BME I5000: Biomedical Imaging

Lecture 9
Magnetic Resonance Imaging (imaging)

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Schedule

1. Introduction, Spatial Resolution, Intensity Resolution, Noise
2. X-Ray Imaging, Mammography, Angiography, Fluoroscopy
3. Intensity manipulations: Contrast Enhancement, Histogram Equalisation
4. Computed Tomography
5. Image Reconstruction, Radon & Fourier Transform, Filtered Back Projection
6. Nuclear Imaging, PET and SPECT
7. Maximum Likelihood Reconstruction
8. Magnetic Resonance Imaging
9. Fourier reconstruction, k-space, frequency and phase encoding
10. Optical imaging, Fluorescence, Microscopy, Confocal Imaging
11. Enhancement: Point Spread Function, Filtering, Sharpening, Wiener filter
12. Segmentation: Thresholding, Matched filter, Morphological operations
13. Pattern Recognition: Feature extraction, PCA, Wavelets
14. Pattern Recognition: Bayesian Inference, Linear classification

MRI – How to generate images using NMR

Nuclear spins resonate at a frequency proportional to the external magnetic field

$$\omega = \gamma B_0$$

Basic idea of MRI: Change the B_0 field with space and the resonance frequency will change with space.

$$\omega(x) = \gamma B_0(x)$$

The detected resonance signal (FID) contains multiple frequency components each giving information about a different portion of space!

MRI – How to generate images using NMR

MRI – Signal detected in MRI

Recall that the signal due to the bulk magnetization precessing at ω detected in the x and y coils can be written as:

$$s(t) = s_x(t) + i\, s_y(t) \propto M_{xy}(0)\, e^{-t/T_2^*}\, e^{-i\omega t}$$

Signal intensity scales with $M_{xy}(0)$ - the magnitude of the transverse magnetization at the end of the RF pulse. $M_{xy}(0)$ is proportional to the number of resonating spins in the material, or the proton density $\rho(\boldsymbol{r})$. It is dependent on the tissue and therefore dependent on space $\boldsymbol{r}$.

MRI generates images of $\rho(\boldsymbol{r})$!

$M_{xy}(0)$ also depends on the specifics of the pulse sequence. By manipulating the pulse sequence MRI can generate images of $\rho(\boldsymbol{r})$ that are modulated by physical properties that affect T_1 or T_2.

MRI – Signal detected in MRI

The main idea is to apply a B_0 field with a magnitude that also depends on space, so that the frequency of the resonance signal relates to space, $\omega(\boldsymbol{r}) = \gamma B_0(\boldsymbol{r})$:

$$s(t) \propto e^{-t/T_2^*} \rho(\boldsymbol{r}) e^{-i\gamma B_0(\boldsymbol{r})t}$$

(where we have ignored the effect of T_1 and T_2). The signal emitted by the entire body is then the sum over space:

$$s(t) \propto e^{-t/T_2^*} \int_{body} d\boldsymbol{r}\, \rho(\boldsymbol{r}) e^{-i\gamma B_0(\boldsymbol{r})t}$$

Note that $B_0(\boldsymbol{r})$ is parallel to the z-axis, only its magnitude may now depend on the location in space $\boldsymbol{r}$.

MRI – Signal detected in MRI

For reconstruction it will be useful to define new signal that is 'demodulated' and without the T_2^* decay:

$$S(t) = s(t) e^{t/T_2^*} e^{i\omega_0 t}$$

Define also $\Delta B_z(\boldsymbol{r})$ as the difference of $B_0(\boldsymbol{r})$ over main B_0:

$$\Delta B_z(\boldsymbol{r}) = B_0(\boldsymbol{r}) - \omega_0/\gamma$$

With this the MRI *imaging equations* becomes

$$S(t) = \int_{body} d\boldsymbol{r}\, \rho(\boldsymbol{r}) e^{-i\gamma \Delta B(\boldsymbol{r}) t}$$

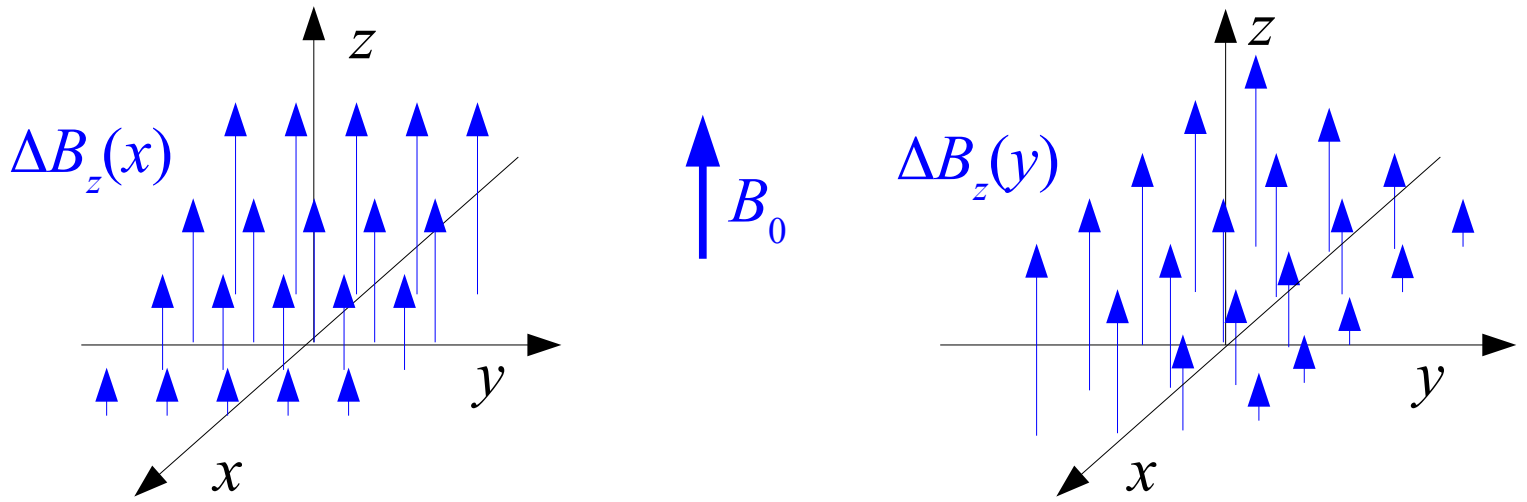
MRI – B_0 gradient, frequency encoding

Lets assume we need spatial resolution in only one direction. For instance x. So we want to recover (ignoring z direction for now):

$$g(x)=\int dy\,\rho(x,y)$$

To do so, we apply a contribution B_0 that changes linearly with x. The strengths of these 'x-gradient' is given by the constants G_x.

$$\Delta B_z(\boldsymbol{r})=G_x x$$



MRI – B_0 gradient, frequency encoding

The imaging equation is now

$$S(t)=\int dx\, g(x) e^{-i\gamma G_x x t}$$

To put this in a more familiar notation lets define a new variable

$$k_x=\gamma G_x t \qquad \gamma=\gamma/2\pi$$

$$S(k_x)=\int dx\, g(x) e^{-i2\pi k_x x}$$

Evidently the detected signal $S(k)$ is a Fourier transform of $g(x)$, and we can recover it with the inverse Fourier transform.

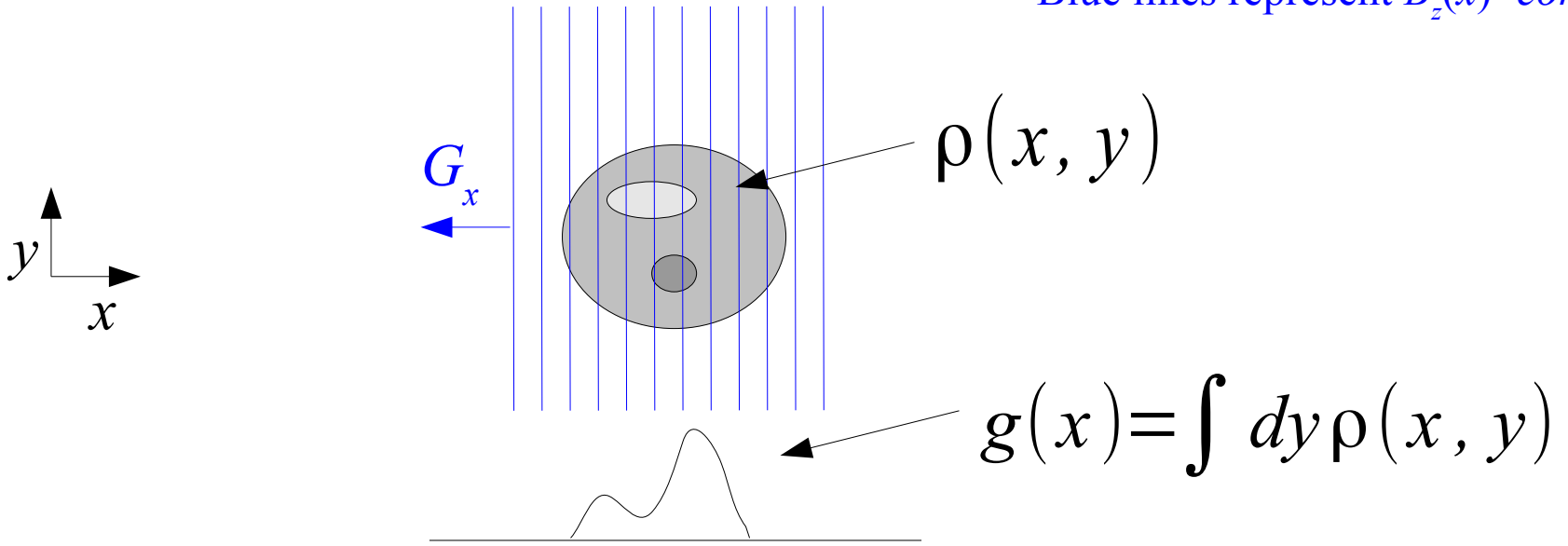
$$g(x)=\int dx\, S(k_x) e^{i2\pi k_x x}$$

This methods is therefore called ***frequency encoding***. Obviously we can also apply a G_y gradient and obtain $g(y)$.

MRI – Axial Reconstruction

Notice that $g(x)$ is the sum of $\rho(x,y)$ along direction y, i.e. the direction orthogonal to the gradient G_x.

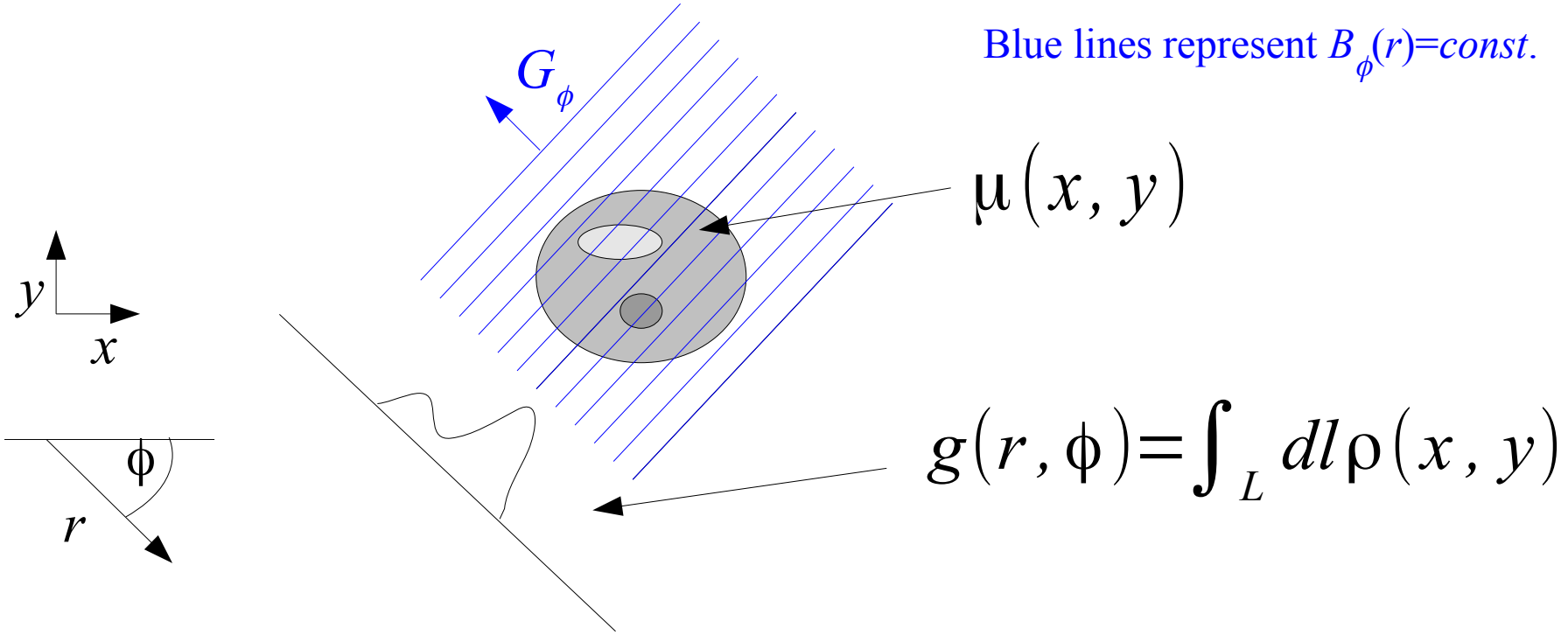
Blue lines represent $B_z(x)=const.$



The signal $S(k_x)$ detected in MRI during a G_x gradient is the Fourier transform of $g(x)$.

MRI – Axial Reconstruction

Obviously one can make the gradient G_ϕ have any orientation ϕ , and measure the corresponding signal $S(k,\phi)$



The signal $S(k,\phi)$ detected in MRI during a G_ϕ gradient is the Fourier transform of the Radon transform $g(r,\phi)$. If we record the signal repeatedly at different orientation ϕ we can therefore apply the same Radon reconstruction as in CAT!

MRI – Axial Reconstruction

By combining x,y gradients linearly we can get gradients that at an arbitrary orientation ϕ:

$$\Delta B_z(\boldsymbol{r}) = G_x x + G_y y = \boldsymbol{G}_\phi \cdot \boldsymbol{r}$$

$$\boldsymbol{G}_\phi = \begin{bmatrix} G_x \\ G_y \end{bmatrix} = G_\phi \begin{bmatrix} \cos\phi \\ \sin\phi \end{bmatrix} \qquad \boldsymbol{r} = \begin{bmatrix} x \\ y \end{bmatrix}$$

The signal we obtain is then a Fourier transform of $\rho(\boldsymbol{r})$ along that direction (the orthogonal directions are summed).

$$\boldsymbol{k}_\phi = \begin{bmatrix} k_x \\ k_y \end{bmatrix} = k \begin{bmatrix} \cos\phi \\ \sin\phi \end{bmatrix}$$

$$k = \gamma G_\phi t$$

$$S(t,\phi) = \int d\boldsymbol{r}\, \rho(\boldsymbol{r}) e^{-i\gamma \boldsymbol{G}_\phi \cdot \boldsymbol{r} t}$$

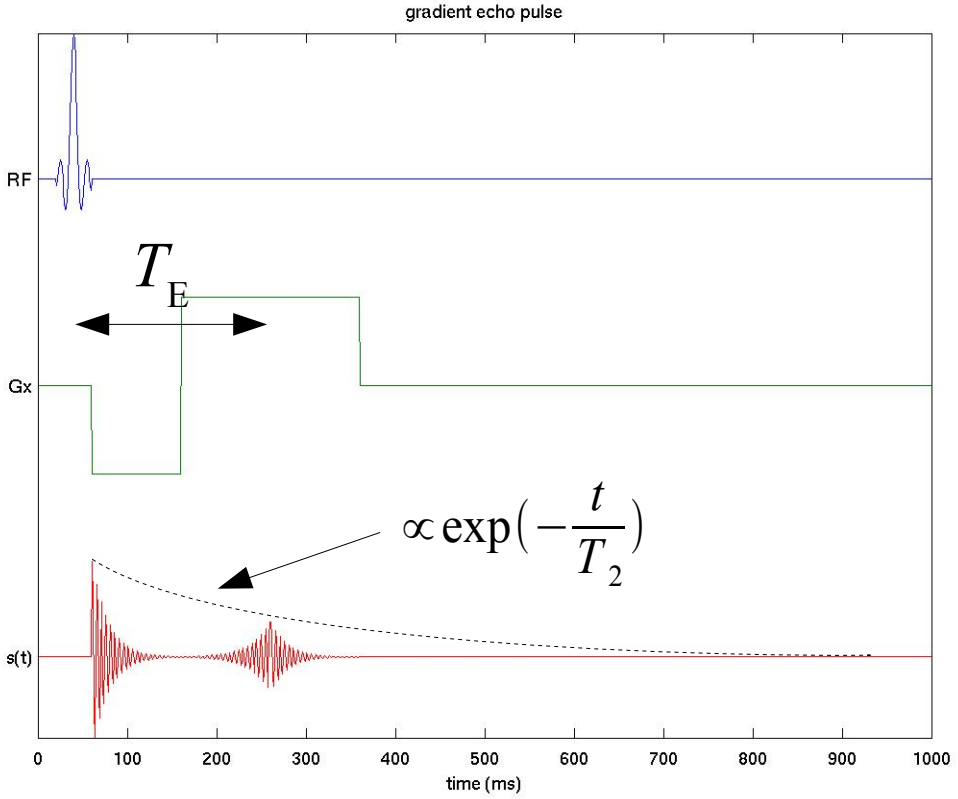
$$\boldsymbol{k}_\phi = \gamma \boldsymbol{G}_\phi t$$

$$S(k,\phi) = \int d\boldsymbol{r}\, \rho(\boldsymbol{r}) e^{-i2\pi \boldsymbol{k}_\phi \cdot \boldsymbol{r}}$$

MRI – Gradient Echo Sequence

90^o and a G_x gradient will also generate a echo sequence. Spins at some locations spin faster than at others due to Gx. After sign reversal of G_x the faster spins become the slower ones, and vice versa. The time it takes the spins to catch up (re-phase) is called echo time (T_E)

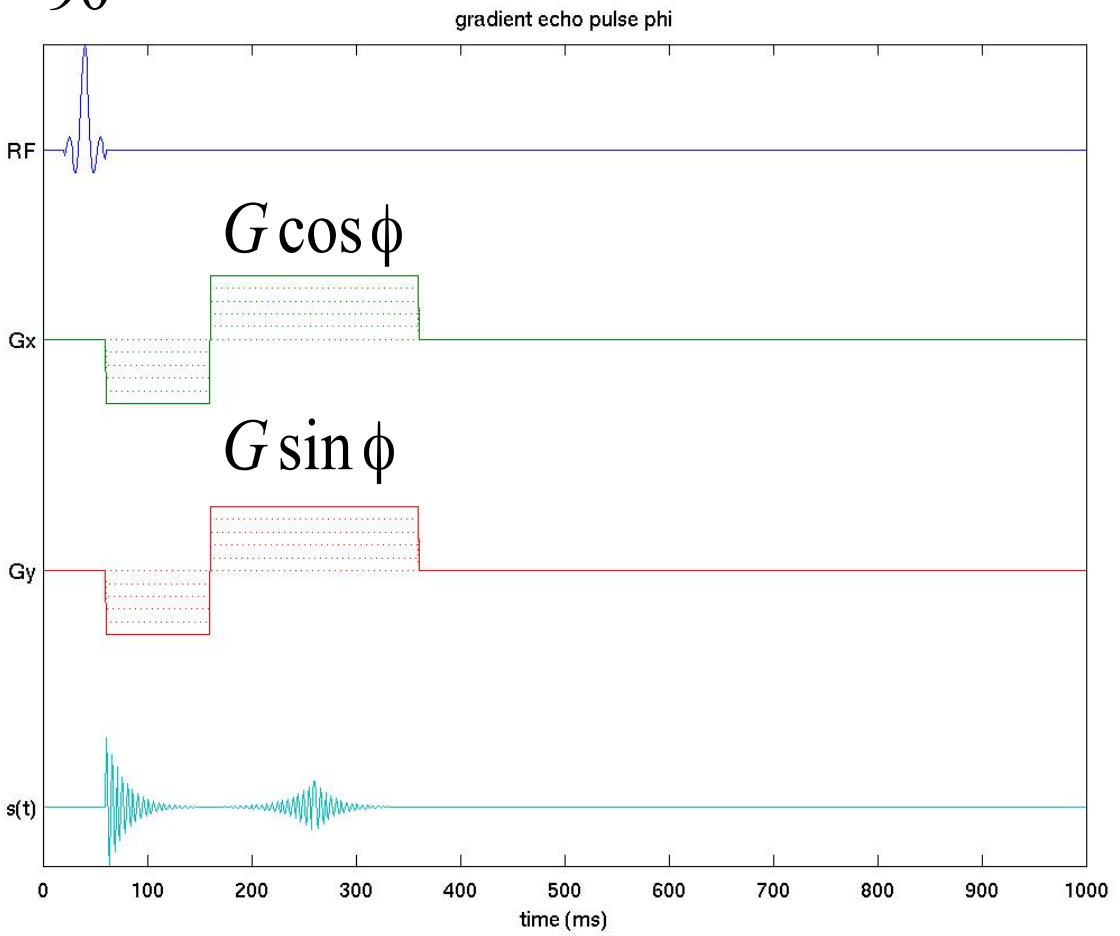
90^o



MRI – Gradient Echo Sequence

We can obtain different angles with different combinations of G_x and G_y:

90^o



The same can be done with at FID or Echo pulse sequence.

MRI – Gradient Echo Sequence

For this pulse sequence the signal we detect is given by

$$S(t,\phi)=\int d\boldsymbol{r}\,\rho(\boldsymbol{r})e^{-i\gamma \boldsymbol{G}_\phi\cdot\boldsymbol{r}(t-T_E)}$$

$$\boldsymbol{k}_\phi=\gamma \boldsymbol{G}_\phi(t-T_E)$$

$$S(k,\phi)=\int d\boldsymbol{r}\,\rho(\boldsymbol{r})e^{-i2\pi \boldsymbol{k}_\phi\cdot\boldsymbol{r}}$$

or in Cartesian coordinates:

$$S(k_x,k_y)=\int dx\,dy\,\rho(x,y)e^{-i2\pi(k_x x+k_y y)}$$

i.e. $S(k_x,k_y)$ is the 2D Fourier transform of $\rho(x,y)$.

MRI – k-space

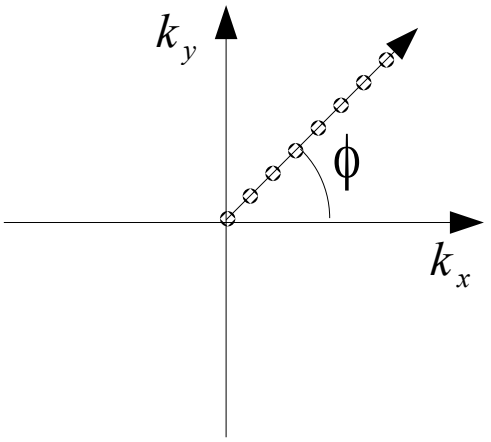
Lets consider the RF signal we measure. It represents the data in the frequency domain, i.e. the "k-space".

For a FID signal

$$\boldsymbol{k}_\phi = \gamma \boldsymbol{G}_\phi t$$

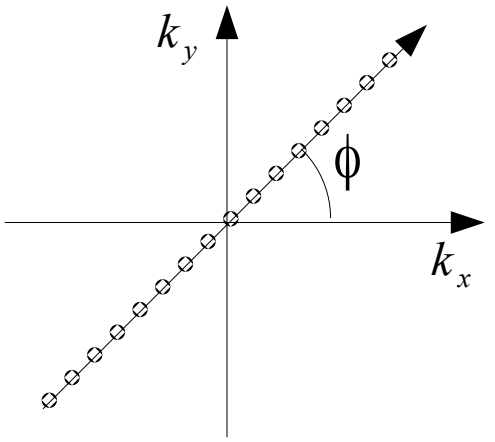
For an ECHO signal

$$\boldsymbol{k}_\phi = \gamma \boldsymbol{G}_\phi \left(t - T_E\right)$$



$$\boldsymbol{k}_\phi = \begin{bmatrix} k_x \\ k_y \end{bmatrix} = k \begin{bmatrix} \cos\phi \\ \sin\phi \end{bmatrix}$$

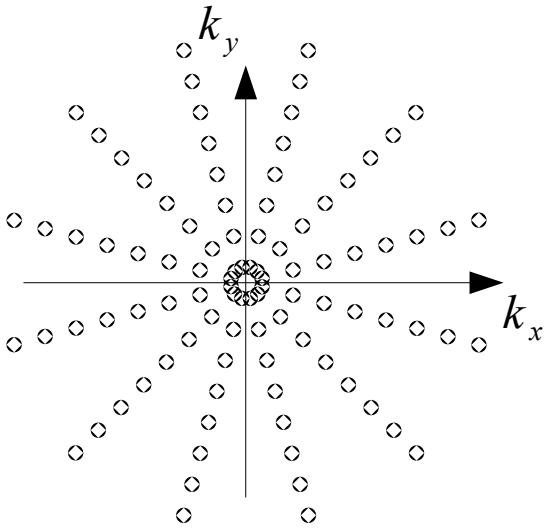
$$\boldsymbol{G}_\phi = \begin{bmatrix} G_x \\ G_y \end{bmatrix} = G_\phi \begin{bmatrix} \cos\phi \\ \sin\phi \end{bmatrix}$$

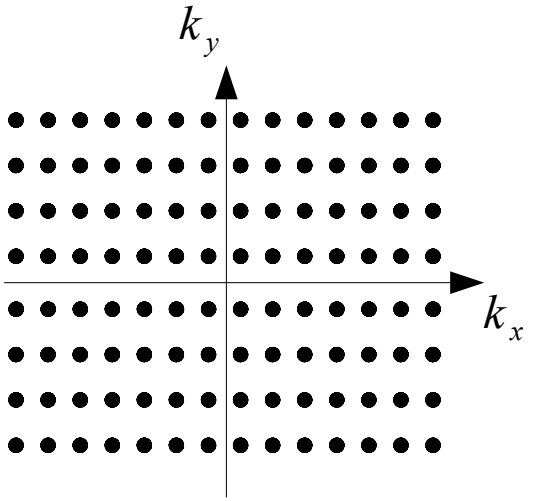


Time starts at t=0 and is sampled in discrete points $t = \Delta t\, n$

MRI – k-space

Signals taken at multiple angles ϕ cover the k-space and allow therefore reconstruction (left).

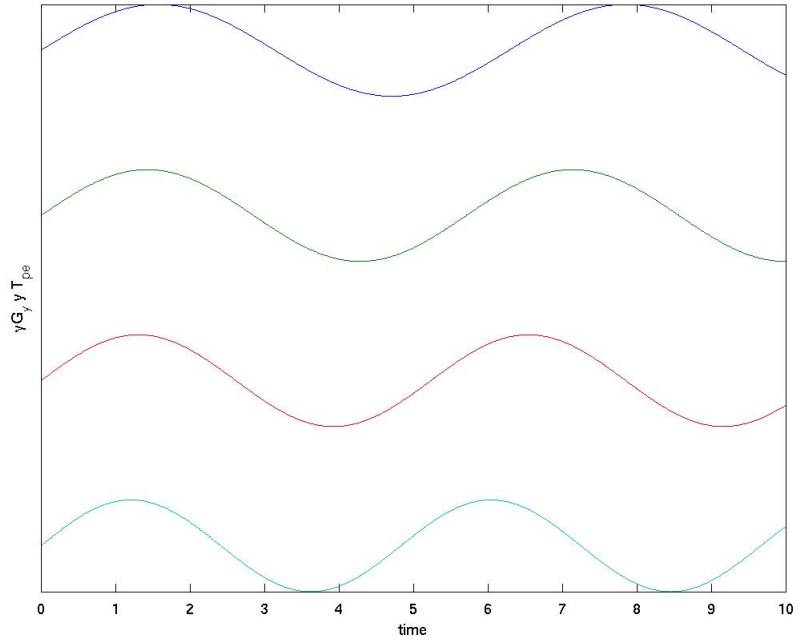




Is there a pulse sequence that can sample the Fourier space evenly as shown on the right so that we can use direct 2D Fourier inverse?

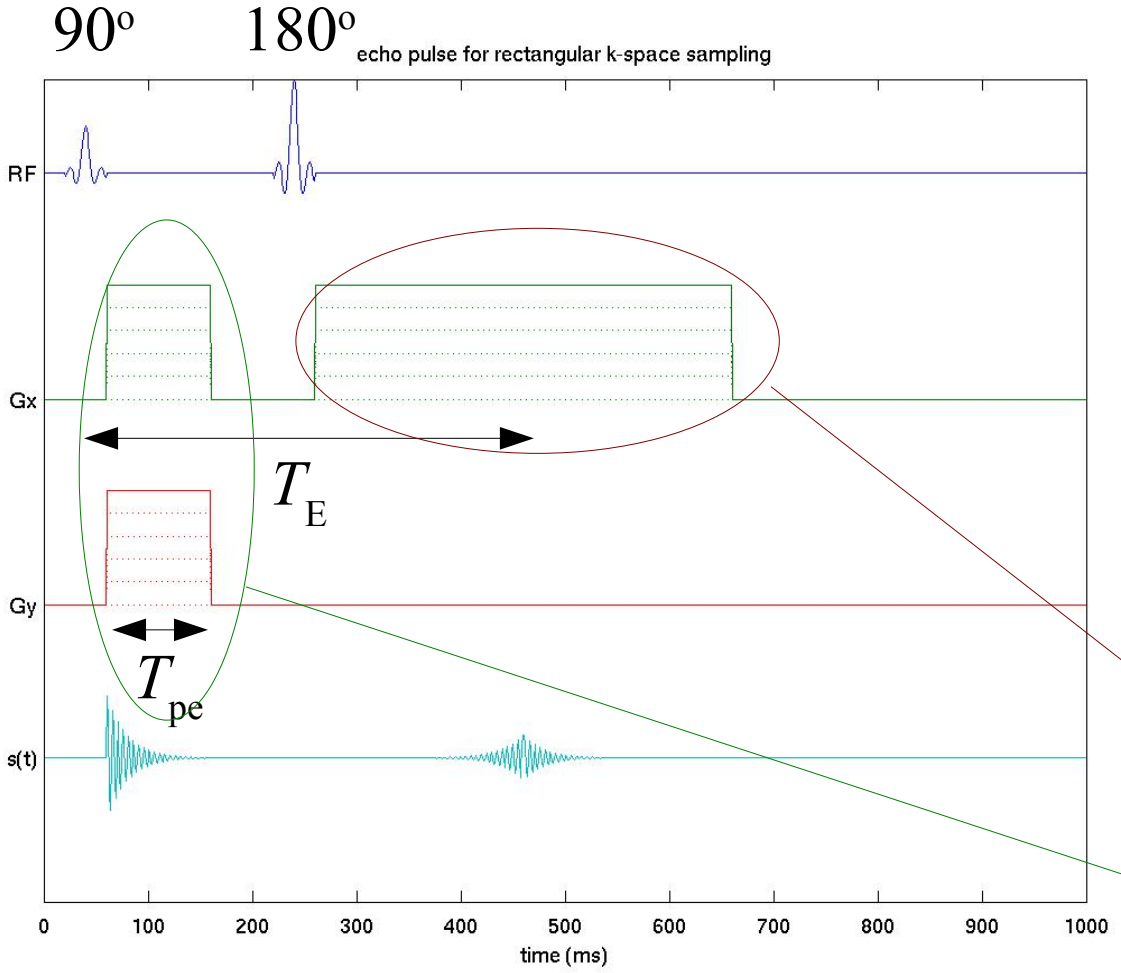
MRI – Phase Encoding

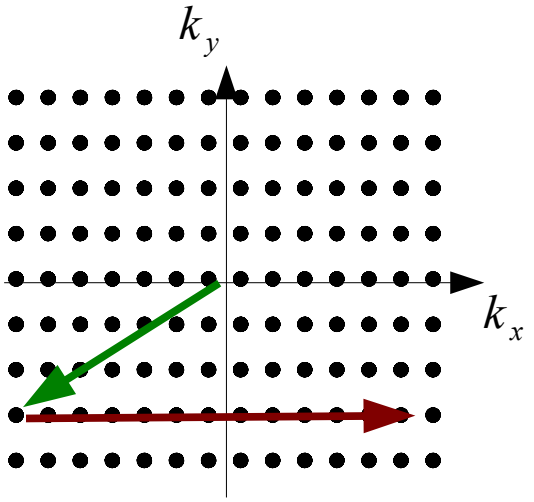
If spins precess at different speeds (due to a variable encoding gradient G_y) during a fixed amount of time T_{pe} they will gain a different phase:



MRI – Phase Encoding

A phase encoding echo pulse sequence, which will sample the k-space along the k_x axis for different values of k_y is as follows:





Frequency encoding

Phase encoding

MRI – Phase Encoding

With a phase encoding gradient G_y in y direction and frequency encoding gradient in G_x in x direction the echo signal would be (ignoring the z direction again):

$$S(t, T_{pe}) = \int dx\, dy\, \rho(x, y)\, e^{-i\gamma G_x x (t - T_E) + \gamma G_y y T_{pe}}$$

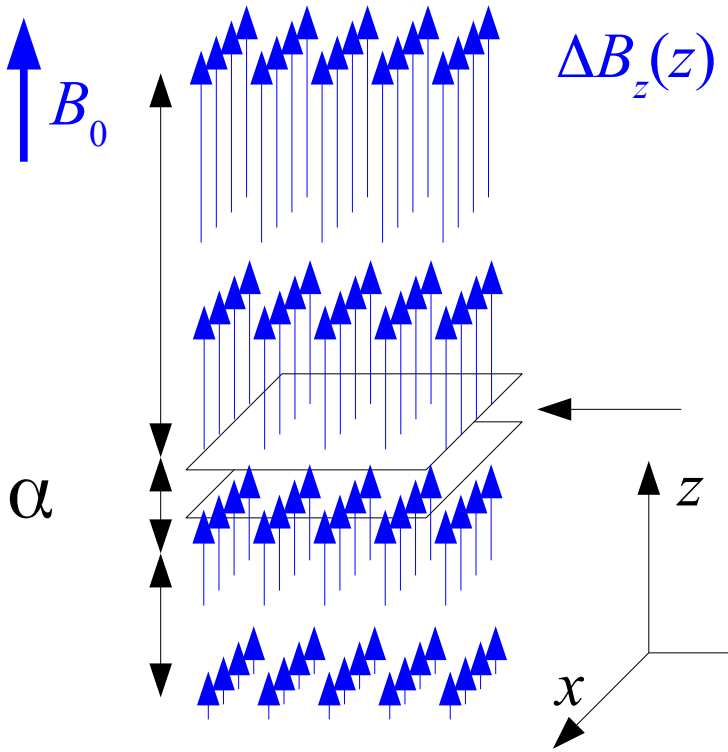
which can be rewritten as a 2D Fourier transform with the following definitions:

$$k_x = \gamma G_x (t - T_E) \qquad k_y = \gamma G_y T_{pe}$$

$$S(k_x, k_y) = \int dx\, dy\, \rho(x, y)\, e^{-i 2\pi (k_x x + k_y y)}$$

MRI – Slice selection

So far we considered gradients applied *after* the RF pulse during free precession. A gradient G_z *during* the RF pulse will select a transversal slice that satisfies the *resonance condition*: The RF pulse affects the spin precession coherently only if the frequency matches the B_z field. For the rest $M_{xy} = 0$ after α pulse.



$$M_{xy}(0)=0$$

$$M_{xy}(0)=M_0 \sin\alpha$$

$$M_{xy}(0)=0$$

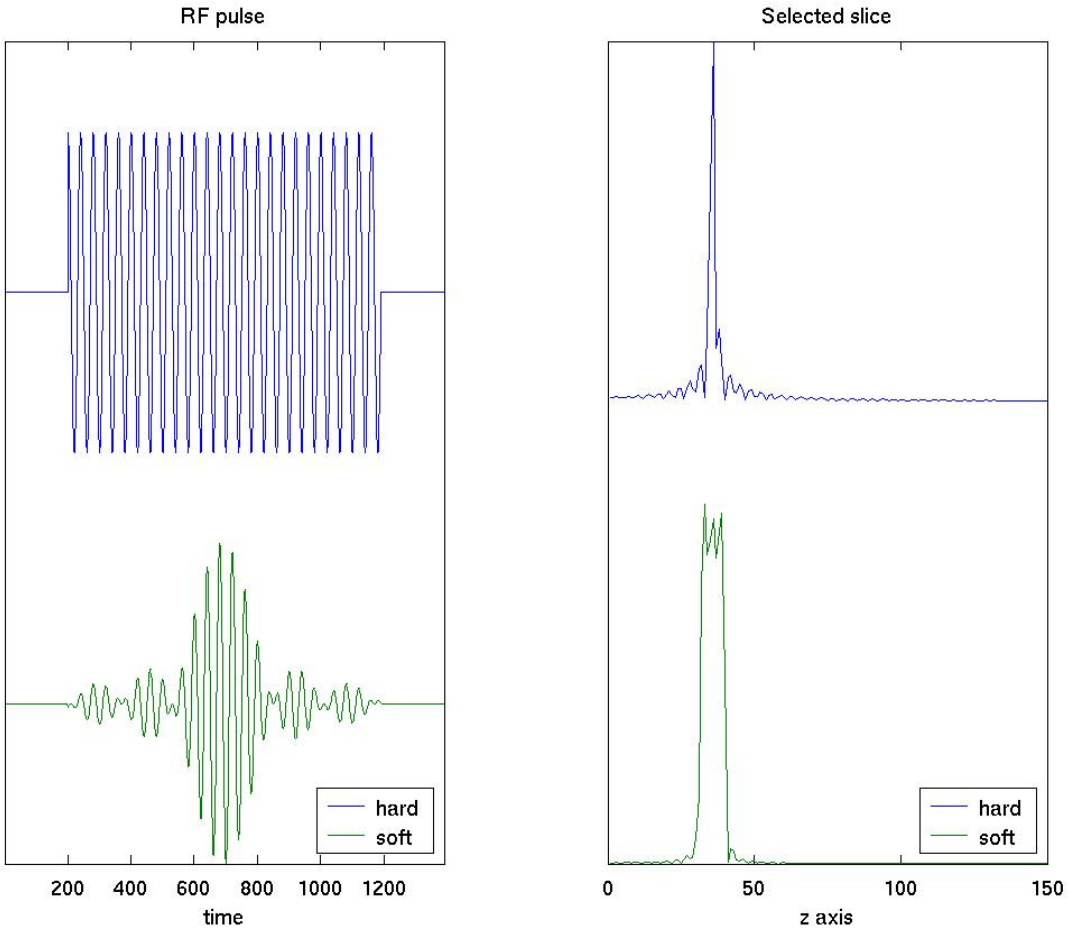
Selected slice at:

$$z=\frac{\gamma B_0-\omega}{\gamma G_z}$$

Only this slice will generate a signal!

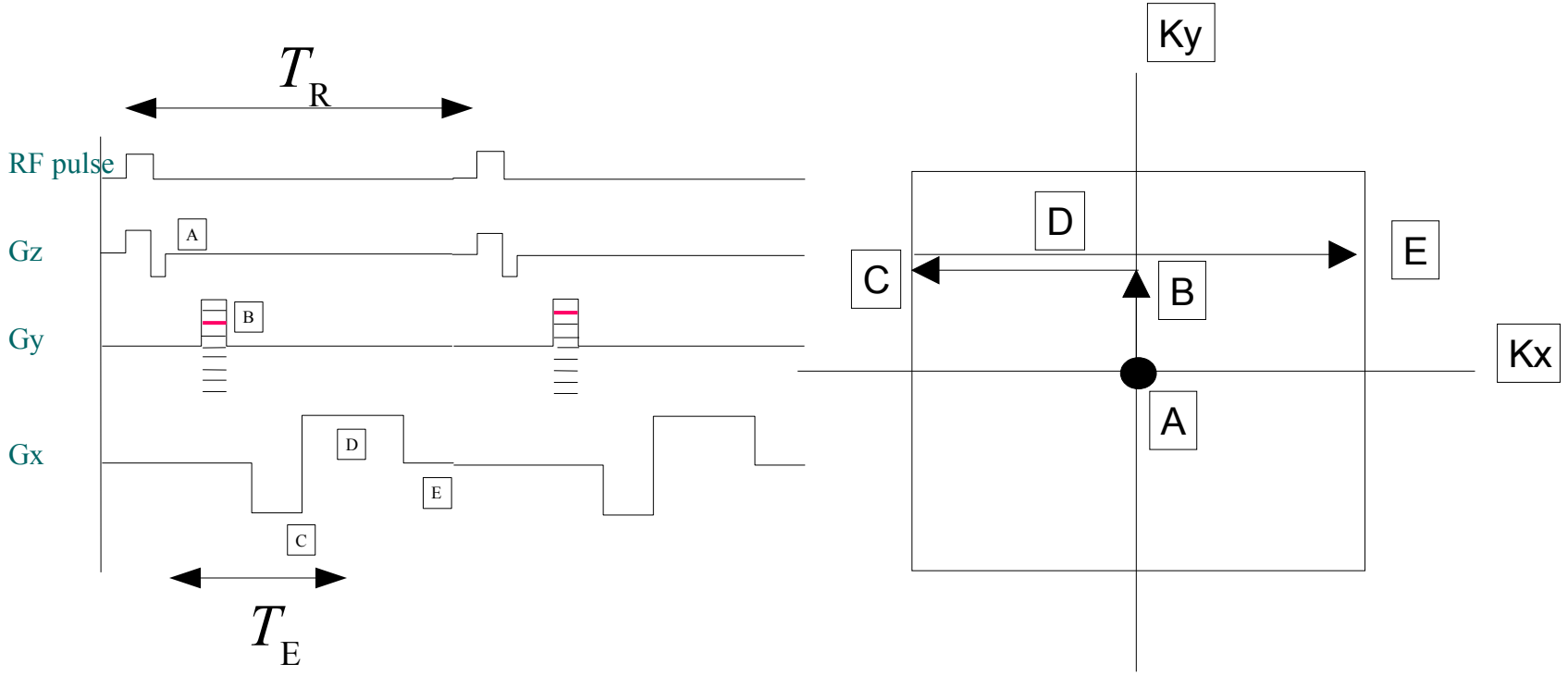
MRI – Slice selection

Note that a "hard" RF pulse contains high frequency components. It is therefore less selective in space as a "soft" pulse (sinusoid modulated by a sync functions - $\sin(\omega_0 t)*\mathrm{sinc}(\omega t)$:

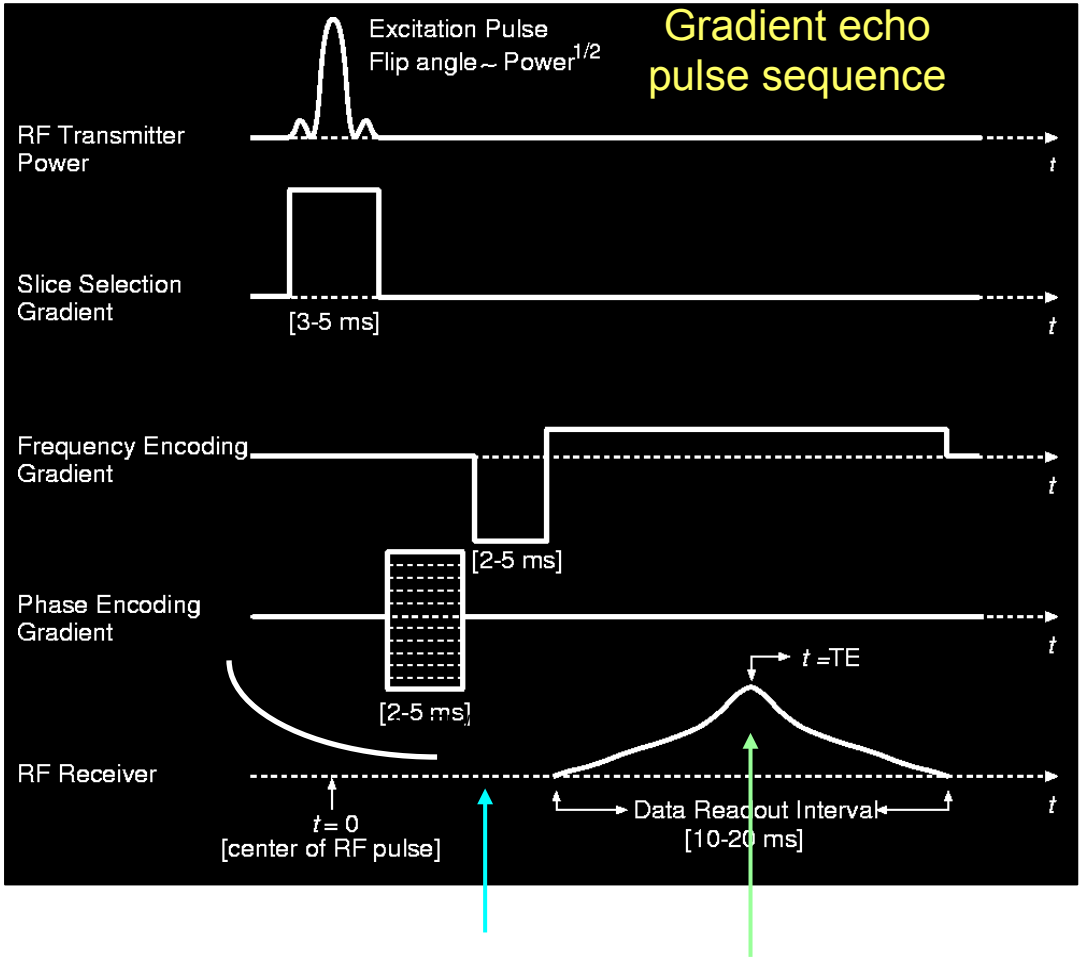


MRI – a pulse sequence example

Example for a full pulse sequence with gradient echo and the corresponding path in k-space:



MRI – a pulse sequence example



A gradient reversal (shown) or 180 pulse (not shown) at this point will lead to a recovery of transverse magnetization

TE = time to wait to measure refocussed spins

Echos – refocussing of signal

Spin echo:

use a 180 degree pulse to "mirror image" the spins in the transverse plane

when "fast" regions get ahead in phase, make them go to the back and catch up

measure T2

ideally TE = average T2

Gradient echo:

flip the gradient from negative to positive

make "fast" regions become "slow" and vice-versa

measure T2*

ideally TE ~ average T2*

Jody Culham, http://defiant.ssc.uwo.ca/Jody_web/fmri4dummies.htm

Source: Mark Cohen

MRI – Summary for Magnetic fields

Main Magnet

High, constant,Uniform Field, B_0.

Gradient Coils

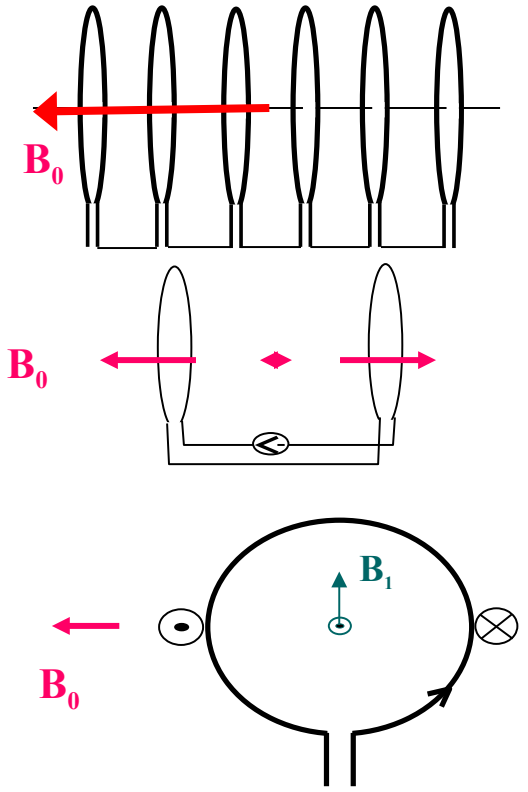
Produce pulsed, linear gradients in this field.

G_x, G_y, & G_z

RF coils

Transmit: B1 Excites NMR signal (FID).

Receive: Senses FID.



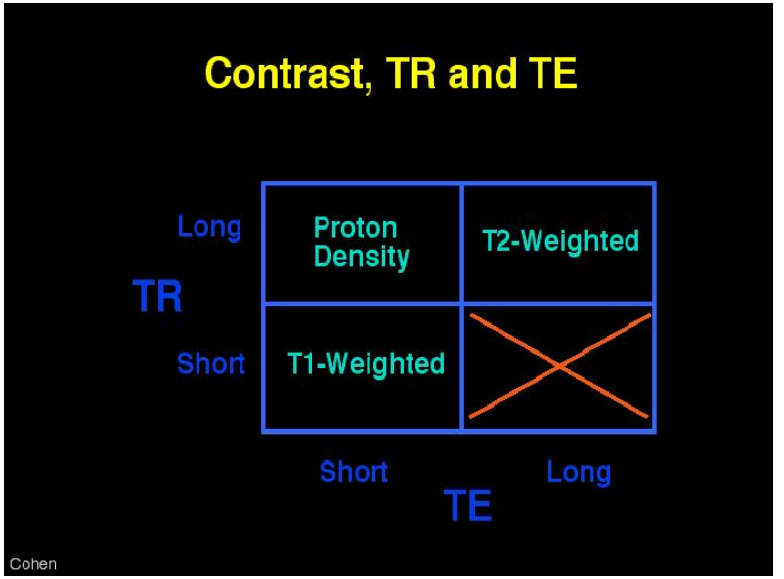
MRI – Contrast properties

The strength of the NMR signal produced by precessing protons in a tissue depends on

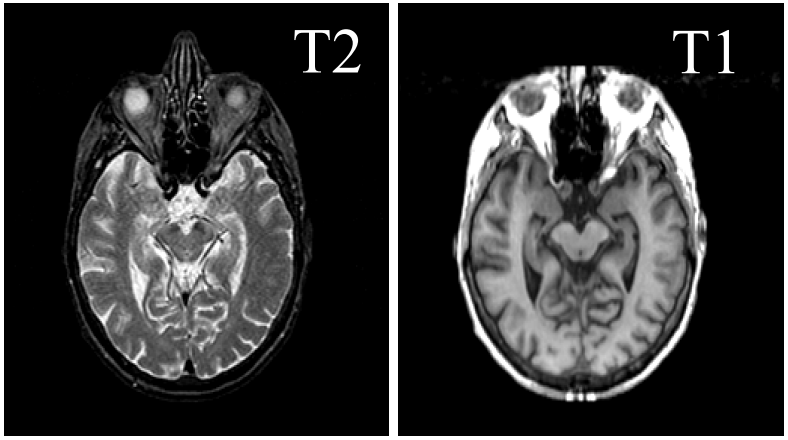
- T1, T2 of the tissue.
- The density of protons in the tissue.
- Motion of the protons (flow or diffusion).
- The MRI pulse sequence used

In a T1 "weighted" image the pulse sequence is chosen so that T1 has a larger effect than T2.

Images can also be made to be T1, T2 proton density or flow/diffusion weighted.



Source: Mark Cohen



MRI – Contrast, T1, T2

MRI Contrast is created since different tissues have different T1 and T2.

Gray Matter: (ms) T1= 810, T2= 101

White Matter: (ms) T1= 680, T2= 92

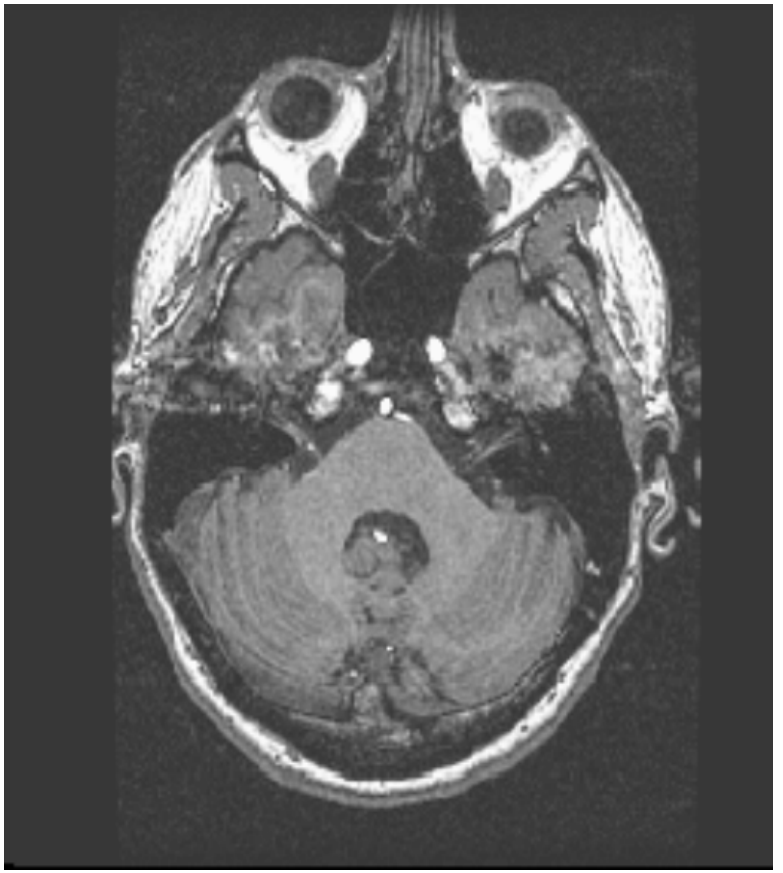
Bone and air are invisible.

Fat and marrow are bright.

CSF and muscle are dark.

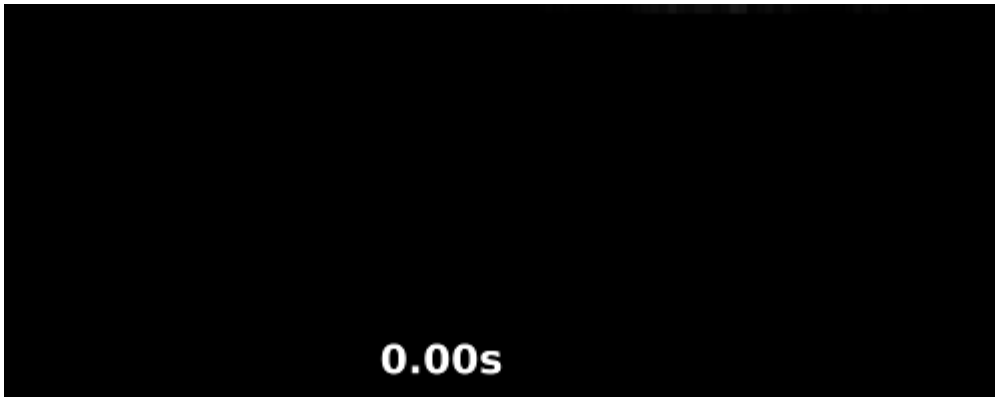
Blood vessels are bright.

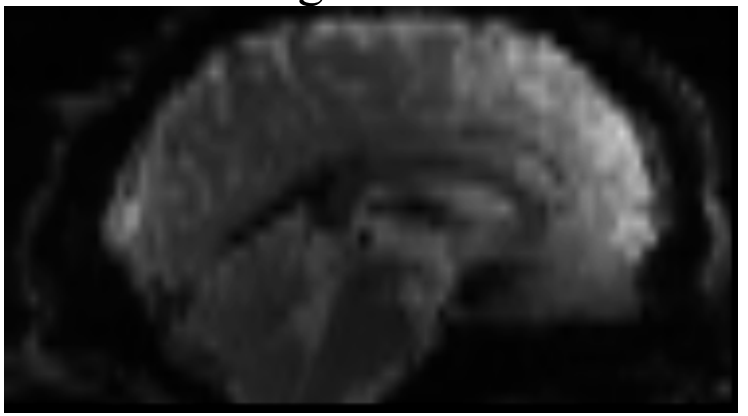
Gray matter is darker than white matter.

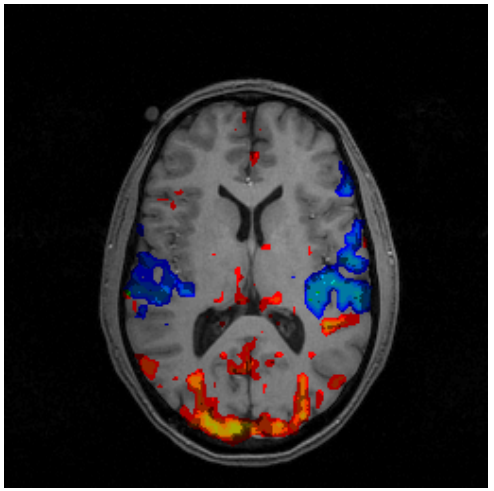


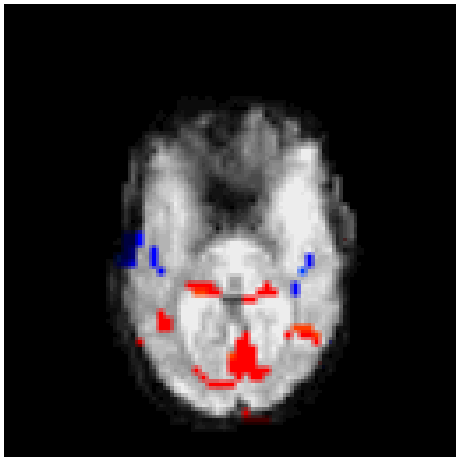
Functional MRI

fMRI measures blood oxigen level difference (BOLD). Measured full volume in about 2s. It is a T2* weighted images. T2* is fast, and dominated by inhomogeneity susceptibility-induced field distortions. Oxigenated blood has higher magnetic susceptibility. In 1990 Ogawa observes BOLD effect with T2*: blood vessels became more visible as blood oxygen decreased. First fMRI image obtained in 1992.







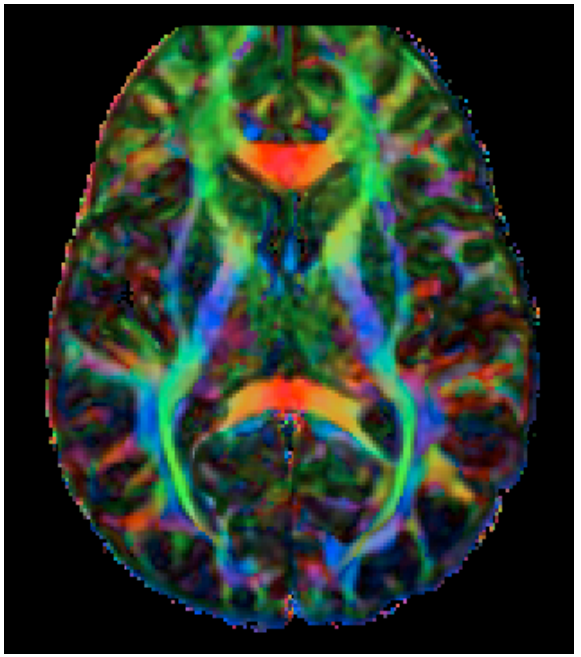


Diffution Tensor Imaging

Measures the diffusion of water molecules (m/s). Has orientation.

Mechanism: During the RF pulse a gradient is applied. This introduces a phase shift that depends on gradient direction. The same is repeated with opposite sign of the gradient, so that the phase shift is removed again, but not for spins (mostly water) which moved. Those spin will not align and the signal will be weaker. So if we take two images and subtract, we get a drop of signal strength where there was more diffusion. This is repeated with pulse gradients in different orientations to measure diffusion in different orientations.

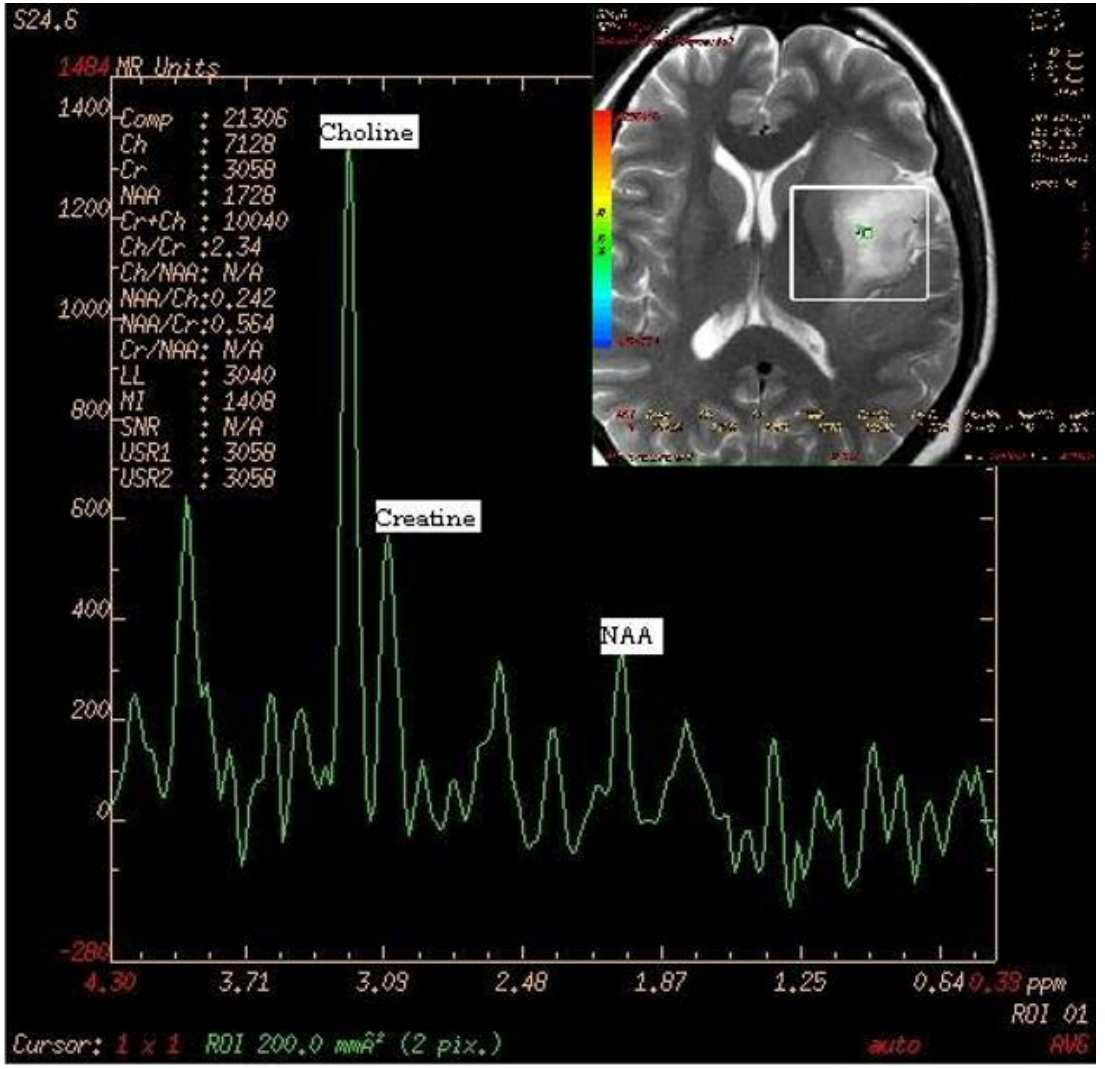
DTI raw data. Color indicates oriantation of diffusion



Fiber track tracing. Color indicates identity of the fiber track.

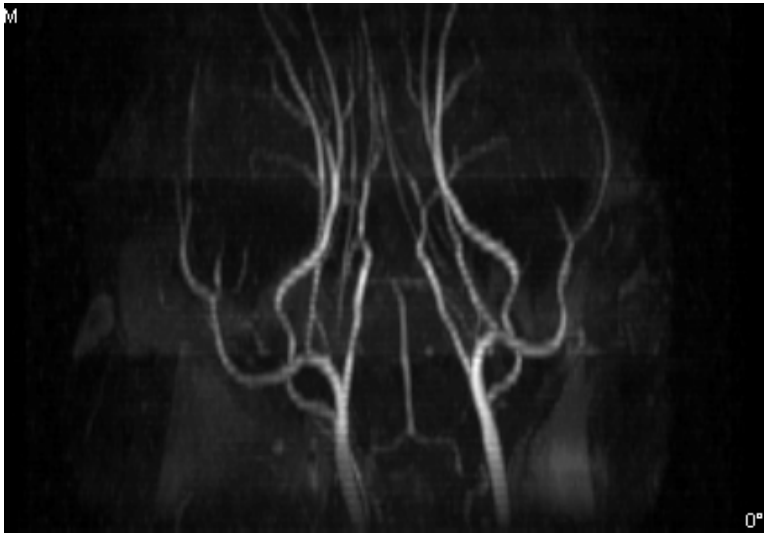
MR Spectroscopy Imaging

MRS, also known as Chemical Shift Imaging, activates the tissue with a somewhat broader band RF pulse (multiple frequencies). Different molecules have slightly different resulting Bz field due to neighboring spins, so they will resonate at slightly different frequencies, typically only parts per million (ppm) of the hydrogen frequency. The detected RF signal has then a spectrum with peaks at resonant frequencies of different molecules.



MR Angiography

MRA in a rat



Rapid MRA time resolved