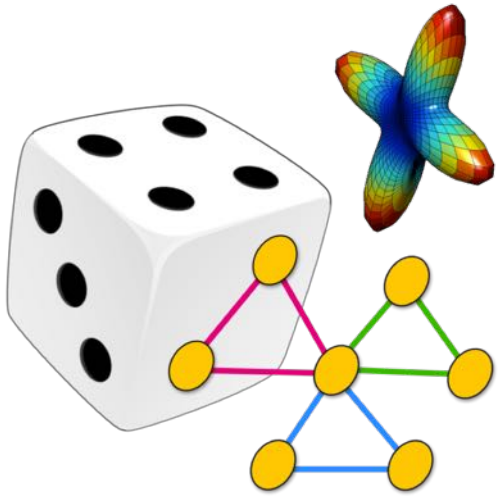

Nuclear Magnetic Resonance and Magnetic Resonance Imaging

Simona Schiavi
DICE LAB, Diffusion Imaging and Connectivity Estimation
Computer Science Department, Univerisity of Verona

Course schedule

- May 31st, 8:30-10:30, alpha: Introduction to python
- June 07th, 8:30-10:30, alpha: How to obtain RM images and python lab (FFT)
- June 10th, 10:30-12:30, H: Bloch Torrey equation and homogenization techniques
- June 11th, 8:30-10:30, gamma: solution of Bloch Torrey equations in simple 2D geometry in FreeFem
- June 12th, 14:30-15:30, F: Numerical Convex Optimization applied to diffusion MRI

Announcement

The challenge of mapping the human connectome based on diffusion MRI tractography

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Speaker: Maxime Descoteaux - Sherbrooke University
Tuesday, June 18, 2019 at 11:00 AM Aula Verde

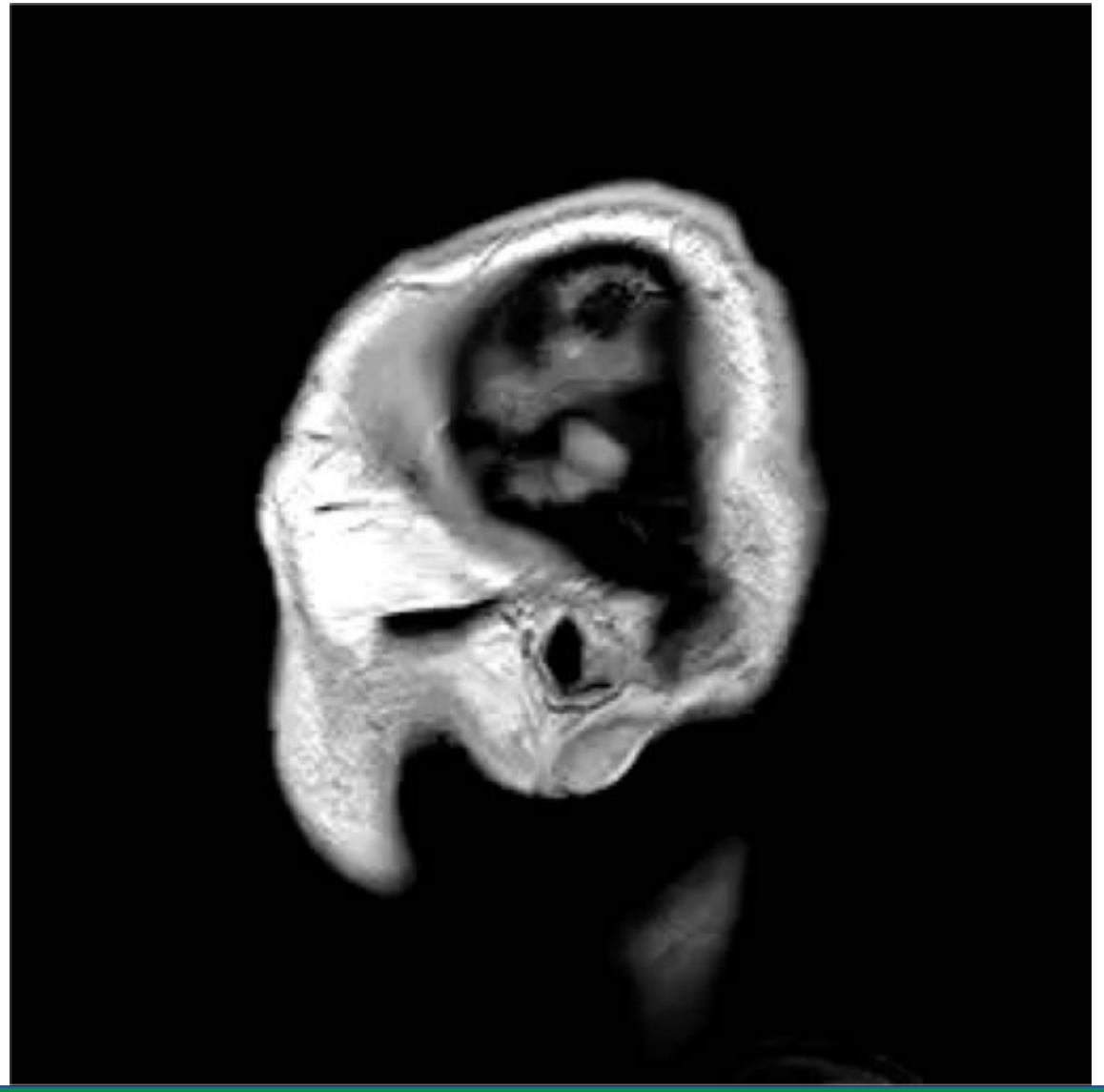
In this talk, I will cover what tractography and tractometry can do well and what are some of the important current tractography limitations such as the length, position, shape and gyral biases, and will present some solutions that start addressing these limitations. The lack of ground truth on long-range connectivity of the human brain makes it hard to quantitatively evaluate results. A key challenge for future tractography algorithms will be to control for false positives, while identifying the full extent of existing fiber bundles. Exiting new solutions are possible with microstructure-informed tractography and machine learning-based techniques.

Maxime DESCOTEAUX, PhD is Professor in Computer Science since 2009 at the Science Faculty of Sherbrooke University. He is the founder and director of the Sherbrooke Connectivity Imaging Laboratory (SCIL) (**http://scil.usherbrooke.ca/**). His research focuses on brain connectivity from state-of-the-art diffusion MRI acquisition, reconstruction, tractography, processing and visualization. The aim of the SCIL is to better understand structural connectivity, develop novel tractography algorithms, validate them and use them for human brain mapping and connectomics applications.Maxime Descoteaux did a post-doctorat fellow at **NeuroSpin** under the supervision of Cyril Poupon and Denis Le Bihan. He also obtained a PhD in Computer Science at **INRIA Sophia Antipolis** - Mediterranée, supervised by R. Deriche after he obtained a M.Sc under the supervision of K. Siddiqi in **Computer Science** at **Center for Intelligent Machines**, McGill University, where he also obtained a B.Sc, graduating from the joint honors **Mathematics** and Computer Science program.

Contact Person: Gloria Menegaz

Example MRI image

Magnetic resonance imaging (MRI) is an imaging technique used primarily in medical settings to produce *in-vivo* high quality images of the inside of the human body.



How the scanner resembles...

A bit of history

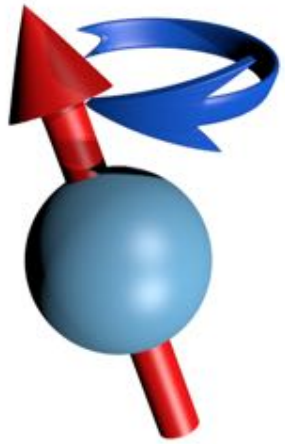
- **1946:** Felix Bloch and Edward Purcell (Nobel laureates in 1952) discovered the phenomenon and applied it in analytic chemistry for identifying chemical compounds in a specimen
- **1967:** first in vivo NMR signal (a finger)
- **1971:** Raymond Damadian showed that the nuclear magnetic relaxation times of tissues and tumours differed
- **1975:** Richard Ernst proposed magnetic resonance imaging using phase and frequency encoding, and the Fourier Transform
- **1980:** clinical Magnetic Resonance Imaging

NMR
Nuclear Magnetic Resonance

Nuclear Magnetism: Spins

NMR: nuclei in a magnetic field absorb and re-emit electromagnetic radiation

An element shows NMR only if it has **non-zero spin** (i.e. the number of protons or neutrons in an atom is odd).



Nuclei	Unpaired Protons	Unpaired Neutrons	Net Spin	γ (MHz/T)
^{1}H	1	0	1/2	42.58
^{2}H	1	1	1	6.54
^{31}P	1	0	1/2	17.25
^{23}Na	1	2	3/2	11.27
^{14}N	1	1	1	3.08
^{13}C	0	1	1/2	10.71
^{19}F	1	0	1/2	40.08

In human tissue, the signal typically derives from the ^{1}H protons of H_2O.

Nuclear Magnetism of ^{1}H

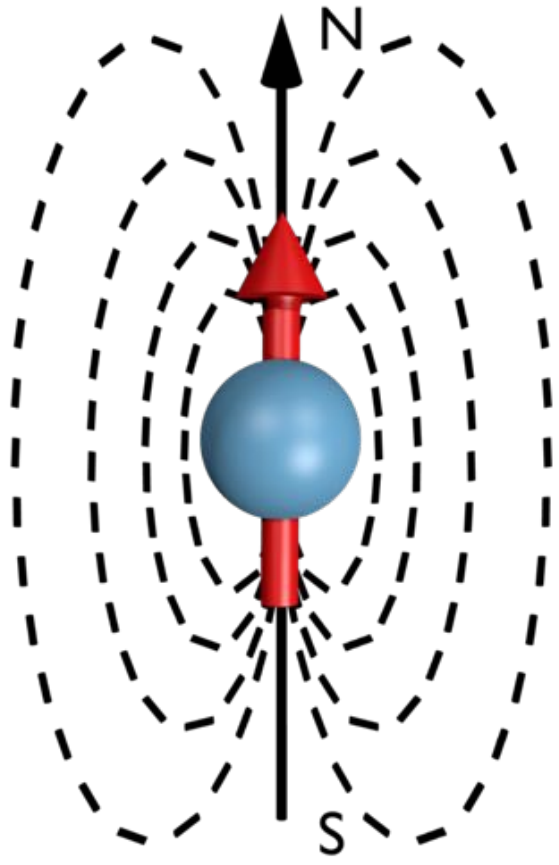
The hydrogen's nucleus is composed of just a single positively charged particle (proton).

This charge induces the nuclear magnetism phenomenon

This nucleus with its charge behaves as a small magnet which has magnetic field lines.

We can describe the magnetic field as a vector:

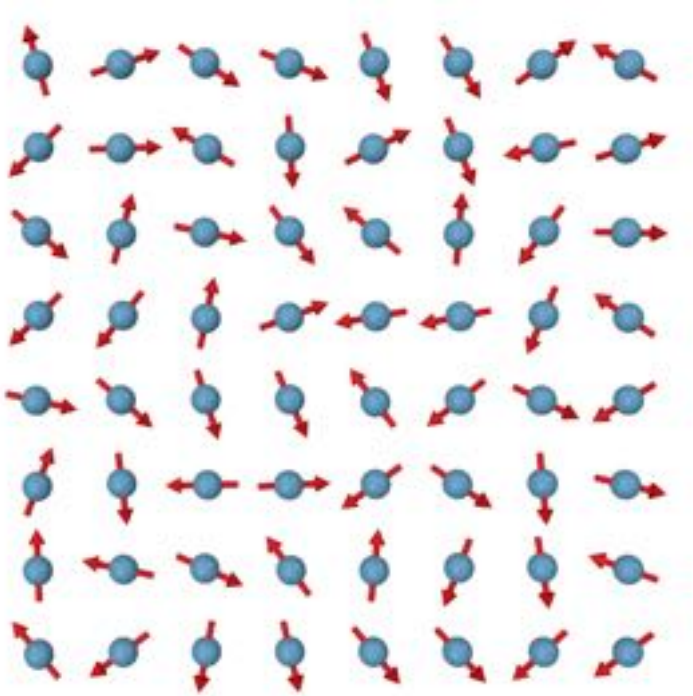
- **length** = magnitude of the generated field
- **orientation and direction** = orientation and direction of the field



Nuclear Magnetism

We cannot observe a single proton, but always a **population**.

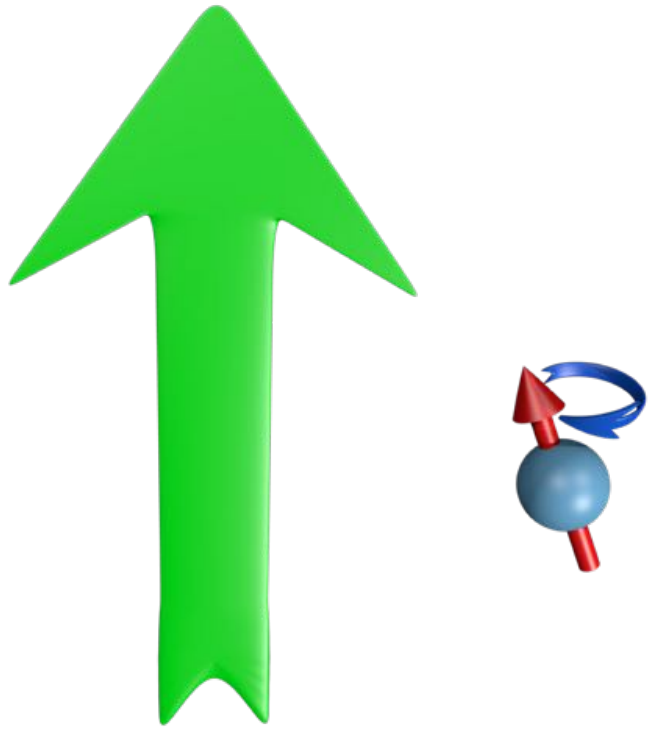
In the absence of any external magnetic field, the orientation of every nuclear proton is random.



The resultant magnetic field is 0

If we put a strong external magnetic field ...

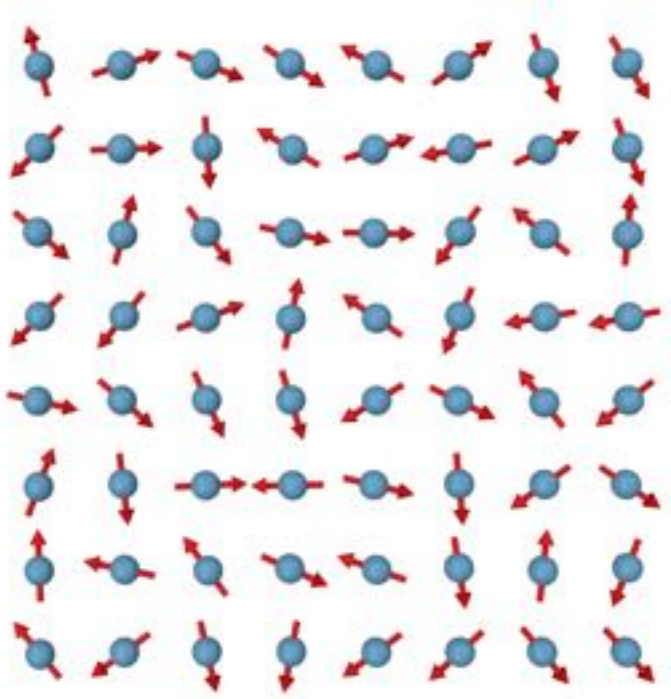
When a proton is put into a magnetic field produced by an external source it is subject to a torque tending to orient its magnetic moment parallel to the field.



The proton's spin will align parallel to the external magnetic field.

Thus for a population of spins ...

All the spins will align parallel to the applied magnetic field but they can choose between two orientations: **parallel** and **anti-parallel**.

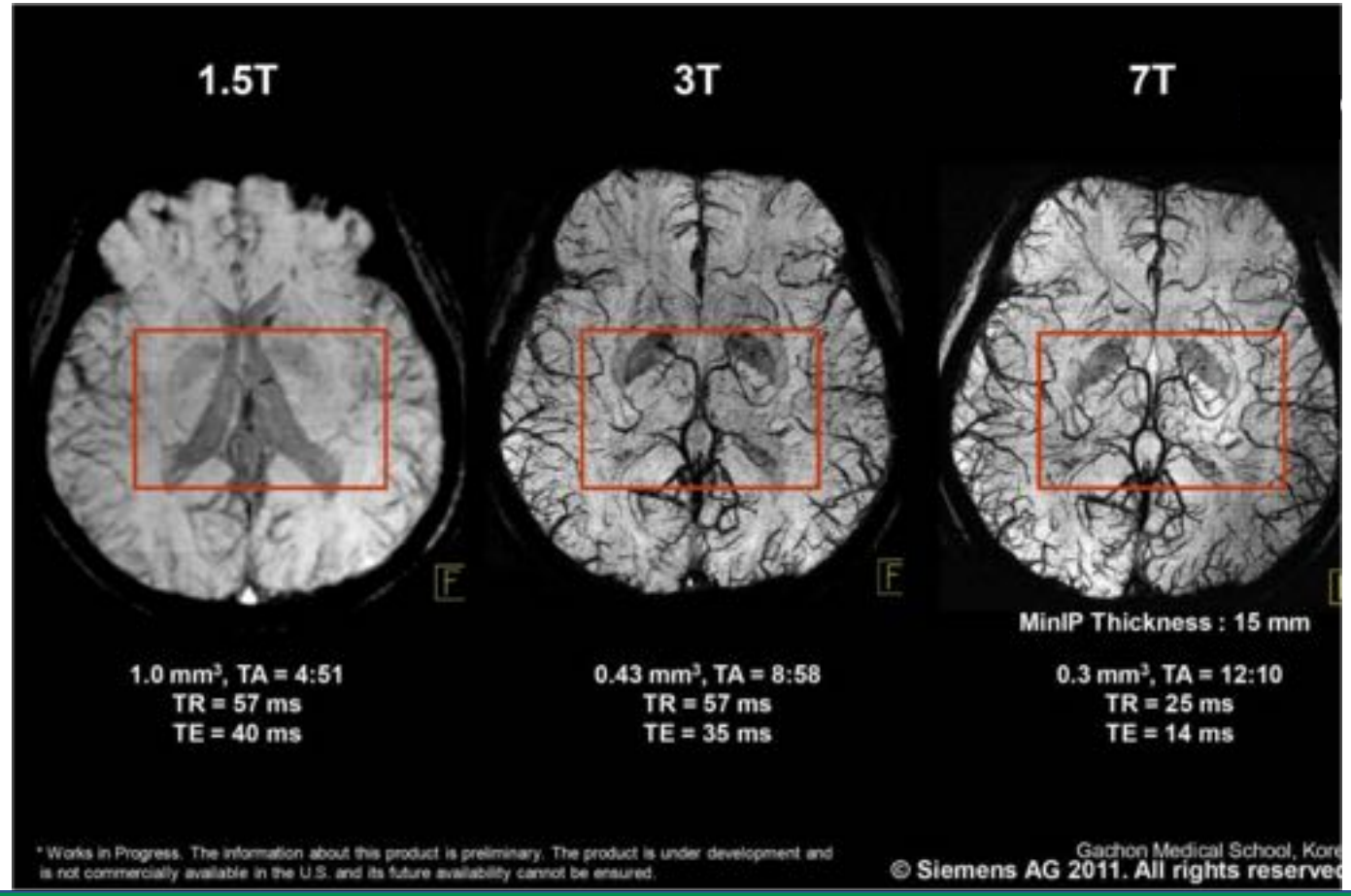


At the common field strength of 1.5 Tesla **only 6/10000 spins contribute to the signal**

Curious fact

You might have probably heard of 1.5, 3, or 7 Tesla MRI scanners…

These values indicate the strength of the magnetic field $\mathbf{B}_0$



Spin angular momentum

When an external magnetic field $\mathbf{B}_0$ is applied, the proton will tend to align with the field, but, because of its spin it will settle out of alignment with the external magnetic field. Its magnetic vector will then **rotate in a cone shape trajectory** around the axis identified by $\mathbf{B}_0$.

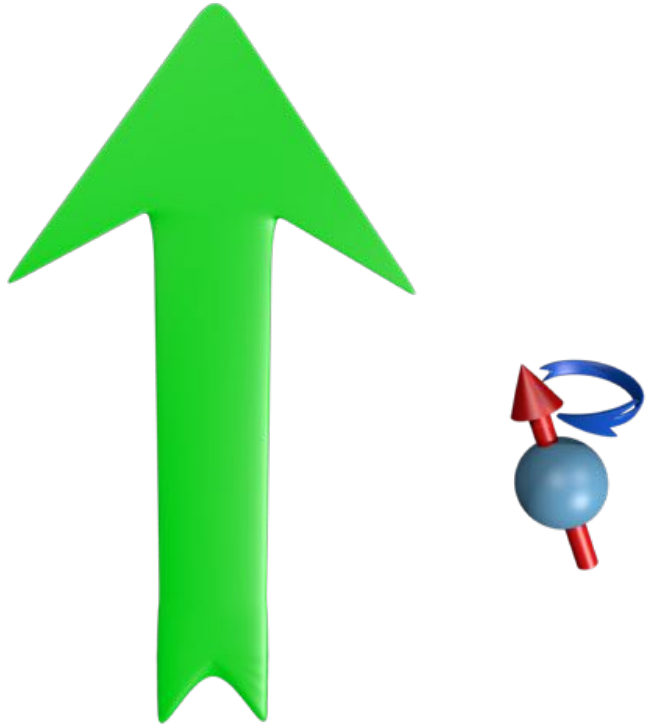
This particular rotation movement is called **precession**

Larmor equation: $\omega_0 = \gamma \boldsymbol{B}_0$

magnetic field

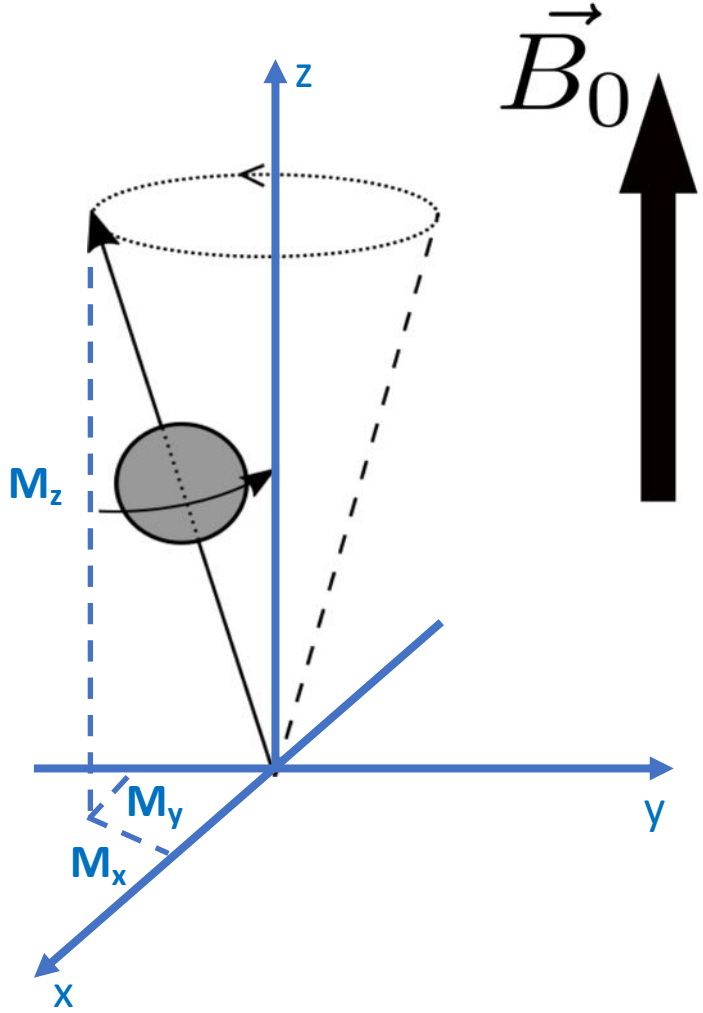
frequency

gyro-magnetic ratio



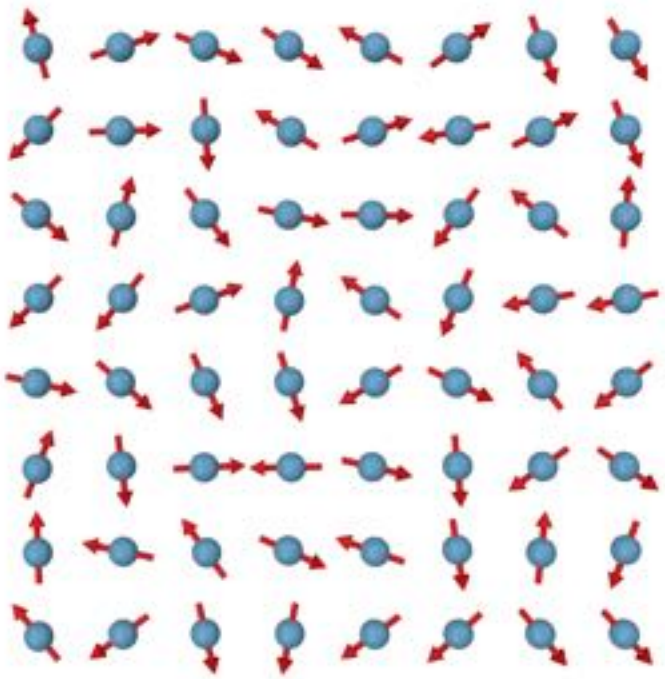
Spin angular momentum

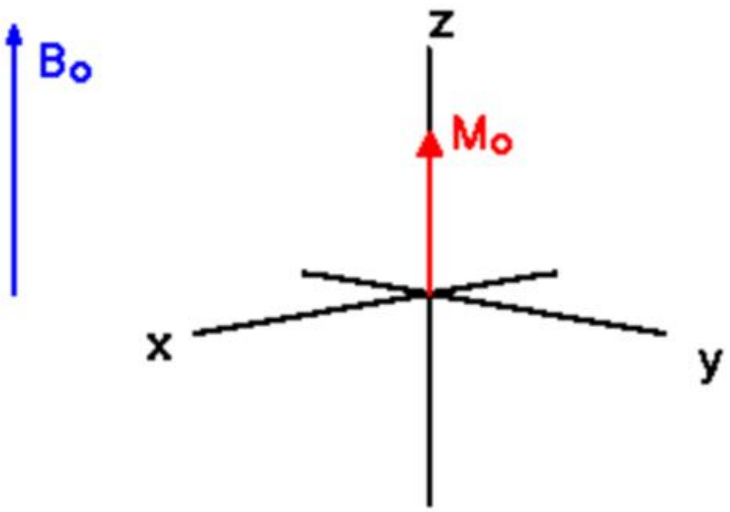
We can decompose each spin into its 3 components along the Cartesian axes



Spins angular momentum

Because of the initial random orientation, a population of spins will have **0 total transverse component** and a **non-zero parallel** component.

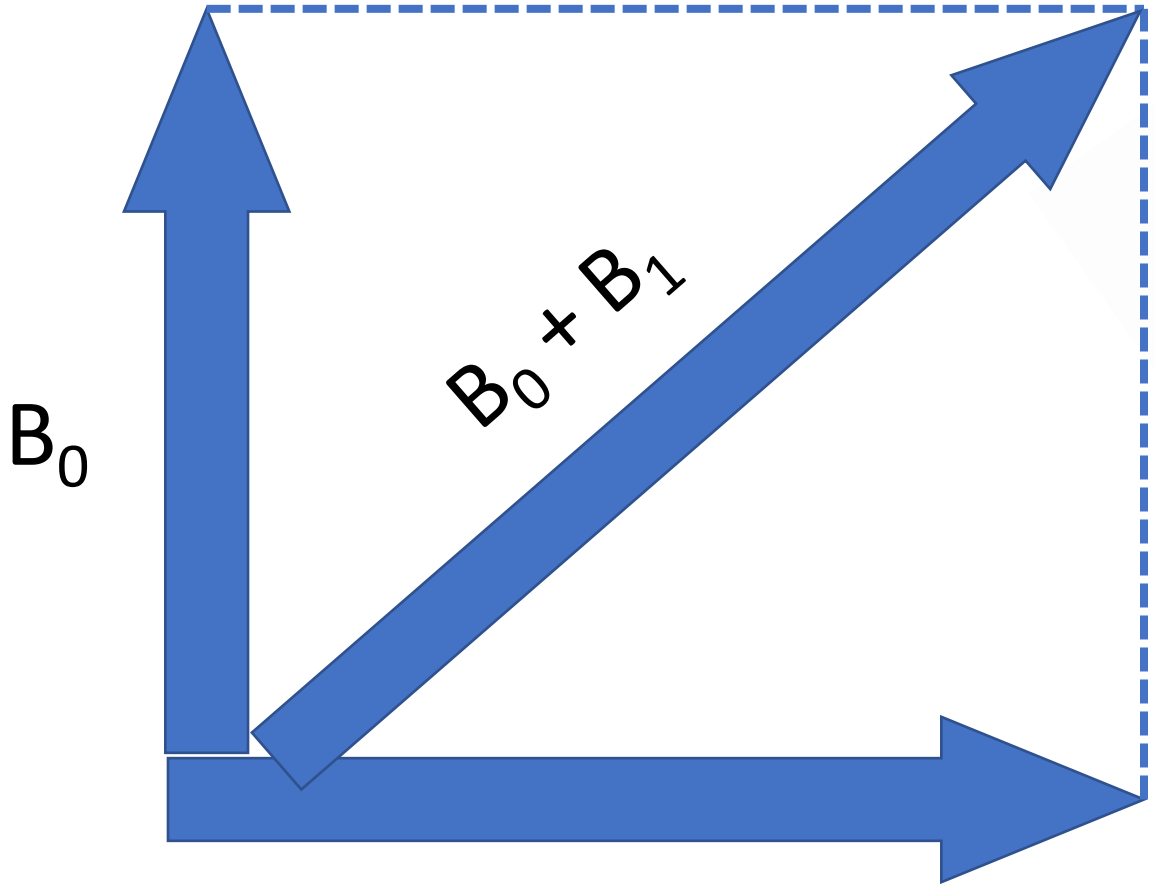


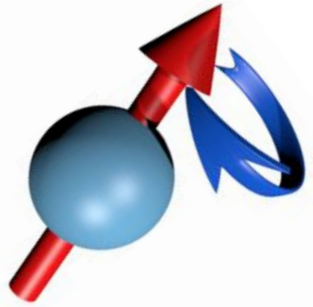


Since only 6/10000 spins contributes to $\mathbf{M}_0$ and their magnetization is very small comparing to $\mathbf{B}_0$ we need to do something to be able to measure a signal!

How do we measure magnetization?

Idea: use the sum of vectors!

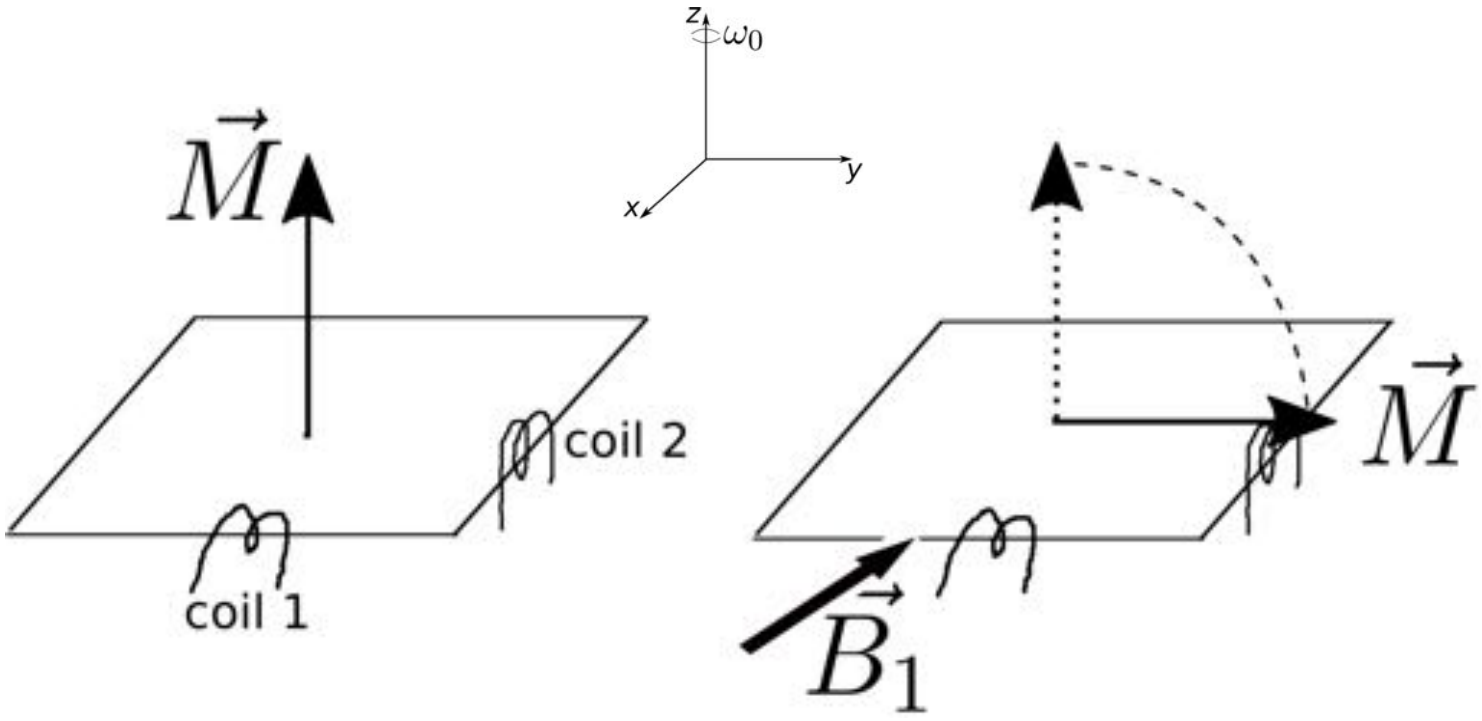




But if B_1 is on, the sum of the magnetizations of the spins will be **too small** to be detected...

How we measure magnetization

We **flip the magnetization** vector into the perpendicular plane, where two quadrature coils (coil 1 and 2) can measure it. This is a **90° flip**



We must use a perpendicular alternate magnetic field $\mathbf{B}_1$ tuned at the **very same frequency** of the **spins precession** (**R**adio **F**requency band) **in resonance** (swing effect)

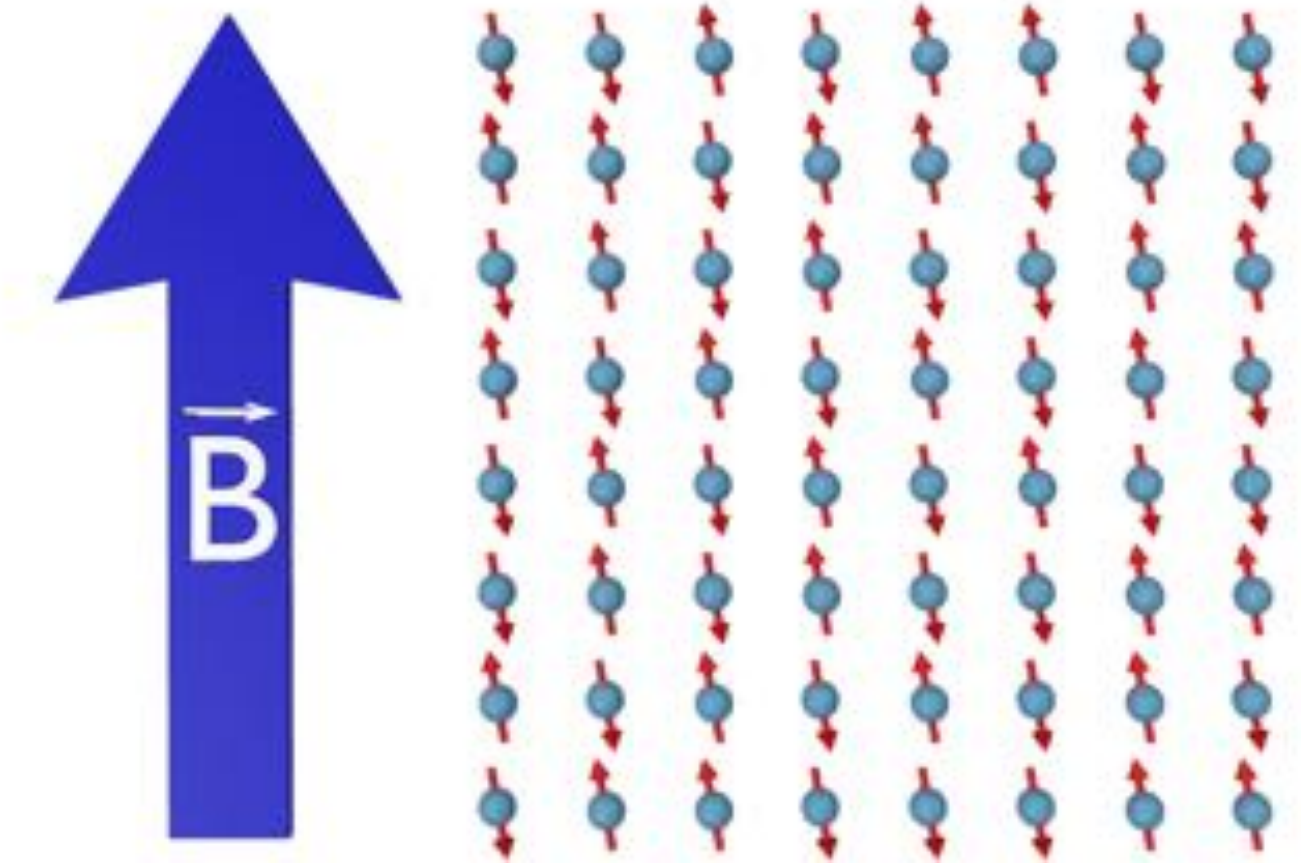
Understanding resonance with a video

Fathoming PHYSICS

Module 3 Pracs and demos

Lesson 2 - Resonant frequency of a swing

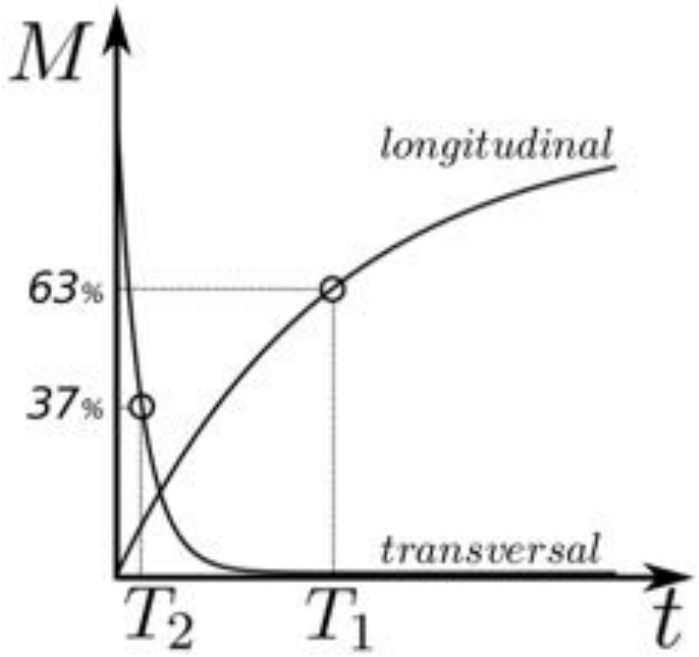
Animation…



Relaxation

When the spins are flipped, the RF pulse is turned off and we use the coils to measure the signal and then infer two different times:

- T_1 amount of time it takes for the system to recover 63% of the longitudinal magnetization (M_z)
- T_2 amount of time it takes for the system to lose 37% of the transversal magnetization (M_{xy})



$$M_z = M_z^{eq}\left[1 - \exp\left(-\frac{t}{T_1}\right)\right]$$

$$M_{xy} = M_z^{eq}\exp\left(-\frac{t}{T_2}\right)$$

T_1: spin-lattice relaxation
is due to the energetic exchange between spins and neighbouring molecules

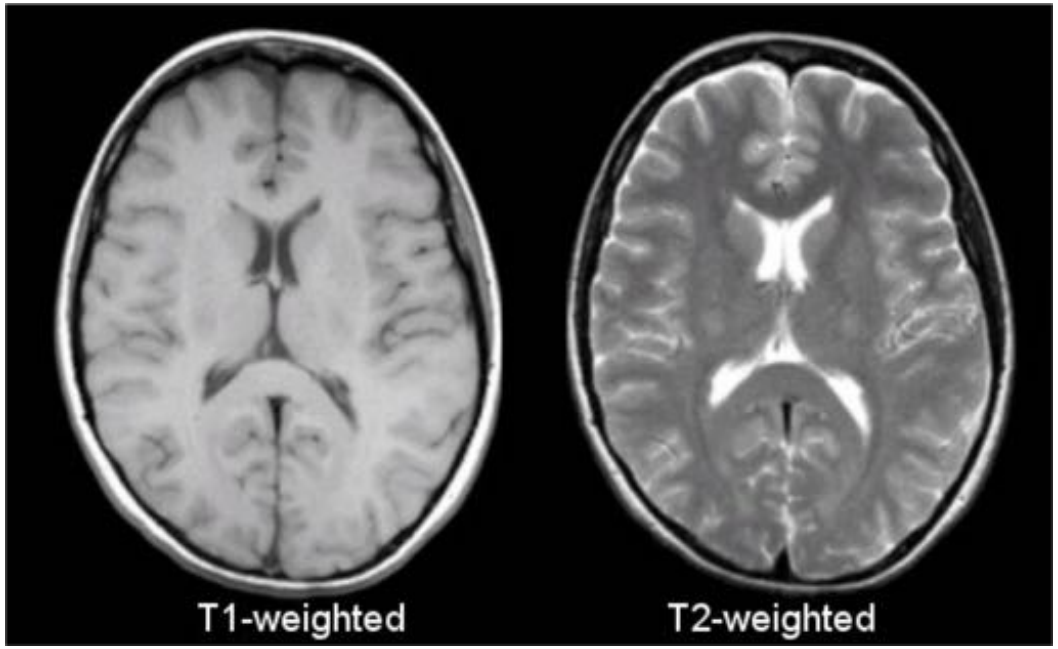
T_2: spin-spin relaxation
is due to the energetic exchange between spins

T1>T2

Tissue	T1 (msec)	T2 (msec)
Water/CSF	4000	2000
Gray matter	900	90
Muscle	900	50
Liver	500	40
Fat	250	70
Tendon	400	5
Proteins	250	0.1- 1.0
Ice	5000	0.001

Approximated values at 1.5T

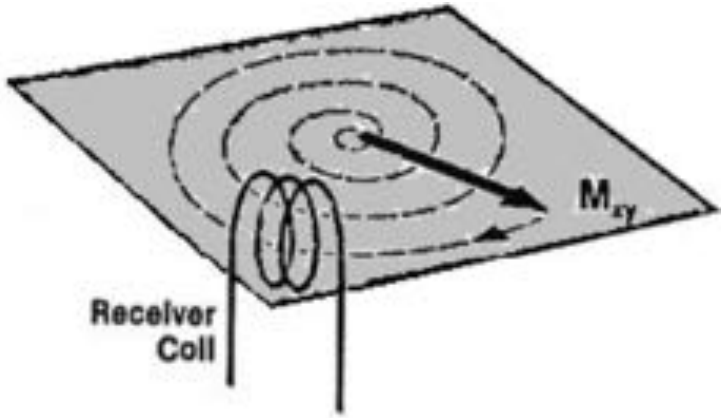
In clinical practice, we acquire images that are "weighted" in T1 or T2, such as these.



Measure the Free Induction Decay (FID)

Different tissues will have different values of T_1 and T_2

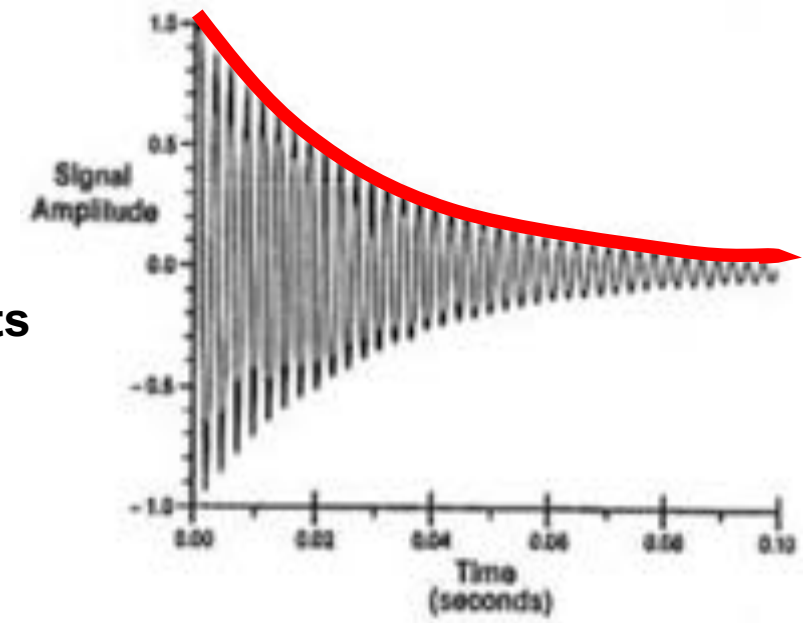
During MRI we actually measure the decrease in the signal amplitude, called the **free induction decay** (FID), of the signal at the echo time T_E. In particular we find that the FID is governed by T2



Magnitude Phase

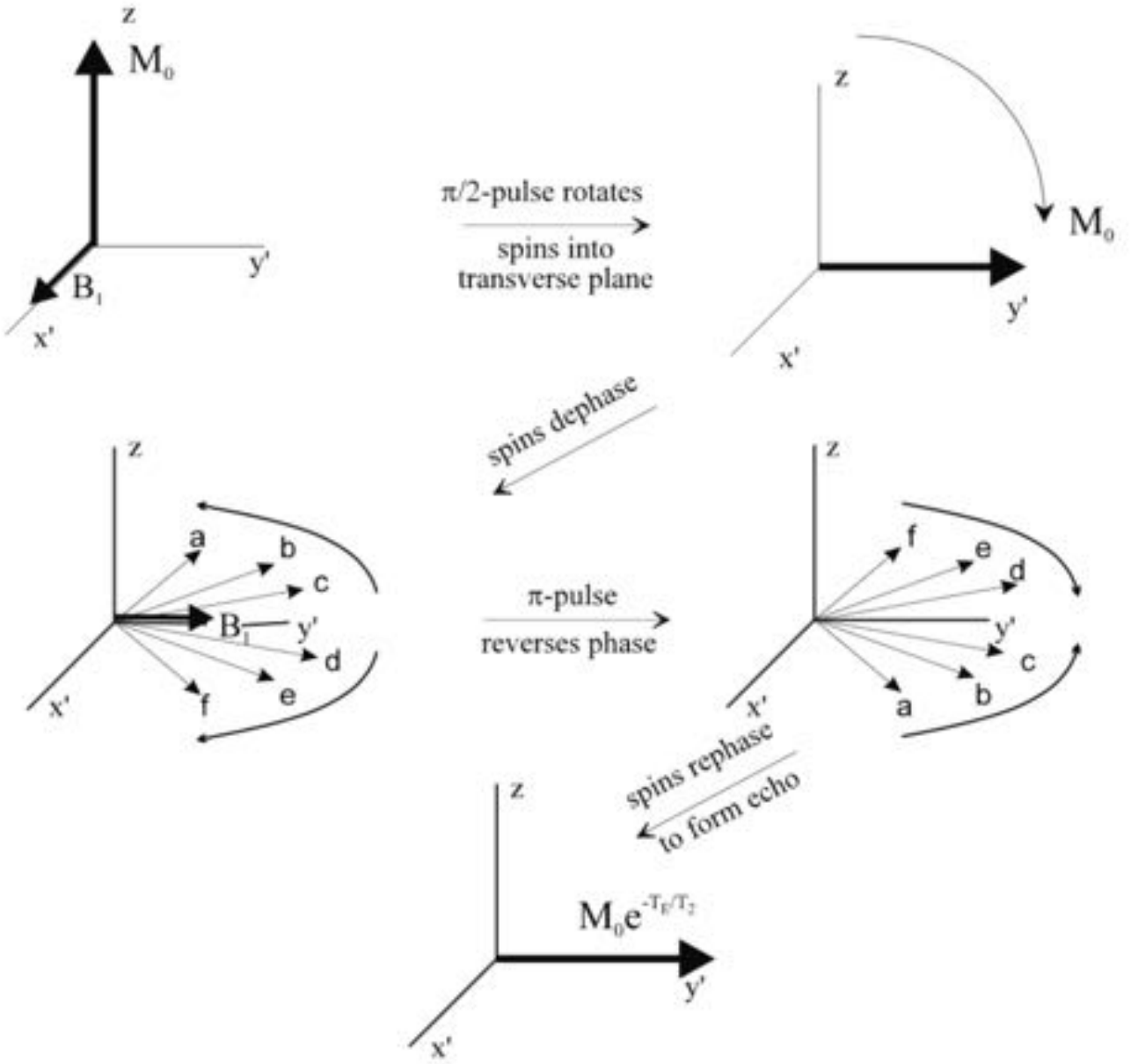
$$M_x(t) = M_{xy}\cos(\Omega_0 t + \varphi)e^{-t/T_2}$$

$$M_y(t) = M_{xy}\sin(\Omega_0 t + \varphi)e^{-t/T_2}$$



In practice…

In general, signals would suffer additional suppression due to dephasing from external field inhomogeneities (T_2 would be replaced by a smaller relaxation time $T_2^* < T_2$).
A re-phasing or echoing of this source of dispersion can be achieved by an additional RF pulse application, where the basic idea is to flip all the spins 180° in the transverse plane. Following the 180° flip the dephasing is reversed, and the phases refocus at what is called **echo time T_E**. The value of T_E can be set by varying the time interval between the initial 90° pulse and the 180° pulse.
A longer wait time will result in a larger T_E, while a short interval between the two pulses will give a small T_E.



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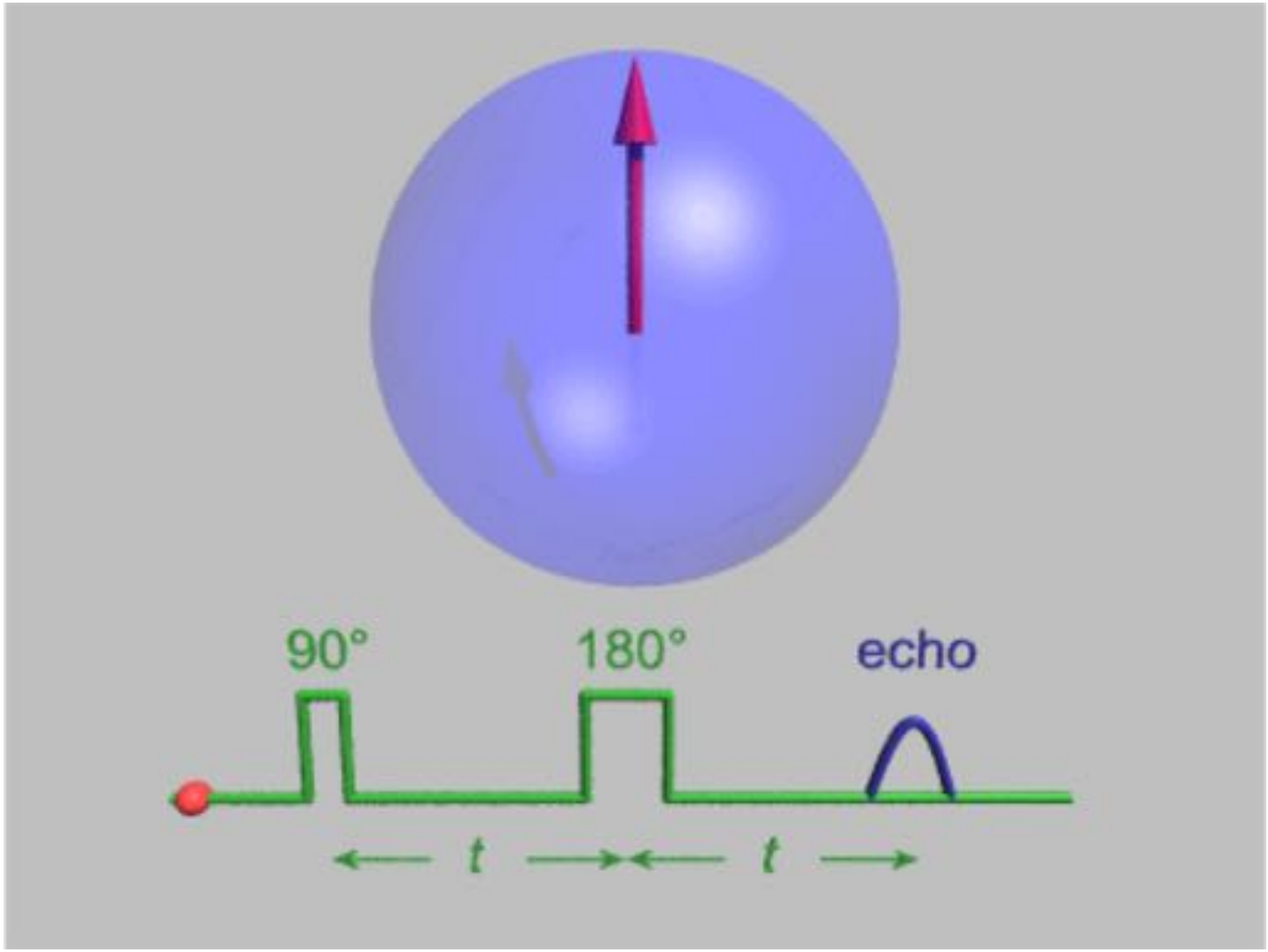
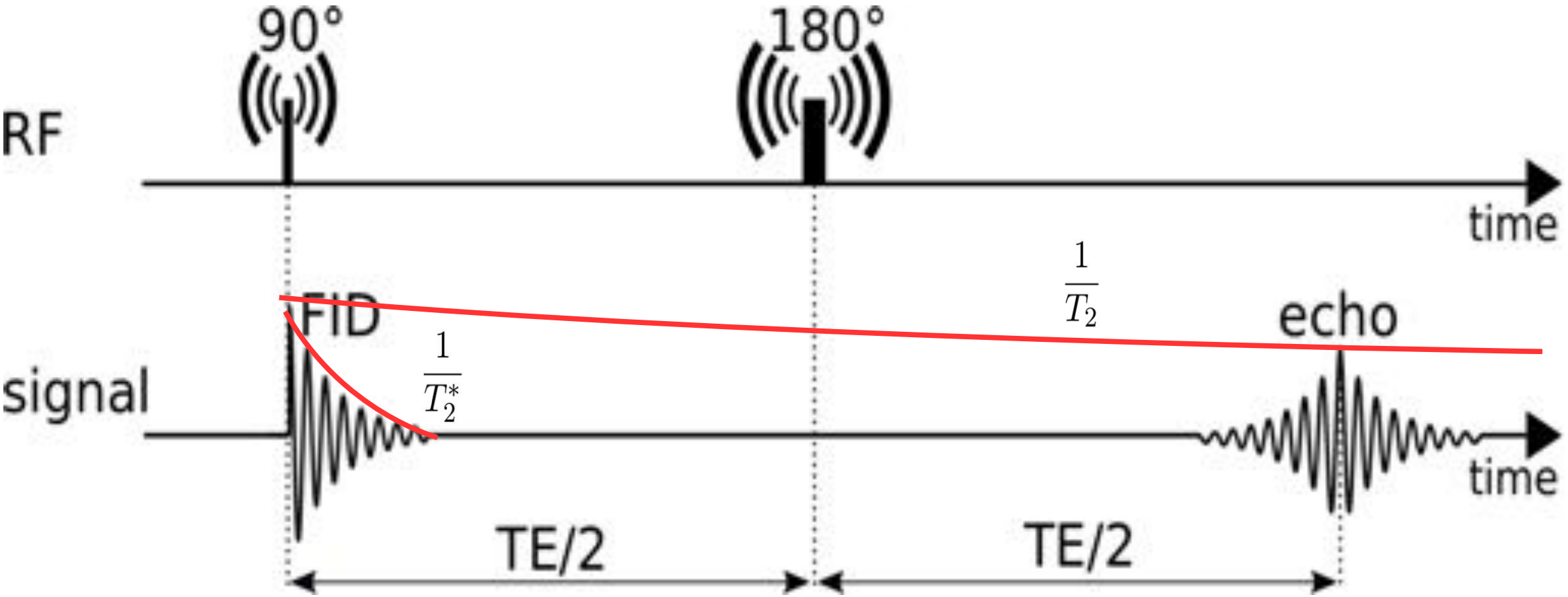
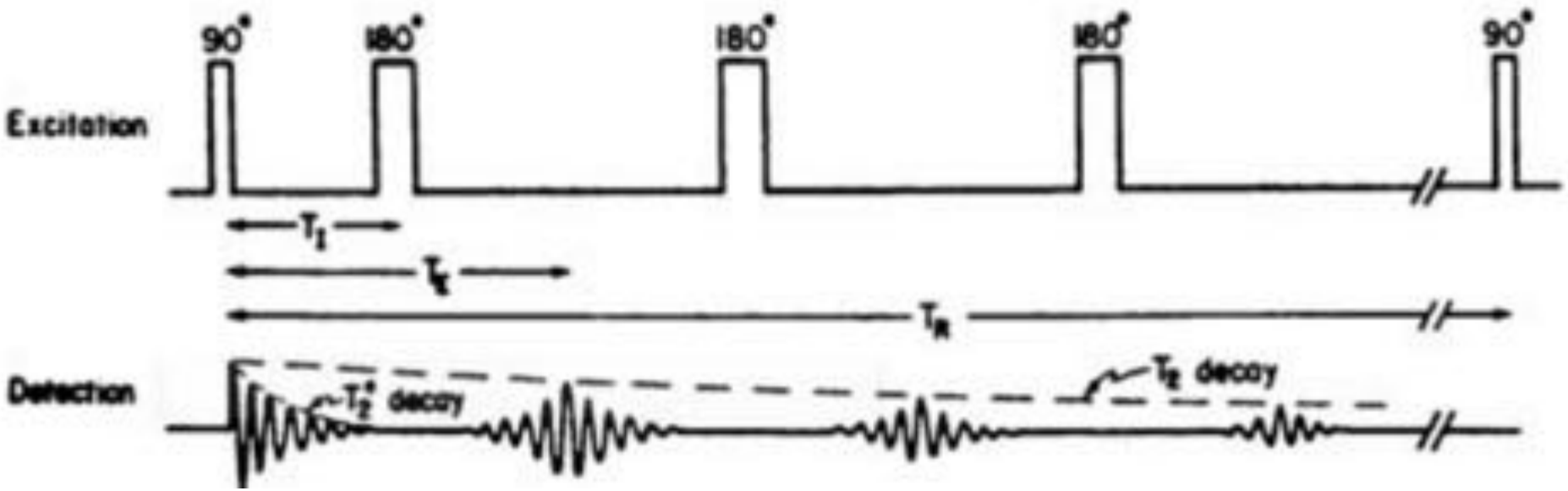


In a schematic way



In practice



$$\underbrace{(90^o - (T_E/2 - 180^o - T_E/2)_m - T')_n}_{T_R}$$

Spin echo signal

protons density

$$S \propto \rho\left(1-2e^{-\frac{(T_R-T_E/2)}{T_1}}+e^{-\frac{T_R}{T_1}}\right)e^{-\frac{T_E}{T_2}}$$

If $T_E/2 << T_R$ (true for clinical applications):

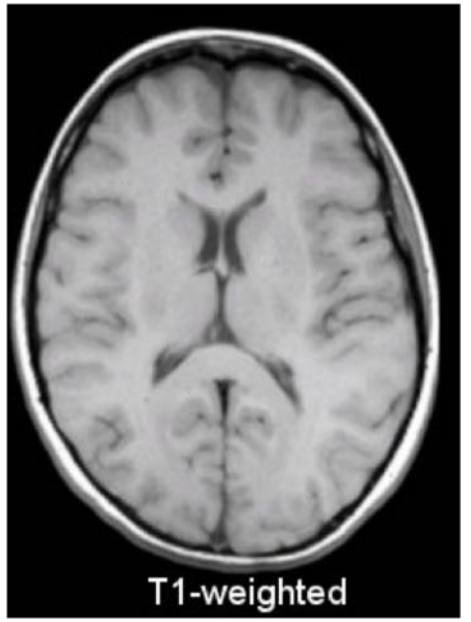
$$S \propto \rho\left(1-e^{-\frac{T_R}{T_1}}\right)e^{-\frac{T_E}{T_2}}$$

T1 or T2 weighted

If $T_E << T_2$

$$S \propto \rho\left(1-e^{-\frac{T_R}{T_1}}\right)e^{-\frac{T_E}{T_2}} \approx \rho\left(1-e^{-\frac{T_R}{T_1}}\right)$$

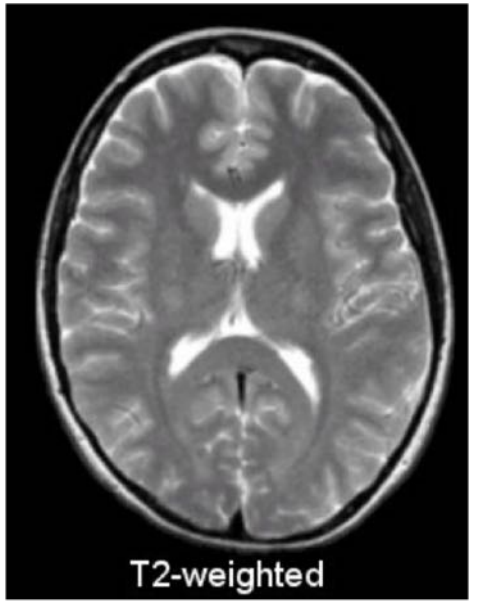
T1-weighted
(no T_2 dependence)



If $T_R >> T_1$

$$S \propto \rho\left(1-e^{-\frac{T_R}{T_1}}\right)e^{-\frac{T_E}{T_2}} \approx \rho e^{-\frac{T_E}{T_2}}$$

T2-weighted
(no T_1 dependence)

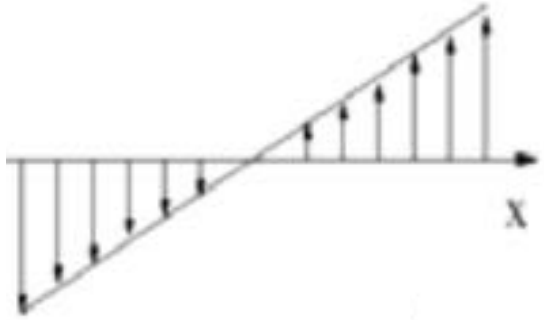


From NMR to MRI …

Gradient magnetic fields

To disentangle signals coming from different spatial locations **we use gradients**.

$$\boldsymbol{B}(x) = \boldsymbol{B_0}(\boldsymbol{x}) + \boldsymbol{G_x} \cdot \boldsymbol{x}$$



$$\omega = \gamma \boldsymbol{B} = \gamma \boldsymbol{B_0} + \gamma \boldsymbol{G_x} \cdot \boldsymbol{x}$$

$$\boldsymbol{x} = \frac{\left(\frac{\omega}{\gamma} - \boldsymbol{B_0}\right)}{\boldsymbol{G_x}} = \boldsymbol{g}(\omega)$$

This equation associates the position $\boldsymbol{x}$ to the angular speed ω.

If we use a gradient and tune the coil to receive at the frequency $f = \frac{\omega}{2\pi}$ we can "listen" to the signal coming from **only** the spatial locations having position $\boldsymbol{x}$

Fourier Method

The most used acquisition method is composed of three steps:

1. **Preparation:** selection of a plane along z direction
2. **Evolution:** codification in phase in y coordinate
3. **Readout:** codification in frequency in x coordinate

1. Preparation

We apply a gradient $\boldsymbol{G_z}$ along the **z-axis** to select an axial plane (**slice**)

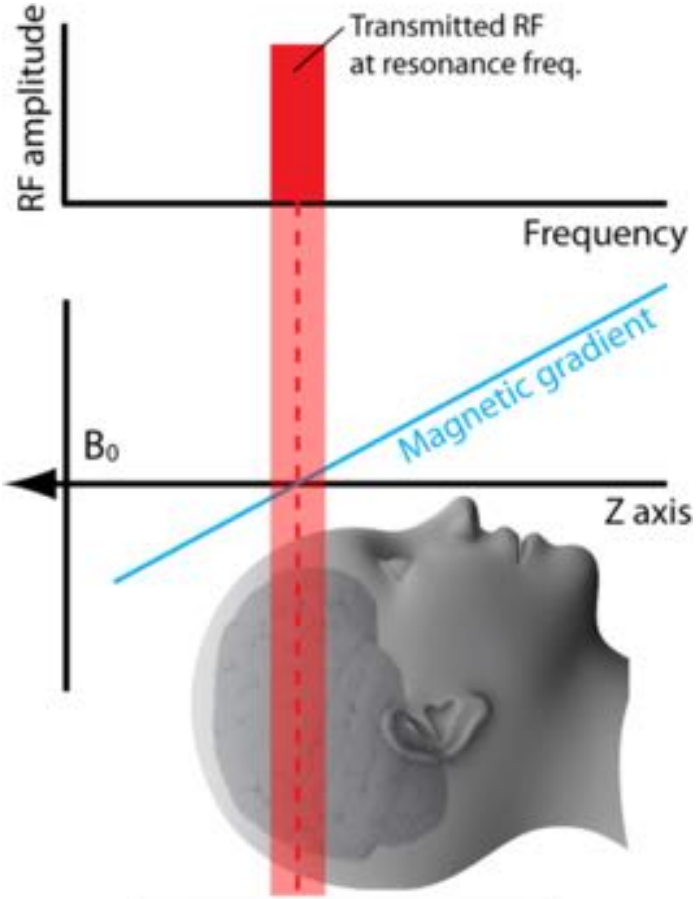
$$\omega(z) = \gamma(B_0 + G_z z) = \omega_0 + \gamma G_z z$$

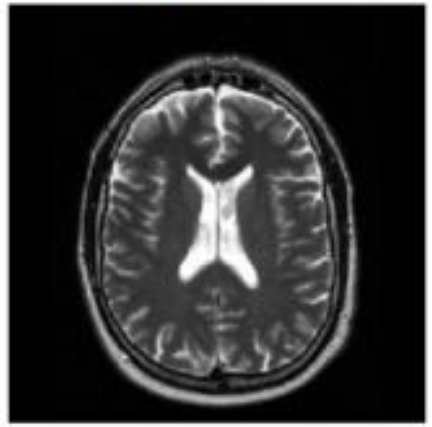
At the same time, we apply a 90° RF pulse with frequency $f = \omega^*/2\pi$ to select only the spins with $z = z_p$ (which coincides with the isocentre)

$$\omega^* = \omega(z_p) = \omega_0 + \gamma G_z z_p$$

In practice we are not able to select only a specific plane z_p , but we achieve a **small volume** identified by Δz (we talk about **voxel** not pixel!)

$$\Delta\omega = \gamma G_z \Delta z \qquad \Delta z = \frac{\Delta\omega}{\gamma G_z}$$



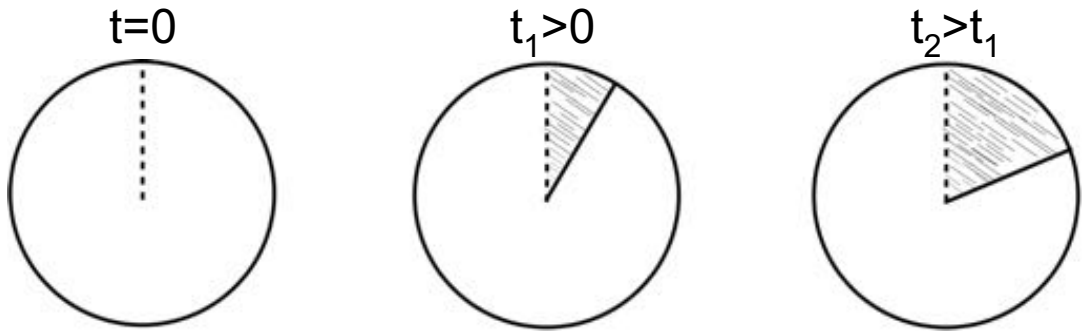


2. Evolution

We apply a gradient $\boldsymbol{G_y}$ along the **y-axis** for a time t_y .

$$\omega(y) = \omega_0 + \gamma \boldsymbol{G_y y}$$

Phase accumulation



After a time t_y , the previously selected (z_p) spins that are located at position *y* will accumulate a phase

$$\varphi_y = \left(\omega_0 + \gamma \boldsymbol{G_y y}\right) t_y$$

At the moment, we have selected the intersection between a plane perpendicular to z and a plane perpendicular to y. We need to do the same for x.

3. Readout

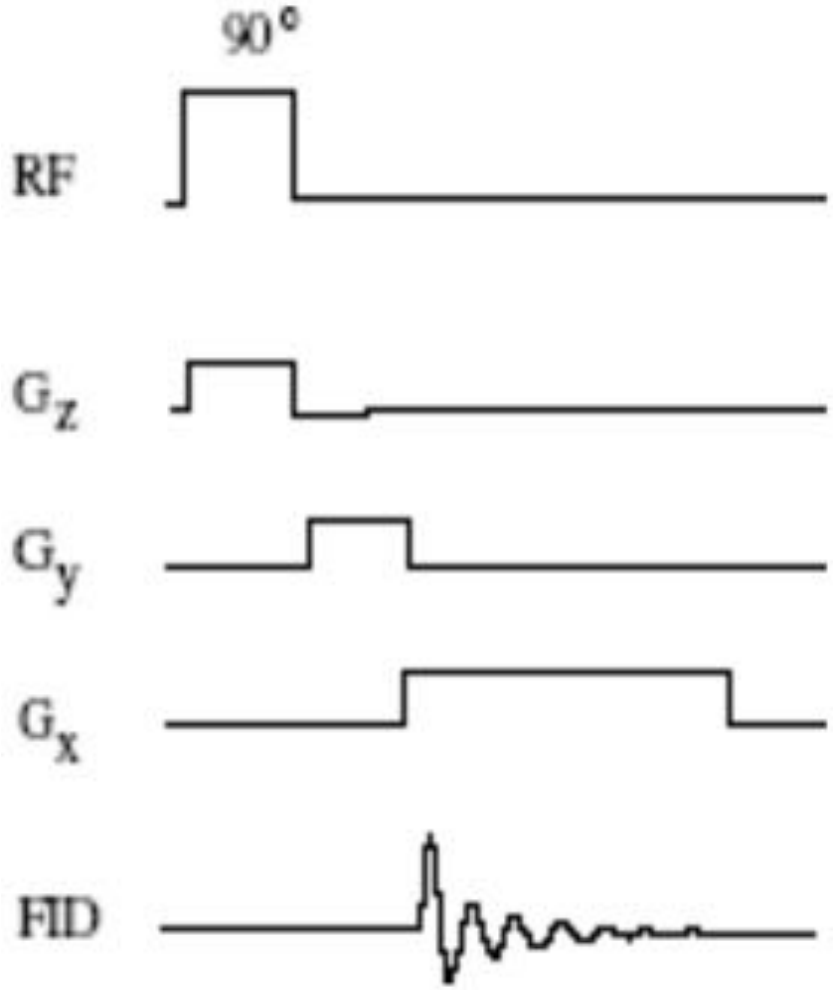
We apply a gradient $\boldsymbol{G_x}$ along the **x-axis**.

$$\omega(x) = \omega_0 + \gamma \boldsymbol{G_x x}$$

Spins with the same *x* have a frequency $f = \omega(x)/2\pi$.

Each voxel in the plane with $z = z_p$ is

uniquely identified by the pair (ω_x, φ_y)



K-space

Each spin inside the magnetic field precesses with frequency ω and phase φ

$$\left.\begin{cases} s_x = s_\perp \cos(\omega t + \varphi) \\ s_y = s_\perp \sin(\omega t + \varphi) \\ s_z = s_\parallel \end{cases}\right\} \; s(t) = s_\perp \exp[i(\omega t + \varphi)]$$

Here $s_\perp$ **is a function that depends on the relaxation times T1 and T2 and on the quantity of spins** present in the volume (protons density).

Substituting the expression of ω and φ given by the applied gradients we have

$$s(t) = s_\perp \exp[i(\omega_0 t + \gamma G_x x t + \gamma G_y y t)]$$

Measured Signal

The measured signal comes from the whole specimen, that is from all of the positions. At the time we sample the FID, typically $t = t_x$ ($= n * T_E$ for spin-echo), the signal is the integral over all of the x and y positions:

$$s(t) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} s_\perp \exp\left[i(\cancel{\omega_0 t} + \gamma G_x x t_x + \gamma G_y y t_y)\right] \mathrm{d}x \mathrm{d}y$$

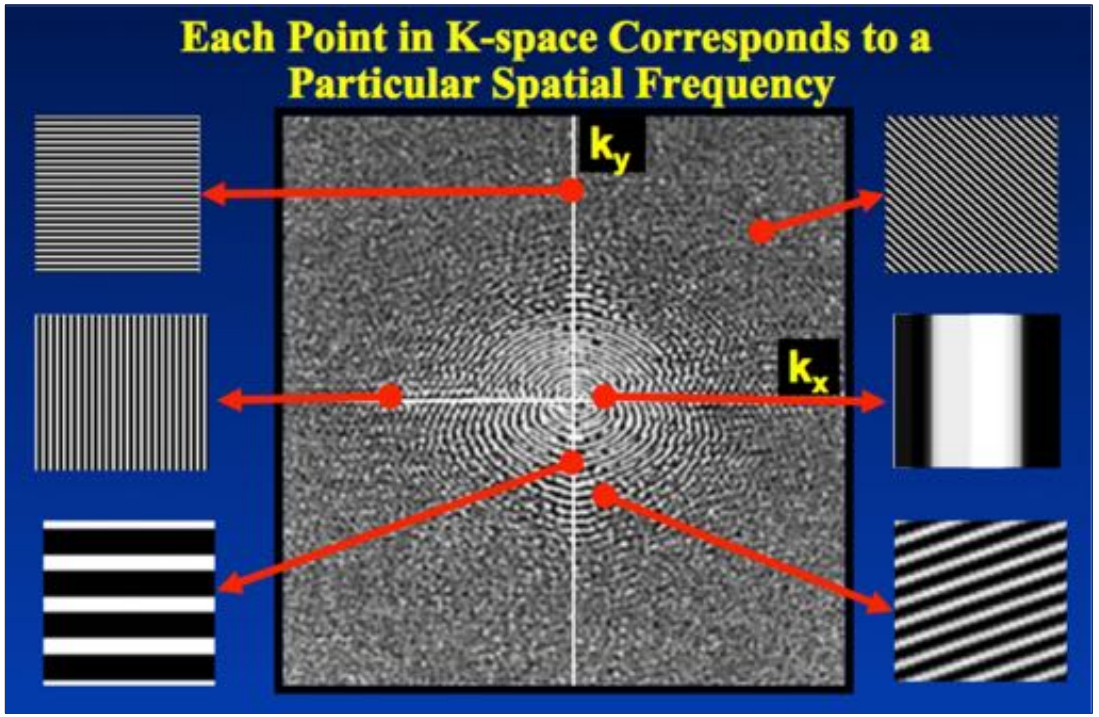
constant so does not depend on (x, y)

Thus defining $k_x = \dfrac{\gamma G_x t_x}{2\pi}$ $\quad$ $k_y = \dfrac{\gamma G_y t_y}{2\pi}$

we obtain $$s(k_x, k_y) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} I(x, y) \exp\left[2\pi i(k_x x + k_y y)\right] \mathrm{d}x \mathrm{d}y$$

By acquiring the signal for a square grid of k_x and k_y we obtain the k-space signal.
The image $I(x, y) = s_\perp$ can be obtained by inverting the equation (**Fourier Transform**).

Interpretation



The measured image is represented in frequency.

Each measured (k_x, k_y) point corresponds to a specific type of pattern, defined by an orientation and a frequency.

We measure how much of these basic blocks contributions the image is made of.

A dark pixel here means that in the image there is not such a basic block. Never!!

On the other hand, the brighter the pixel (in k-space) the more the real image $I(x, y)$ contains the corresponding basic block..

Notably, the brightest pixel is usually the one in the middle (k_x=0,k_y=0), corresponding to a constant (zero frequency, no directionality).

Understanding the K-space

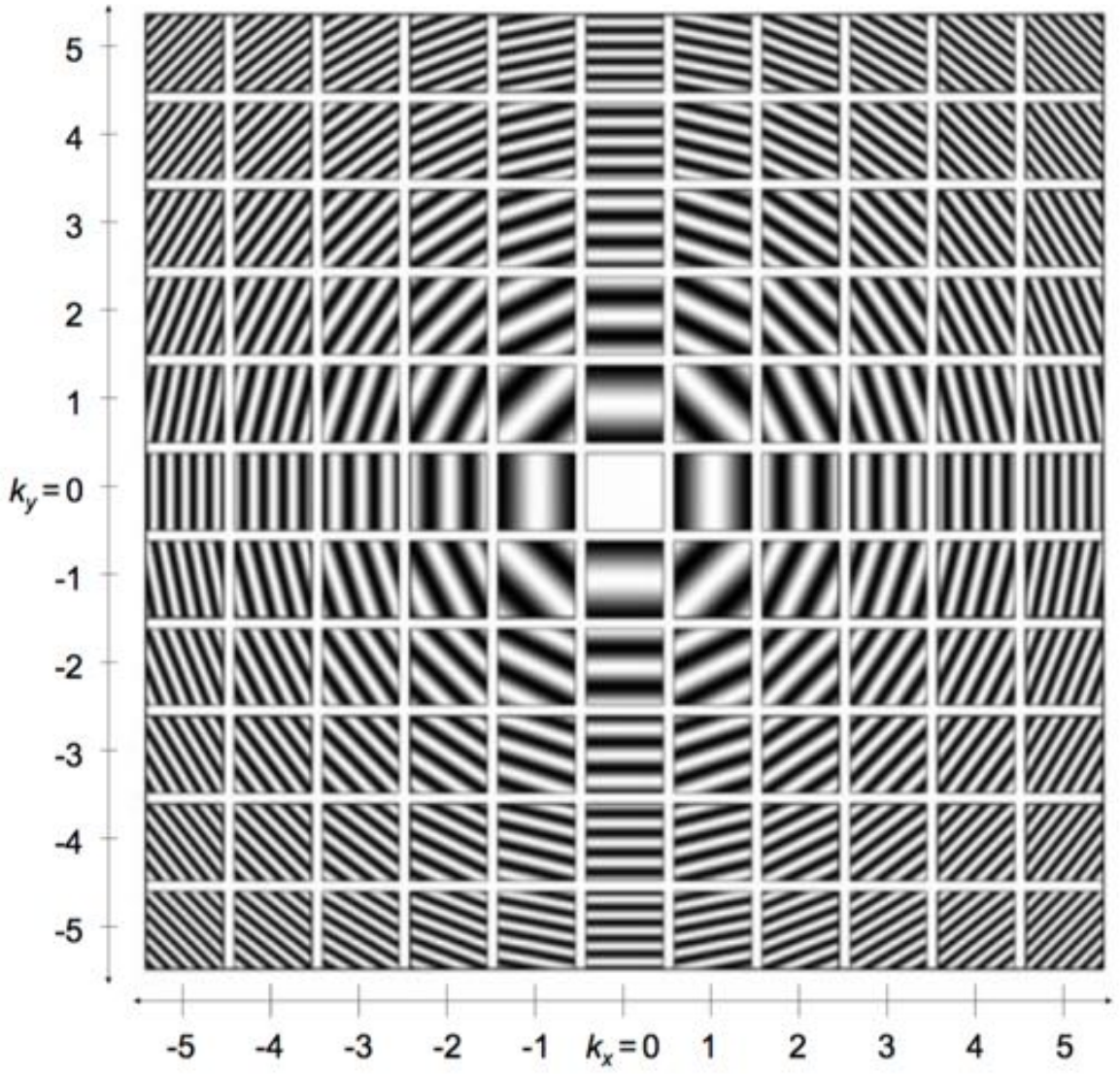


Fig. 1.10. Illustration of spatial frequencies: Each *square* illustrates a single point in (two-dimensional) k-space. These points correspond to those spatial oscillations in image space that are shown in the *squares*. Low spatial frequencies are found in the center of k-space; image intensity varies slowly in image space. High spatial frequencies correspond to peripheral points in k-space.

From k-space to images

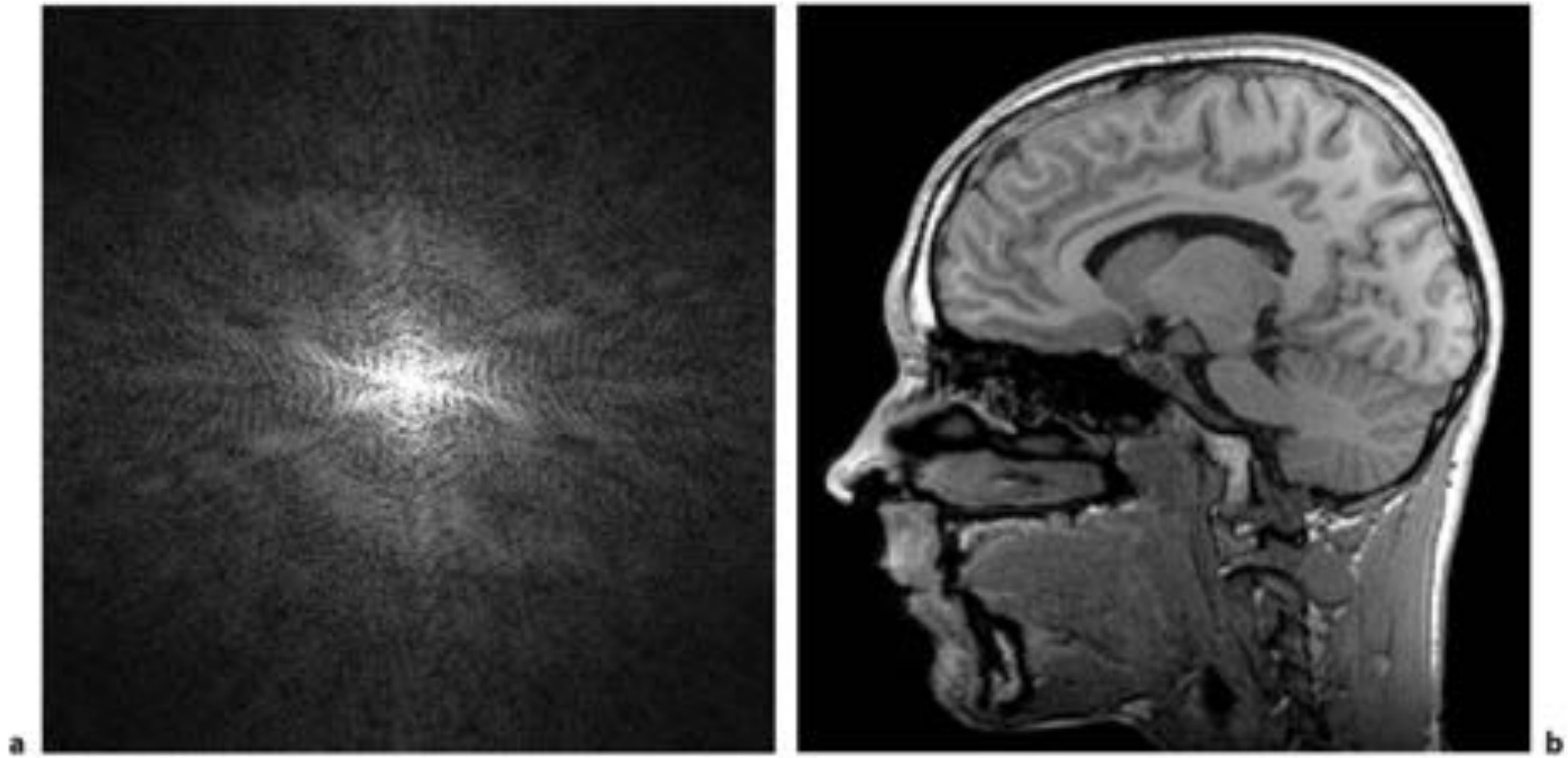


Fig. 1.11a,b. **a** Raw data in k-space (magnitude presentation of complex data) and **b** corresponding reconstructed MR image. Highest intensities in k-space are found in the center, corresponding to slowly varying image intensities in b.

Basic contrast → low frequencies

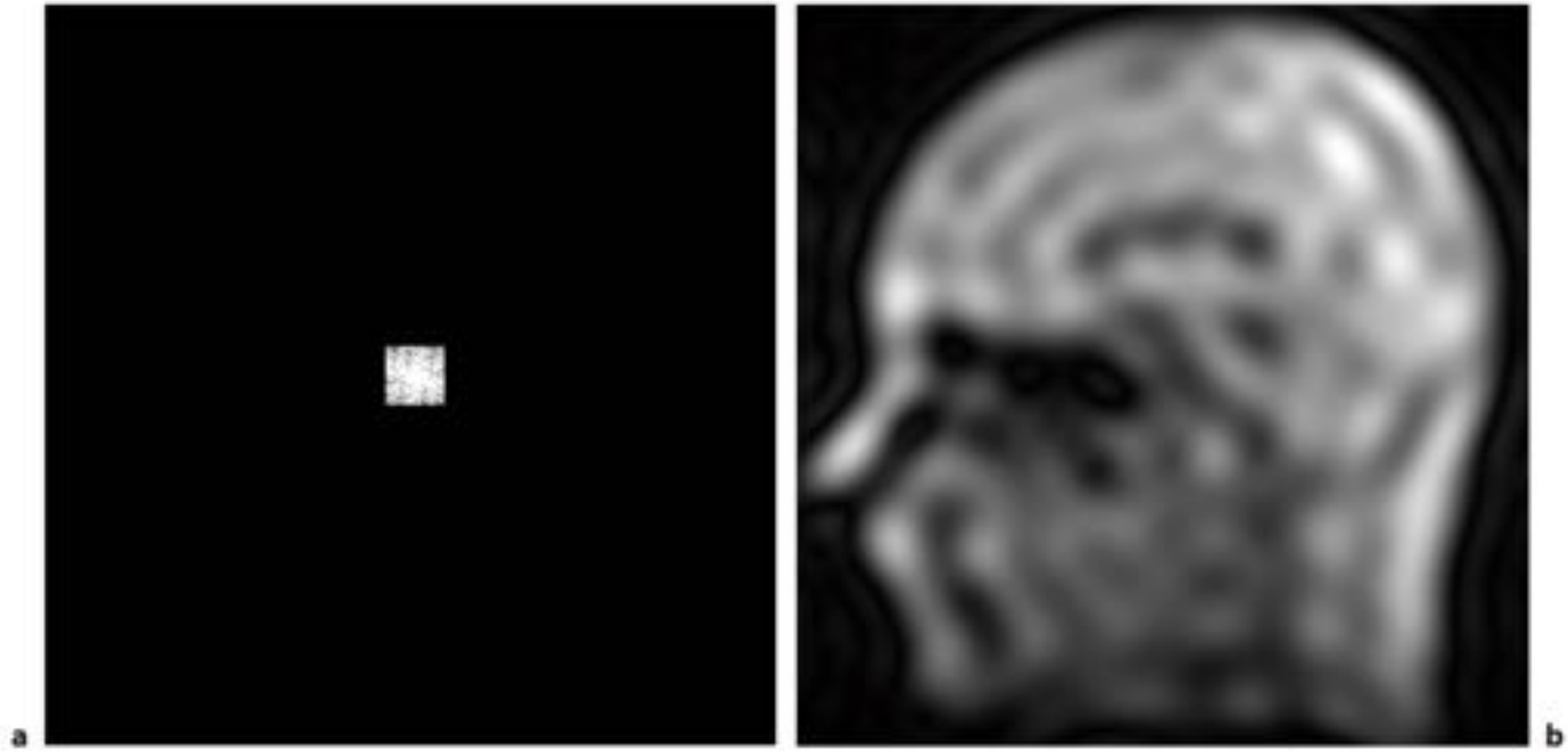


Fig. 1.12a,b. **a** Raw data in the center of k-space and **b** corresponding reconstructed MR image. The image appears filtered; only large-scale changes of intensity corresponding to low spatial frequencies remain.

Details → high frequencies

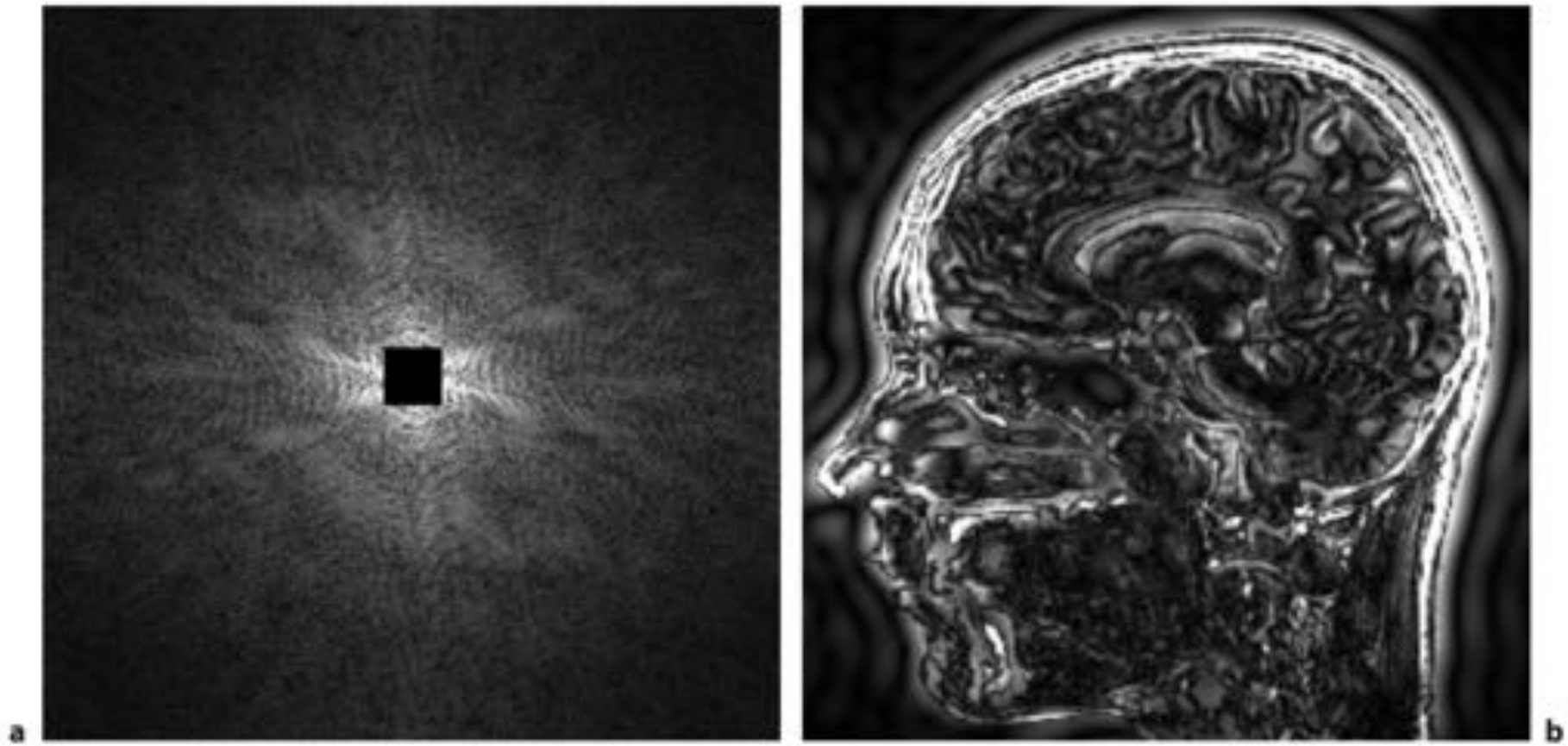
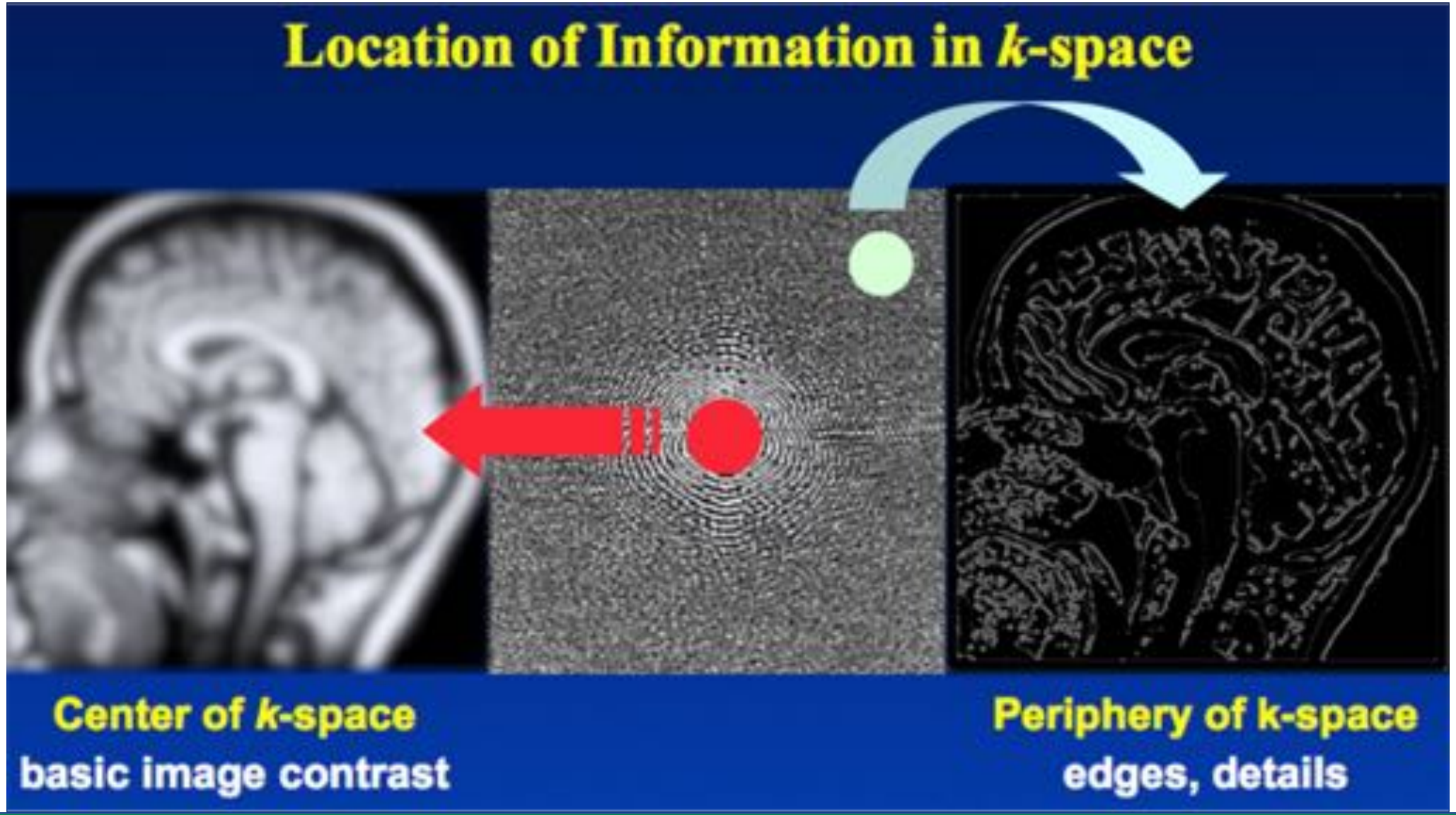


Fig. 1.13a,b. **a** Raw data in the periphery of k-space and **b** corresponding reconstructed MR image. The image contains information complementary to Fig. 1.12a. Only changes of intensity in small spatial scales corresponding to high spatial frequencies remain.

Interpretation



Artefacts due to wrong k-space sampling

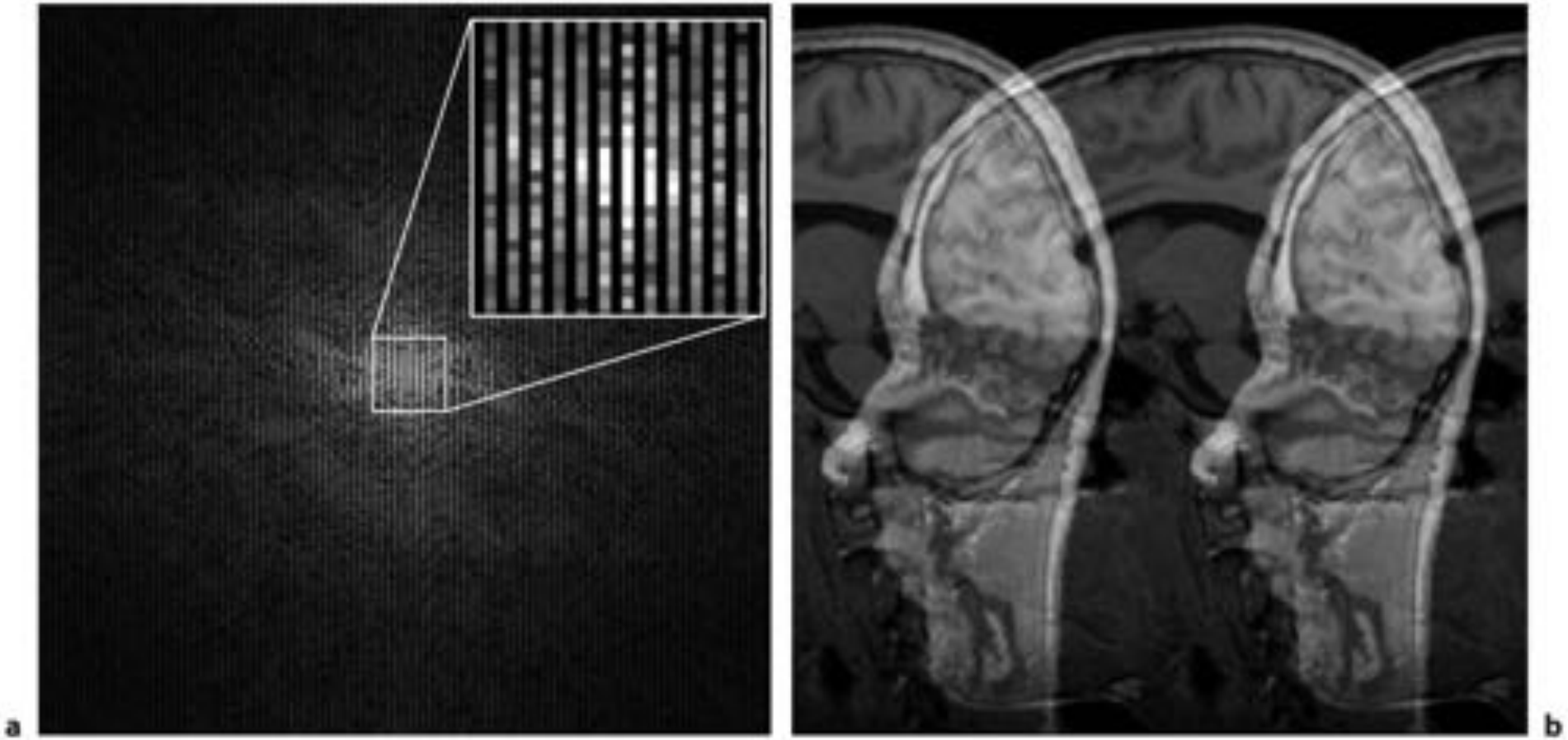


Fig. 1.14a,b. a Raw data after removing every other line in k-space and b corresponding reconstructed MR image. The effective field of view of the image is reduced to 50%, resulting in aliasing artefacts in anterior-posterior direction.

Artefacts due to wrong k-space sampling

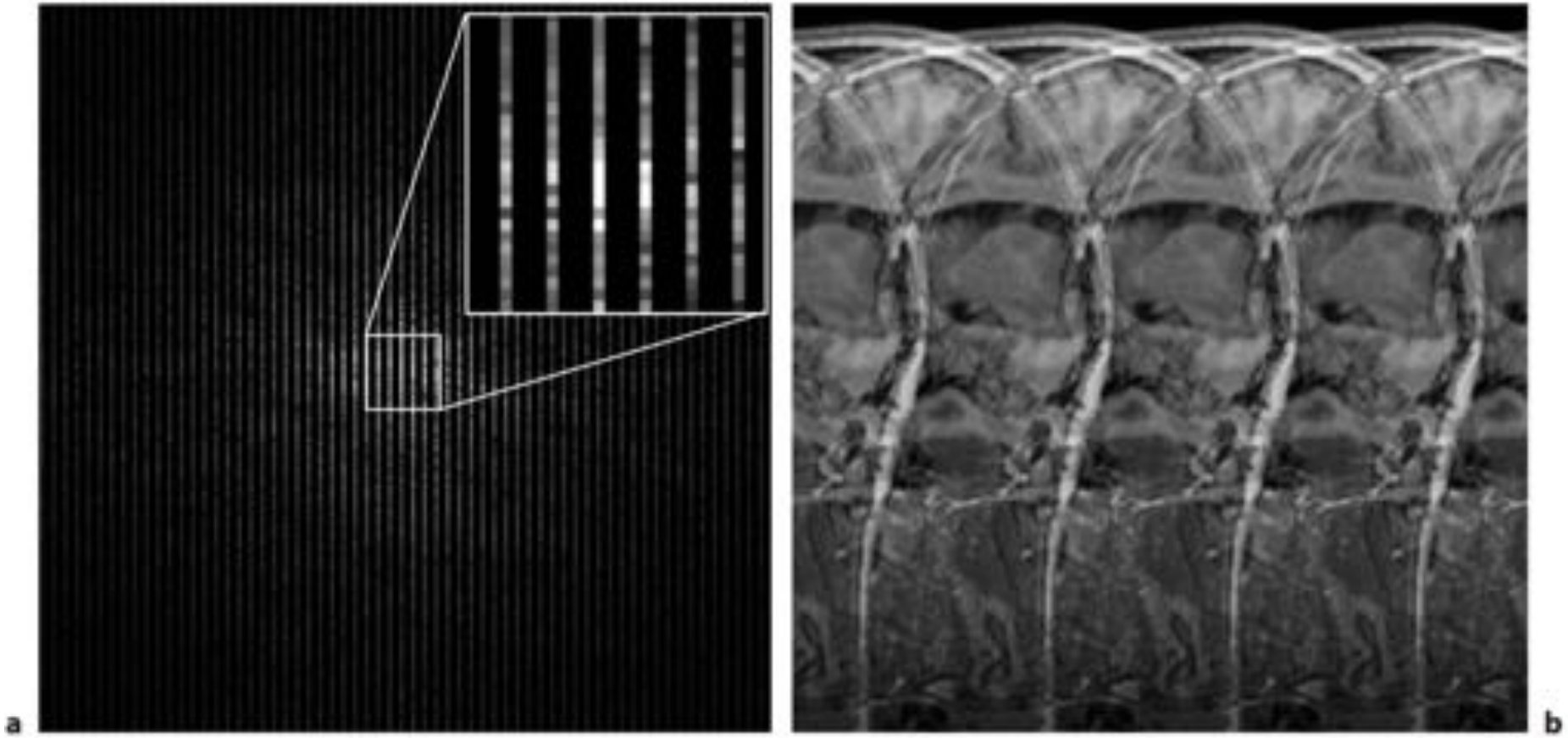


Fig. 1.15a,b. **a** Raw data after removing three of four neighboring lines and **b** corresponding reconstructed MR image. The effective field of view of the image is reduced to 25%, resulting in severe aliasing artefacts in anterior-posterior direction.

Artefacts due to modifications of k-space

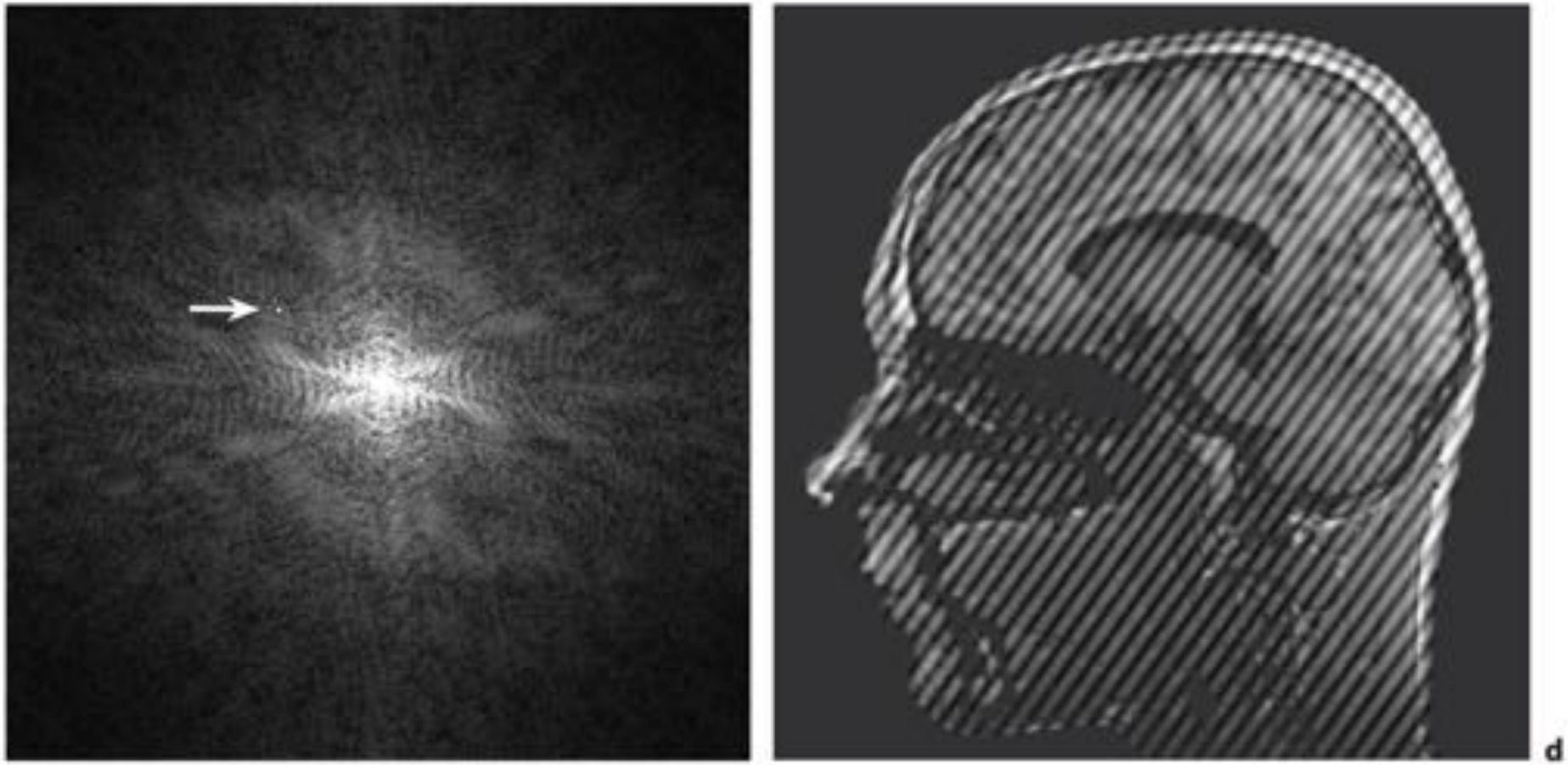


Fig. 1.16a,b. **a** Raw data after adding a single point with artificially increased intensity (*arrow*) and **b** corresponding reconstructed MR image. The image is affected from artefacts extending over the complete field of view.

Some references

- https://www.cis.rit.edu/htbooks/mri/index.html
- http://mriquestions.com
- Michael L. Lipton. *Totally accessible MRI: a user's guide to principles, technology, and applications*. Springer Science & Business Media, 2010
- Robert W. Brown, Y-C Norman Cheng, E. Mark Haacke, Michael R. Thompson, Ramesh Venkatesan. *Magnetic resonance imaging: physical principles and sequence design*. John Wiley & Sons, 2014
- http://eknygos.lsmuni.lt/springer/339/3-17.pdf (images of k-space)

Questions?