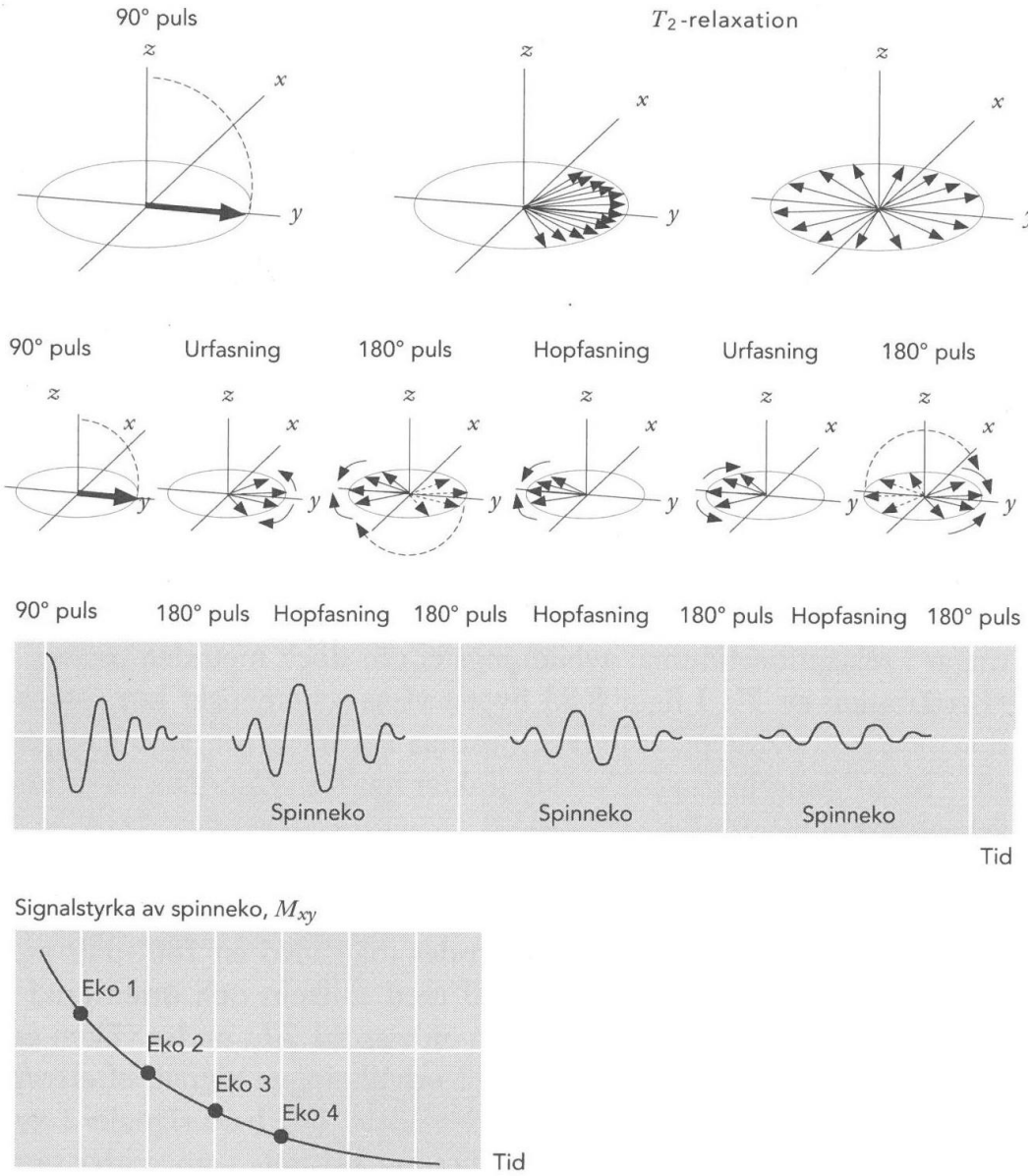


■ FIGURE 12-16 **A.** The loss of M_{xy} phase coherence occurs exponentially caused by intrinsic spin-spin interactions in the tissues and extrinsic magnetic field inhomogeneities. The exponential decay constant, T2, is the time over which the signal decays to 37% of the initial transverse magnetization (e.g., after a 90-degree pulse). **B.** T2 is the decay time resulting from *intrinsic* magnetic properties of the sample. T2* is the decay time resulting from *both intrinsic and extrinsic magnetic field variations*. T2 is always longer than T2*.

Bushberg et al, The essential physics of medical imaging

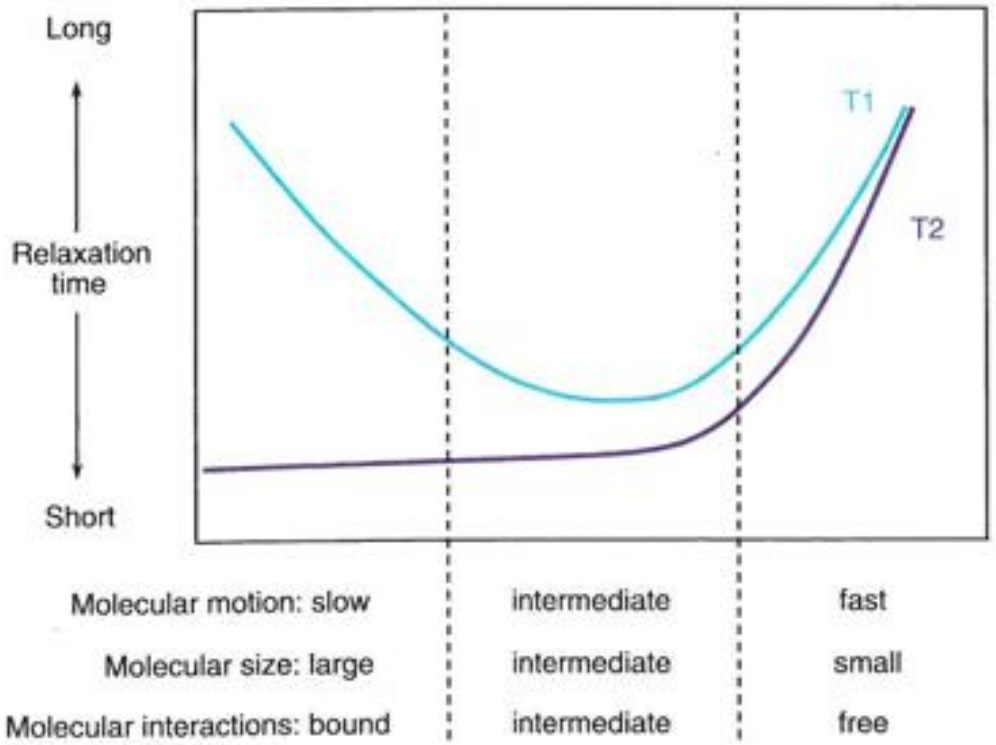
UiO

T_2 recording



Figur 8:84. Överst visas hur T_2-relaxationen sker i xy-planet. Därunder visas hur T_2 mäts med spinnekometoden. Först ges en 90° puls, som vrider spinnvektorerna ner till xy-planet. Urfasning börjar omedelbart. Efter vändning med en 180° puls kommer de snabbaste roterande vektorerna att hinna upp de långsammare, varför ett maximum uppstår i utsänd signal. Med hjälp av upprepade 180° pulser kan relaxationskurvan bestämmas.

Comparison of T1 and T2

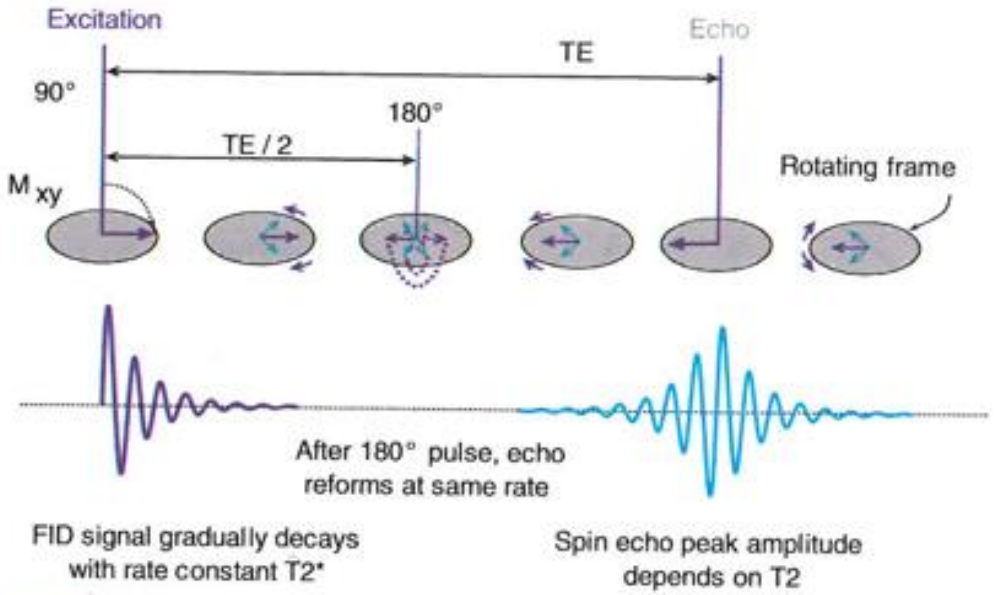


■ FIGURE 12-20 Factors affecting T1 and T2 relaxation times of different tissues are generally based on molecular motion, size, and interactions that have an impact on the local magnetic field variations (T2 decay) and structure with intrinsic tumbling frequencies coupling to the Larmor frequency (T1 recovery). The relaxation times (vertical axis) are different for T1 and T2.

Bushberg et al, The essential physics of medical imaging

Spin echo contrast weighting

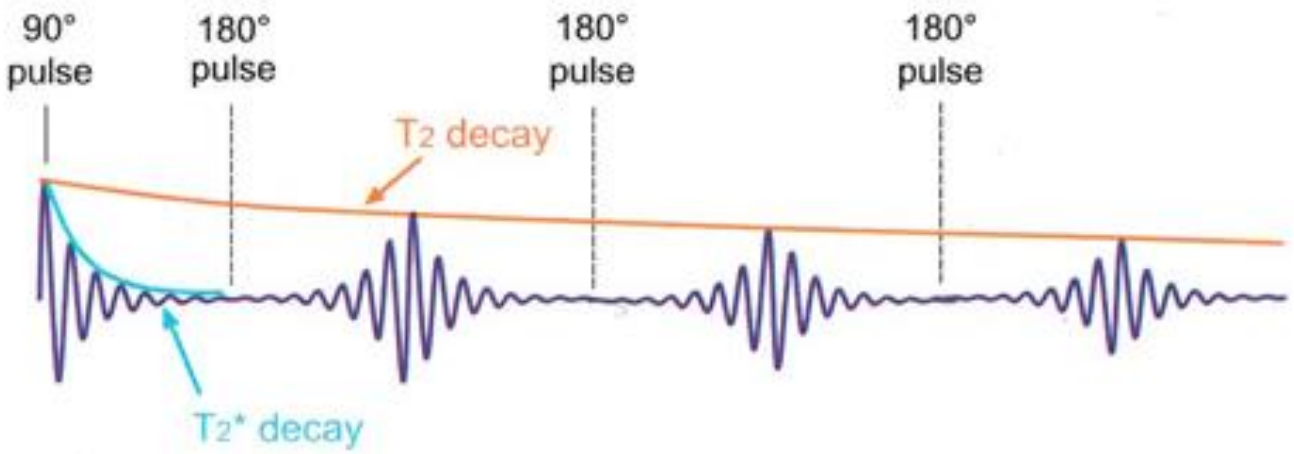
$$S \propto r_H \left[1 - e^{TR/T1}\right] e^{-TE/T2}$$



■ FIGURE 12-22 The SE pulse sequence starts with a 90-degree pulse and produces an FID that decays according to T2* relaxation. After a delay time TE/2, a 180-degree RF pulse inverts the spins that re-establishes phase coherence and produces an echo at a time TE. Inhomogeneities of external magnetic fields are canceled, and the peak amplitude of the echo is determined by T2 decay. The rotating frame shows the evolution of the echo vector in the opposite direction of the FID. The sequence is repeated for each repetition period, TR.

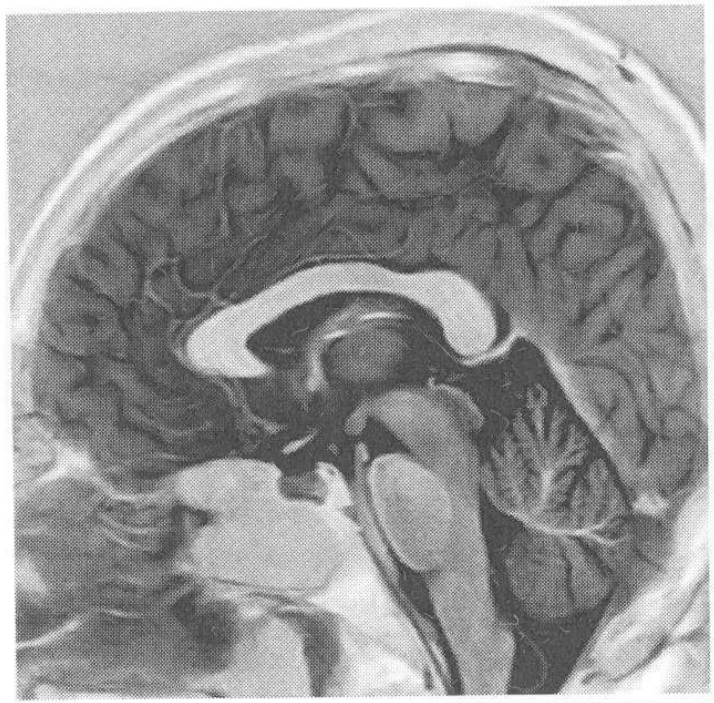
Bushberg et al, The essential physics of medical imaging

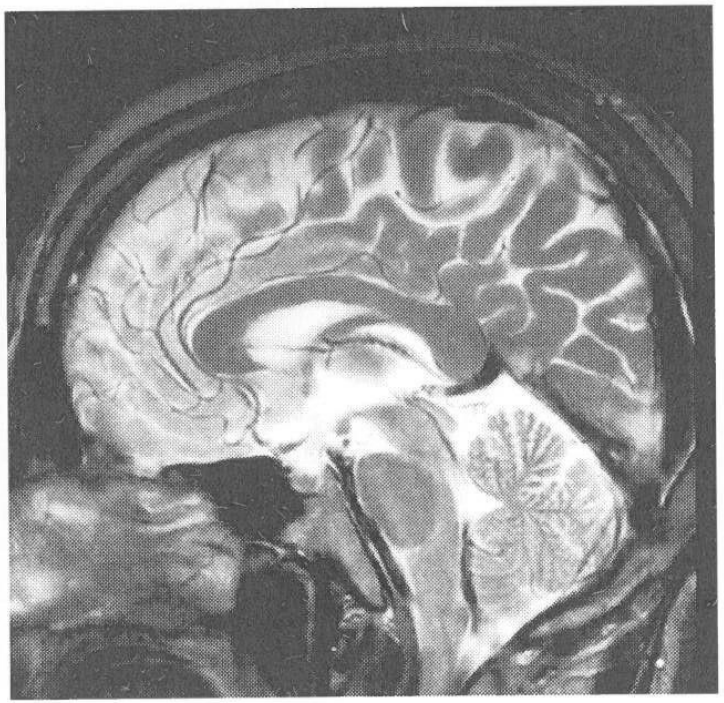
T2-spin echo decay

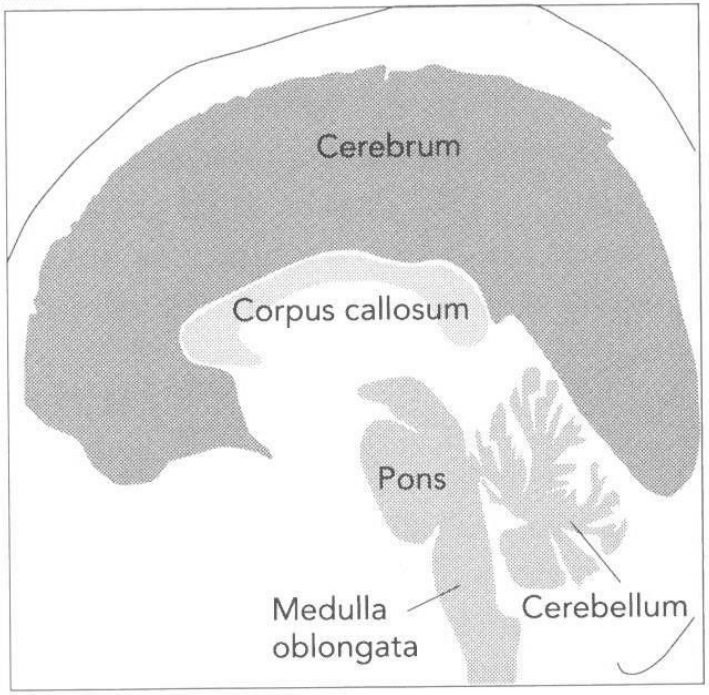


■ FIGURE 12-23 "True" T2 decay is determined from multiple 180-degree refocusing pulses acquired during the repetition period. While the FID envelope decays with the T2* decay constant, the peak amplitudes of subsequent echoes decay exponentially according to the T2 decay constant, as extrinsic magnetic field inhomogeneities are cancelled.

Bushberg et al, The essential physics of medical imaging







Figur 8:75. Hjärnor avbildade genom utnyttjande av uppmätta relaxationstider. Till vänster T_1-viktad bild och till höger T_2-viktad. (MR-bilder från Philips Medical Systems.)

MR images T1 and T2 contrast

Exercise

Form groups of 2 or 3 students, and explain to each other:

a. How the MRI-machine use T1 and T2 relaxation to find the number of protons in a tissue

b. Explain spin-echo weighting

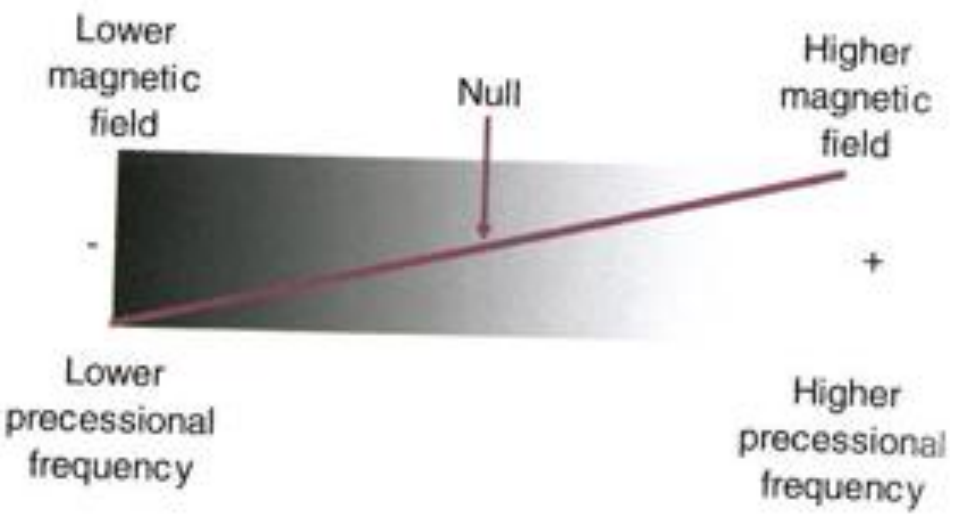


http://integrativeosteopathy.com.au/wp-content/uploads/2015/08/Exercise.jpg

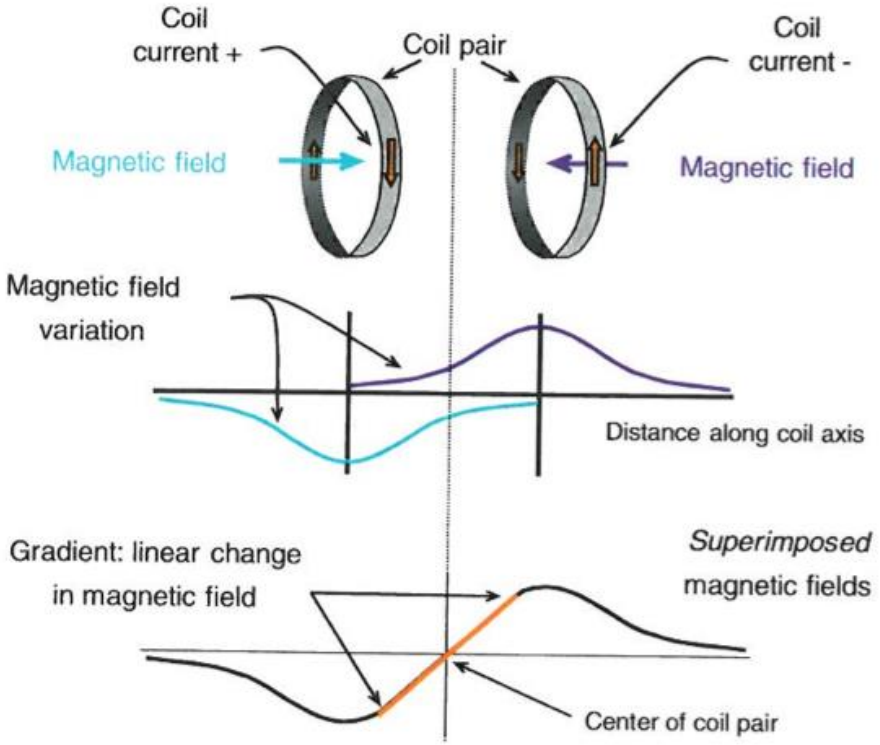
Construction of the MRI

Gradient magnetic field

■ FIGURE 12-47 The gradient magnetic field creates a net positive and negative magnetic environment that adds to and subtracts from the main magnetic field. Associated with the local change in magnetic field is a positionally dependent change in precessional frequencies, per the Larmor equation. The precessional frequencies directly vary across the field in proportion to the applied gradient strength.



Gradients



■ **FIGURE 12-4** Gradients are produced inside the main magnet with coil pairs. Individual conducting wire coils are separately energized with currents of opposite direction to produce magnetic fields of opposite polarity. Magnetic field strength decreases with distance from the center of each coil. When combined, the magnetic field variations form a linear change between the coils, producing a linear magnetic field gradient, as shown in the lower graph.

Bushberg et al, The essential physics of medical imaging

Superconducting magnet

To achieve high magnetic field strength, greater than 1T, require the electromagnet core wires to be superconductive. Shim coils interact with the main magnetic field in order to improve homogenity (minimize magnetic flux density). RF-coils are used to transmit energy to the patient and pick up return signal

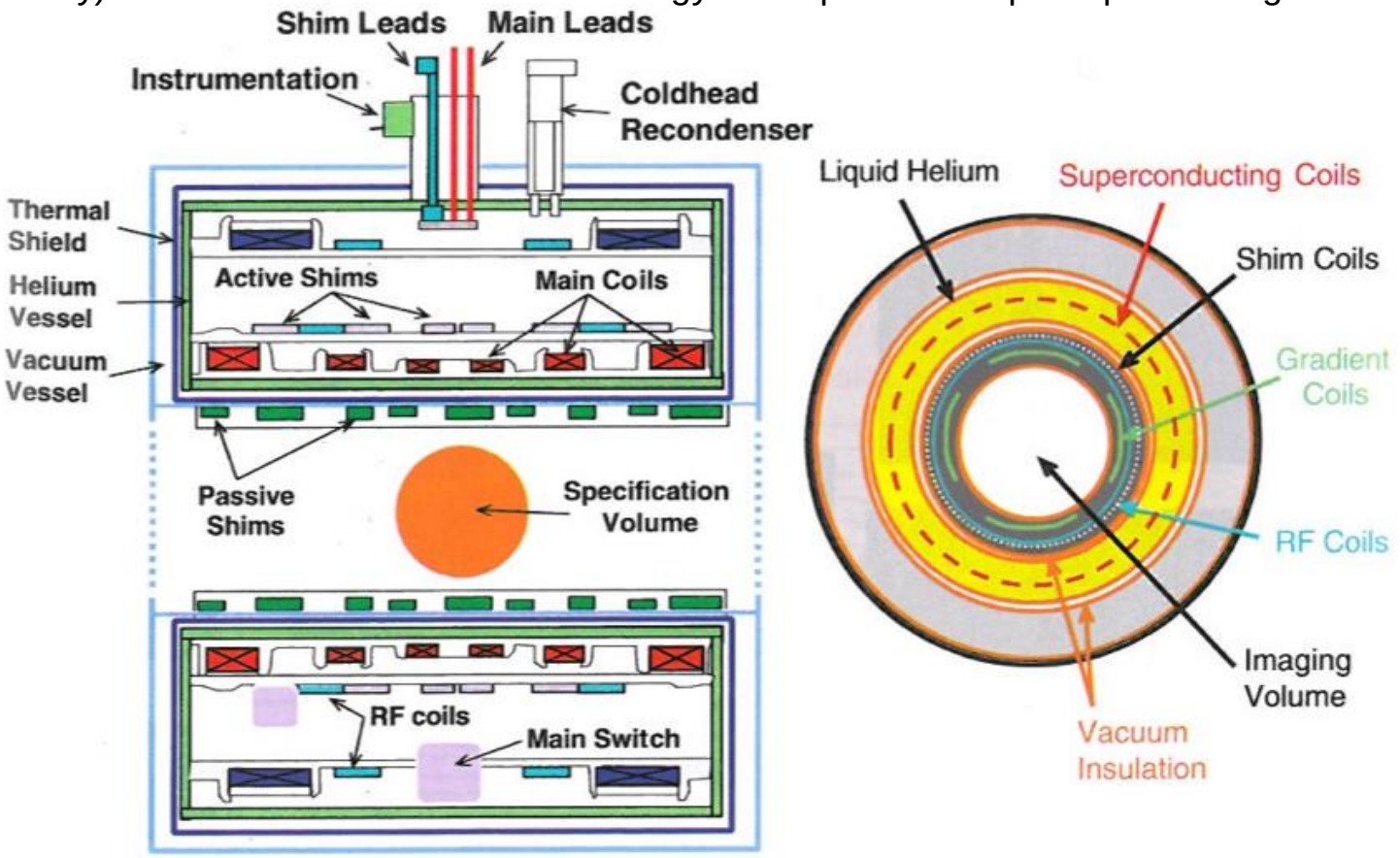
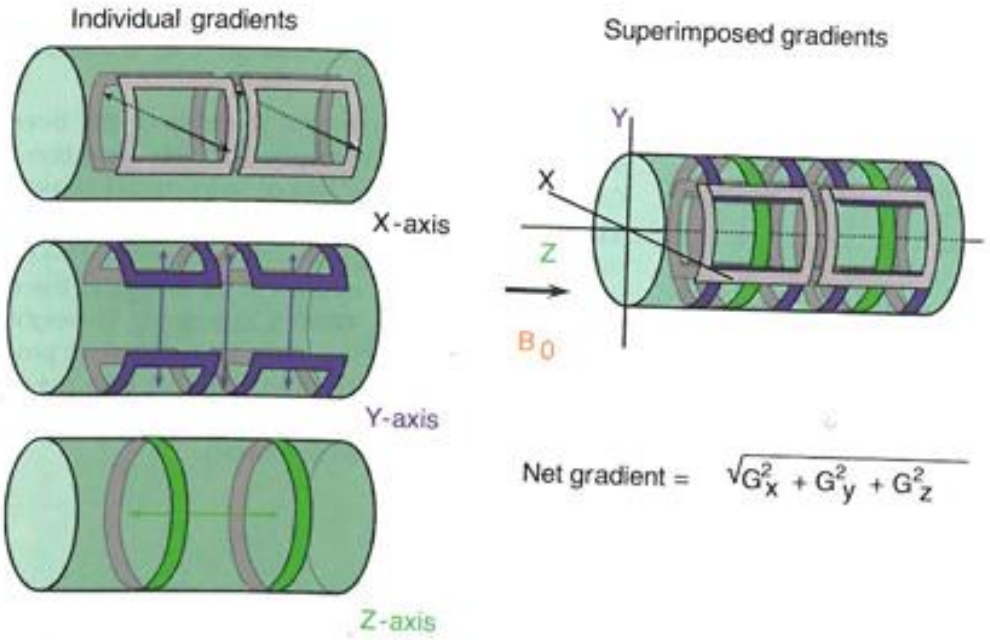


FIGURE 12-3 Internal components of a superconducting air-core magnet are shown. On the left is a cross section through the long axis of the magnet illustrating relative locations of the components, and on the right is a simplified cross section across the diameter.

Bushberg et al, The essential physics of medical imaging

MR Signal Localization

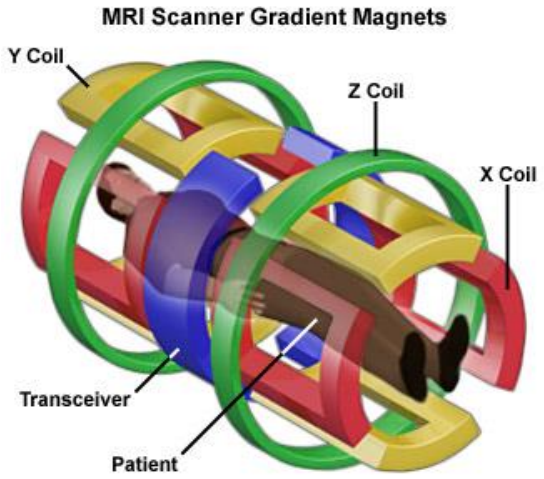


■ FIGURE 12-48 Within the large stationary magnetic field, field gradients are produced by three separate coil pairs placed within the central core of the magnet, along the *x*, *y*, or *z* directions. In modern systems, the current loops are distributed across the cylinders for the x-, y- and z- gradients, which generates a lower, but more uniform gradient field. Magnetic field gradients of arbitrary direction are produced by the vector addition of the individual gradients turned on simultaneously. Any gradient direction is possible by superimposition of magnetic fields generated by the three-axis gradient system.

Bushberg et al, The essential physics of medical imaging

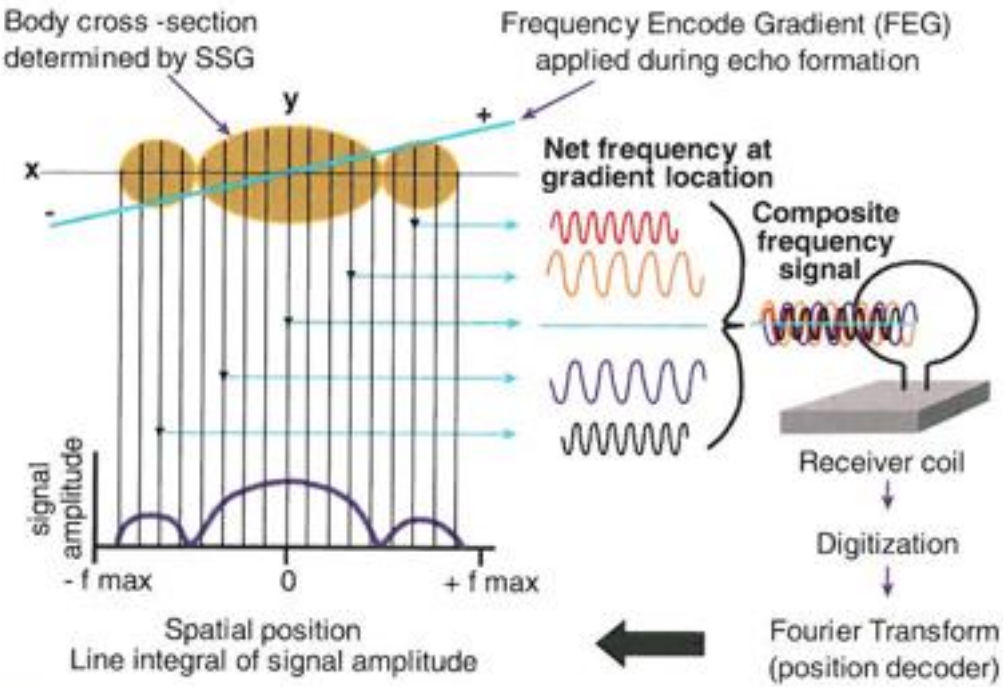
To record an MRI signal

- Needs a receive coil tuned in to the same RF-requency as the excitation coil
- Measure net magnetization
- The signal oscillates at the resonanse frequency when the net magnetization vector rotates in the room
- Signalamplitude will be weakened when the netto magnetization returns to the B-direction



Bushberg et al, The essential physics of medical imaging

Frequency encode gradient

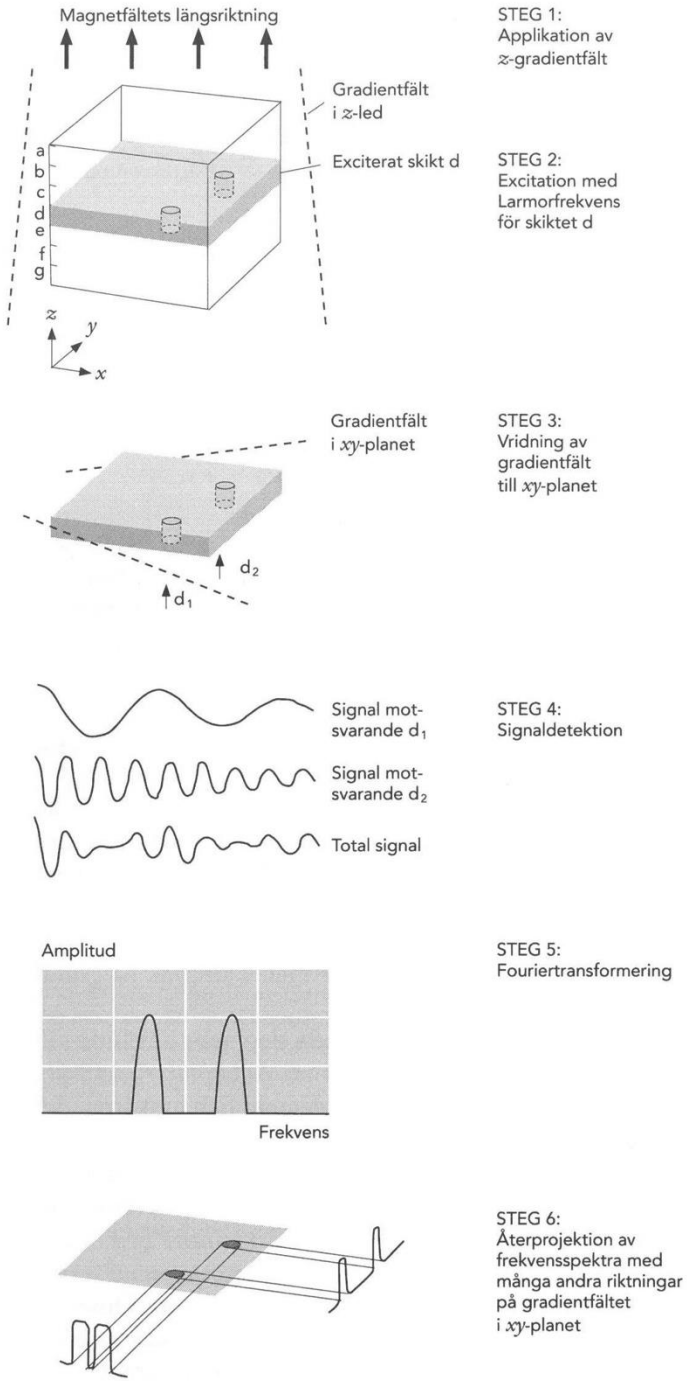


■ FIGURE 12-52 The *FEG* is applied in an orthogonal direction to the SSG, and confers *a spatially dependent variation* in the precessional frequencies of the protons. Acting only on those protons in a slice determined by the SSG excitation, the composite signal is acquired, digitized, demodulated (Larmor frequency removed), and Fourier transformed into frequency and amplitude information. A one-dimensional array represents a *projection* of the slice of tissue (amplitude and position) at a specific angle. (Demodulation into net frequencies occurs *after* detection by the receiver coil; this is shown in the figure for clarity only.)

Bushberg et al, The essential physics of medical imaging

3D picture construction

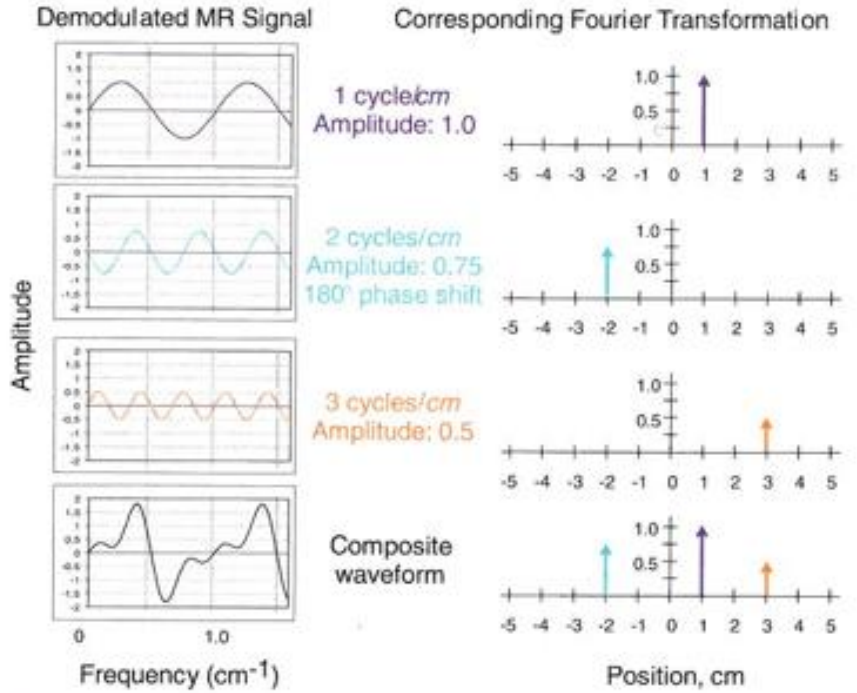
$$\omega = \gamma B$$



Figur 8:85. En möjlig metod för rekonstruktion av MR-bilder.

A rotating FEG in the x-y plane will give the acquisition of individual projections as a function of angle.

Fourier transform of spatial freq. signals



■ FIGURE 12-53 Spatial frequency signals (cycles/cm) and their Fourier transforms (spatial position) are shown for three simple sinusoidal waveforms with a specific amplitude and phase. The Fourier transform decodes the frequency, phase and amplitude variations in the spatial frequency domain into a corresponding position and amplitude in the spatial domain. A 180-degree phase shift (second from the top) is shifted in the negative direction from the origin. The composite waveform (a summation of all waveforms, lower left) is decoded by Fourier transformation into the corresponding positions and amplitudes (lower right).

Bushberg et al, The essential physics of medical imaging

Functional MRI

Diffusion Weighted Imaging

Diffusion relates to the random motion of water molecules in tissues. Interaction of the local cellular structure with the movement of water molecules produces anisotropic, directionally dependent diffusion (e.g., in the white matter of brain tissues).

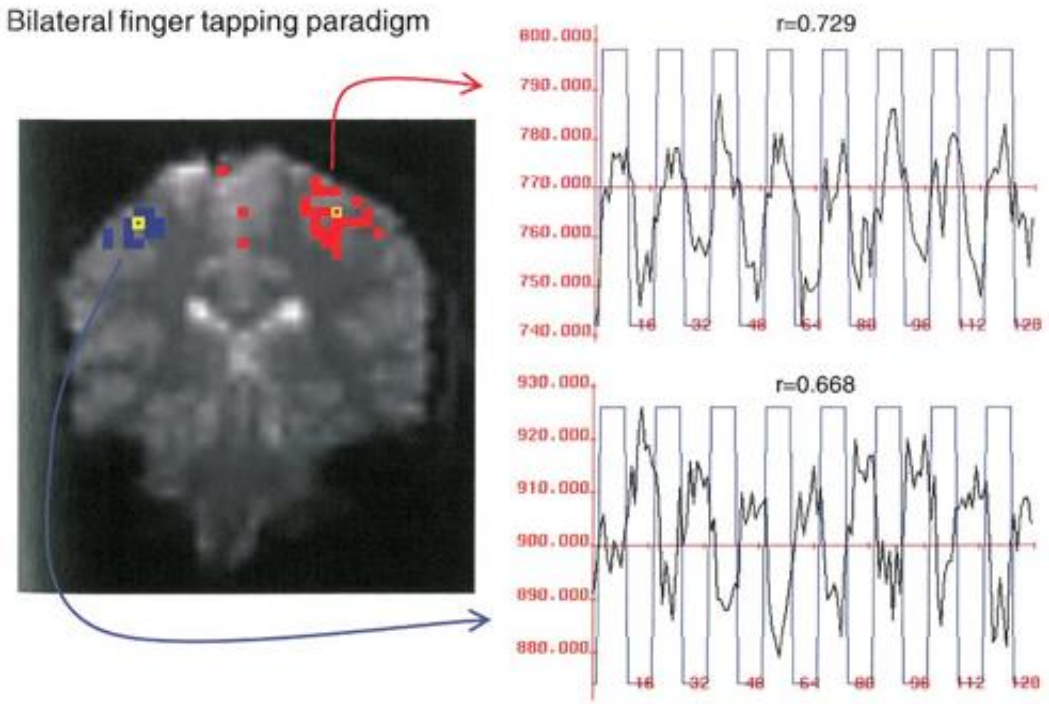
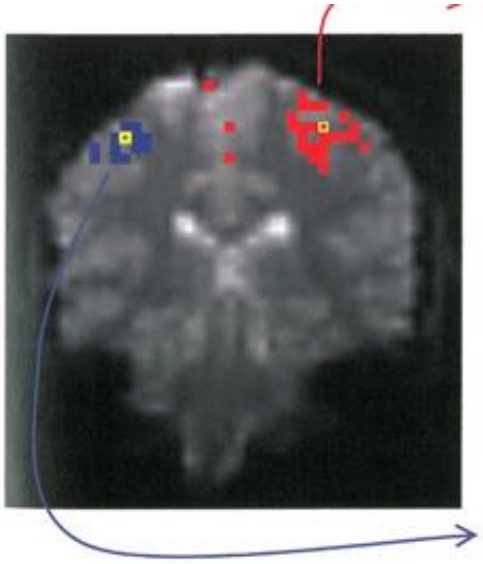


FIGURE 13-20 Functional MR image bilateral finger tapping paradigm shows the areas of the brain activated by this repeated activity. The paradigm was a right finger tap alternated by a left finger tap (time sequence on the right side of the figure) and the correlated BOLD signals (*black traces*) derived from the echo planar image sequence. A voxel-by-voxel correlation of the periodic stimulus and MR signal is performed, and when exceeding a correlation threshold, a color overlay is added to the grayscale image. In this example, *red* indicates the right finger tap that excites the left motor cortex, which appears on the right side of the image, and blue the left finger tap. (Figure compliments of MH Buonocore, MD, PhD University of California Davis.)

Bushberg et al, The essential physics of medical imaging

Functional MRI



Hb differs in how it responds to the magnetic field, the deoxygenated Hb molecule is slightly more attracted to magnetic field.

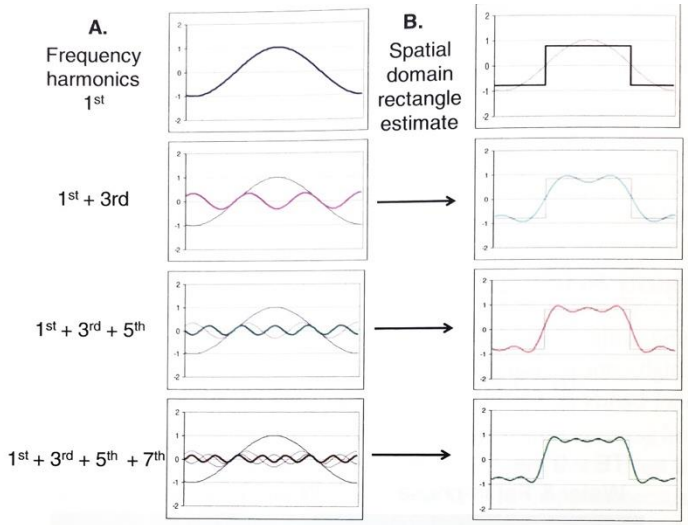
In consequence, the magnetic field is distorted by causing the nuclei to loose magnetization faster in the *T2 decay = MR pulse sequences sensitive to *T2 will have more signals when the blood is oxygenated than for the deoxygenated blood.

(The effect increases with the square of the magnetic field strength) = Minimum 1.5T

MRI-artefacts

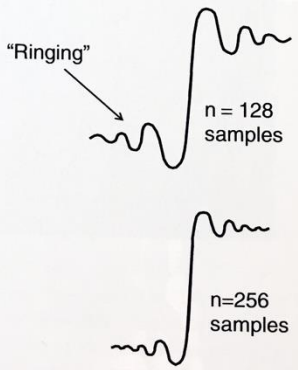
Caused by insufficient sampling, could be reduced with a larger number of samples.

Ringing artefacts



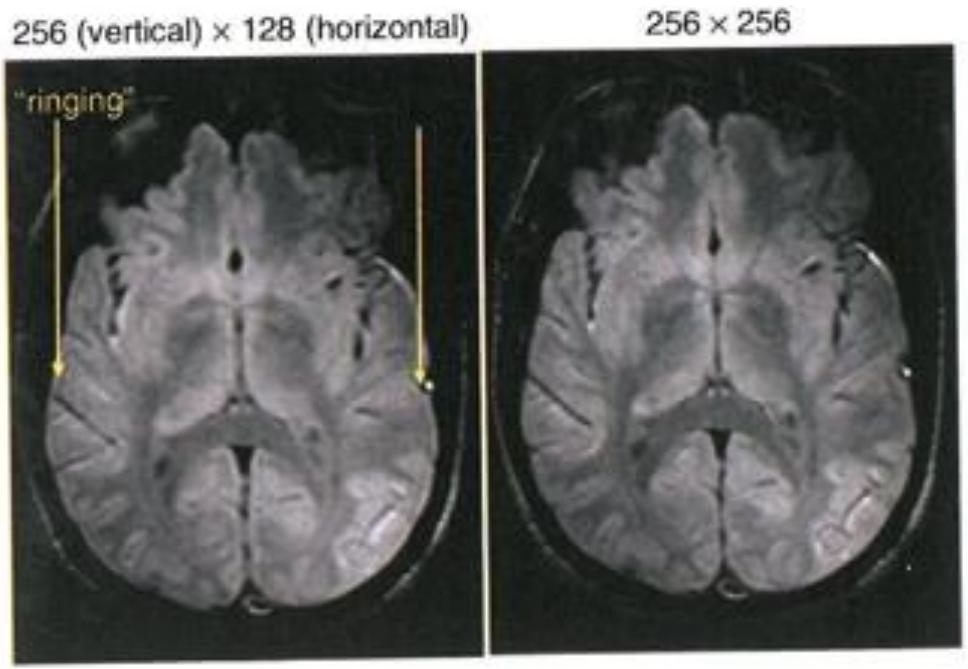
■ **FIGURE 13-36** The synthesis of a spatial object occurs by the summation of frequency harmonics in the MR image. **A. Left column**: frequency harmonics that estimate a rectangle function with progressively higher frequencies and lower amplitudes are shown. **B. Middle column**: As higher frequency harmonics are included, the summed result more faithfully represents the object shape, in this example a rectangle with two vertical edges. The number of frequencies encoded in the MR image is dependent on the matrix size. **C. Right column**: A sharp transition boundary in an MR image is represented with 256 samples better than with128 samples (frequency harmonics in k-space). The amount of ringing caused by insufficient sampling is reduced with a larger number of samples.

C. Sharp transition in MR image:



Bushberg et al, The essential physics of medical imaging

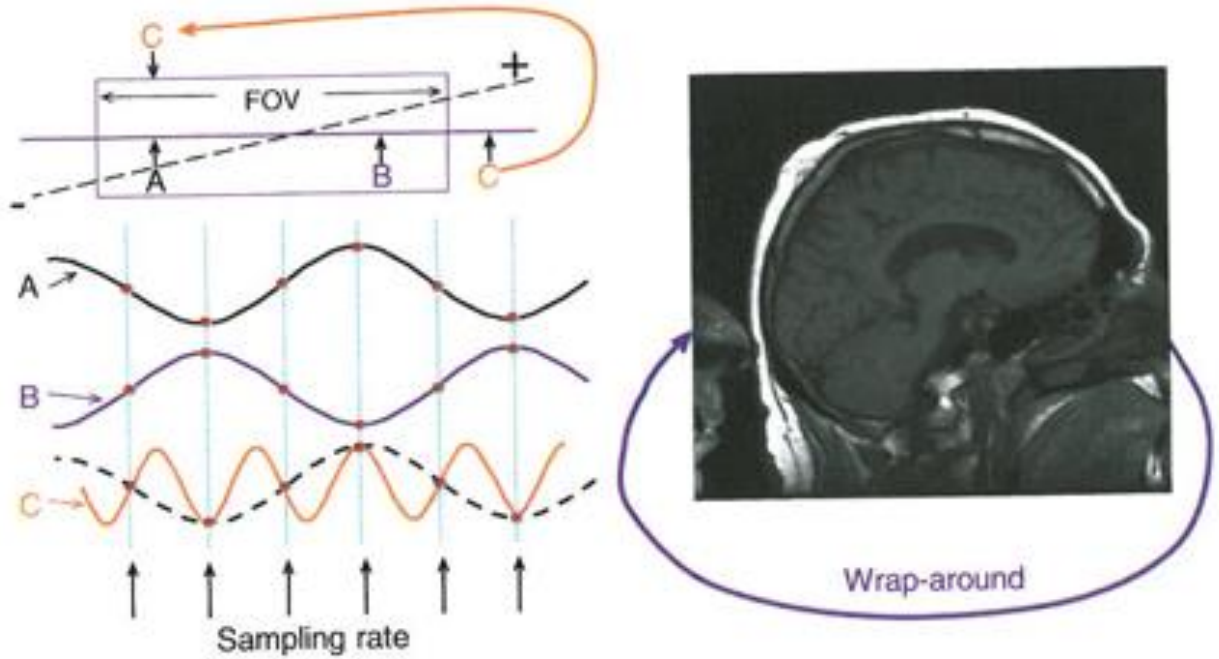
Ringing artefacts



■ FIGURE 13-37 Example of ringing artifacts caused by a sharp signal transition at the skull in a brain image for a 256 × 128 matrix (left) along the short (horizontal) axis, and the elimination of the artifact in a 256 × 256 matrix (right). The short axis defines the PEG direction.

Caused by insufficient sampling, could be reduced with a larger number of samples.

Wraparound artefacts

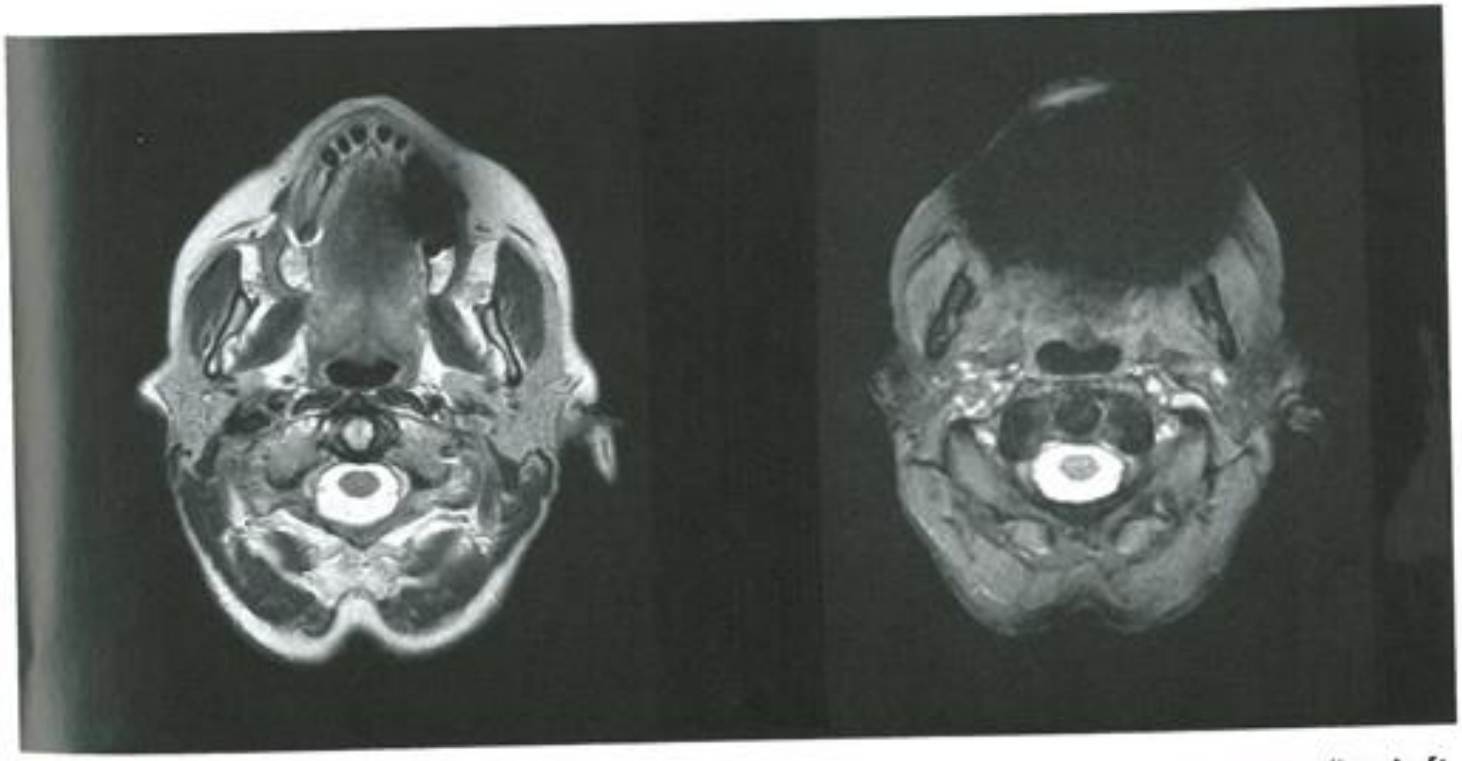


■ FIGURE 13-38 **Left.** Wraparound artifacts are caused by aliasing. Shown is a fixed sampling rate and net precessional frequencies occurring at position *A* and position *B* within the FOV that have identical frequencies but different phase. If signal from position *C* is at *twice* the frequency of *B* and insufficiently sampled, the same frequency and phase will be assigned to *C* as that assigned to *A*, and therefore will appear at that location. **Right.** A wrap-around artifact example displaces anatomy from one side of the image (or outside of the FOV) to the other side.

Bushberg et al, The essential physics of medical imaging

Susceptibility artefacts

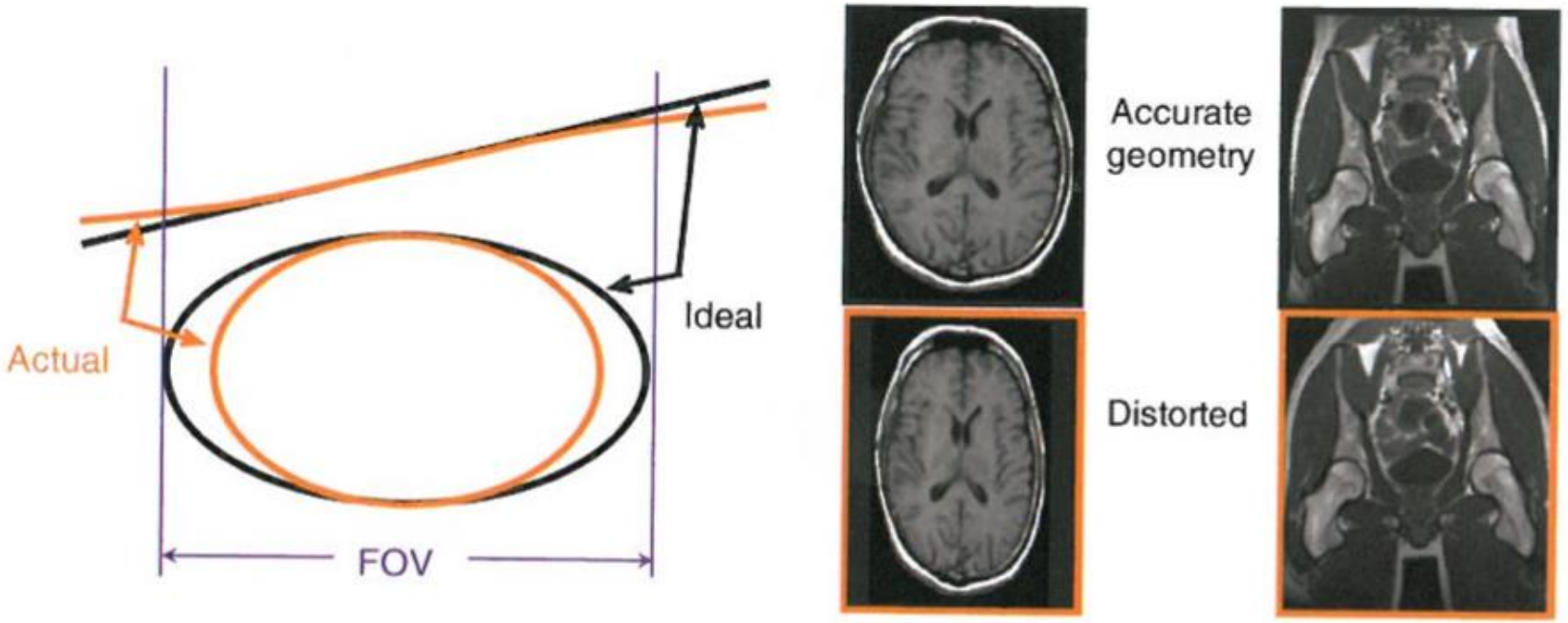
Any drastic change in the magnetic susceptibility will distort the magnetic field. Changes in susceptiblity like tissue-air interface, or metals (ferrous or not) may have a significant effect on the local tissues adjacent to changes in susceptibility. Can also be helpful, like diagnosing the age of a hemorrhage based on signal characteristics of the blood degeneration products



■ FIGURE 13-24 Susceptibility artifacts due to dental fillings are shown in the same axial image slice. **Left.** Axial T2-weighted fast spin echo image illustrates significant suppression of susceptibility artifacts with 180-degree refocusing pulse. **Right.** Axial T2*-weighted gradient echo image illustrates significant image void exacerbated by the gradient echo, where external inhomogeneities are not canceled in the reformed echo.

Bushberg et al, The essential physics of medical imaging

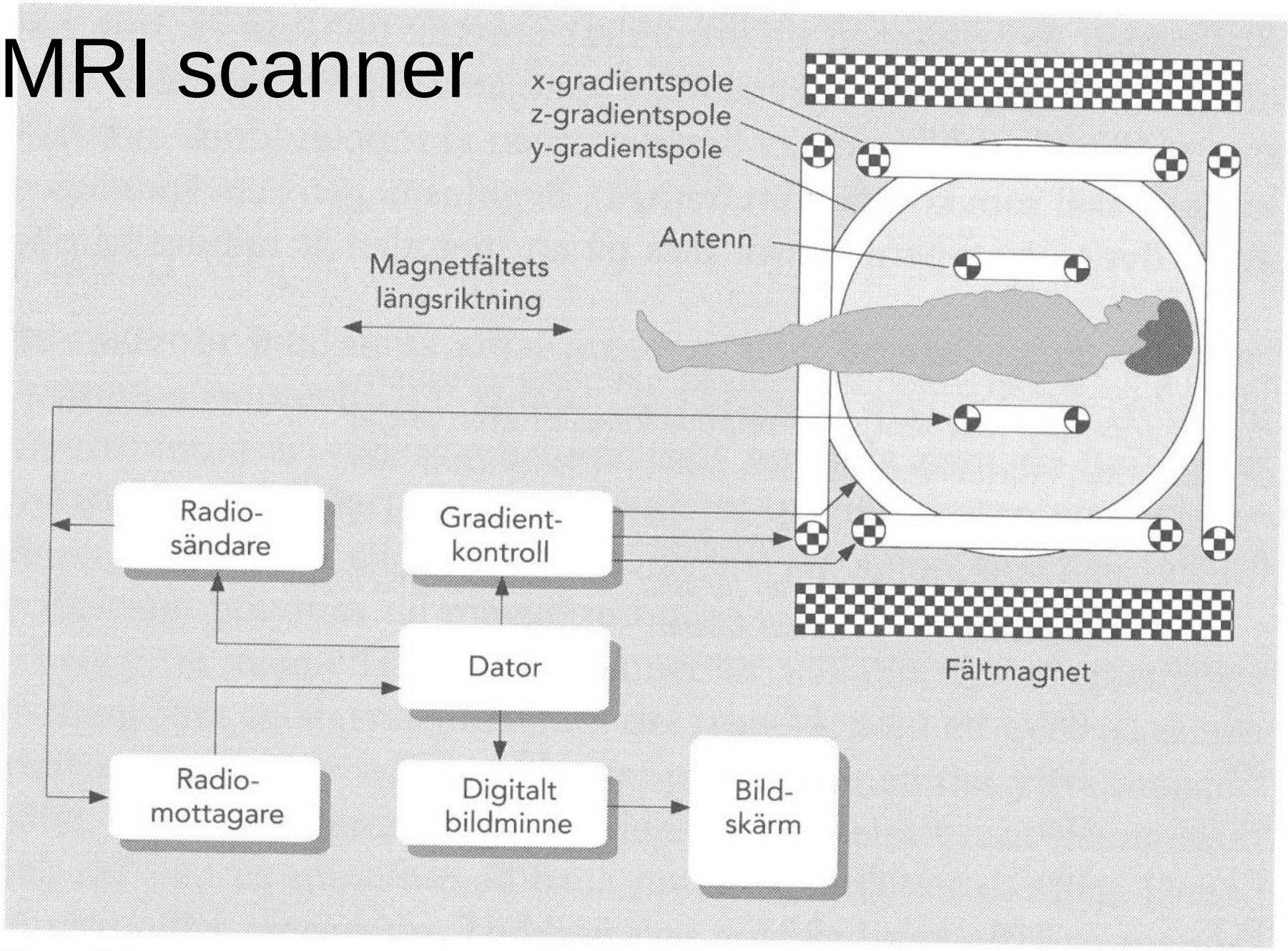
Gradient nonlinearity artefacts



■ **FIGURE 13-25** Gradient nonlinearity causes image distortions by mis-mapping anatomy. In the above examples, the strength of the gradient at the periphery is less than the ideal (*orange line* versus *black line*). This results in a compression of the imaged anatomy, with inaccurate geometry (images with *orange* border). For comparison, images acquired with linear corrections are shown above.

Any deviation or temporal instability in linear gradients will give a distortion. With gradient calibration and limitation of FOV (Lowering gradient field strength) or holding the gradient field strength and number of samples constant while decreasing the frequency bandwidth, minimize spatial distortion

MRI scanner



Figur 8:74. Apparatur för magnetresonanstomografi.

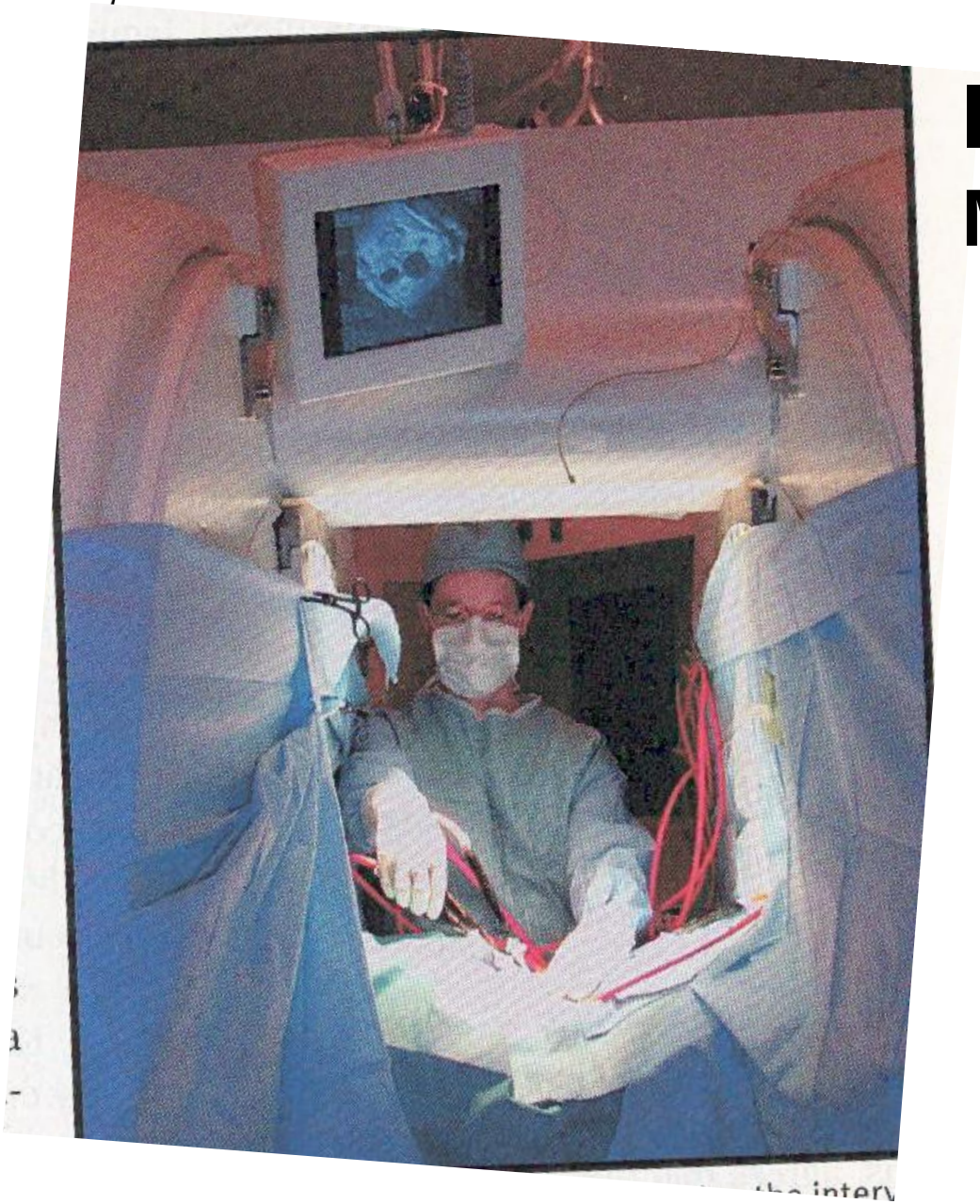
Vertical main field



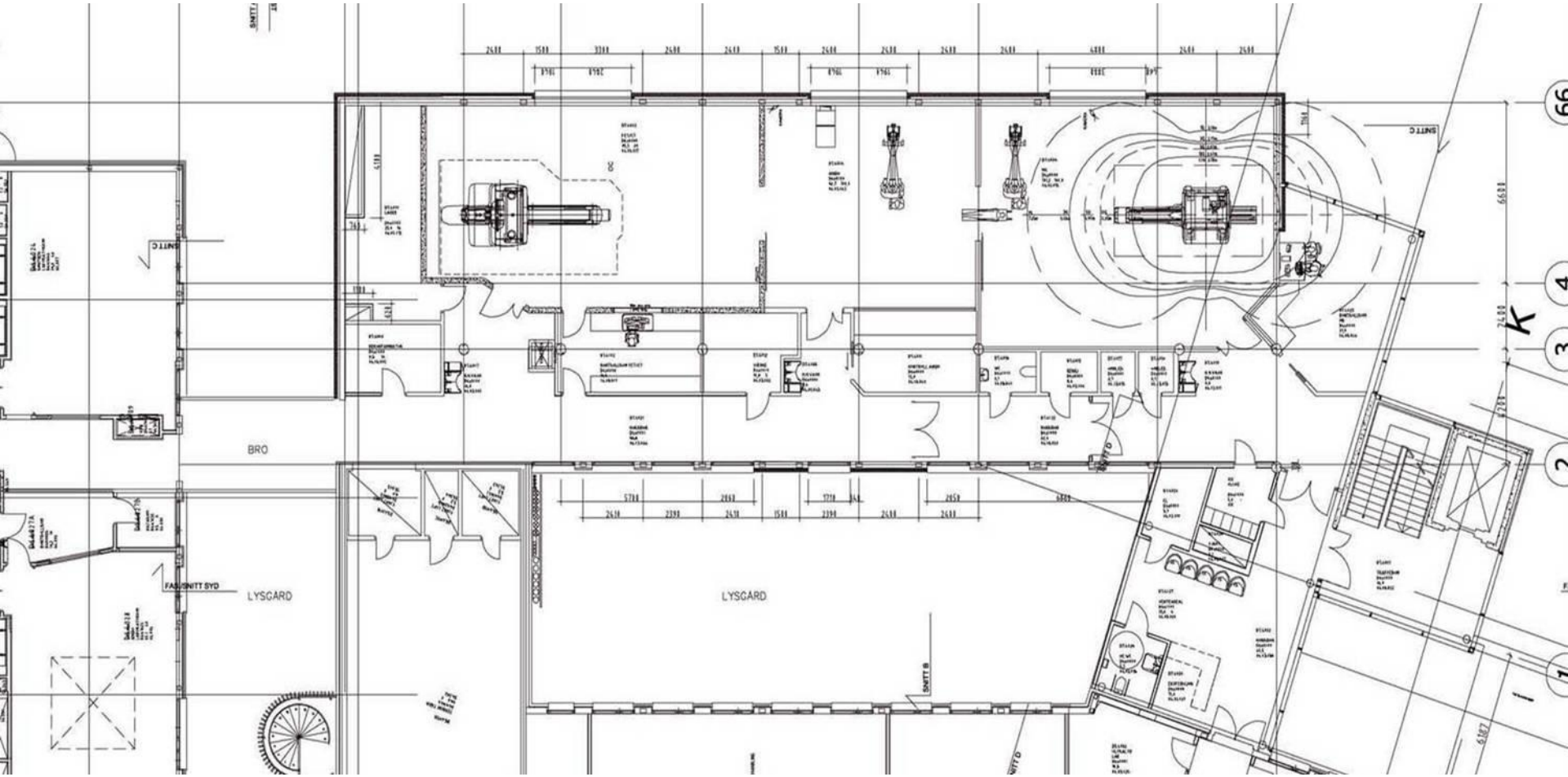
Source: Oulun Yliopisto

Extremity MRI





Interventional MRI


BRO
LYSGARD
LYSGARD

Adv/disadv MRI

- Advantages
- No harmful radiation
- Soft tissue imaging
- High resolution images of T1 or T2 preferences

- Disadvantages
- Expensive, large installation with superconducting magnets++
- Very strong magnetic field
- Claustrophobic
- Not for frozen tissue
- Time consuming

MRI and patient safety



What happened?



