Intensive Programme 2011 | Design, Synthesis and Validation of Imaging Probes

NMR and MRI : an introduction

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IMAGING PRINCIPLES | Introduction

Magnetic Resonance Imaging is an imaging modality which is primarily used to construct pictures of the NMR signal from the hydrogen atoms of water molecules contained in an object

The resonance frequency n of a spin is proportional to the magnetic field it is experiencing:

$$\omega = \gamma B_o$$

where γ is the gyromagnetic ratio of the spin and B_o the magnetic field strenght.

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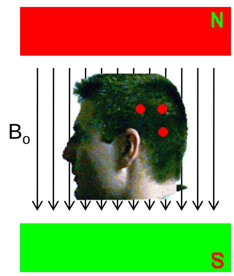
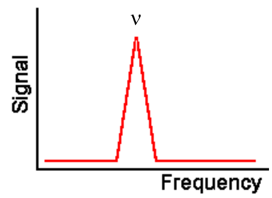
Introduction

Assume that a human head contains only three small distinct regions where there is hydrogen spin density



When these regions of spin are experiencing the same general magnetic field strength, there is only one peak in the NMR spectrum



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Magnetic Field Gradients

Gradients are small perturbations of the main magnetic field Bo taht are linearly dependent on the position within the magnet, with a typical imaging gradient producing a total field distorsion of less than 1%.

In presence of a gradient an expanded version of the Larmor equation applies:

$$\omega = \gamma (B_o + \mathbf{G} \bullet \mathbf{r}_i)$$

where $\mathbf{r}_i$ represents the position of the generic proton and **G** is a vector representing the total gradient amplitude and direction

The dimensions of **G** are usually expressed in mT/m or gauss/cm, where:

10 mt/m = 1 G/cm.

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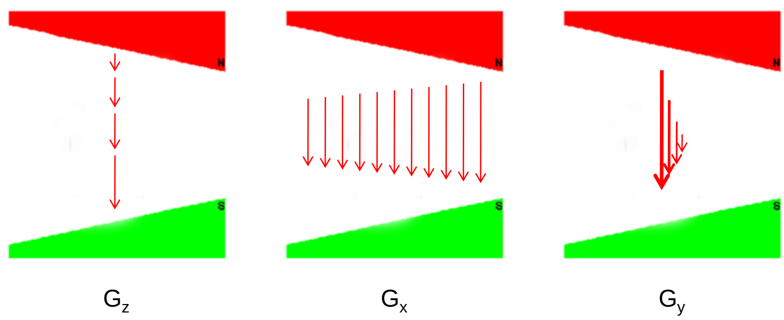
Magnetic Field Gradients

In presence of a magnetic field gradient, each proton will resonate at a unique frequency that depends on its exact position within the magnetic field.

Three physical gradients are used in imaging, one in each of the x, y, and z directions and are denoted as G_x, G_y and G_z.



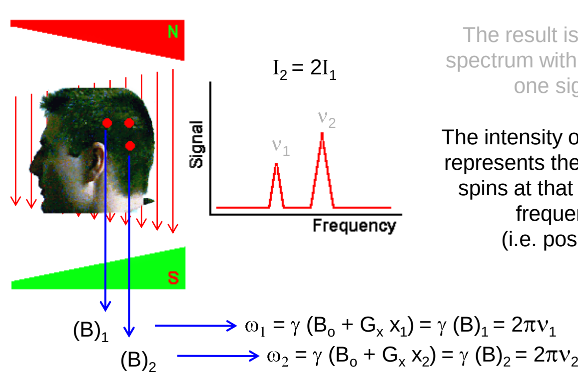
The length of the arrows represents the magnitude of the magnetic field (which is always directed along the z direction).

IMAGING PRINCIPLES

Magnetic Field Gradients : Frequency Encoding

Frequency Encoding

If a linear magnetic field gradient is applied (e.g. G_x) to our hypothetical head with three spin containing regions, the three regions experience different magnetic fields



The result is an NMR spectrum with more than one signal

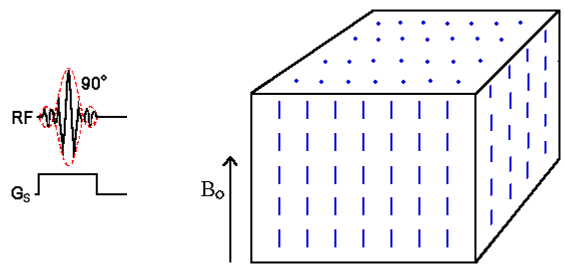
The intensity of the signal represents the amount of spins at that particular frequency (i.e. position)

IMAGING PRINCIPLES | **Magnetic Field Gradients : Slice Selection**

Slice selection

Slice selection in MRI is the selection of spins in a plane through the object

Slice selection is achieved by applying a one-dimensional,
linear magnetic field gradient during the period that the RF pulse is applied



IMAGING PRINCIPLES | **Reconstruction**

How could an MR image be acquired?

Two basic techniques

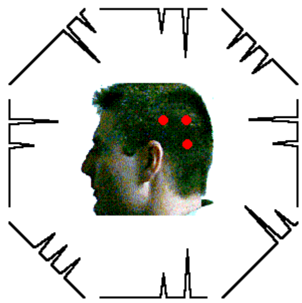
- Back-projection Imaging
- FT-Imaging (spin warp imaging)

IMAGING PRINCIPLES | **Reconstruction : Back Projection Imaging**

Backprojection imaging is a form of magnetic resonance imaging.
It is an extension of the frequency encoding procedure.

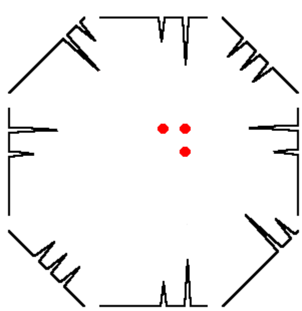
The object is first placed in a magnetic field

A one-dimensional field gradient is applied at several angles,
and the NMR spectrum is recorded for each gradient



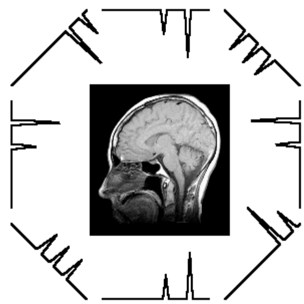
IMAGING PRINCIPLES | **Reconstruction : Back Projection Imaging**

Once the data has been acquired and recorded
they can be backprojected through space in computer memory



IMAGING PRINCIPLES | **Reconstruction : Back Projection Imaging**

In a real situation this is what could obtain



IMAGING PRINCIPLES | **Reconstruction : Back Projection Imaging**

Back projection imaging is also called
projection acquisition, projection imaging or radial acquisition.

It was the first kind of MRI acquisition
(Lauterbur 1973, Lai and Lauterbur 1981)

Advantages: - small Field Of View is possible
Disadvantages: - very dependent on inhomogeneities of B_0.
- very dependent on the quality of the magnetic filed gradients.

The technique was given up

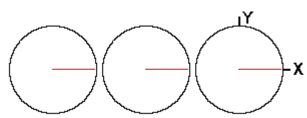
IMAGING PRINCIPLES | Magnetic Field Gradients : Phase Encoding

Phase Encoding

A phase encoding gradient is used to give a specific phase angle to a transverse magnetization vector.

The phase angle depends on the location of the transverse magnetization vector

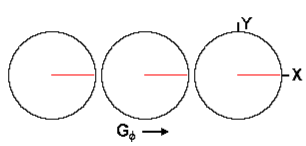
Let's imagine to have three regions with spins; after the 90° slice selection pulse the three magnetization vectors precess with the same frequency and phase



IMAGING PRINCIPLES | Magnetic Field Gradients : Phase Encoding

Phase Encoding

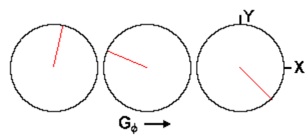
If a magnetic field gradient is applied along the X direction the three vectors will precess about the direction of the applied magnetic field (i.e. Z) at a frequency that depends on X



IMAGING PRINCIPLES | **Magnetic Field Gradients : Phase Encoding**

Phase Encoding

When the gradient in the X direction is then turned off the spins have acquired a phase angle and then continue to precess with the same frequency



IMAGING PRINCIPLES | **Magnetic Field Gradients : Phase Encoding**

Phase Encoding

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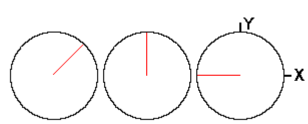
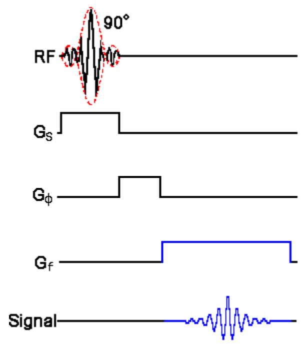


IMAGING PRINCIPLES | **Fourier Transform Tomographic Imaging**

The best way to understand a new imaging sequence
is to examine a timing diagram for the sequence

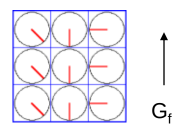


a radio frequency pulse
&
a slice selection gradient

a phase encoding gradient

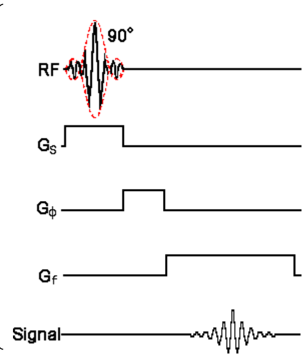
a frequency encoding gradient

an echo signal to be acquired



IMAGING PRINCIPLES | **Fourier Transform Tomographic Imaging**

The experiment has to be repeated with different values of the G_ϕ gradient



phase encoding gradient

signal to be acquired

n (e.g. 128)

For example, to obtain a 128x128 image, the whole experiment
has to be repeated 128 times with 128 different values of G_ϕ

IMAGING PRINCIPLES | Fourier Transform Tomographic Imaging

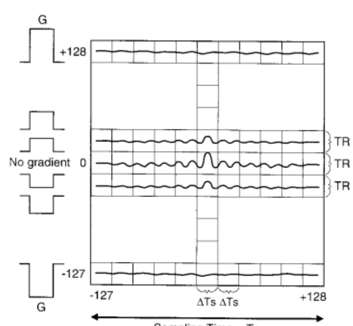
In every phase encoding step the phase is increased by

$$\Delta\phi = 360°/n$$

where n is the total number of phase encoding steps.

Each time we do a separate phase encode, it is a new excitation and a new value for the phase encoding magnetic gradient .

IMAGING PRINCIPLES | Fourier Transform Tomographic Imaging

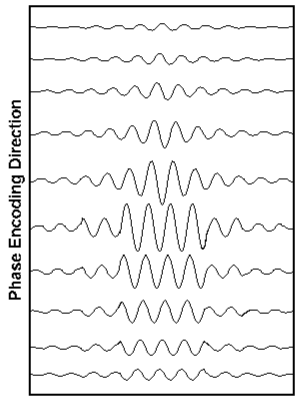


Each phase encoding step require a time TR

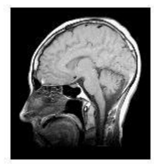
Phase encoding takes time

IMAGING PRINCIPLES | **Fourier Transform Tomographic Imaging**

The echo signals described above must be Fourier transformed in order to obtain an image or picture of the location of spins



2D - FT



IMAGING PRINCIPLES | **Fourier Transform Tomographic Imaging**

Each of the signals acquired fills one line in a set of rows called:

DATA SPACE

Its digitalized version is called:

K-SPACE

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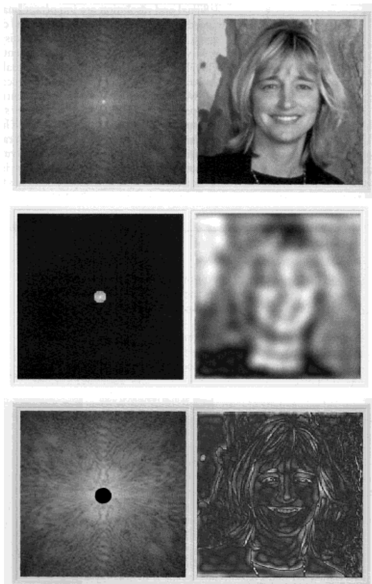
Fourier Transform Tomographic Imaging



All points of k-space are used

If the centre of k-space is used

most of the signal
low resolution

If the periphery of k-space is used

low signal
high resolution

IMAGING PRINCIPLES

Acquisition Parameters

The **dwell time** (DW) is the time between sampled points and determines

the **receiver spectral width** or **receiver spectral window** (SW) in the frequency domain.

$$SW = \frac{1}{2\,DW}$$

The SW is the largest frequency *difference* that we can distinguish

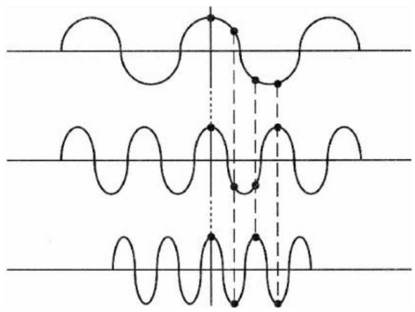
IMAGING PRINCIPLES | Acquisition Parameters

Frequency Encoding

The **Nyquist Theorem** states that in order for a frequency difference of SW to be measured, the time domain data (FID) has to be sampled at a frequency not less than 2*SW

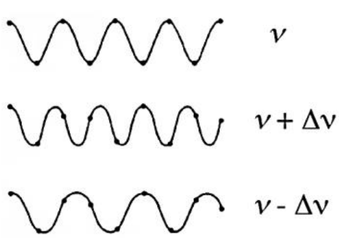
This frequency is called the **Nyquist frequency**

In other words, at least two samples per cycle are required to avoid **aliasing**



IMAGING PRINCIPLES | Acquisition Parameters

Aliasing



Frequencies greater than or less than ν cannot be discriminated from one another

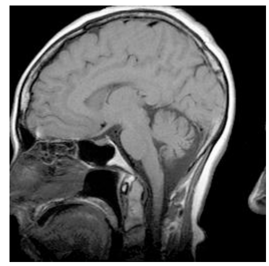
IMAGING PRINCIPLES | Acquisition Parameters

Aliasing

• at frequency ν, sampling occurs at 1/(2SW)

• if the same sampling frequency is used to digitize two signals, one at $\nu + \Delta\nu$ and one at $\nu - \Delta\nu$, we cannot discriminate between the faster and the slower signal

• for signals outside of SW (for signals digitized at a frequency less than their Nyquist frequency), the peaks corresponding to the signals will be folded in the image (wrapping artifact)



IMAGING PRINCIPLES | Image parameters

Spatial resolution

The dimensions (in pixel) of the image are determined by the number of phase encoding steps and the number of points used to acquire the NMR signal during frequency encoding

Both square and rectangular matrices can be used

The dimensions (in cm/mm) of the image is called the Field-Of-View (FOV)

Both square and rectangular FOVs can be used

The spatial resolution in both dimensions is given by:

$$resolution = \frac{FOV}{matrix\ size}$$

IMAGING PRINCIPLES | **Image parameters**

Spatial resolution

What is the minimum spatial resolution possible?

$$\omega = 2\pi\upsilon = \gamma(B_{o+}G_R r) \qquad \upsilon = \frac{\gamma}{2\pi}(B_o + G_R r) \qquad \Delta r = \frac{2\pi\Delta\upsilon}{\gamma G_R}$$

with a reasonable width of the water signal of ~ 200Hz it gives:

with G_R = 500 mT/m (μ-Imaging) → r > 10 μm

with G_R = 25 mT/m (human MRI) → r > 200 μm

IMAGING PRINCIPLES | **Image parameters**

Frequency dimension of the pixels

The water signal on the extremes of the FOV differ in frequency of SW Hz

The "dimension" in Hz of the pixel is:

$$\frac{SW}{matrix\ size}$$

The most abundant substance in MRI studies is water,
but in some districts a strong signal arise from fat protons.

Between fat and water there is a chemical shift difference of ≈ 3.5 ppm

IMAGING PRINCIPLES | **Image parameters**

Frequency dimension of the pixels

This separation (in Hz) depends on the strength of the main magnetic field:

Es.: B_o = 1,5 T

$$\omega_o = \gamma B_o = 42{,}6\,\frac{MHz}{T}\,1{,}5\,T \cong 64\,MHz$$
$$\Delta\omega = \Delta\delta\omega_o = 3{,}5\;10^{-6}\;64\;10^{6} \cong 220Hz$$

With a receiver bandwidth (SW) of 32 kHz and a square matrix of 256 pixels:

$$\frac{SW}{pixel} = \frac{32\,kHz}{256} = 125\,Hz$$

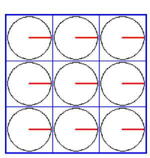
$$pixel\ difference = \frac{220}{125} \cong 2$$

There will be a difference of ca 2 pixels between signals arising from water and fat that belong to the same region

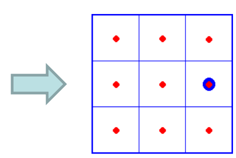
IMAGING PRINCIPLES | **Image parameters**

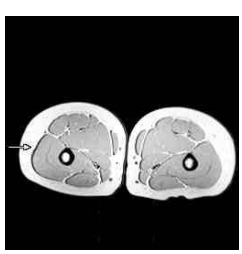
Chemical shift artifact

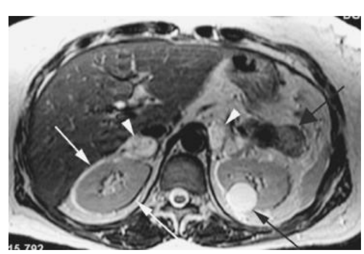
The signal from the fat (in blue) is precessing at the same frequency of the water in the adjacent pixel



This give rise to an artifact in the image called chemical shift artifact:

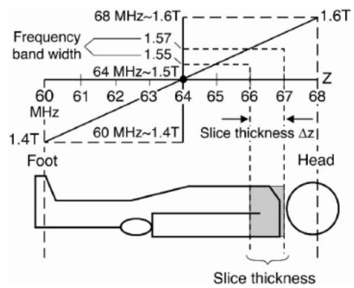






IMAGING PRINCIPLES | **Image parameters**

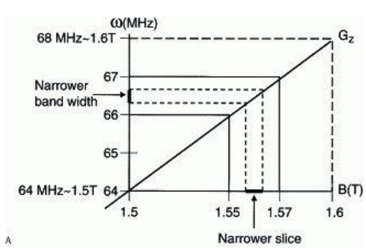
The slice thickness and slice position is determined by the interplay of gradient strength, frequency and pulse bandwidth

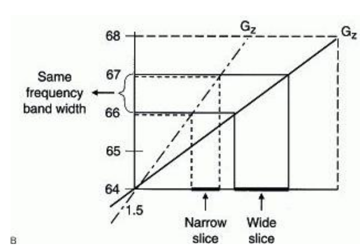


slice thickness ↔ pulse bandwidth, gradient strength
position ↔ pulse frequency

IMAGING PRINCIPLES | **Image parameters**

The slice thickness can be varied either by changing the pulse bandwidth or the gradient strength





narrow slice ↔ narrow bandwidth / steep gradient
wide slice ↔ wide bandwidth / shallow gradient

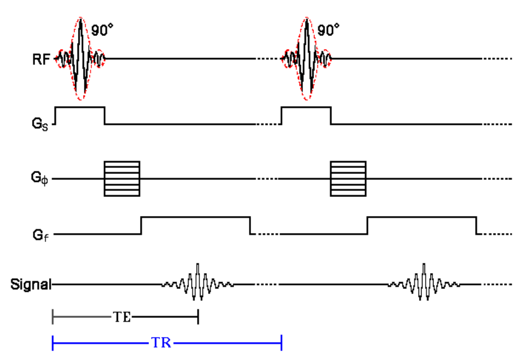
PULSE SEQUENCES | **Introduction**

The different steps that make up an *MR pulse sequence* are:

- Excitation of the target area
  - Switching on the *slice-selection gradient*,
  - Delivering the *excitation pulse* (RF pulse),
  - Switching off the *slice-selection gradient*.
- Phase encoding
  - Switching on the *phase-encoding gradient* repeatedly, each time with a different strength, to create the desired number of phase shifts across the image.
- Formation of the echo or MR signal
  - *Generating an echo*
- Collection of the signal
  - Switching on the *frequency-encoding or readout gradient*,
  - *Recording* the echo.

These steps are repeated many times, depending on the desired image quality.

PULSE SEQUENCES | **Pulse sequences**



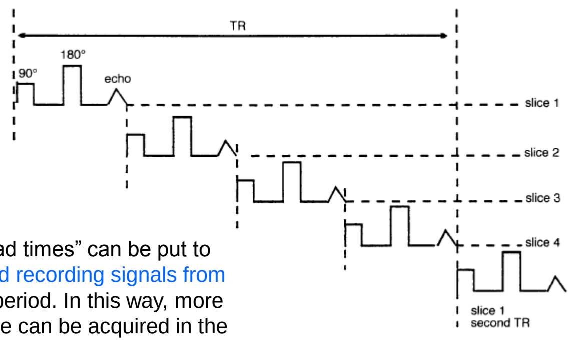
TE is the time between the first pulse and the maximum of the acquired echo

TR is the time between two successive repetitions of the pulse sequence

PULSE SEQUENCES | **Multislice Imaging**

Multislice Imaging

- Conventional imaging with *"inactive"* repetition times (TR) between two successive excitation pulses is highly inefficient, especially when using sequences with long scan times and long TR (e.g. scan time of more than 4 min for acquisition of a T_{1w} SE image with 256 excitations and a TR of 1000 ms).



- The "wait times" or "dead times" can be put to good use by exciting and recording signals from other slices during this period. In this way, more slices instead of only one can be acquired in the same time.

PULSE SEQUENCES | **Classification**

A wide variety of sequences are used in medical MR imaging.

The main classification between the pulse sequences is:

- Spin echo (SE) sequences
- Gradient echo (GRE) sequences.

In SE sequences the spin-echo sequence 90° - τ - 180° - τ - echo is applied.

In GRE an echo is generated by the inversion of the frequency-encoding gradient.

These sequences are the basic MR pulse sequences.

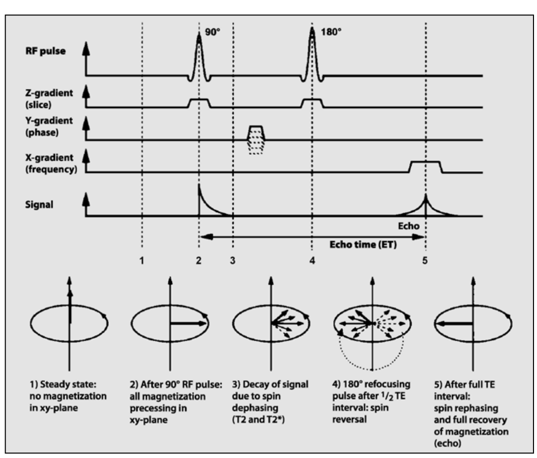
PULSE SEQUENCES | Spin echo

Spin Echo (SE) Sequences

- Spin echo sequences use a *slice-selective 90° RF pulse* for excitation.
- Then dephasing occurs because some spins precess faster than others as a result of the static magnetic field inhomogeneities.
- A *180° RF pulse* is then delivered to reverse and refocus the spins.
- It serves to eliminate the effects of static magnetic field inhomogeneities but cannot compensate for *variable* field inhomogeneities that underlie spin-spin interaction (T_2).

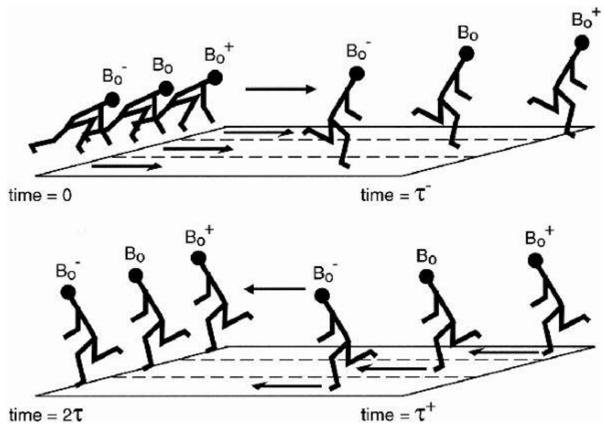
PULSE SEQUENCES | Spin echo

Spin Echo Pulse Sequence Diagram



PULSE SEQUENCES | **Spin echo**

Spin Echo : the analogy of the runners



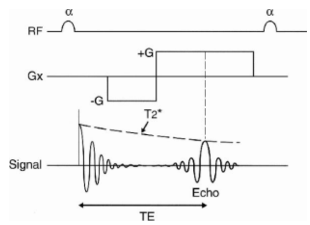
PULSE SEQUENCES | **Spin echo**

- Spin echo sequences are characterized by an excellent image quality because the effects of static field inhomogeneities are eliminated by the 180° RF pulse.
- The tradeoff is a fairly long scan time (because of the 90° excitation pulse)

PULSE SEQUENCES | **Gradient echo**

Gradient Echo (GRE) Sequences

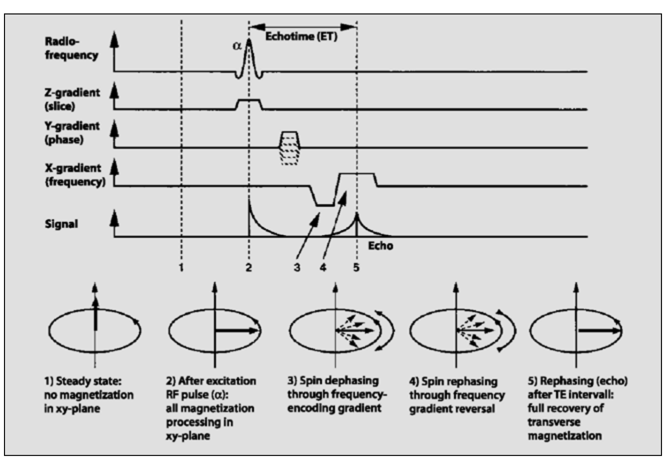
- They are also known as gradient-recalled echo or fast field echo (FFE) sequences.
- The frequency-encoding gradient coils are used to produce an echo rather than a pair of RF pulses. This is done by first applying a negative pulse to destroy the phase coherence of the precessing spins (dephasing). Subsequently, the gradient is reversed and the spins rephase to form a gradient echo.



- Flip angles α different than 90° can be used.

PULSE SEQUENCES | **Gradient echo**

Gradient Echo Pulse Sequence Diagram



PULSE SEQUENCES **Gradient echo**

- Since the flip angle can be set to less than 90° it does not take too much for the magnetization to recover, so very short repetition times (TR) can be achieved.
- *Faster imaging* is possible compared with SE.
- Static field inhomogeneities are not compensated so the images show a decreased quality (with respect to SE).

REFERENCES

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Ray Hashman Hashemi, William G. Bradley, Christopher J. Lisanti.

Intensive Programme 2011 | Design, Synthesis and Validation of Imaging Probes

Contrast and T_1 / T_2 measurement

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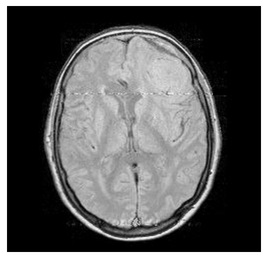
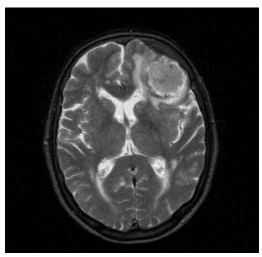
IMAGE CONTRAST | Introduction

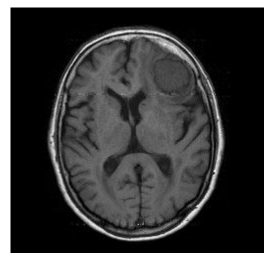
In order for a structure (physiological or pathological) to be clearly visible in a magnetic resonance image there must be contrast (i.e. a difference in signal intensity) between it and the adjacent tissues



Low contrast



High contrast



High contrast

IMAGE CONTRAST | Parameters Influencing Image Contrast

The signal intensity S, is determined by the signal equation for the specific pulse sequence used.

It depends on intrinsic and instrumental (sequence specific) variables

Some intrinsic variables are:

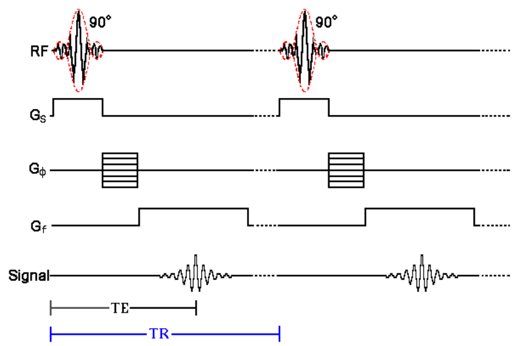
Spin-Lattice Relaxation Time, T_1
Spin-Spin Relaxation Time, T_2
Spin Density, ρ

Some instrumental (sequence) variables are:

Repetition Time, TR
Echo Time, TE
Flip Angle, θ
Inversion Time, TI

IMAGE CONTRAST | Parameters Influencing Image Contrast

The acquisition of an image requires the acquisition of a number of echoes



TE is the time between the first pulse and the maximum of the acquired echo

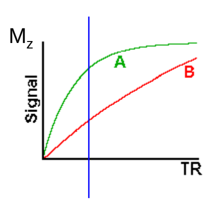
TR is the time between two successive repetitions of the pulse sequence

By varying the values of TR ad TE it is possible to modify the amount of contrast in the image

IMAGE CONTRAST | T_1 and T_2 weighted Images : Effect of TR

T_1 contrast (T_1 weighted images)

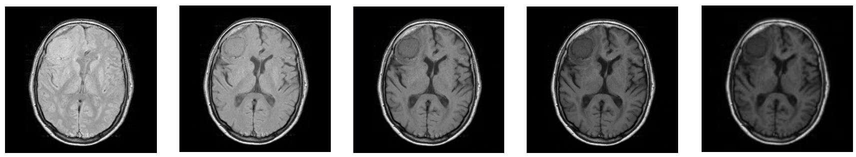
$(T_1)_A < (T_1)_B$



With short values of TR the signal coming from regions with short T_1 will be more intense (image more white) than that coming from regions with long T_1 (image mostly black)

IMAGE CONTRAST — T_1 and T_2 weighted Images : Effect of TR

T_1 contrast (T_1 weighted images)



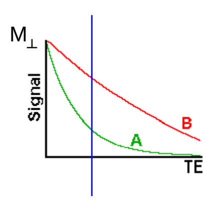
TE = 1 ms TR = 5000 ms	TE = 1 ms TR = 2000 ms	TE = 1 ms TR = 1000 ms	TE = 1 ms TR = 500 ms	TE = 1 ms TR = 200 ms

Increased T_1 weighting →

IMAGE CONTRAST — T_1 and T_2 weighted Images : Effect of TE

T_2 contrast (T_2 weighted images)

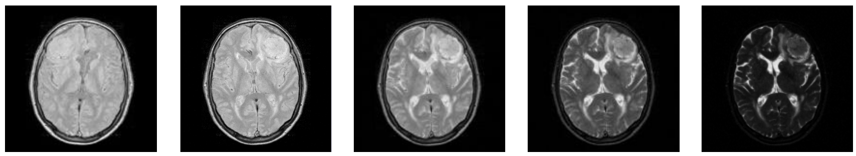
$(T_2)_A < (T_2)_B$



With short values of TE the signal coming from regions with short T_2 will be less intense (image more black) than that coming from regions with long T_2 (image mostly white)

IMAGE CONTRAST — T_1 and T_2 weighted Images : Effect of TE

T_2 contrast (T_2 weighted images)

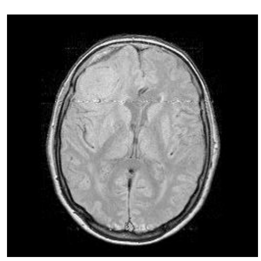


TE = 1 ms TR = 5000 ms	TE = 20 ms TR = 5000 ms	TE = 50 ms TR = 5000 ms	TE = 100 ms TR = 5000 ms	TE = 200 ms TR = 5000 ms

Increased T_2 weighting

IMAGE CONTRAST — Proton Density Images

Long values of TR together with short values of TE give raise to images in which the signal is proportional to the amount of spins inside the region



TE = 1 ms
TR = 5000 ms

A proton density image: everything is almost the same color, except for the skull

T_1 AND T_2 MEASUREMENT — Introduction

The spin-lattice relaxation time (T_1), spin-spin relaxation time (T_2), and the spin density (ρ) are properties of the spins in tissues.

The value of these quantities change from one normal tissue to the next, and from one diseased tissue to the next.

The calculation of T_1 and T_2 starts with the collection of a series of images.

Then, best fitting procedures are used to extract T_1 and T_2 values for every pixel using the equations of motion for the magnetization

T_1 AND T_2 MEASUREMENT — T_1 measurement

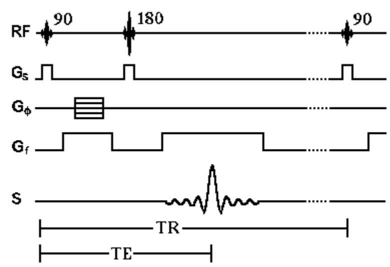
The main sequences that can be used to measure T_1 are:

- Spin Echo Saturation Recovery
- Spin Echo Inversion Recovery
- Gradient Echo Inversion Recovery FLASH (Fast Low Angle Shot)
- Gradient Echo Inversion Recovery SNAPSHOT FLASH

T_1 AND T_2 MEASUREMENT

T_1 measurement: SE Saturation Recovery

Spin Echo Saturation Recovery

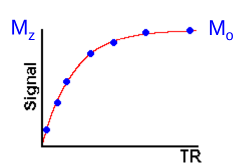


For a 128x128 image the sequence has to be repeated 128 times for every value of TR, with TR values ranging from 0 to at least 5 times the longest T_1 in the object.

TE has to be kept fixed and at the minimum possible value.

T_1 AND T_2 MEASUREMENT

T_1 measurement: SE Saturation Recovery



The signal intensity measured from the images can be plotted against TR.

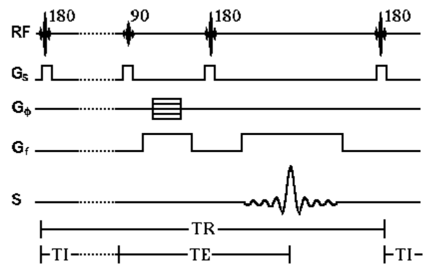
The data can be fitted against equation:

$$M_z = M_o [1 – \exp(- TR / T_1)]$$

In order to evaluate the value for T_1

T_1 AND T_2 MEASUREMENT | T_1 measurement: SE Inversion Recovery

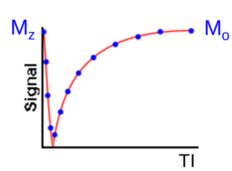
Spin Echo Inversion Recovery



For a 128x128 image the sequence has to be repeated 128 times for every value of TI, with TI (inversion time) values ranging from 0 to at least 5 times the longest T_1 in the object.

TE has to be kept fixed and at the minimum possible value, while TR has to be fixed and at least 5 times the longest T_1 in the image.

T_1 AND T_2 MEASUREMENT | T_1 measurement: SE Inversion Recovery



The signal intensity measured from the images can be plotted against TI.

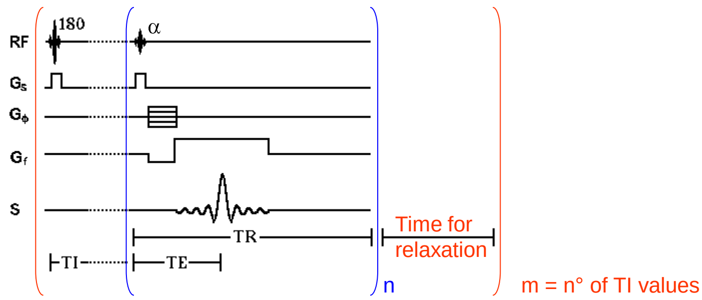
The data can be fitted against equation:

M_z = abs (M_o [1 – 2 exp(- TI / T_1)])

In order to evaluate the value for T_1

T_1 AND T_2 MEASUREMENT | T_1 measurement: GE IR FLASH

Gradient Echo IR FLASH

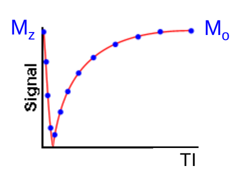


For every single value of TI (ranging from 0 to 5 time the longest T_1 in the object) the sequence in blue brackets has to be repeated n times (for example128).

TE and TR has to be kept fixed and at the very low value, while α is a very short pulse angle (5° - 10°).

T_1 AND T_2 MEASUREMENT | T_1 measurement: GE IR FLASH

The data analysis is the same used for Spin Echo Inversion Recovery

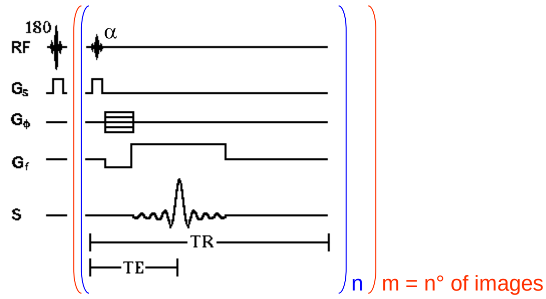


The signal intensity measured from the images can be plotted against their time of acquisition (i.e. TI if acquisition is centric).

$$M_z = \text{abs} (M_o [1 – 2 \exp(- t_{image} / T_1)])$$

T_1 AND T_2 MEASUREMENT | T_1 measurement: GE IR SNAPSHOT FLASH

Gradient Echo IR SNAPSHOT FLASH

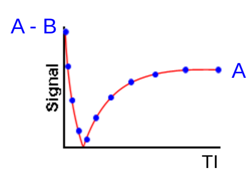


After a 180° pulse, a train of GE images is acquired for a time span covering at least 5 times the longest T_1 in the object.

TE and TR has to be kept fixed and at the very low value, while α is a very short pulse angle (5° - 10°).

T_1 AND T_2 MEASUREMENT | T_1 measurement: GE IR SNAPSHOT FLASH

The continuing pulsing (α) causes the return to equilibrium of the magnetization to be faster and the equilibrium value M_o is no longer reached because of saturation



The signal intensity measured from the images can be plotted against their time of acquisition (i.e. TI if acquisition is centric).

The data can be fitted against equation:

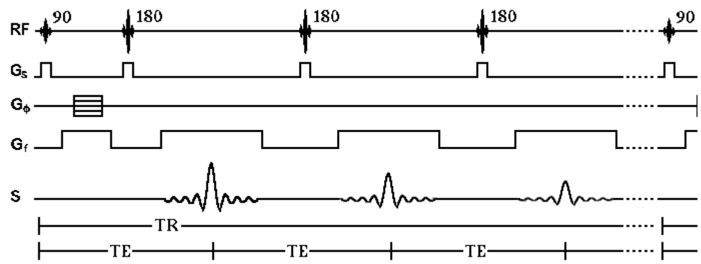
$$M_z = \text{abs} [A – B \exp(- t_{image} / T_1^*)]$$

$$\text{where } T_1^* = T1 / (B/A -1)$$

T_1 AND T_2 MEASUREMENT

T_2 measurement: Multi SE

Sequence used to measure T_2 : Multi Spin Echo

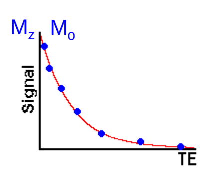


In a Multi Spin Echo experiment a 90° pulse is followed by a train of TE-separated 80° pulses that give raise to echoes of decreasing intensity due to T_2 relaxation.

TE has to be chosen according to the expected T_2, while TR has to be long enough to permit the recovery of the equilibrium magnetization.

T_1 AND T_2 MEASUREMENT

T_2 measurement: Multi SE



The signal intensity measured from the images can be plotted against TE.

The data can be fitted against equation:

$M_z = M_o \exp(- TE / T_2)$

In order to evaluate the value for T_2

REFERENCES

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