Lecture 11: Magnetic Resonance Imaging I

Contents

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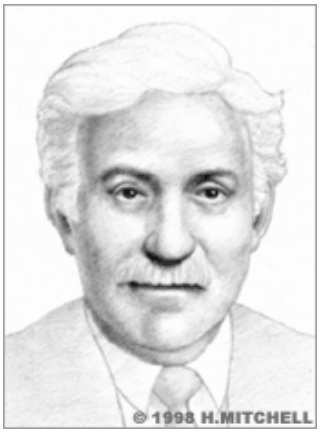
Nuclear-Magnetic Resonance (NMR)

- Measurement principle based on magnetic properties of atomic nuclei.
- Applicable for atomic nuclei with odd number of nucleons (protons and neutrons). Hydrogen and phosphorus are the most important representatives in imaging tissues.
- NMR has been used for measurements without spatial resolution since the 1950s, for imaging since 1972.

The physical description provided here is one model. It neglects e.g. quantum mechanical aspects. Quantum mechanical descriptions may look completely different at first glance.

Nuclear-Magnetic Resonance (2)

History



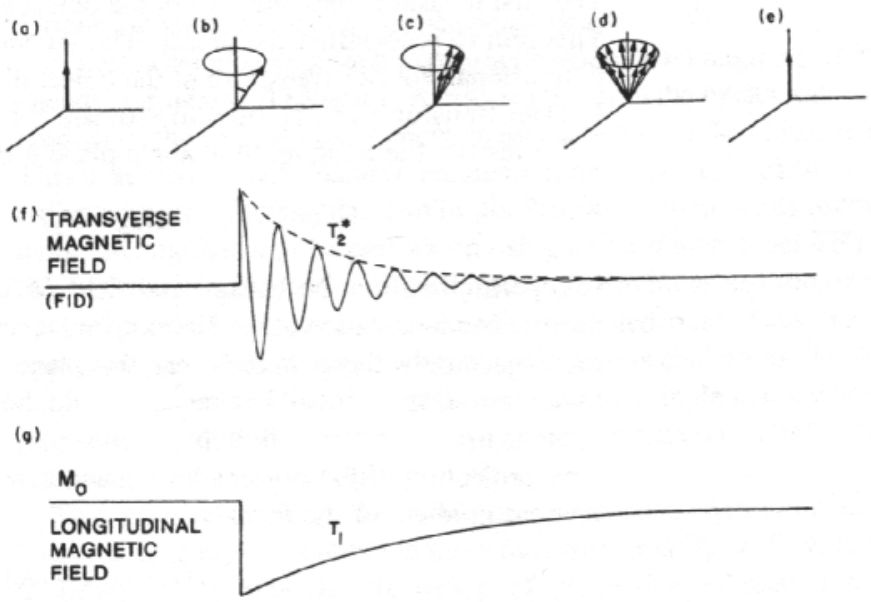

Left: Raymond V. Damadian (b. 1936). **Middle:** Paul C. Lauterbur (b. 1929). **Right:** Peter Mansfield (b. 1933). Lauterbur and Mansfield laid the foundations of modern MR imaging and received a 2003 Nobel Prize for this work. Damadian seems to have been the first to suggest an application of NMR to detect cancer and later made the first whole-body MR scan. However, his early works had no follow-ups in later MR development.
Images: http://web.mit.edu/invent/iow/damadian.html,
http://www.scs.uiuc.edu/chem/lauterb.htm,
http://www.nottingham.ac.uk/~ppzwww/staff/Mansfield_P_t.html

Nuclear-Magnetic Resonance (3)

Physical Principle

- An atomic nucleus with an odd number of nucleons possesses a *magnetic moment*. In the presence of a strong magnetic field, it behaves like a small spinning magnetic dipole.
- Given an external magnetic field H, the atom performs gyroscopic precessions at the **Larmor frequency**
$$\omega = \gamma H$$
where γ is the *gyromagnetic constant*. This constant depends primarily on the physical properties of the nucleus, secondarily it is modulated by its chemical environment.
- In a strong magnetic field, the magnetic dipoles align. By radio waves with their Larmor frequency, they can be *excited* to precessions. Once excitation is over, precession decays exponentially (*relaxation*).
- During relaxation, atoms emit exponentially decaying radio waves at the Larmor frequency. This radiation is called *free induction decay (FID)* or *spin echo*. It can be measured by coils around the object. Immediately after excitation, all excited atoms are in phase. The FID is therefore proportional to the number of atoms.

Physical Principle, cont.

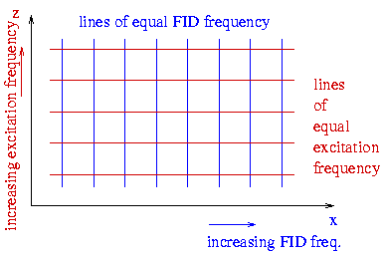


Excitation and relaxation of precessing nuclear magnetic moment. *(Kak/Slaney 2001)*

Introducing Spatial Resolution

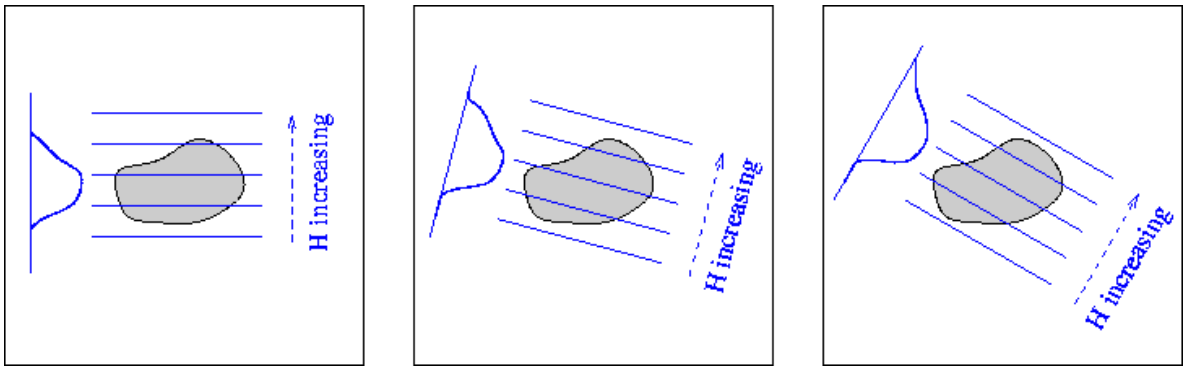
In **magnetic resonance imaging (MRI)**, one considers only atoms with identical gyromagnetic constant γ, e.g. hydrogen atoms (in water molecules).

- In a homogeneous magnetic field, all atoms have equal Larmor frequency.
- In a spatially variant magnetic field, only atoms exposed to equal field magnitude H have equal Larmor frequency.
- By using a magnetic field with gradient along the z axis (central axis of scanner), each Larmor frequency becomes characteristic to one cross-section.
- Excitation can therefore be constrained to one cross-section, w.l.o.g. $z = 0$.
- Using a different gradient during readout, e.g. in x direction, a longitudinal plane can be addressed. Combining both, it is possible to measure hydrogen concentrations along lines.



MR Tomography Via Line Integrals

- Repeating this with different longitudinal sections, it is therefore possible to measure a similar set of 1-D projections as in classical CT. Since different longitudinal sections according to some gradient are distinguished by different FID frequencies, the recorded FID is in fact the Fourier transform of a projection.
- The same reconstruction techniques can be used to reconstruct 3-D images from these data.



However, in magnetic resonance imaging this technique is rarely used.

Phase Encoding and Direct Fourier Measurement

By introducing a third magnetic field gradient, it is possible to measure the full 2-D Fourier transform of a cross-section.

- After excitation, all excited atoms (within one cross-section plane) oscillate in equal phase.
- A gradient in y direction applied for a defined time interval splits up precession frequencies in y direction for this time, leading to a phase shift proportional to the y coordinate (**phase encoding**).
- Finally, a gradient in x direction is applied while the FID is measured.
- The measured FID is a superposition of signals of different frequency and phase, corresponding to different locations (x, y). If $\varrho(x, y)$ is the hydrogen density within the plane $z = 0$, then one measures

$$\hat{p}(q_y, t) = \iint \varrho(x, y) \mathrm{e}^{\mathrm{i}(xq_x + yq_y + \omega_0 t)} \,\mathrm{d}x \,\mathrm{d}y$$

with ω_0 being the Larmor frequency without perturbation by the gradients, and $q_x = G_x \gamma t$ and $q_y = G_y \gamma T$ determined by the gradients G_x and G_y.

Magnetic Resonance Imaging (3a)

Explanatory Remark. In the measurement of the FID, one in fact records the emitted radio frequency oscillation as a function of time.

This means that not only echos at one specific (Larmor) frequency are detected. Instead, echos from locations with different x are emitted at different Larmor frequencies. Without the y phase encoding, one would therefore receive a superposition of oscillations of these different frequencies:

$$\hat{p}(t) = \int \varrho(x) \mathrm{e}^{\mathrm{i}(xG_x\gamma+\omega_0)t}\,\mathrm{d}x = \int \varrho(x) \mathrm{e}^{\mathrm{i}(xq_x+\omega_0 t)}\,\mathrm{d}x\ ,$$

which is (up to the factor $\mathrm{e}^{\mathrm{i}\omega_0 t}$ which can easily be compensated by a division) a 1-D Fourier transform of $\varrho(x)$, with t encoding the spatial frequency in x direction. So far, only one excitation–measurement cycle is needed.

The y gradient applied between excitation and measurement then introduces the additional phase encoding of the y location, i.e. the factor $\mathrm{e}^{\mathrm{i}qy}$ in the integrand. The resulting FID

$$\hat{p}(q_y, t) = \iint \varrho(x, y) \mathrm{e}^{\mathrm{i}(xq_x+yq_y+\omega_0 t)}\,\mathrm{d}x\,\mathrm{d}y$$

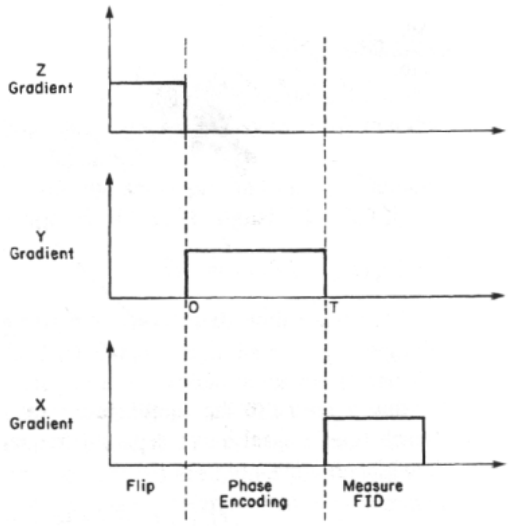
is (again, up to a factor $\mathrm{e}^{\mathrm{i}\omega_0 t}$) the 2D Fourier transform of the density $\varrho(x, y)$, with t encoding the spatial frequency in x direction and q_y encoding the spatial frequency in y direction. To measure this function of q_y and t requires to repeat the excitation–measurement cycle for as many different values of q_y, as given by the desired y resolution.

The density ϱ is finally recovered by a 2D inverse Fourier transform.

Magnetic Resonance Imaging (4)

Phase Encoding and Direct Fourier Measurement, cont.

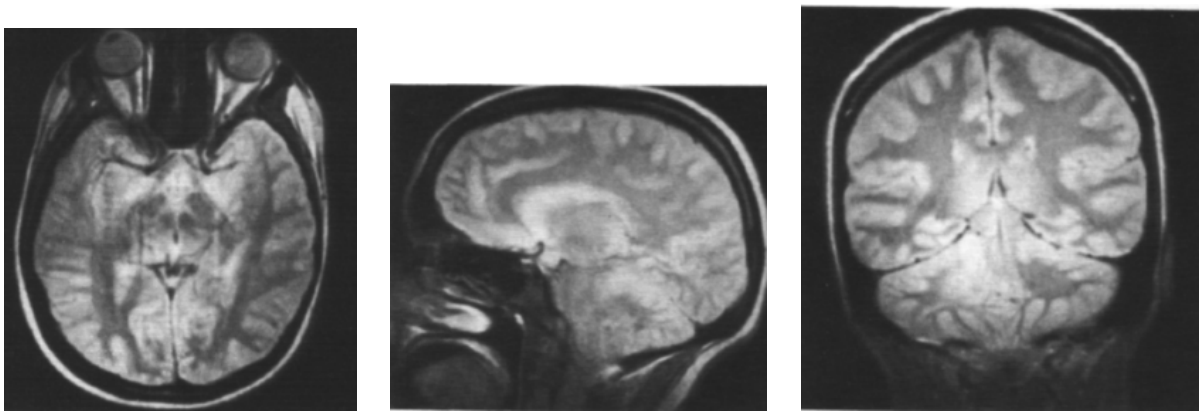
- A full measurement includes single measurements of this type with stepping through a full range of different y gradients. Then one obtains the 2-D Fourier transform of ϱ up to the factor $\exp(\mathrm{i}\omega_0 t)$ in the integrand. The density ϱ is then found essentially by inverse Fourier transform.



Sequence of three different spatial gradients used for measuring the Fourier transform of an object in MRI. *(Kak/Slaney 2001)*

Magnetic Resonance Imaging (5)

Example



Three sections in orthogonal planes of a human head imaged by MRI. *(Webb 1988)*

Remarks:
Because of its sensitivity for hydrogen atoms, MR imaging is particularly appropriate for imaging soft tissue.

Imaging phosphorus instead of hydrogen allows to map metabolistic activity since phosphorus is involved in energy transport (ADP, adenosine triphosphate).

MR Imaging Equipment (1)

Components of an MR Imaging System

- **Main magnet:** generates static magnetic field
- **Radiofrequency (RF) system:**
  - generates oscillating magnetic fields for excitation
  - detects oscillating magnetic fields during relaxation
- **Magnetic field gradient system:** generates controllable variable fields
- **Shielding**

The Main Magnet

- Generates a static homogeneous magnetic field
- High field intensities:
  - $(0.5 \ldots 3)\,\mathrm{T}$ (Tesla) for medical applications.
    Such fields are considered not dangerous for the human organism.
  - in laboratories even up to $18\,\mathrm{T}$.
    However, such strong fields are not consented to be harmless.

  For comparison: A household refrigerator magnet has about $0.01\,\mathrm{T}$.
- Field homogeneity requirements: tolerable deviations $10 \ldots 50\,\mathrm{ppm}$
- Typically, *supraconducting coils* are used in the main magnet; cooling by liquid helium $(4\,\mathrm{K} \mathrel{\hat{=}} -269\,^{\circ}\mathrm{C})$ is required.
- Homogeneity is improved by further correction coils.

MR Imaging Equipment (2)

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Radio Frequency (RF) System

- Generates fields oscillating at radio frequencies $((1 \ldots 100)\,\mathrm{MHz})$ for excitation
- Detects fields oscillating at radio frequencies during relaxation
- Consists of various *transmitter* and *receiver* coils or of *transceiver* coils which are used for both transmission and reception
- Typical frequencies are in the range $(1 \ldots 100)\,\mathrm{MHz}$. Note that the wavelengths of the corresponding radio waves are $(3 \ldots 300)\,\mathrm{m}$, thus much larger than the object details being detected!
- Transmitter power is typically up to $20\,\mathrm{kW}$ for a whole-body medical imaging system

MR Imaging Equipment (3)

Magnetic Field Gradient System

- Generates controllable variations (inhomogeneities) in the magnetic field
- Crucial for spatial localisation
- Three sets of gradient coils (in x, y and z directions)
- Gradients are vectorial quantities; a gradient with components G_x, G_y, G_z possesses total magnitude

$$|G| = \sqrt{G_x^2 + G_y^2 + G_z^2}$$

- Gradient fields can reach $10\,\mathrm{mT/m}$
- Gradients must be switched on and off quickly: Currents of up to $200\,\mathrm{A}$ need to be switched with rise times of $(0.5 \ldots 1)\,\mathrm{ms}$ in clinical systems

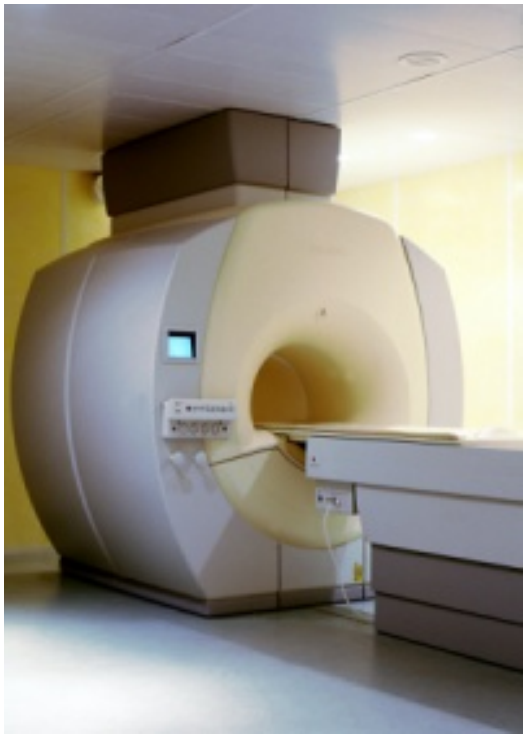
MR Imaging Equipment (4)

Shielding

- *Magnetic field shielding*
  - Constrains the extremely strong magnetic field to the measurement room
  - Passive: Iron shielding in the walls
  - Active: Additional coils surrounding the primary ones that compensate the exterior field, i.e. "bend back" magnetic field lines
- *Radio-frequency shielding*
  - Electrically conductive shielding (*Faraday cage*) surrounds the entire imaging system
  - Prevents the generated radio waves from leaving the machinery
  - Protects the imaging equipment from external radio waves

MR Imaging Equipment (5)

MR Scanner



MR scanner. *(http://www.meb.uni-bonn.de/radiologie/Patienteninformation/MRT.html)*

MR Imaging Sequences (1)

Decay Times

After excitation by an RF pulse at Larmor frequency, the relaxation process is characterised mainly by two time constants. These time constants depend on the internal structure of the matter in which the water molecules (e.g.) are bound.

- **Longitudinal relaxation time** or *spin-lattice relaxation time*, **T1**
  - Half-value time of the exponential decay process in which the magnetic moments return to alignment with the external magnetic field, thus the longitudinal magnetisation returns to its stable value
  - Indicates vibrational motion, i.e. whether water is able to tumble or rotate, thus whether it is free or absorbed in tissue
- **Transversal relaxation time**, *dephasing time* or *spin-spin relaxation time*, **T2**
  - Half-value time of the exponential decay of transversal (oscillating) polarisation
  - Cannot be longer than T1
  - Due to dephasing – loss of phase correlation between atoms – this decay is even faster than the longitudinal one

MR Imaging Sequences (2)

Decay Times, cont.

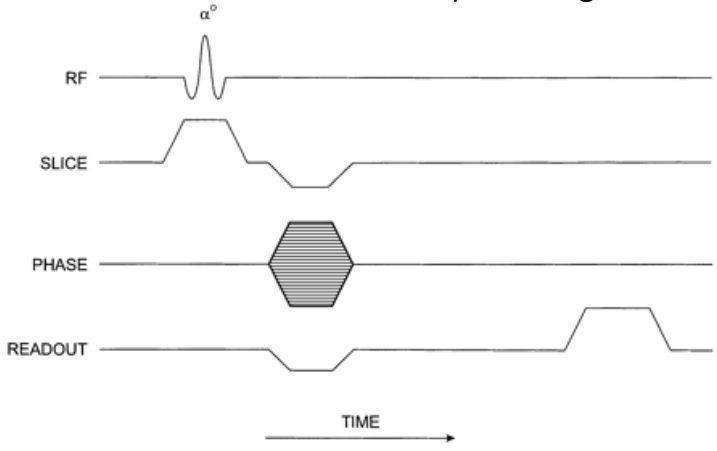
Tissue type	T1 (ms)	T2 (ms)
White brain matter	510	67
Grey brain matter	760	77
Cortico-spinal fluid	2650	280
Edema	900	126

Typical longitudinal and transversal relaxation times for different types of tissue. *(Webb, 1988)*

MR Imaging Sequences (3)

Parameters of MR Sequences

MR measurements can generate different types of imaging by different **MR sequences**, i.e. different combinations of time-dependent gradients and radio pulses.



Simple MR sequence. *(Image: http://radiology.rsnajnls.org)*

MR Imaging Sequences (4)

Parameters of MR Sequences, cont.

- Pulse durations and intervals between pulses vary widely, between a few milliseconds and ca. 2 seconds
- Most important time parameters are
  - T_{R}, repetition time, time between subsequent excitation pulses, typically a few hundred ms
  - T_{E}, echo time, time between excitation and echo measurement, often $(10 \ldots 100)\,\mathrm{ms}$

  Examples: $T_{\mathrm{R}} = 500\,\mathrm{ms}$, $T_{\mathrm{E}} = 15\,\mathrm{ms}$; $T_{\mathrm{R}} = 130\,\mathrm{ms}$, $T_{\mathrm{E}} = 4.5\,\mathrm{ms}$
- Readout time, the time during which RF echoes are recorded, typically ranges between 30 and $100\,\mathrm{ms}$

MR Imaging Sequences (5)

Basic Types of MR Sequences

Variation of the parameters allows for different relative weightings of the influences of tissues with high/low T1 or T2 on the signal.

- **T1-weighted** images
  - Short repetition time T_{R} suppresses the signal from tissues with long T1 (they had not enough time to restore their spin orientations) – strong weighting of T1 contrast
  - Long T_{R} – low T1 contrast
  - In T1-weighted images, tissues with long T1 appear darker than those with short T1

MR Imaging Sequences (6)

Basic Types of MR Sequences, cont.

- **T2-weighted** images
  - Remember that T2-relaxation is faster than T1-relaxation
  - In measurements with long T_{E}, tissues with short T2 have already lost much of their signal intensity due to T2-relaxation while tissues with long T2 have retained it – strong weighting of T2 contrast
  - In measurements with short echo time T_{E}, T2-decay has just started – low T2 contrast
  - In T2-weighted images, tissues with long T2 appear brighter than those with short T2
- **Proton density** images
  - Parameters are chosen such that neither T1 nor T2 dominates the overall contrast

MR Imaging Sequences (7)

Basic Types of MR Sequences, cont.

	T_{R}	T_{E}
T1-weighted	short	short
T2-weighted	long	long
proton-density	long	short

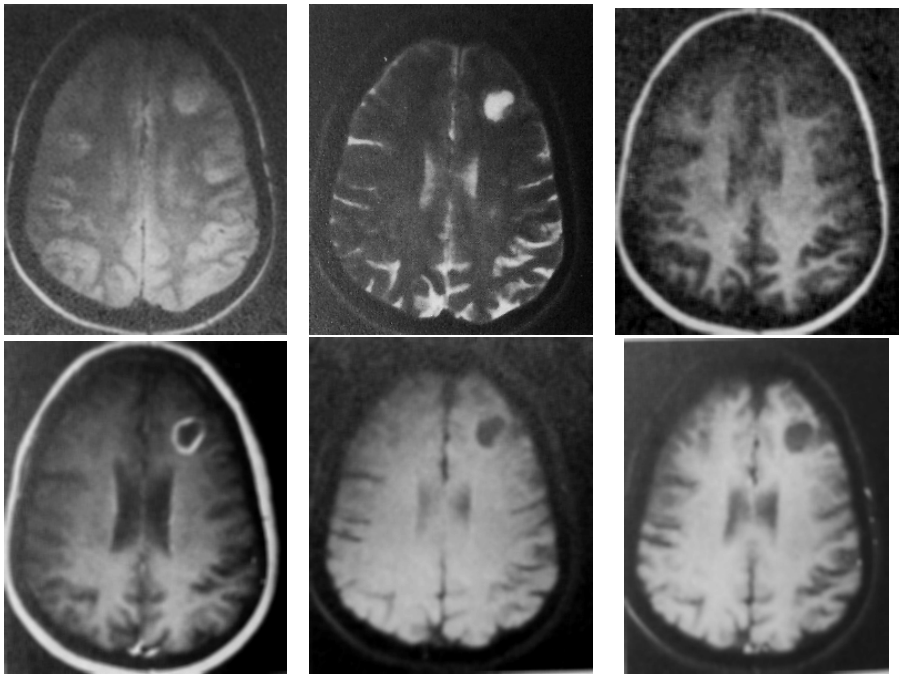
MR contrast types depending on sequence parameters. *(Weishaupt et al. 2006)*

- Proton-density, T1- and T2-weighted sequences allow to distinguish various types of tissues
- In some cases, contrast agents based on gadolinium are used
- There exist many other, more sophisticated sequences for specific purposes
- For instance, repeated and inverse pulses can be used
- Speedups can be achieved by refreshing magnetisation with further radio pulses between subsequent rows, etc.

Remark: In contrast to X-ray tomography, no absolute scale exists for MR measurements (of any type).

MR Imaging Sequences (8)

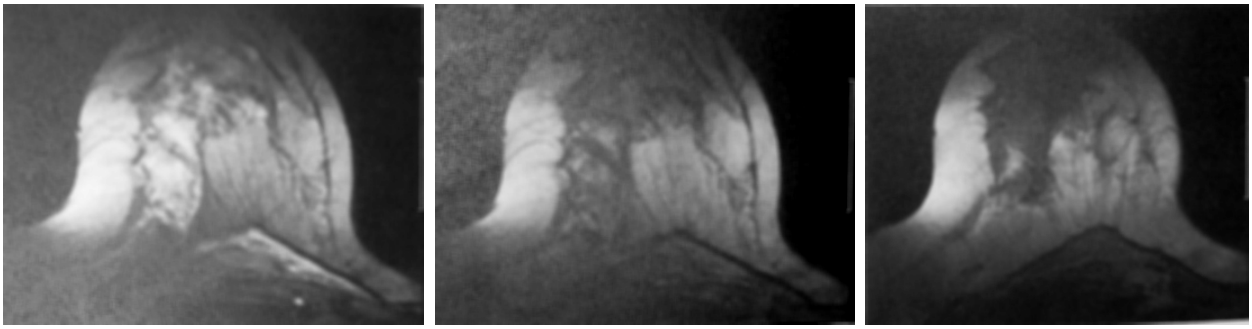
Example



Transaxial brain images, with secondary lung cancer deposits. **Top, left to right:** Proton density image – T2-weighted image – T1-weighted image. **Bottom, left to right:** T1-weighted image with gadolinium contrast agent – Flash 90 image – Flash 50 image. Flash sequences are fast measurement sequences which give mixtures of T1 and T2 contrast with different weights. *(Webb 1988)*

MR Imaging Sequences (9)

Example



Breast MR images. **Left to right:** Flash 20 – Flash 50 – Flash 70. The T1 contrast increases in weight from left to right. *(Webb 1988)*

Summary

- NMR uses magnetic properties of atomic nuclei to measure hydrogen concentrations.
- Hydrogen atoms in a magnetic field respond to a characteristic radio frequency (Larmor frequency) that depends on the field intensity.
- Spatial resolution is achieved by modulating the static base field with gradient fields in different directions.
  Different gradient fields are applied while exciting the atoms, during readout of the response, and inbetween.
- The measurements made for one slice amount to the Fourier transform of the actual image.
- Different time regimes of the measurement allow to bias the image contrast towards different materials / tissues (T1-weighted, T2-weighted, proton density sequences).
- MR scanners require an involved construction, and need extensive shielding due to the extreme electromagnetic field intensities.

References

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