MRI physics for SPM users

SPM course

11/2013

Antoine Lutti

antoine.lutti@chuv.ch

Outline

- **General principals**
  - Origin of the signal
  - RF excitation
  - Relaxation (T1, T2, …)
- **Anatomical imaging**
  - Image contrast
  - Standard acquisition methods
  - Advanced acquisition methods
- **Functional imaging**
  - BOLD effect
  - Limitations of fMRI acquisitions
  - Advanced methods

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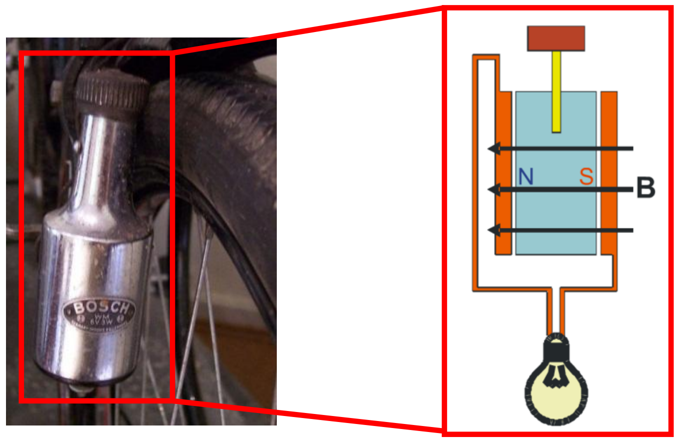
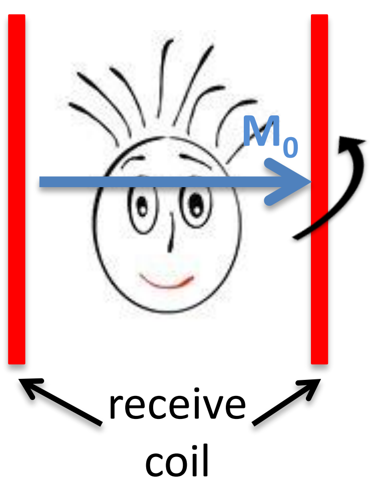
Origin of the signal

Bicycle dynamo



Rotating magnet induces an electric current in the coil

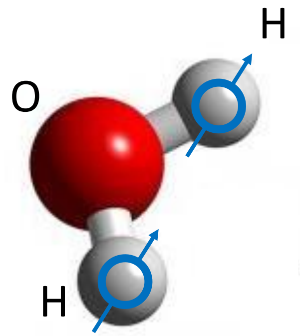
MRI

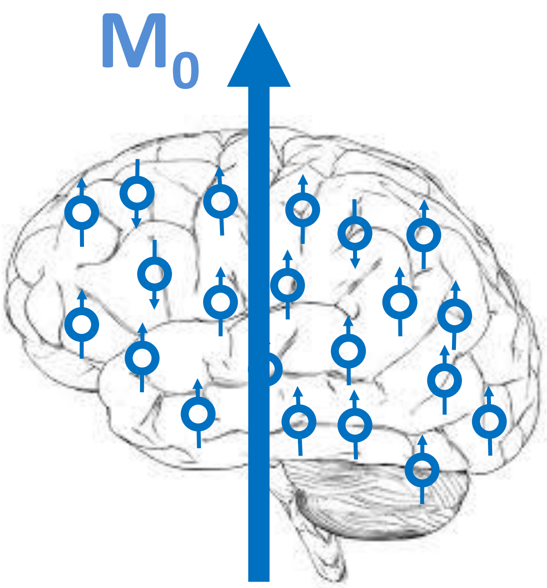


Rotating magnetization M_0 induces an signal in the head coil

Origin of the signal

Water molecule

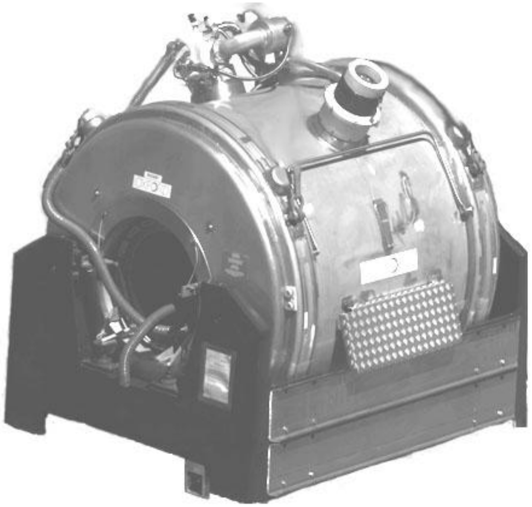




- **MRI signal arises from water molecules** surrounding brain tissue **NOT** from tissue itself

- The **higher** the water concentration (*proton density*) the **stronger** the signal

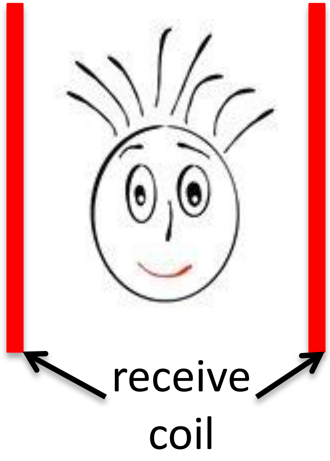
Hardware



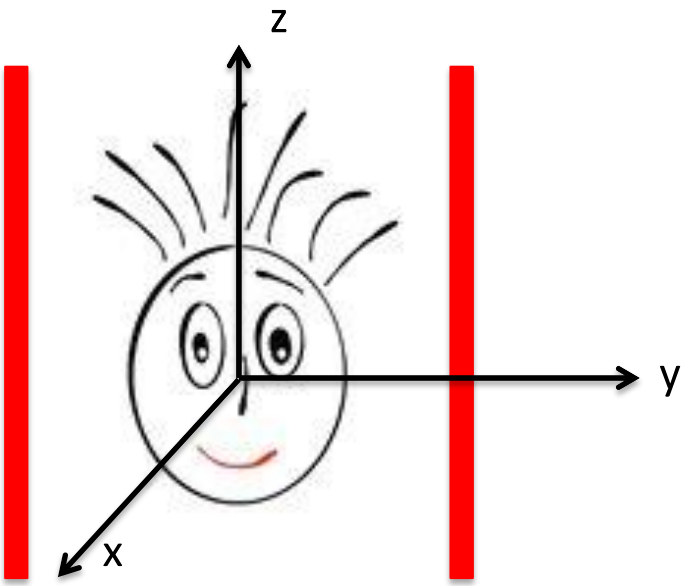
Magnetic field $\mathbf{B_0}$ created by superconducting magnet

$\mathbf{B_0}$ is oriented along the main direction of the bore

The receive coil detects signal arising from the magnetization



Layout - orientation



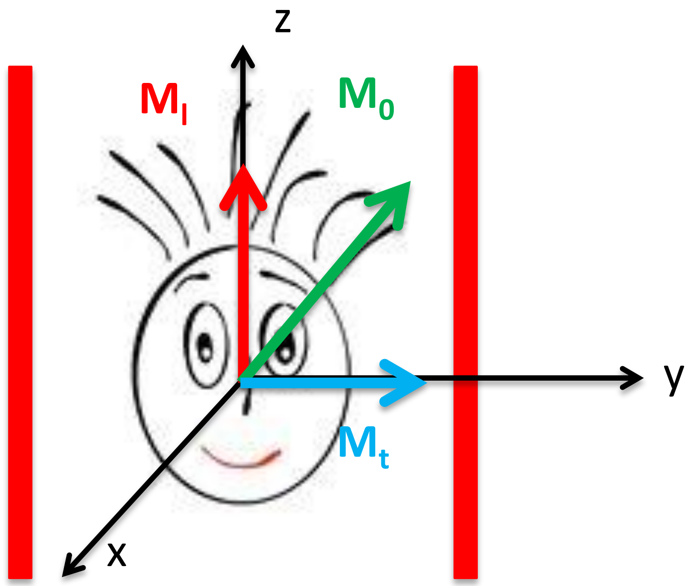
z direction: aligned with receive coil

Longitudinal direction

(x,y) plane: perpendicular to receive coil

Transverse plane

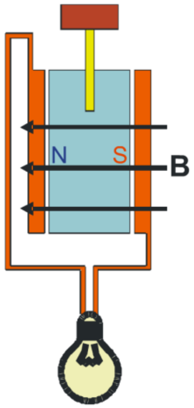
Layout - orientation

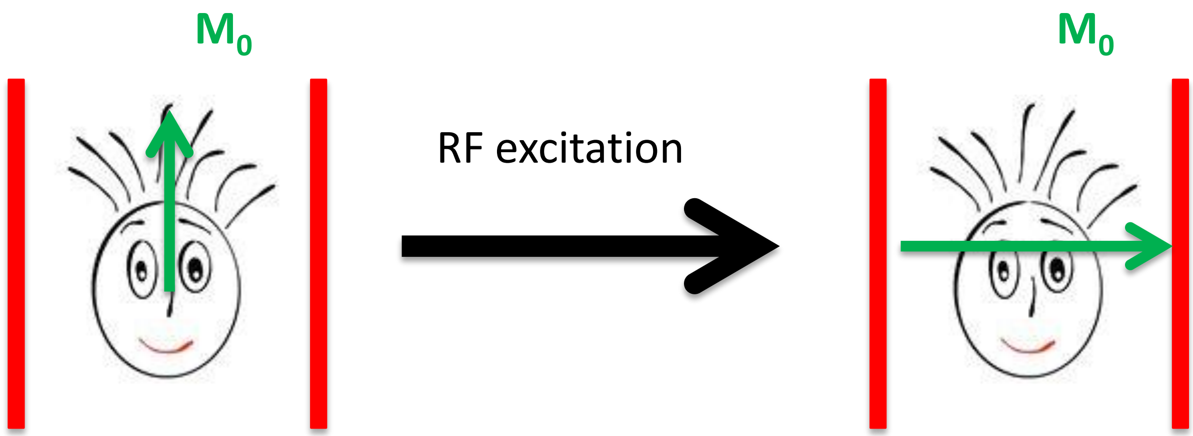


Magnetization M_0 has a **longitudinal component** along the z-direction

Magnetization M_0 has a **transverse component** in the x-y plane

RF excitation



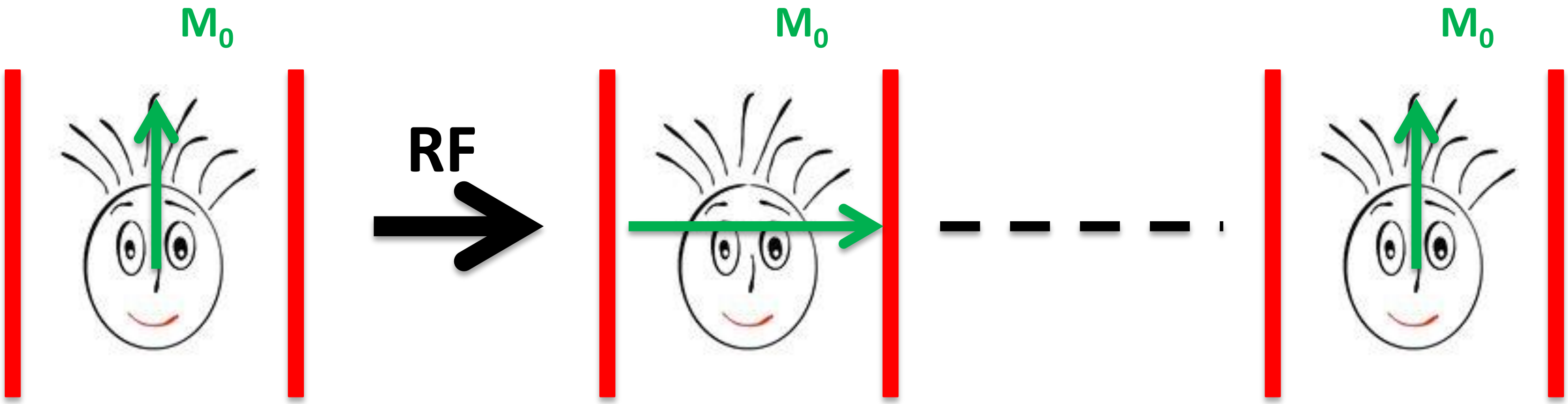


At rest:
M_0 is along the longitudinal direction
Signal cannot be detected

After RF excitation:
M_0 is in the transverse plane
Signal can be detected

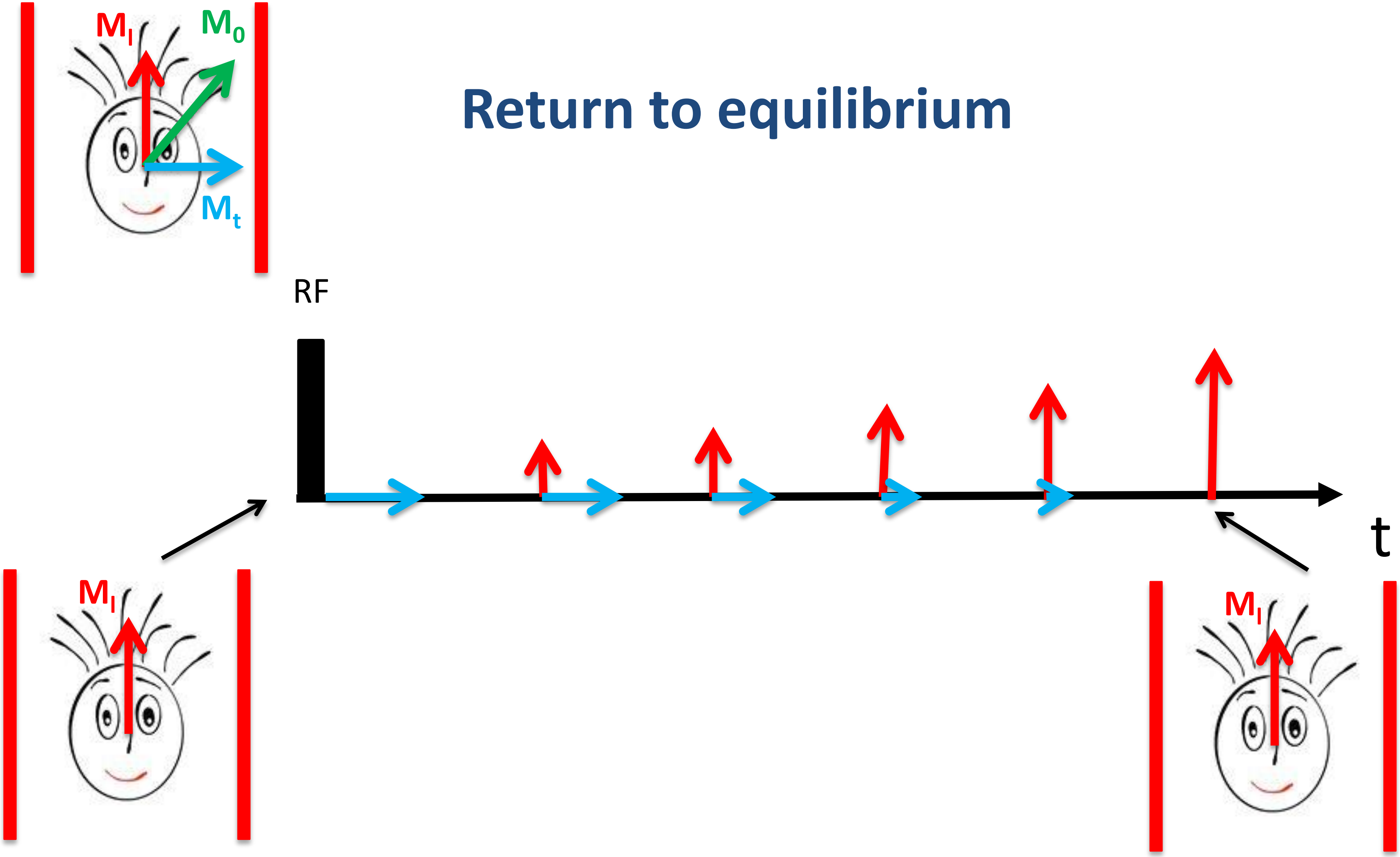
All MR sequences require RF excitation

Return to equilibrium



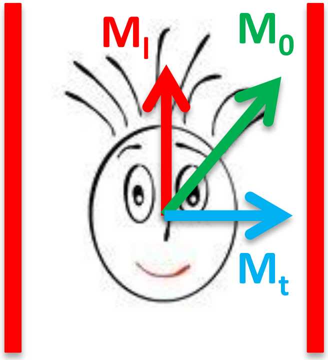
After RF excitation M_0 returns to its initial state (*equilibrium*)

Return to equilibrium

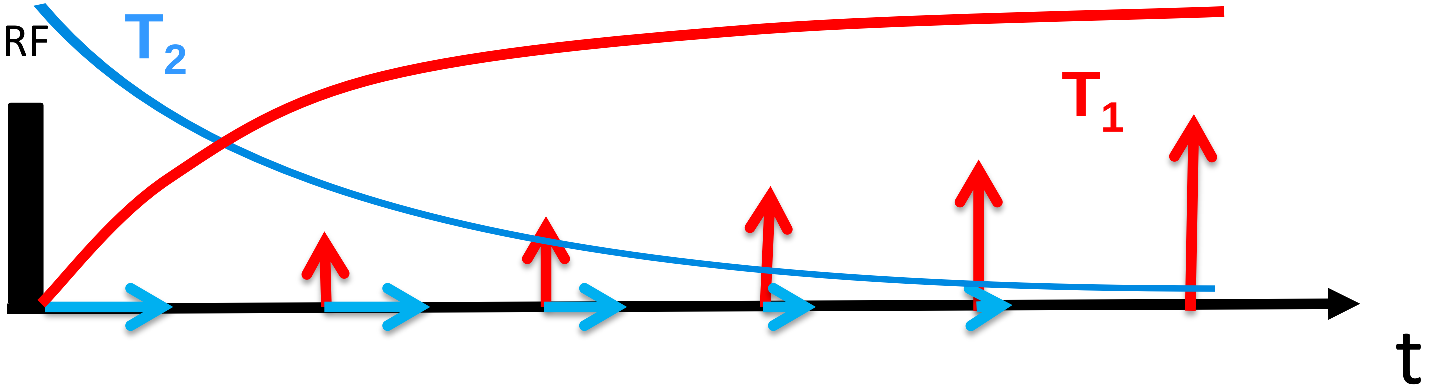


Equilibrium

Equilibrium



Return to equilibrium



Following RF excitation M_0:

- **Longitudinal** component of M_0 **increases**. Recovery time **T1**
- **Transverse** component of M_0 **decreases**. Decay time **T2**

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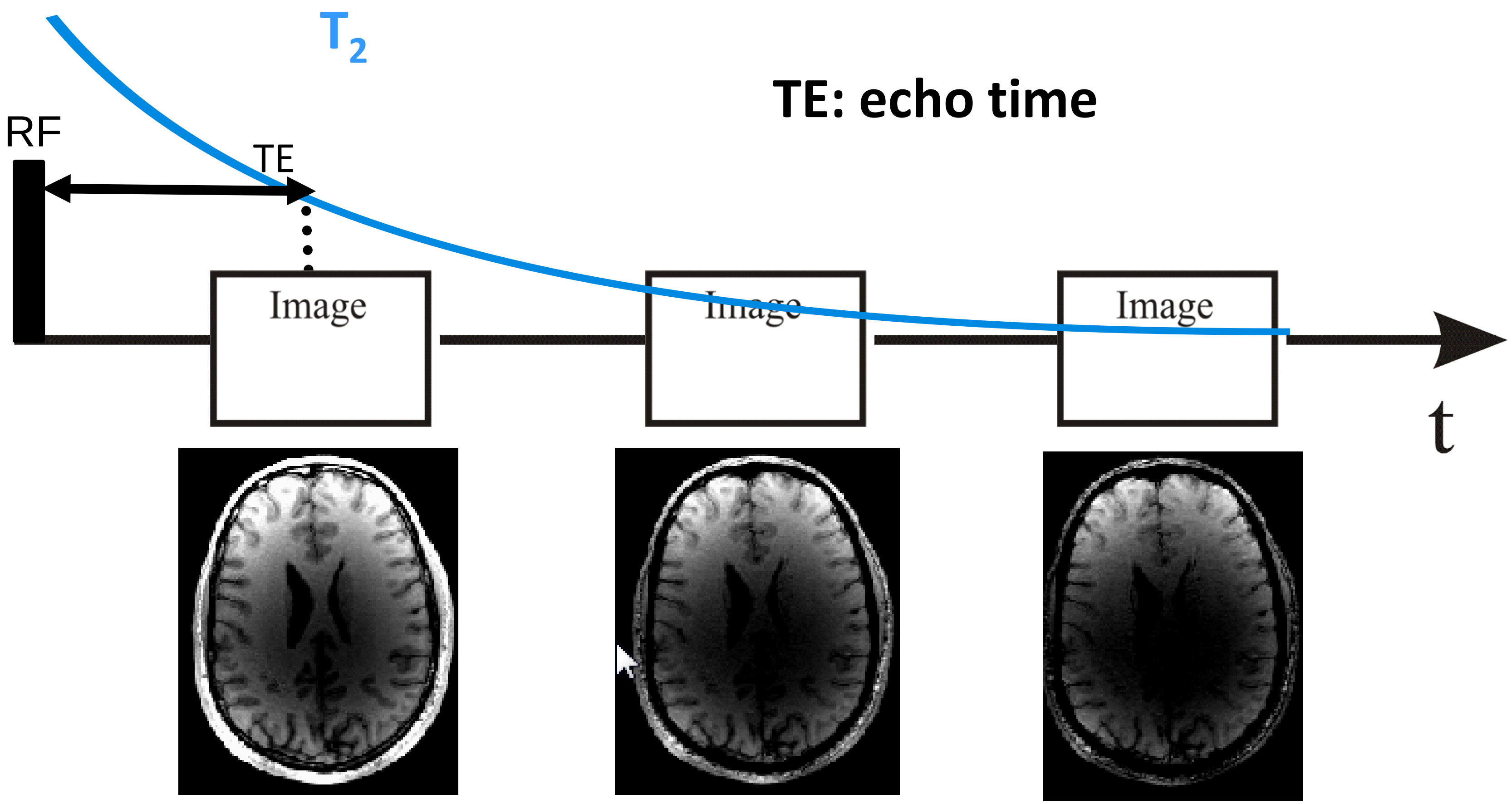
Anatomical imaging requirements

- **Optimal image contrast**

$$CNR = \frac{(S_{WM} - S_{GM})/(S_{WM} + S_{GM})}{\sigma}$$

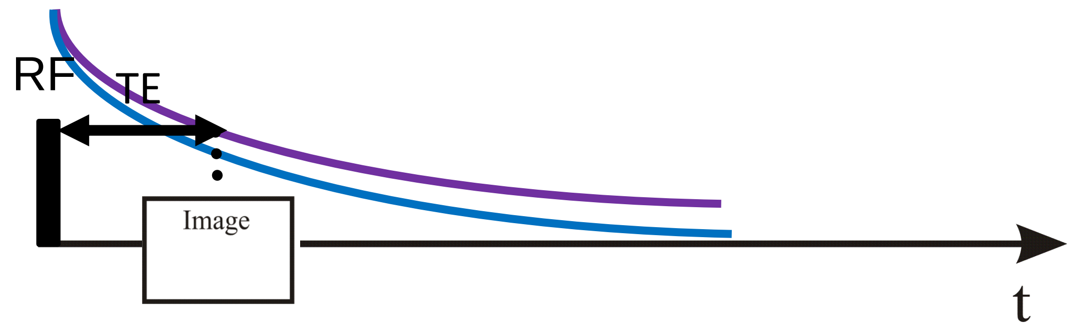
- **High image resolution**
- **Preserve brain morphology**
- **Avoid signal losses**

T2 relaxation & signal intensities

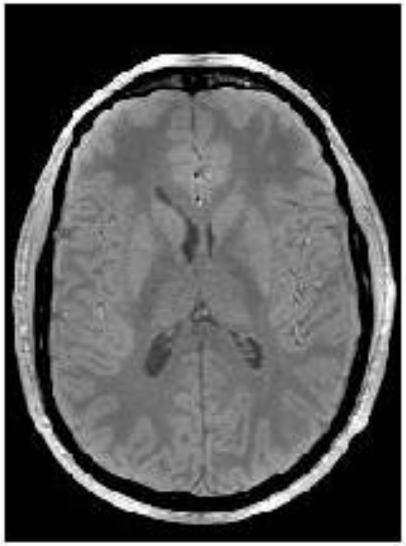


SIGNAL INTENSITIES DECREASE WITH INCREASING ECHO TIME

T_2 contrast



TE << T2
proton density-weighted image



TE ~ T2
T2-weighted image

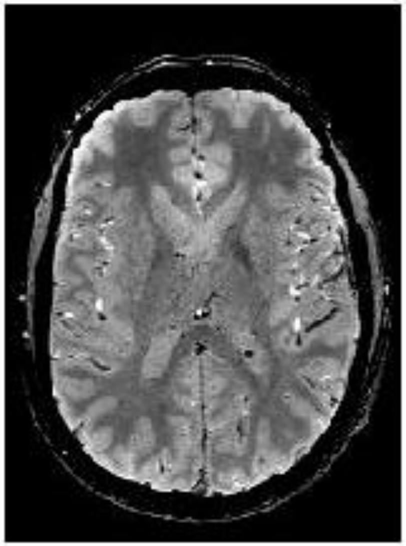
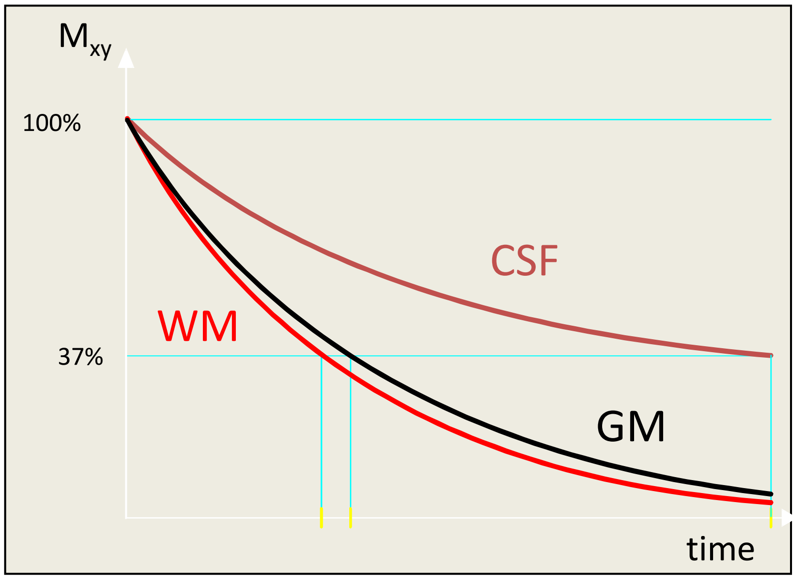
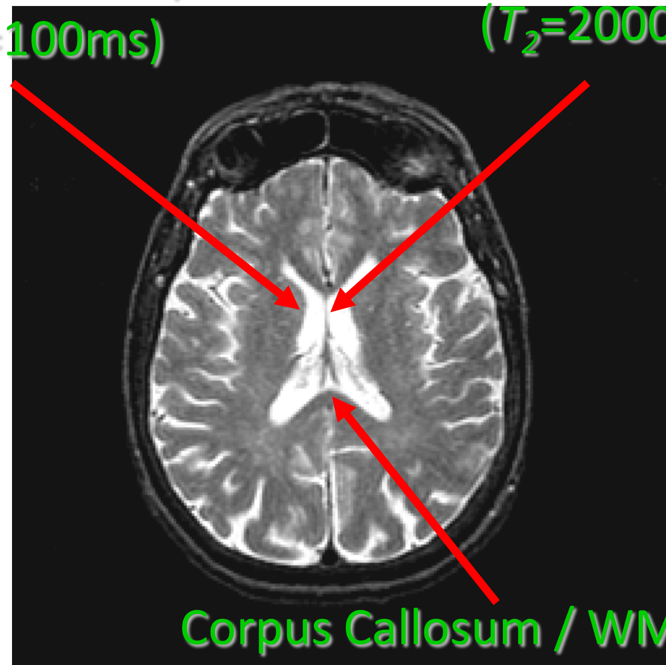


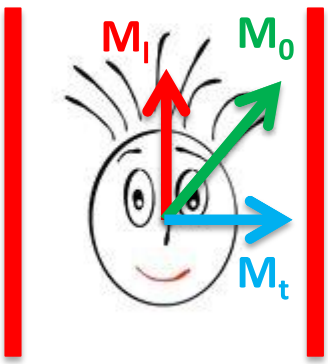
Image contrast is **TE-dependent**

T_2 contrast





- $T_{2,CSF}$>$T_{2,GM/WM}$ => On T_2-weighted images, CSF appears bright
- WM and GM have similar T_2 values => low WM/GM contrast in T_2-weighted images

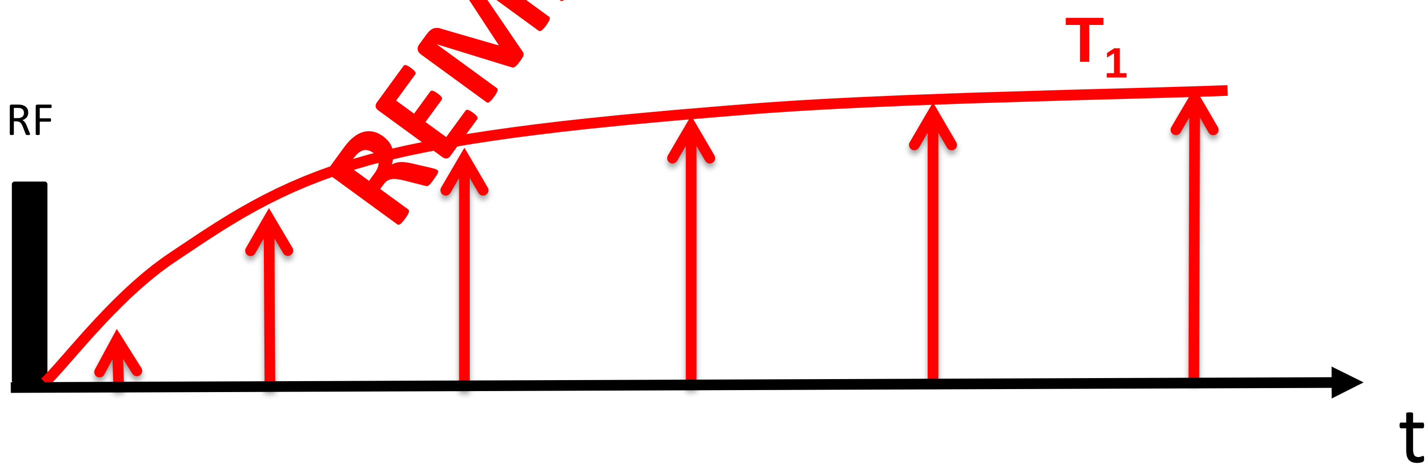


Longitudinal relaxation

REMINDER

Return to equilibrium:

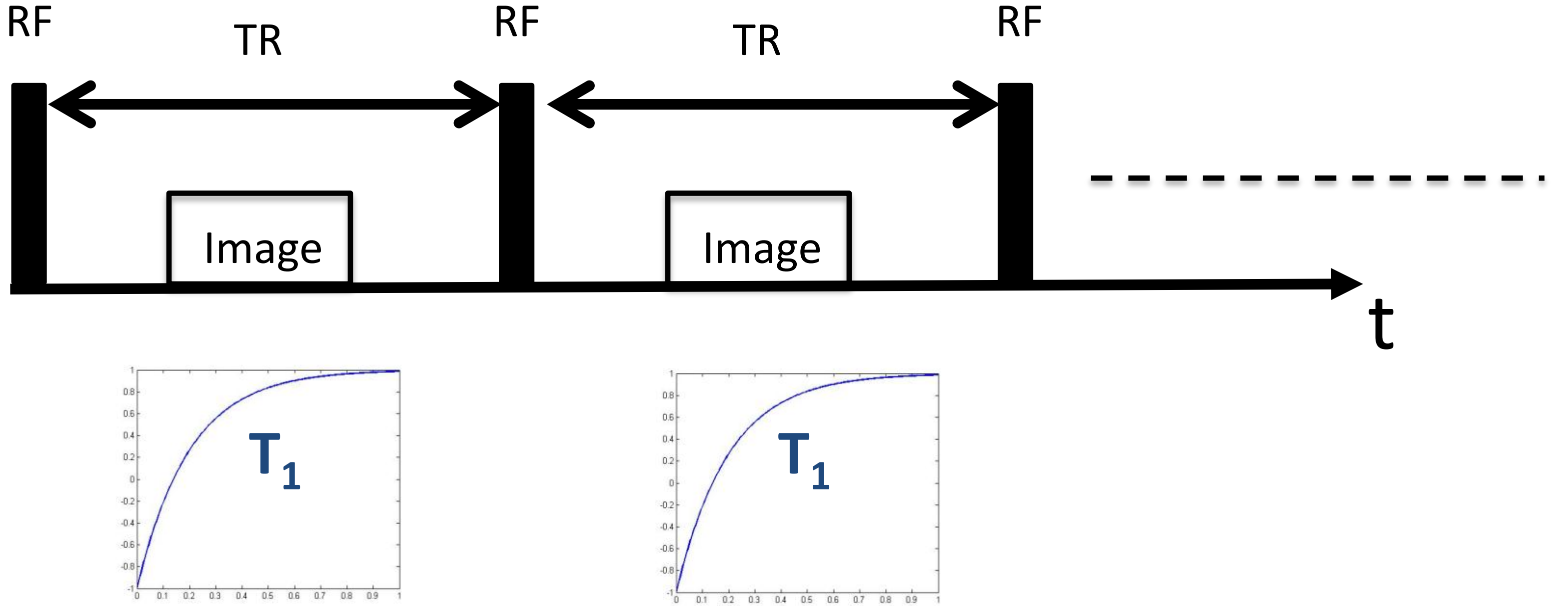
Increase of longitudinal component time constant $\mathbf{T_1}$



The **recovered longitudinal component will be flipped into the transverse plane** when RF excitation is repeated

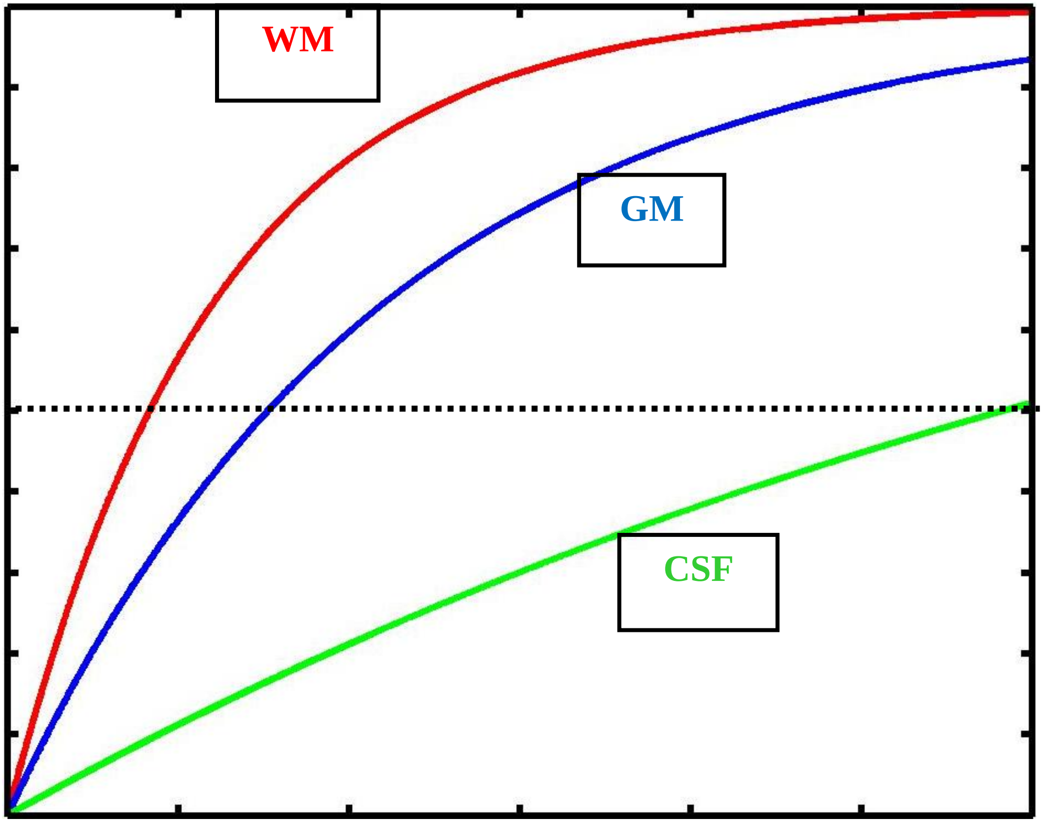
Longitudinal relaxation

A simple imaging acquisition:



T1 relaxation during TR governs amount of magnetization available for next excitation

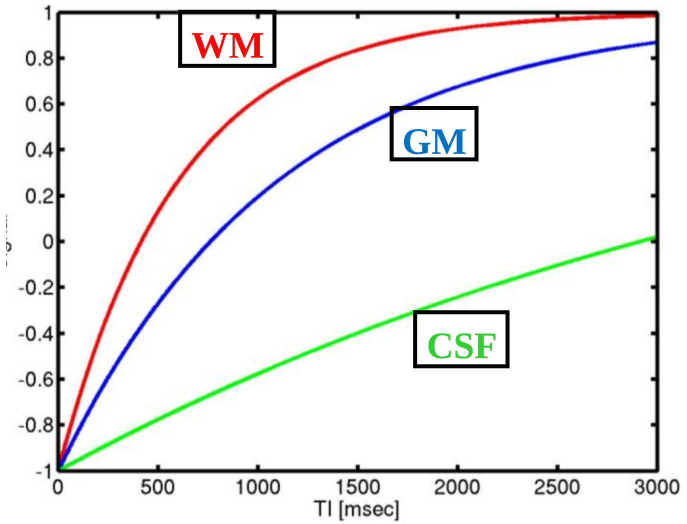
T1 contrast



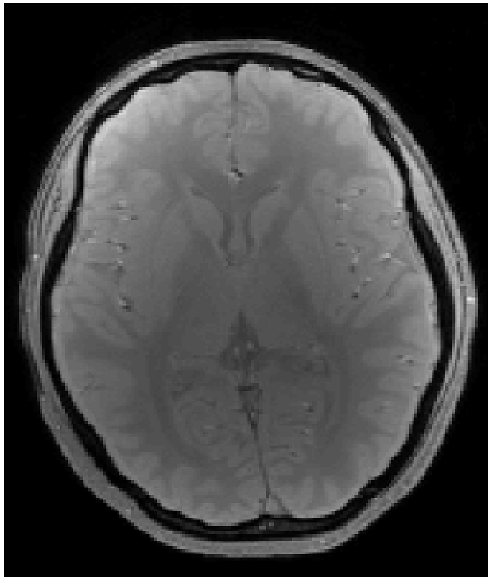
T1 differences between brain tissues yield image contrast in anatomical imaging

PD contrast – long TR

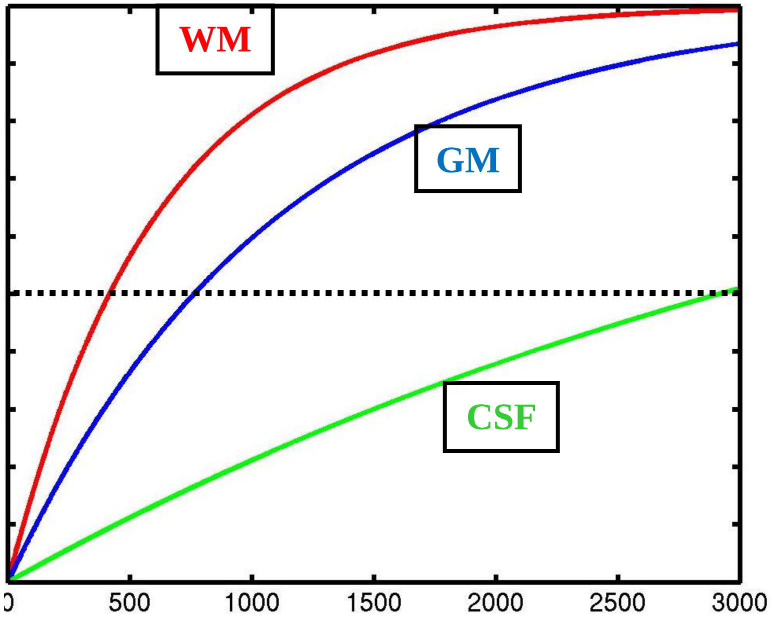
- TR >> T1:
  - All tissues fully relax
    ⟶ No T1w contrast
  - Image contrast: water density
    ⟶ PDw contrast



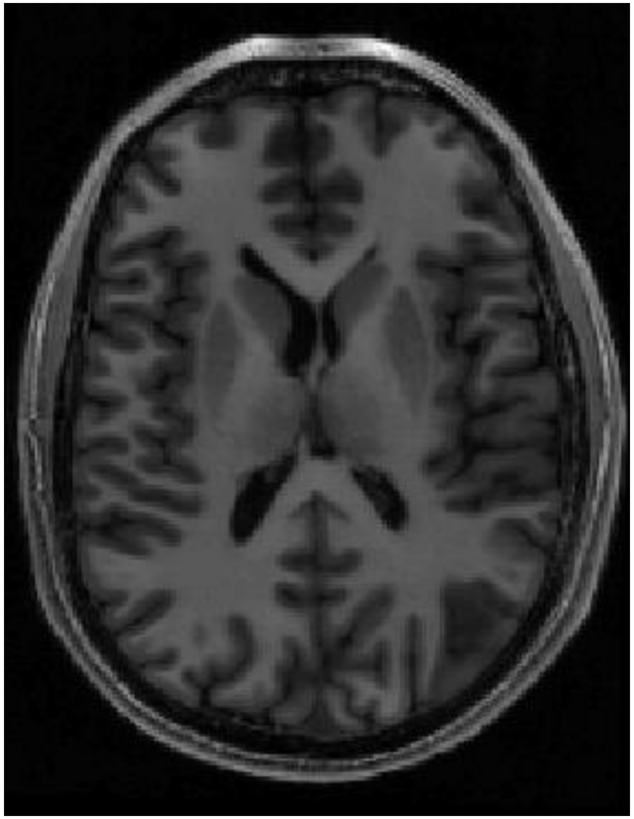
- Inconveniences:
  - Very time consuming
  - Fairly poor GM/WM contrast



T1 contrast – short TR



TR<<T1

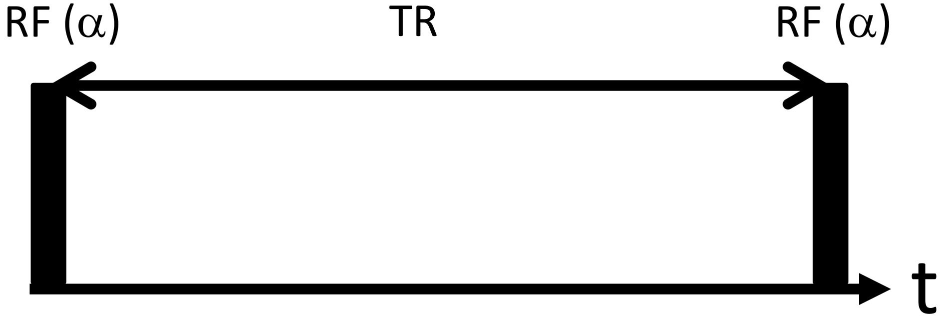


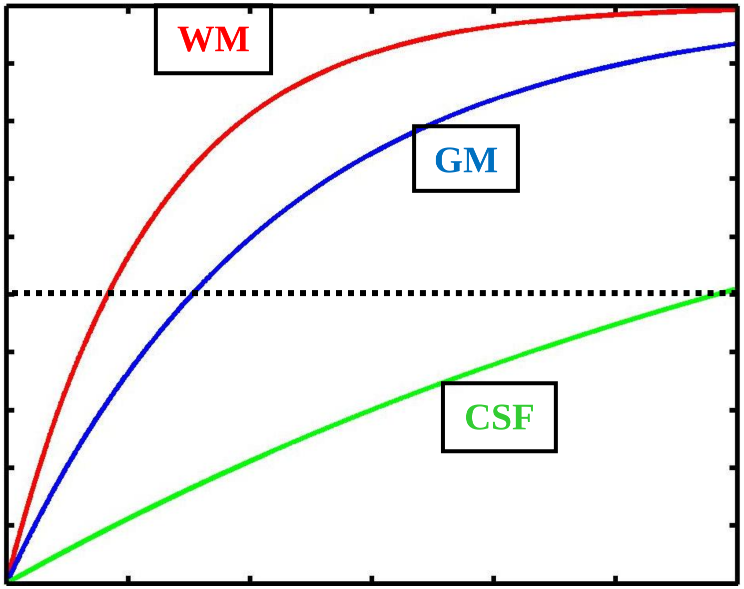
Optimal GM/WM contrast

Generally preferred for anatomical imaging

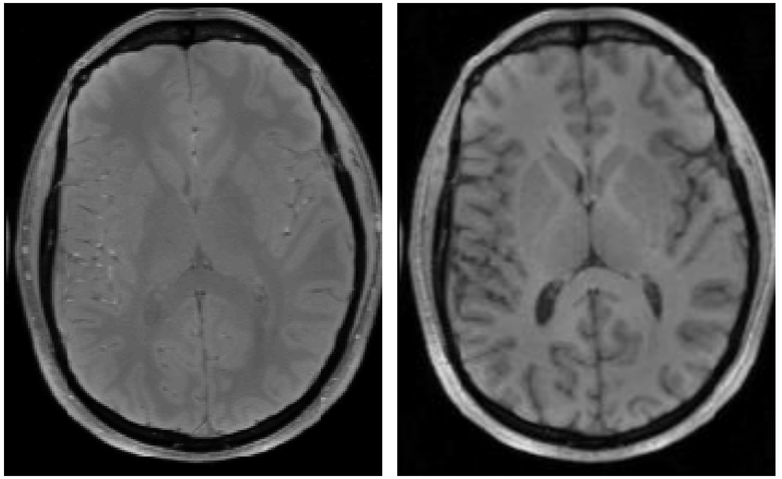
Frahm J. et al. MRM 1986

T1 contrast – short TR





TR=20ms



α=6°
PDw

α=20°
T1w

At short TR, image contrast depends on nominal flip angle of RF excitation

Anatomical sequences

- FLASH

  Frahm J. et al. MRM 1986

- Inversion Recovery (time consuming)

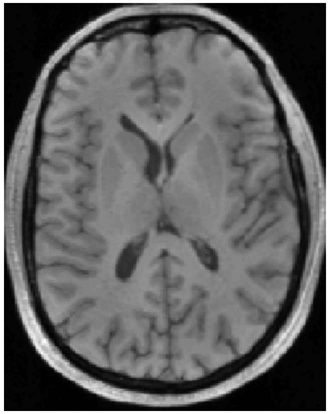
- MPRAGE

  Mugler & Brookeman MRM 1990; Mugler & Brookeman JMRI 1991 ; Look D.C., Locker D.R., Rev. Sci. Instrum, 1970 ;

- MDEFT

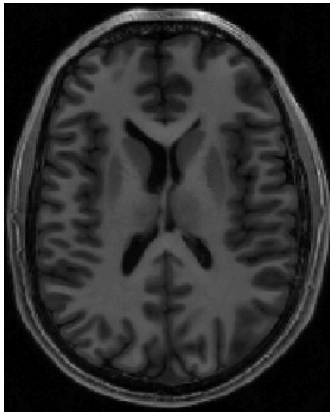
  Deichmann R. et al Neuroimage 2006

FLASH



FLASH: ~6-7mins

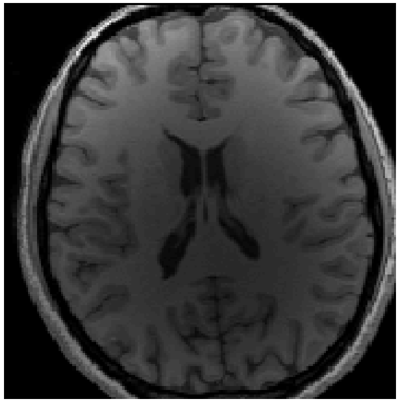
MDEFT



MDEFT:~12mins

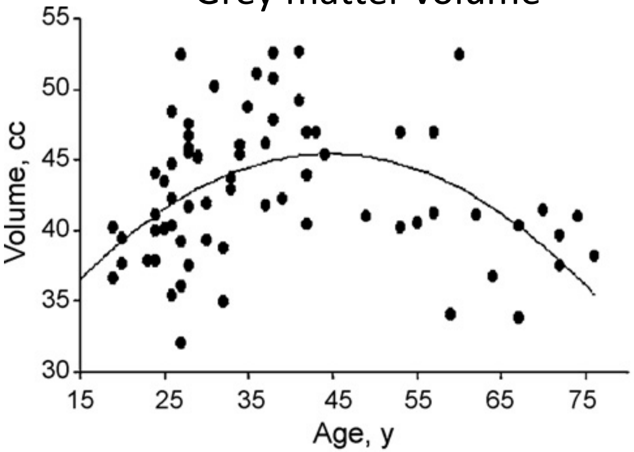
Standard anatomical imaging applications

Anatomical images yields estimates of **grey matter volume**



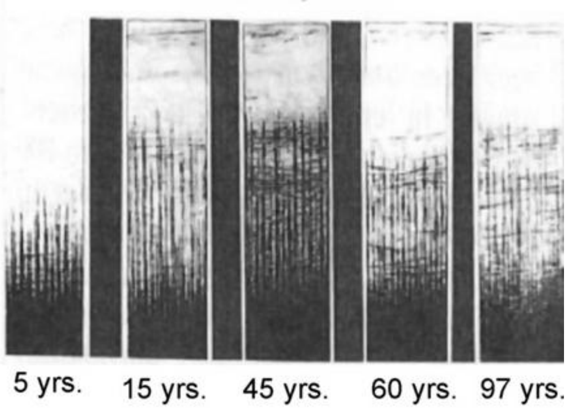
Ashburner & Friston Neuroimage 2000;

Grey matter volume



Frontal lobe white matter with age.

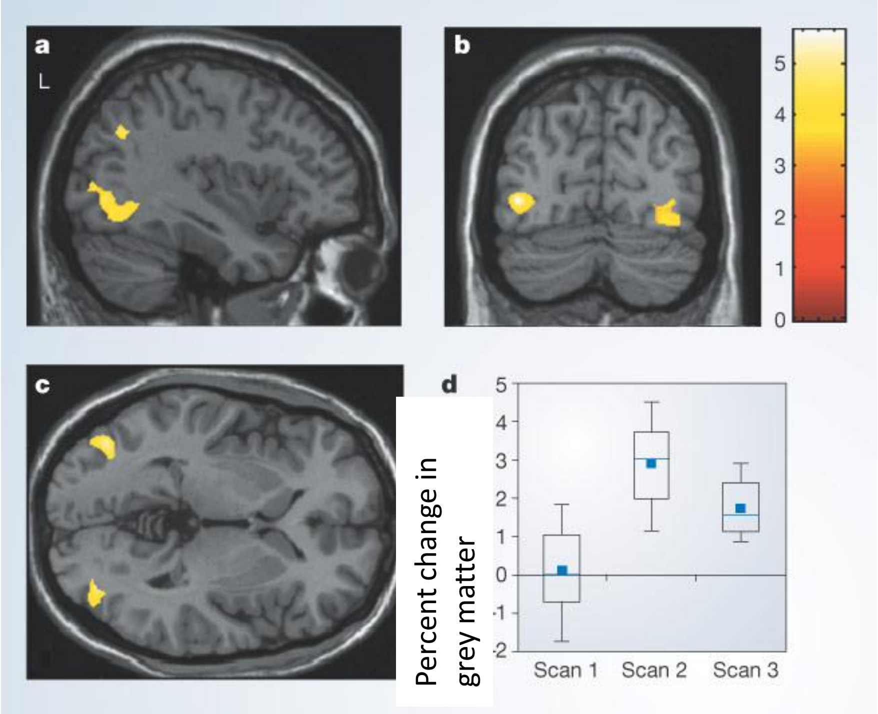
Intracortical myelination



Bartzokis G Neurobiol. Aging 2011

Standard anatomical imaging applications

Transient changes in brain structure due to juggling

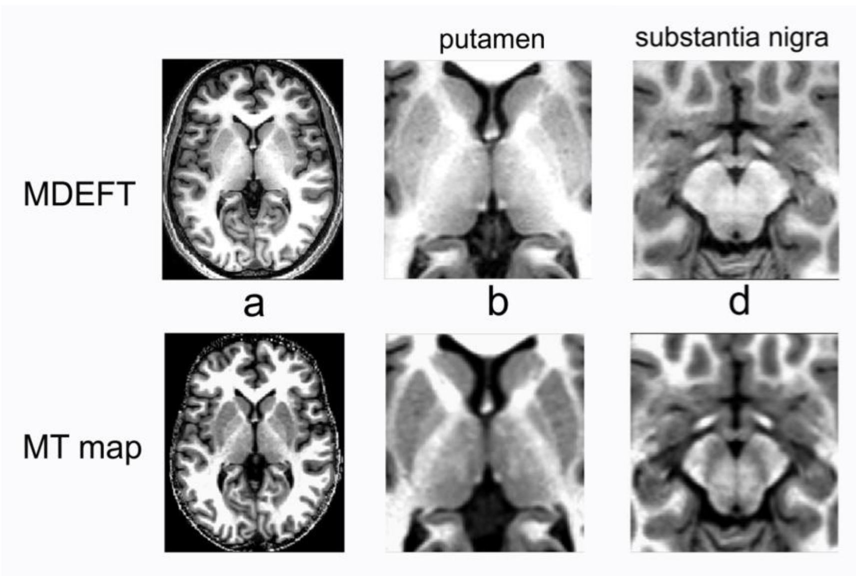


Standard anatomical imaging allows insight into brain plasticity

Draganski B et al. Nature 2004

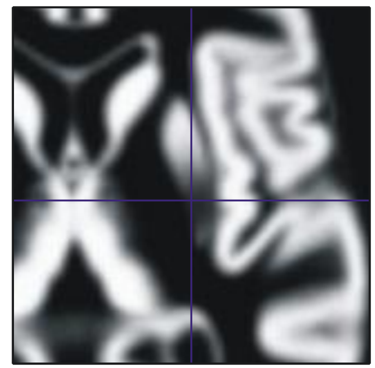
Improved morphometry: MT based VBM

Image contrast

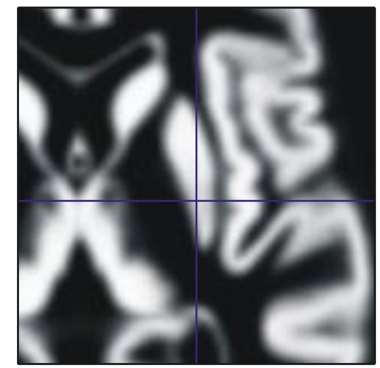


Grey matter volumes

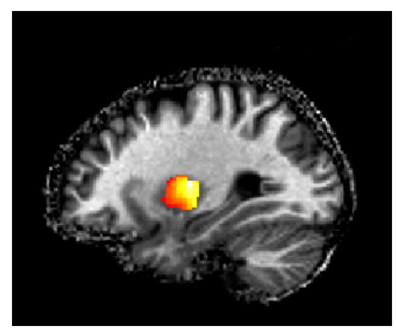
MDEFT

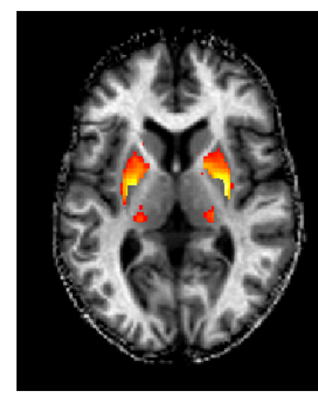


MT



MT > MDEFT



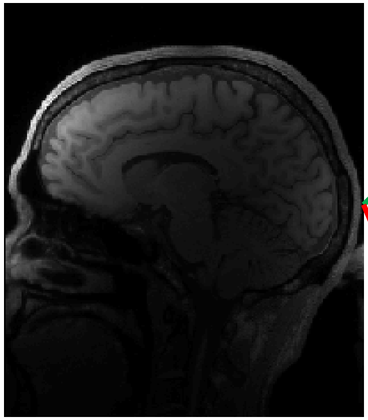


Enhanced image contrast yields improved grey matter volume estimates

Helms et al., Neuroimage 2009

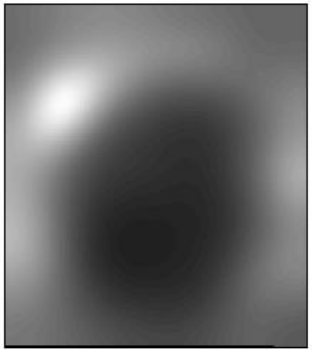
Standard limitations
Spatially-varying bias

Standard T1w image

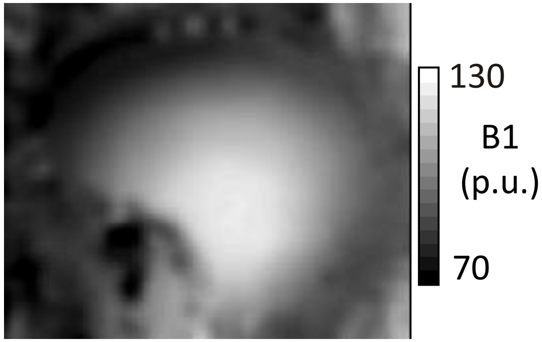


receive bias

Receive head coils with spatially varying sensitivities



transmit bias

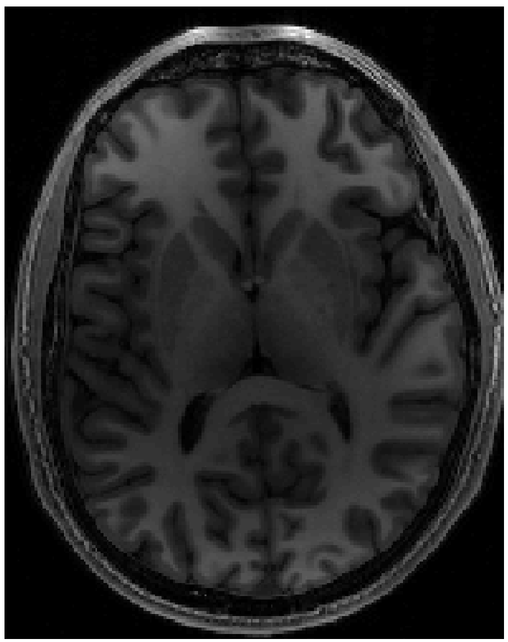


Non uniform RF excitation:

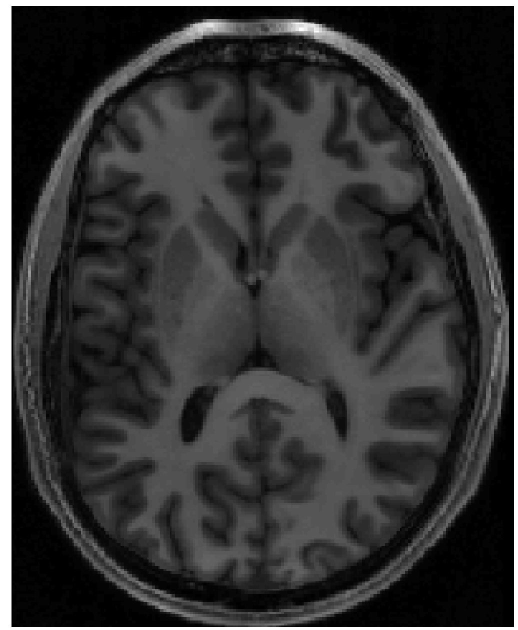
$\alpha = B1 x \alpha_{nom}$

Standard limitations
receive bias

Original image



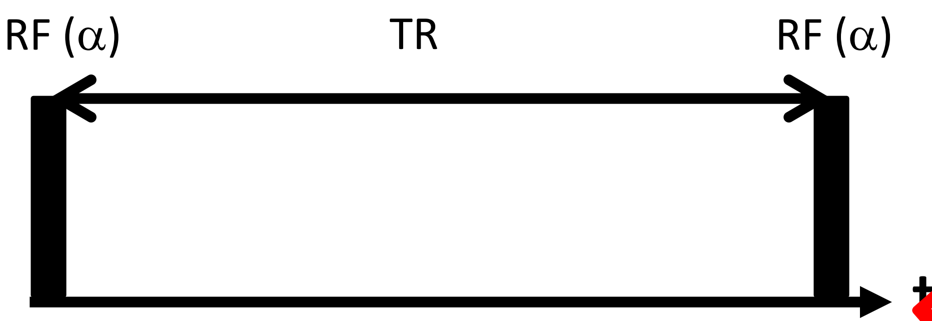
Corrected image



Receive bias corrected by bias field correction of SPM 's unified segmentation

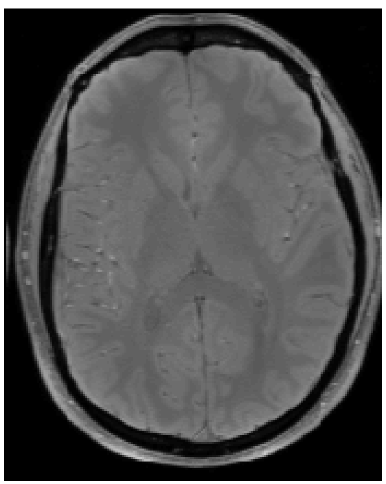
Ashburner J., Friston K., Neuroimage, 2005

T1 contrast – short TR



REMINDER

TR=20ms



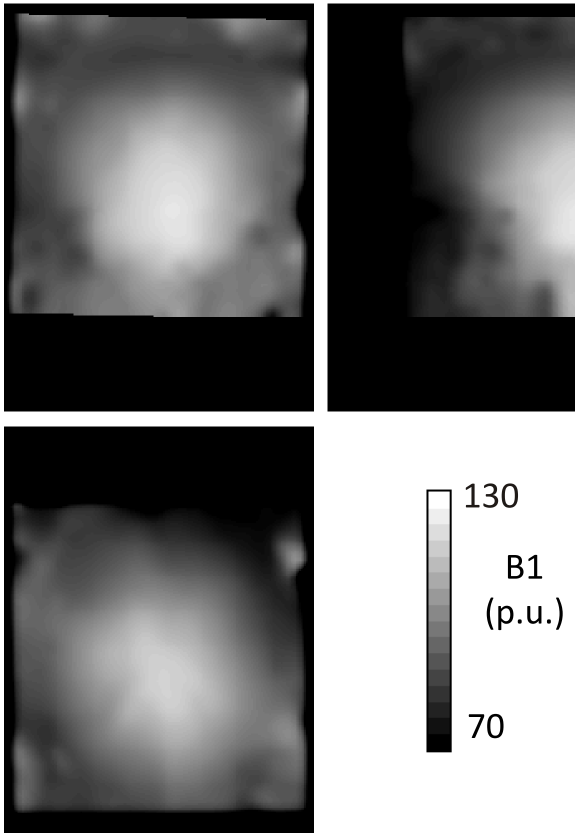
α=6°
PDw

α=20°
T1w

Nominal value of RF excitation affects image contrast

Frahm J. et al. MRM 1986

Standard limitations transmit bias



Non uniform RF excitation:

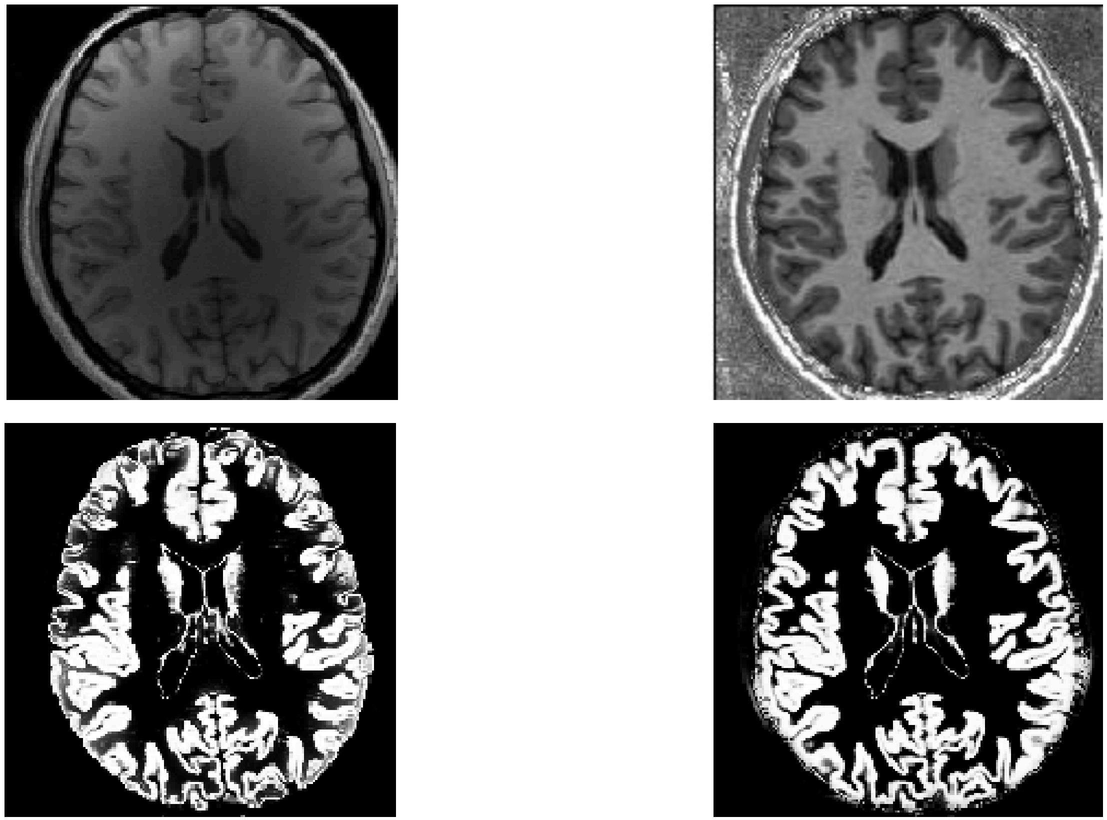
$$\alpha = B1 x \alpha_{nom}$$

- Non uniform RF excitation leads to inhomogeneous **contrast over the image**
- Cannot be corrected at post-processing
- Map of B1 field must be acquired in-vivo

Lutti A. et al MRM 2010, Lutti A. et al PONE 2012

Standard imaging limitations
transmit bias

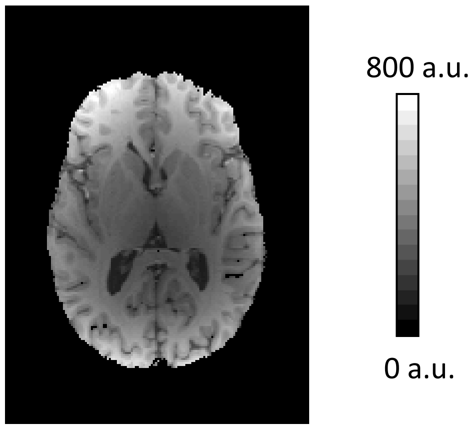
Standard T1w image | Bias-free image



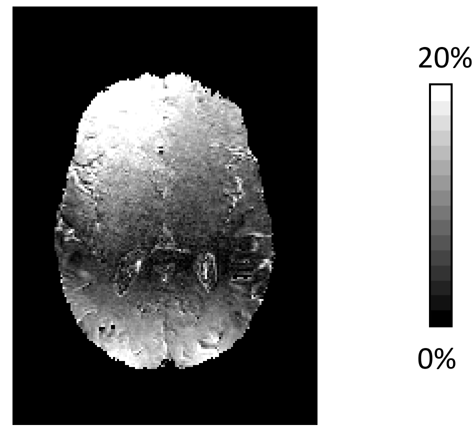
Contrast bias affect **grey matter volume estimates**

Standard imaging limitations comparability

Mean



Inter-site variance

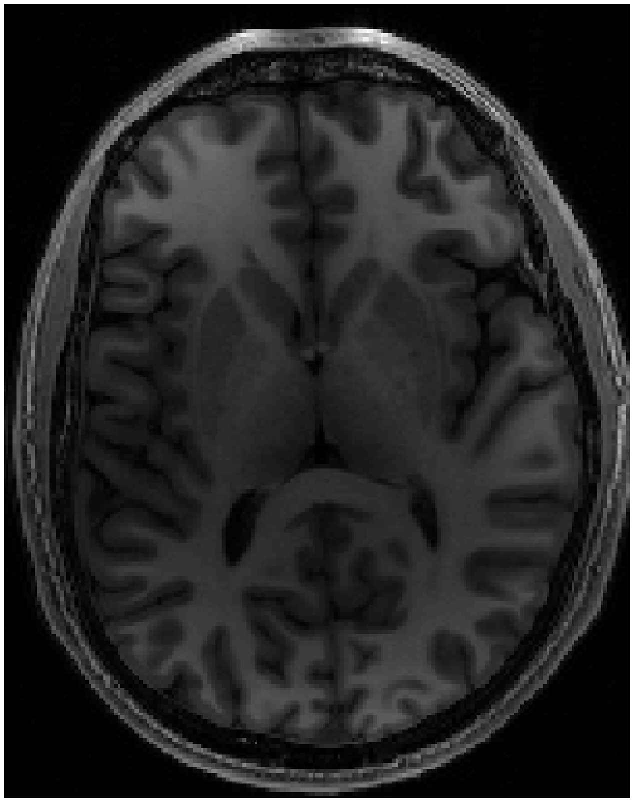


High variability across multiple scans – low comparability

Low sensitivity in cross-sectional/longitudidal studies

Weiskopf N. et al Front. Neurosci. 2013

Standard imaging limitations - summary



Inaccuracy
Hardware bias

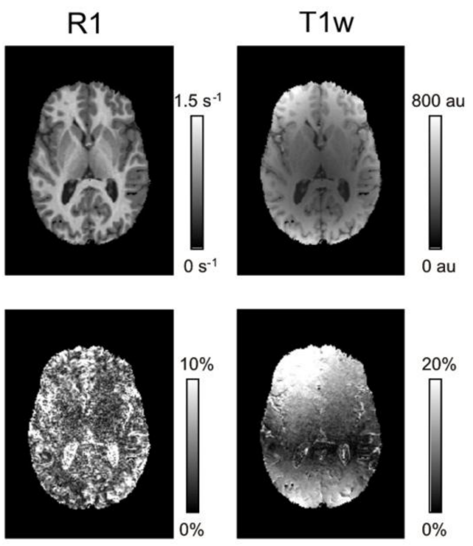
Comparability
Varies with imaging sequence and across scans

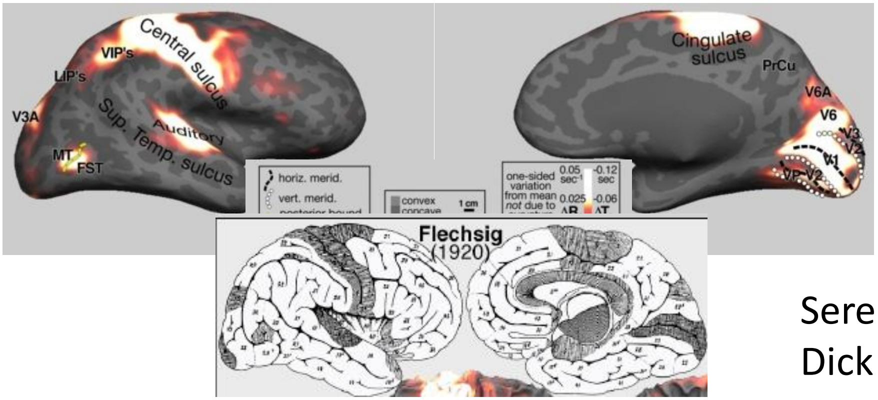
Interpretability
Mixed effect of multiple MR parameters

Qualitative
Arbitrary units. No insight into microarchitecture

Quantitative mapping - motivations

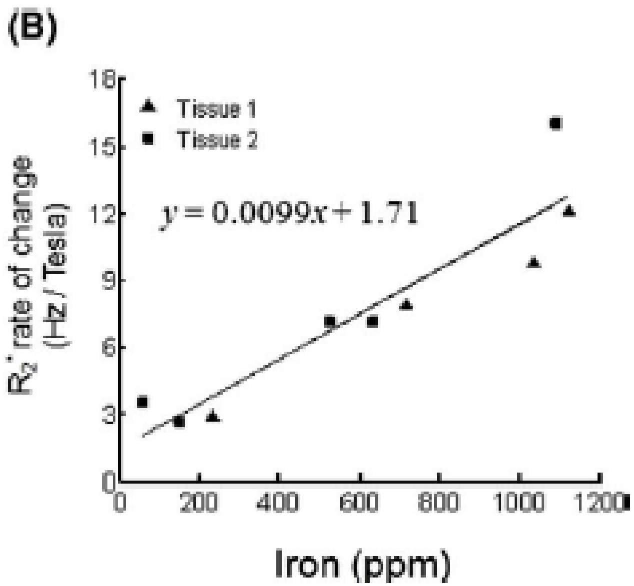
- Quantitative MRI provides **quantitative** and **specific biomarkers** of brain tissue properties (myelination, iron concentration, water concentration,...)
- **No bias** between brain areas (transmit/receive field)
- Data quantitatively comparable across scanners. **Optimal sensitivity** in longitudinal and multi-centre studies



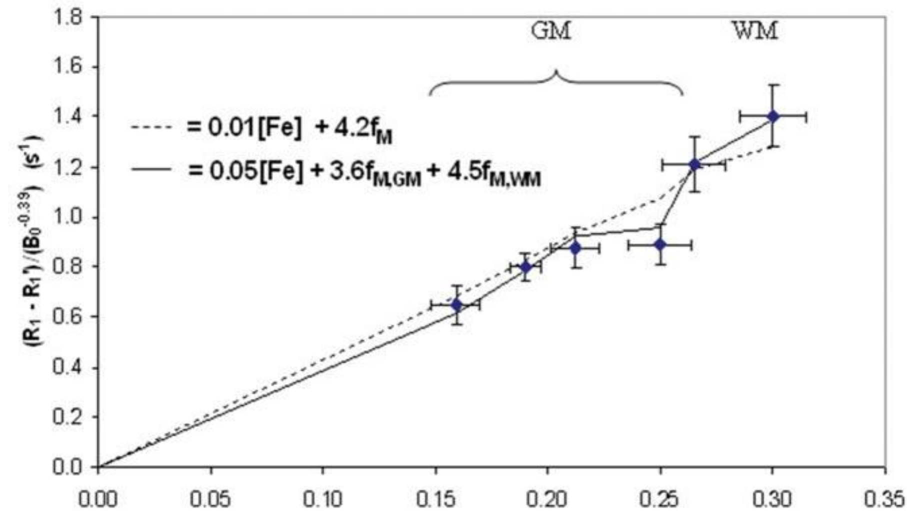


Sereno M.I. et al., Cereb. Cortex 2013;
Dick F. et al J. Neurosci. 2012

Quantitative mapping - motivations



Yao B. et al. NI 2009

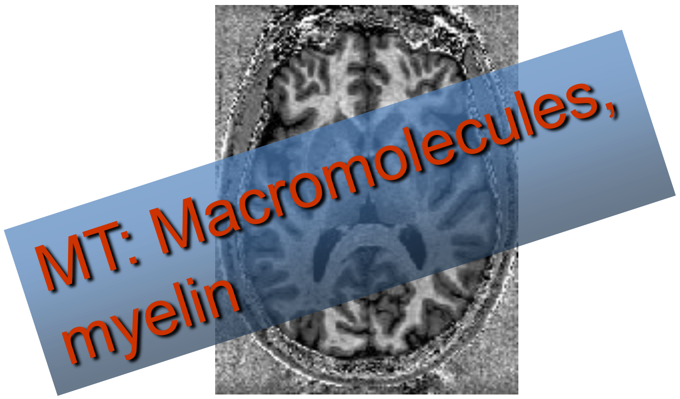


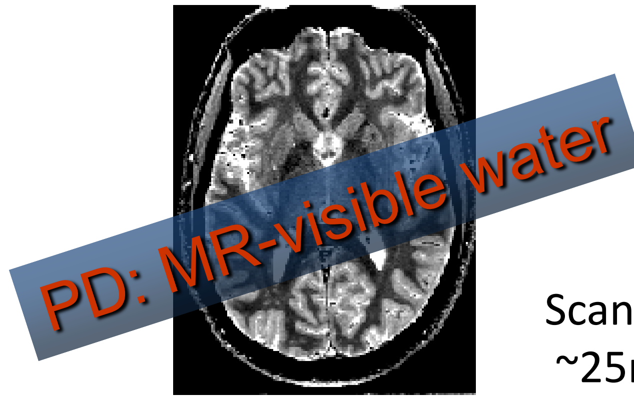
Macromolecular concentration

Rooney W.D. et al MRM 2007

Quantitative estimates of MRI parameters are **biomarkers of tissue properties**

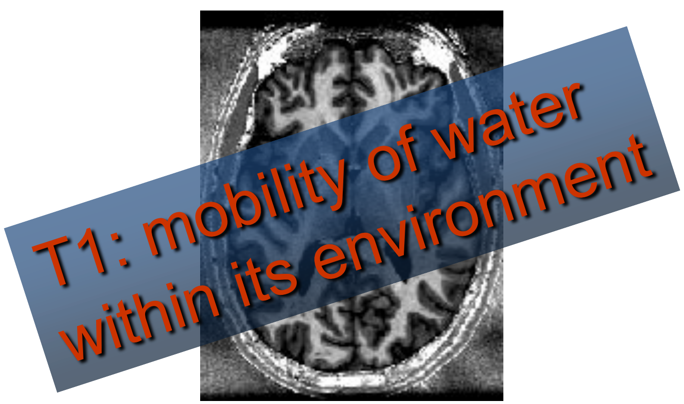
MPM protocol for quantitative mapping

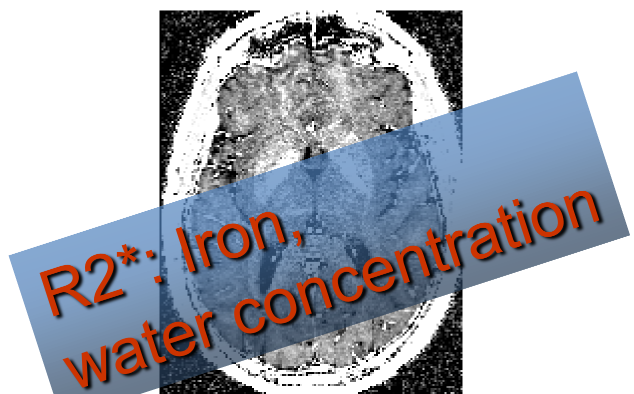




Scan time:
~25min (1mm^3 resolution)

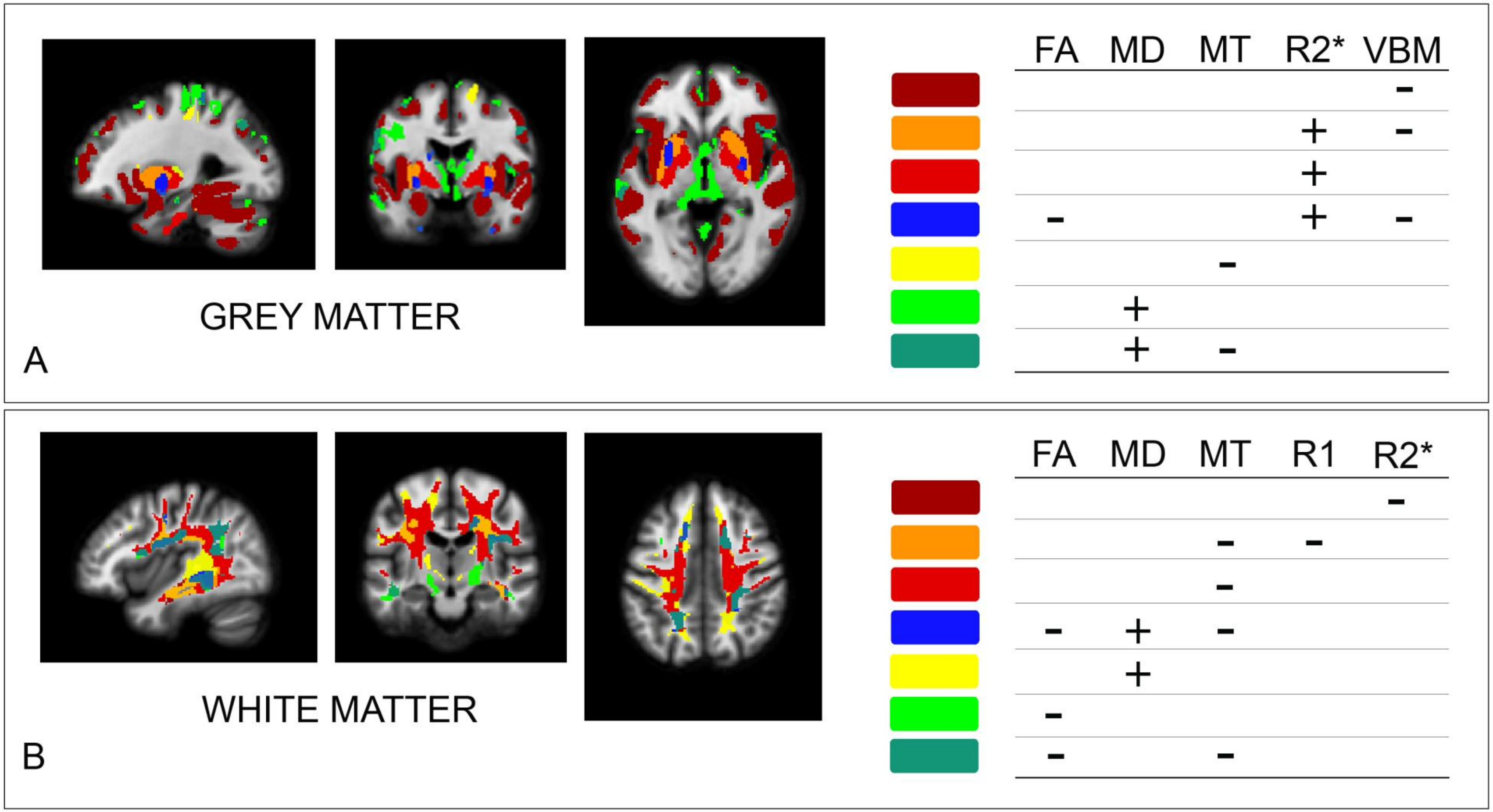
~35min (800um^3 resolution)





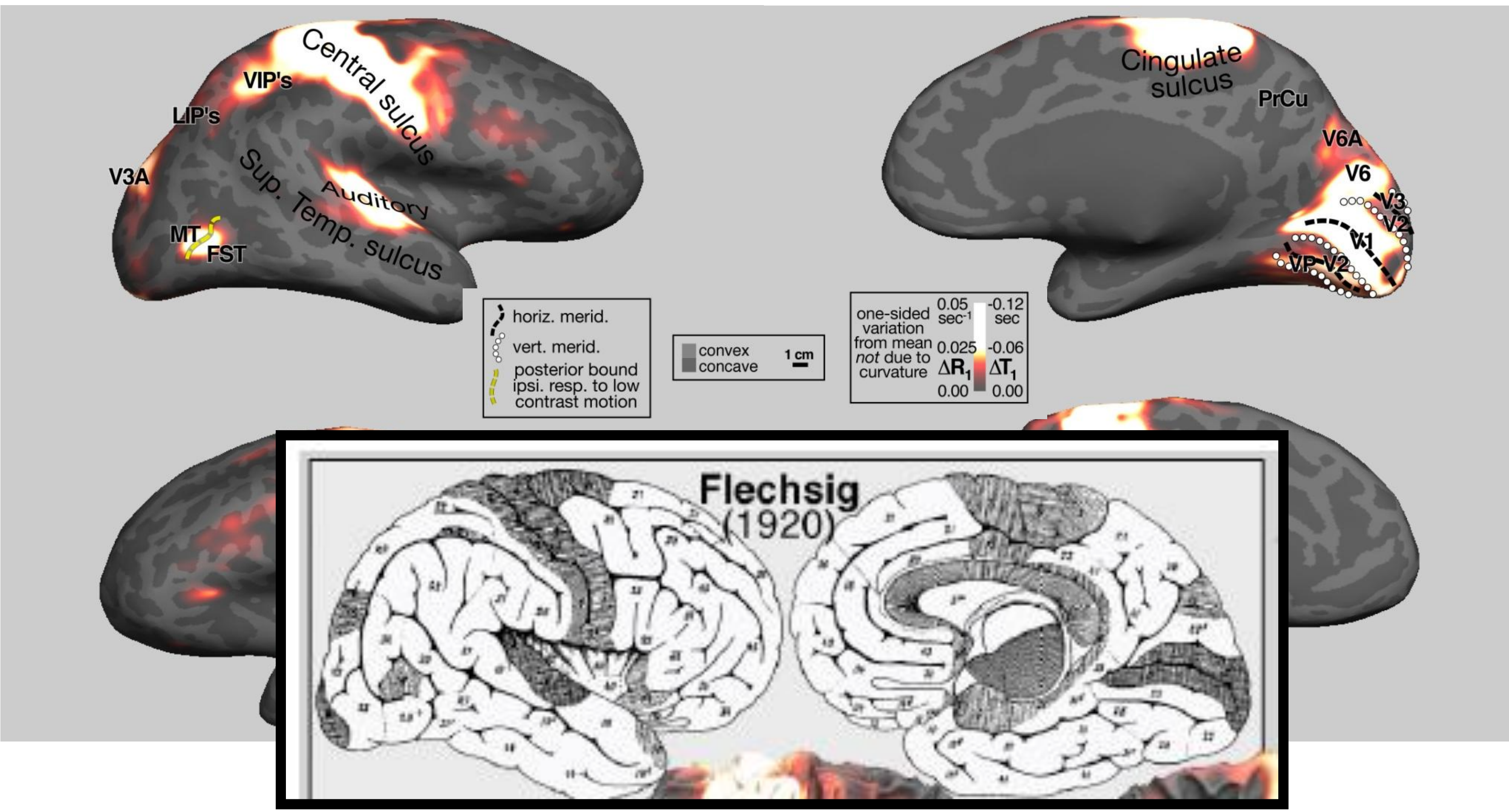
Helms G., et al MRM 2008; Helms G., et al MRM 2009;
Lutti A. et al MRM 2010, Lutti A. et al PONE 2012;

VBQ: fingerprint of tissue changes in ageing



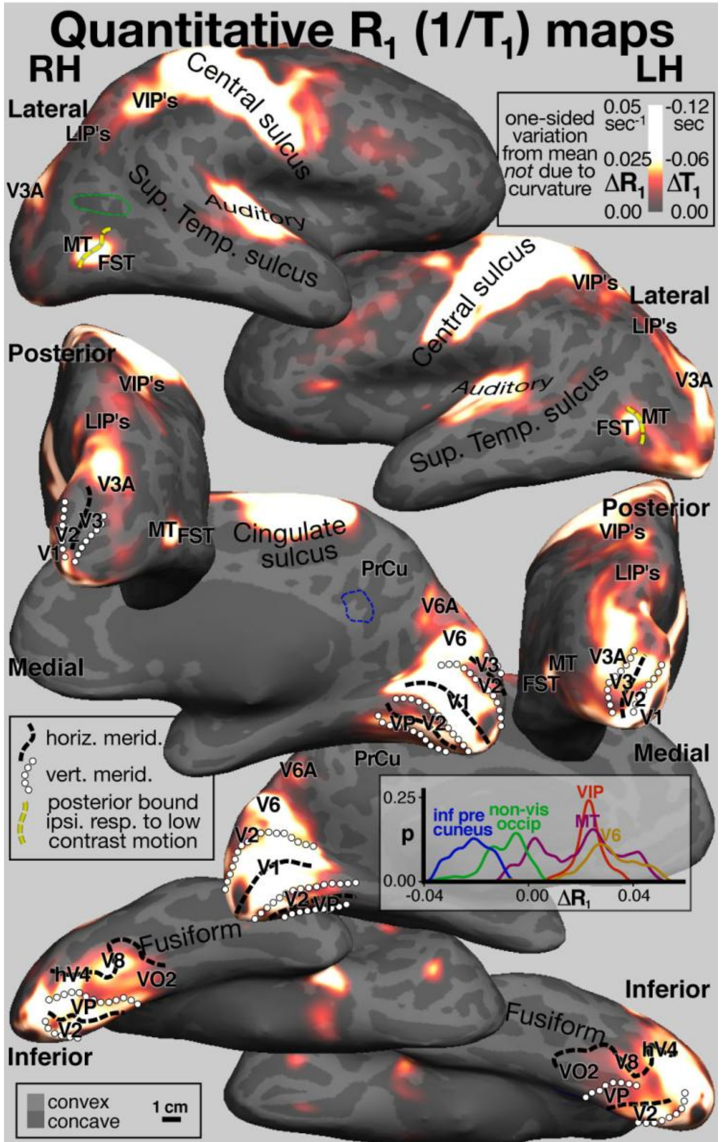
Draganski et al., Neuroimage 2011

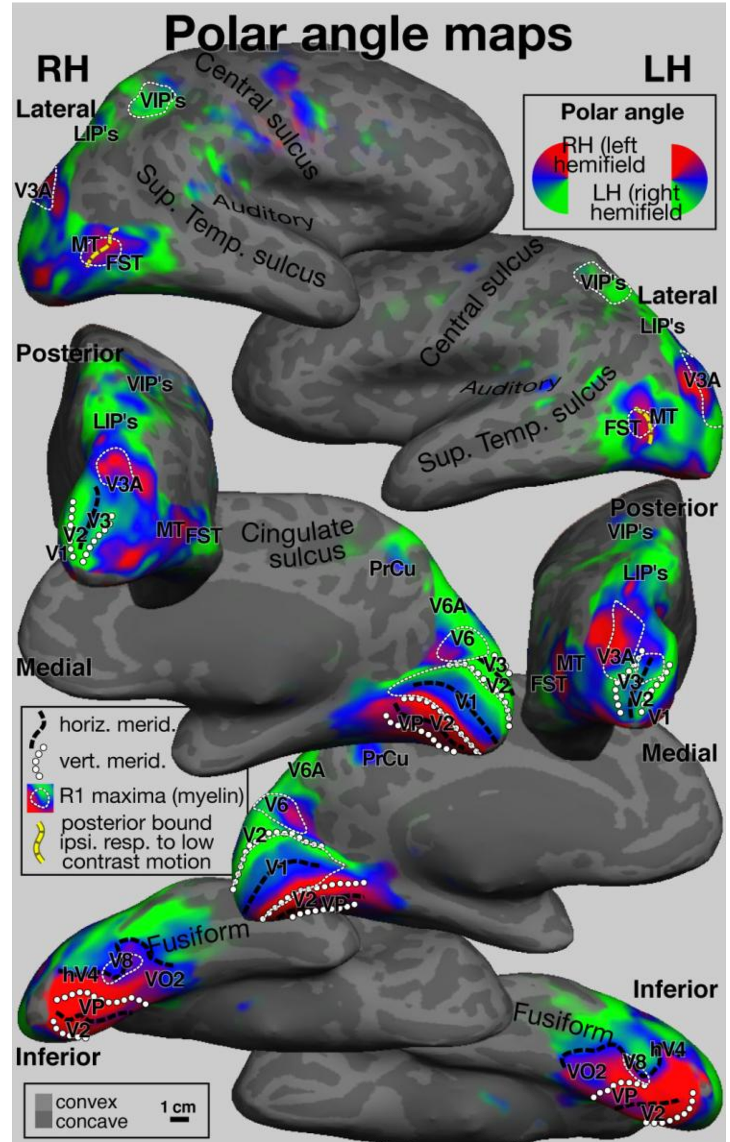
Myelin mapping: towards in-vivo histology



Sereno M.I. et al., Cereb. Cortex 2013; Dick F. et al. J.Neurosci. 2012

Structure/function relationship





Sereno M.I. et al., Cereb. Cortex 2013; Dick F. et al. J.Neurosci. 2012

Anatomical imaging - summary

Standard anatomical imaging

- Provides estimates of grey matter volumes. Study of brain plasticity, neurodegeneration,…
- Limited accuracy, sensitivity and specificity.

Quantitative MRI

- Provides **quantitative** estimates of MRI parameters
- Enhanced **accuracy**, **sensitivity**, **specificity**
- Provides biomarkers of tissue microstructure - insight into biological processes underlying tissue change.

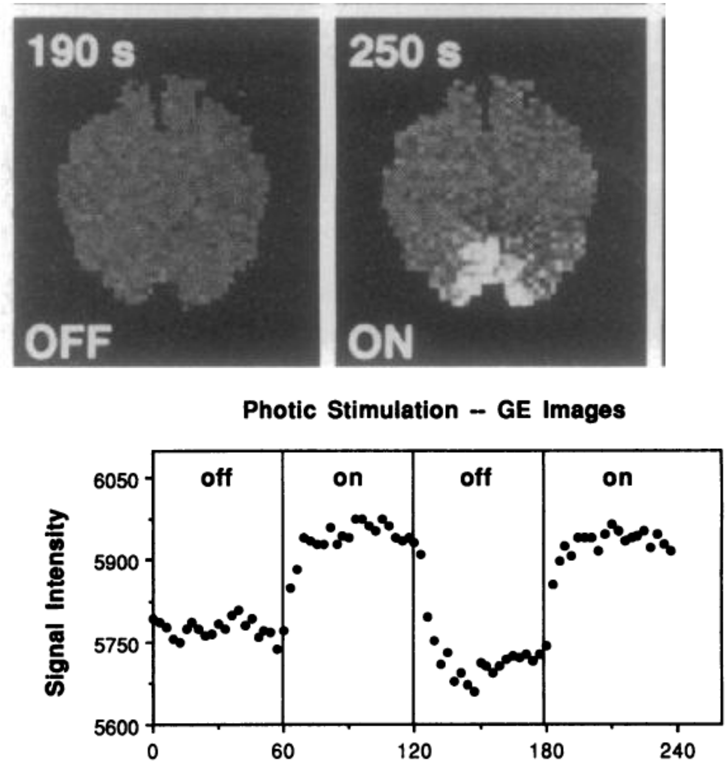
Outline

- **General principals**
  - Origin of the signal
  - RF excitation
  - Relaxation (T1, T2, …)
- **Anatomical imaging**
  - Image contrast
  - Standard acquisition methods
  - Advanced acquisition methods
- **Functional imaging**
  - BOLD effect
  - Limitations of fMRI acquisitions
  - Advanced methods

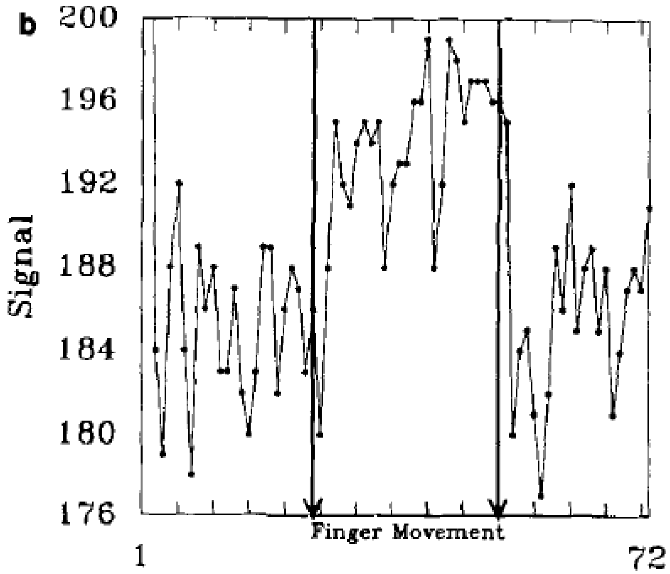
Blood Oxygen Level Dependent (BOLD) effect

- Ogawa et al., 1990: “static” BOLD effect in rat brain
- Kwong et al., Bandettini et al., Ogawa et al., 1992: BOLD fMRI in human

Note: localized changes, delayed/dispersed BOLD response

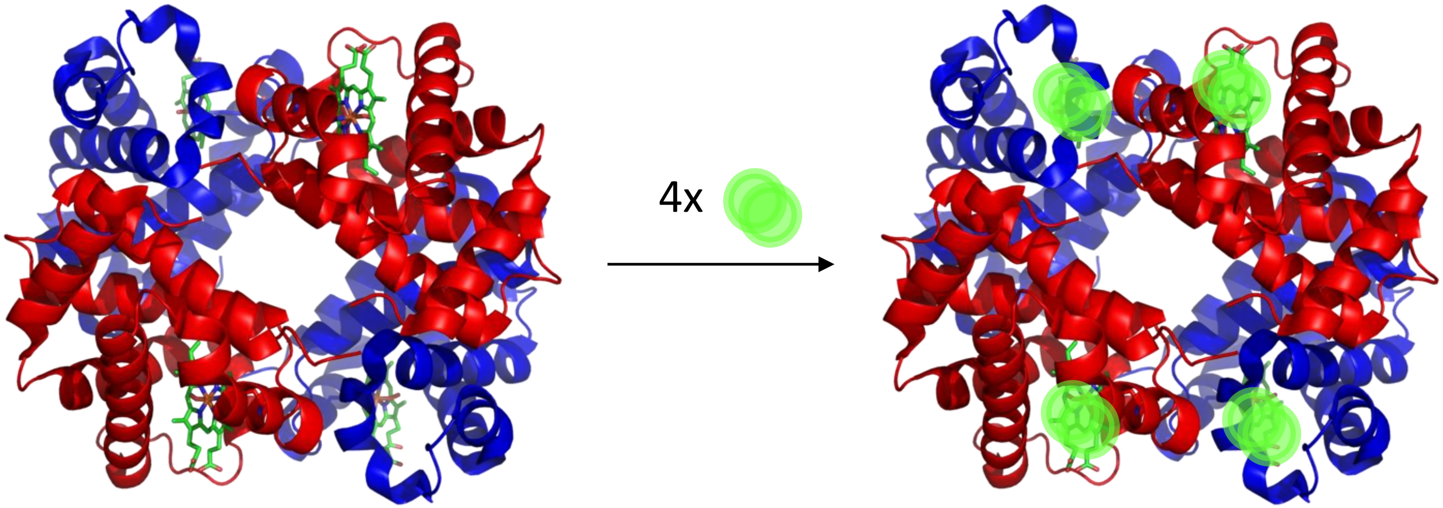


Kwong et al., PNAS 1992



Bandettini et al., MRM 1992

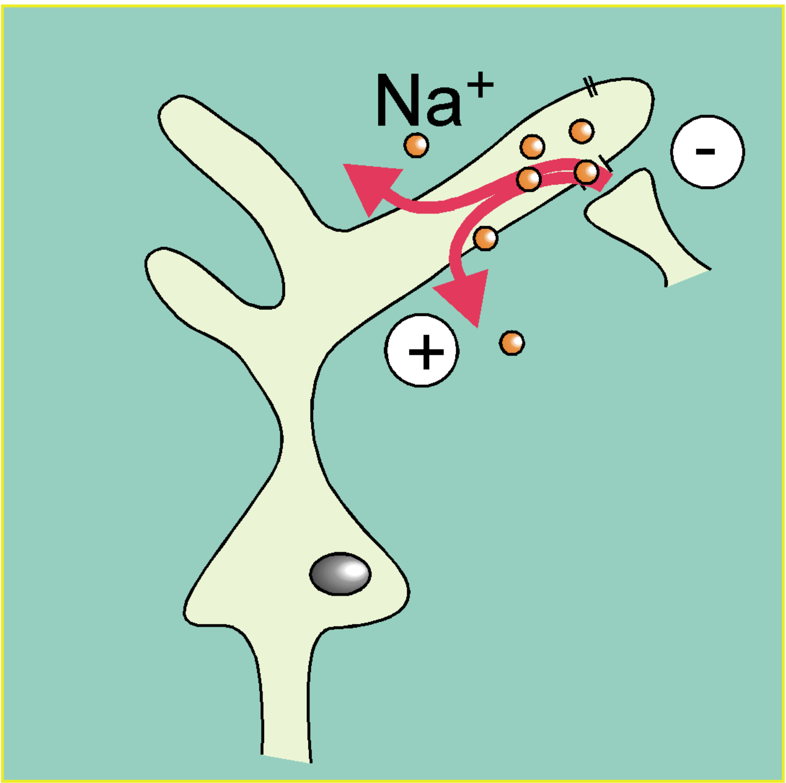
Magnetic susceptibility of hemoglobin



Deoxygenated hemoglobin (Hb)
- paramagnetic
- different to tissue (H_2O)
- Changes local magnetic field and **reduces signal in MRI images**

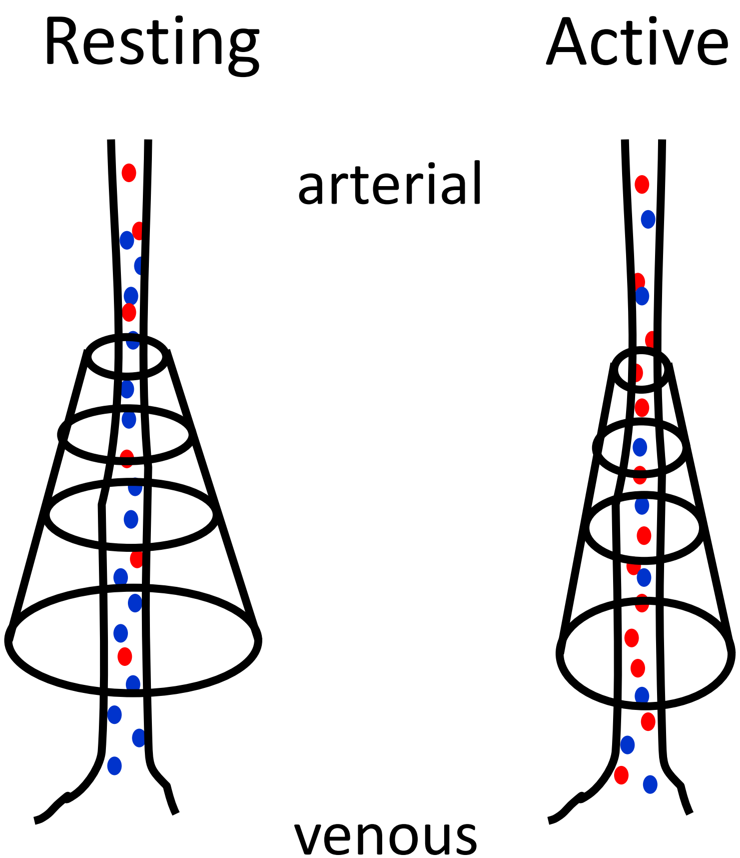
Oxygenated Hb:
- diamagnetic
- same as tissue (H_2O)

BOLD contrast in a nutshell
(Blood Oxygen Level Dependent)

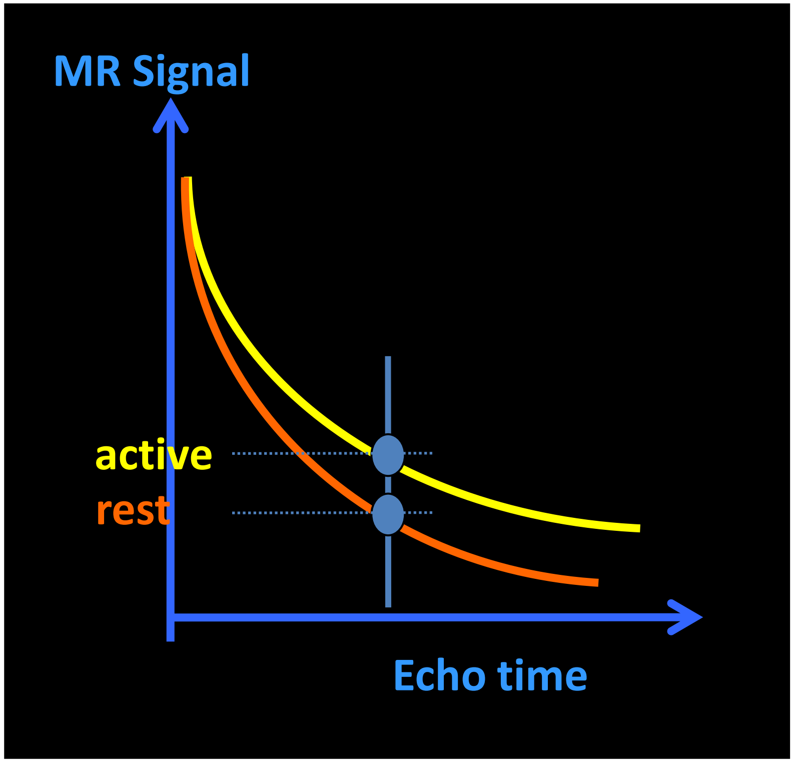


- Synaptic activity increases metabolism
- Increased cerebral blood flow (neurovascular coupling) and oxyhemoglobin concentration

The BOLD effect



Oxygenated / deoxygenated hemoglobin
= endogenous contrast agent



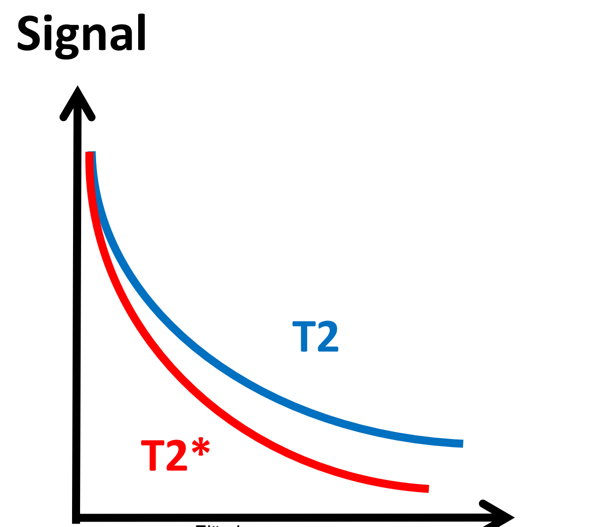
BOLD EFFECT
Change in oxygenated / deoxygenated hemoglobin concentration leads to **detectable signal change**

Functional imaging requirements

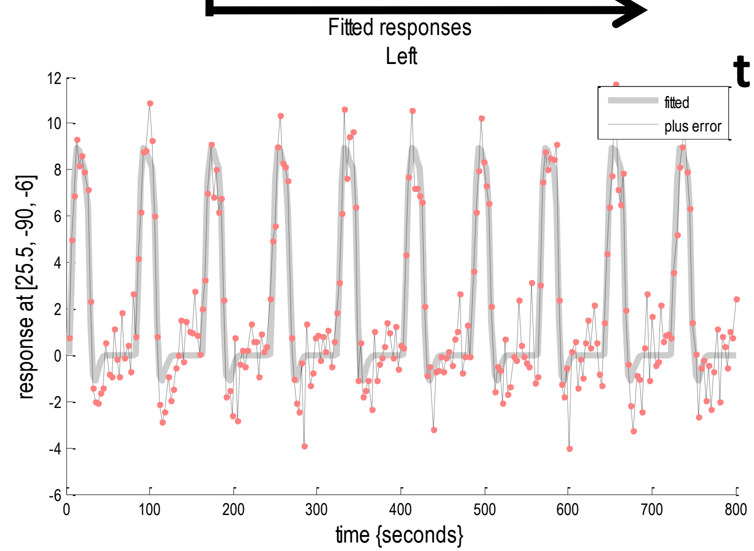
- **Optimal BOLD sensitivity** – T2* weighted contrast

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'}$$

Field inhomogeneities



- **Rapid sampling of BOLD response**
  - Short acquisition time per image volume



Echo-Planar Imaging - EPI

EPI acquisitions: whole 2D plane (*slice*)
acquired following one RF excitation - FAST

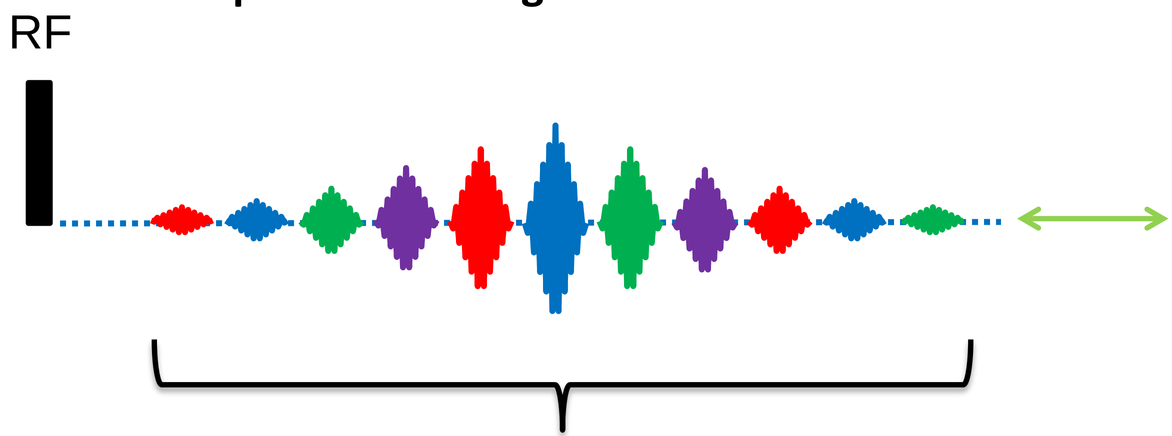
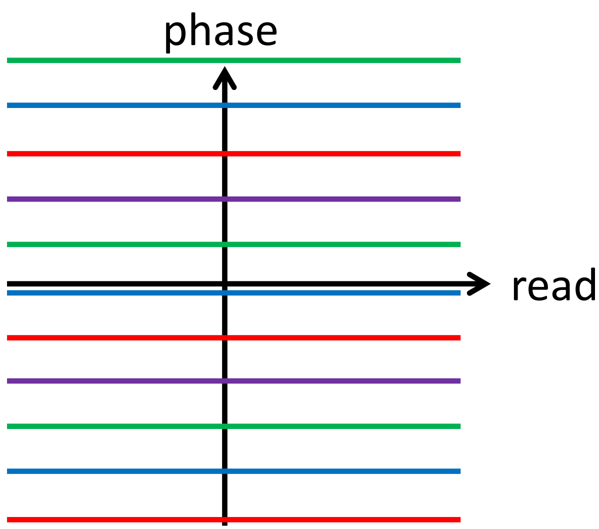


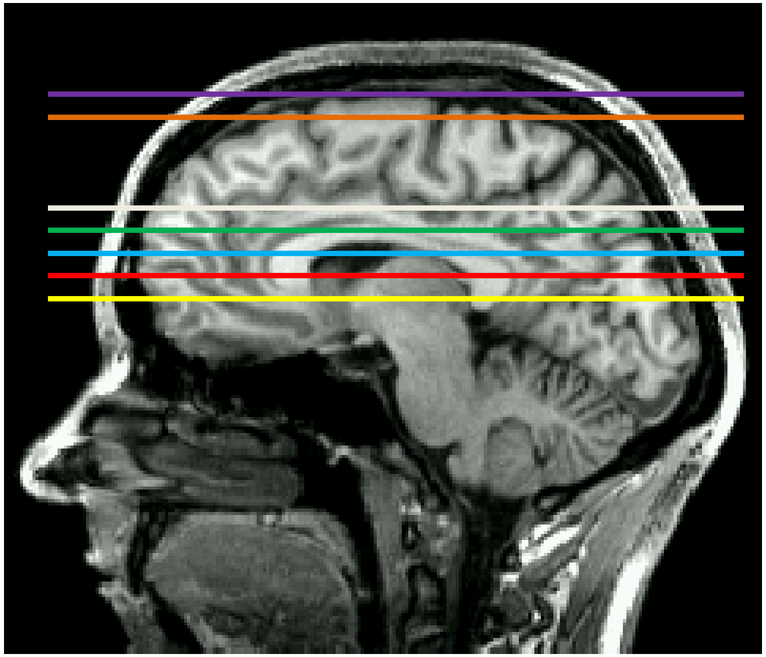
Image acquisition
(*readout*)

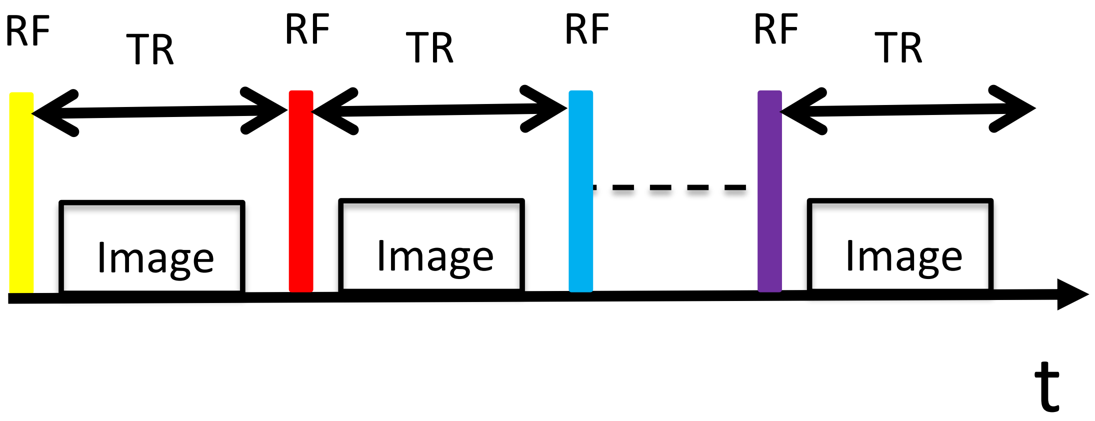


Typical protocol: 64 voxels along read & phase, 3mm resolution

- read direction: 500us per line – **fast**
- phase direction: 500usx64=32ms – **slow** (*low bandwidth*)

Echo-Planar Imaging - EPI





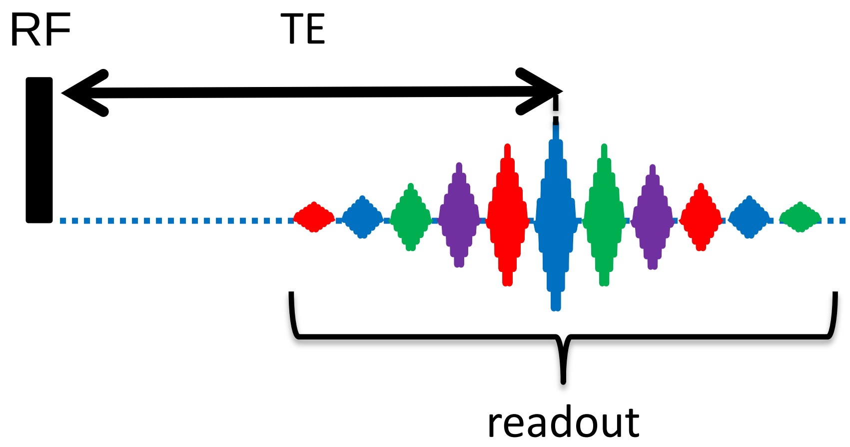
Acquisition time per volume:

$$TR_{volume} = Nslices \times TR$$

Slice ordering: ascending, descending, interleaved

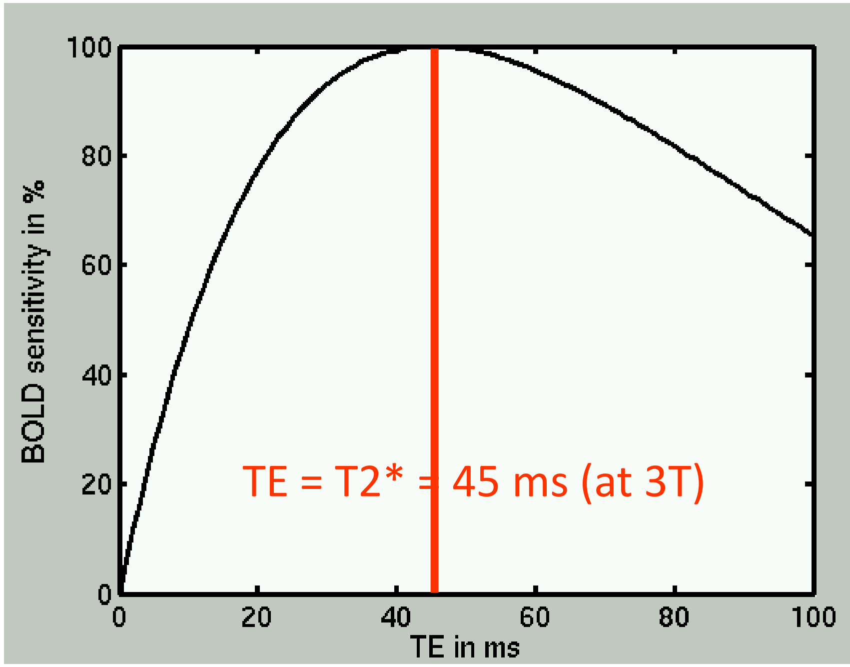
3mm resolution: TR~60ms

Echo-Planar Imaging - EPI



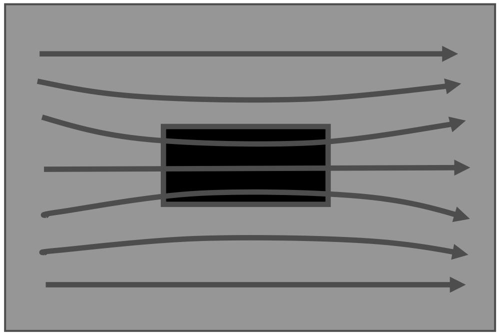
Optimal echo time TE for fMRI

BS(TE) = C · TE · exp(-TE/T2*)

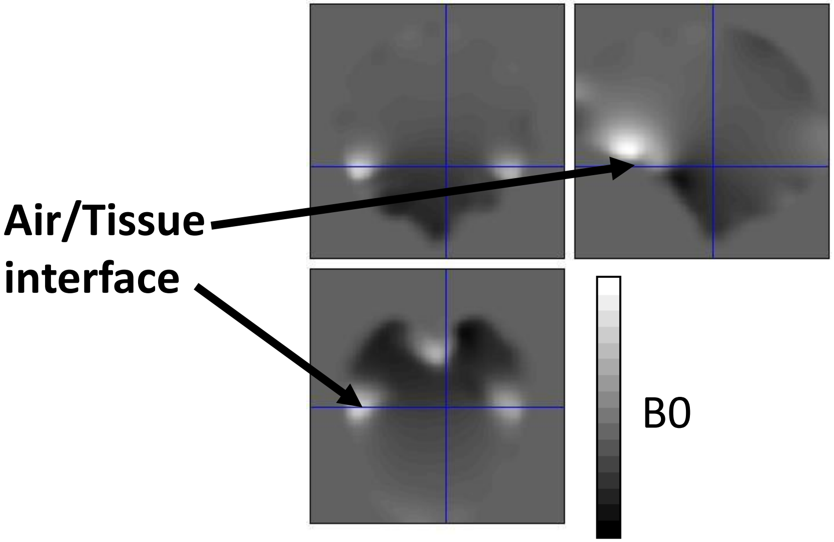


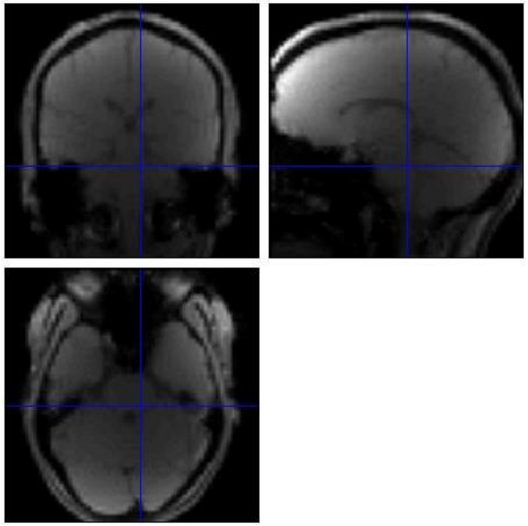
At 3T TE = 30 ms: - Good trade-off between high BOLD sensitivity and low susceptibility-related signal dropout
- Optimal time-efficiency

Echo-Planar Imaging - EPI

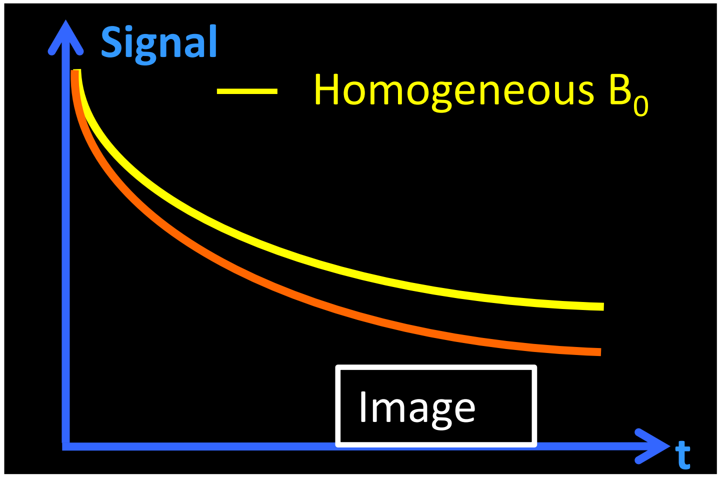


- Variation in magnetic susceptibility distorts the static magnetic field (B0)
- **Strong B0 inhomogeneities at the air/tissue interface lead to artefacts in EPI images**

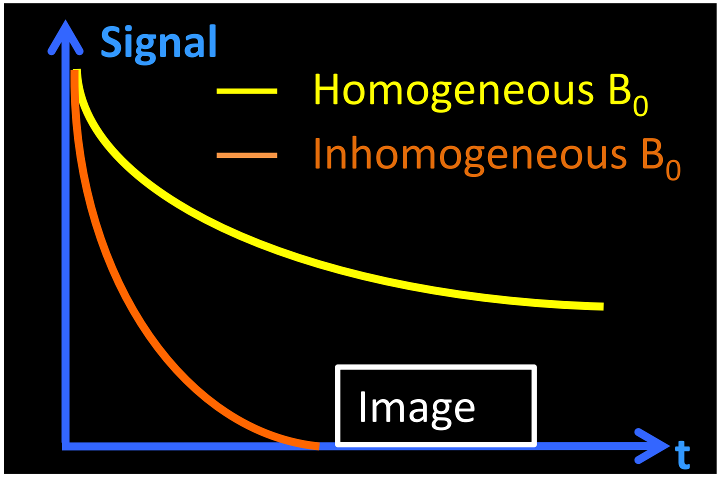




Susceptibility effects in EPI: distortion and dropout



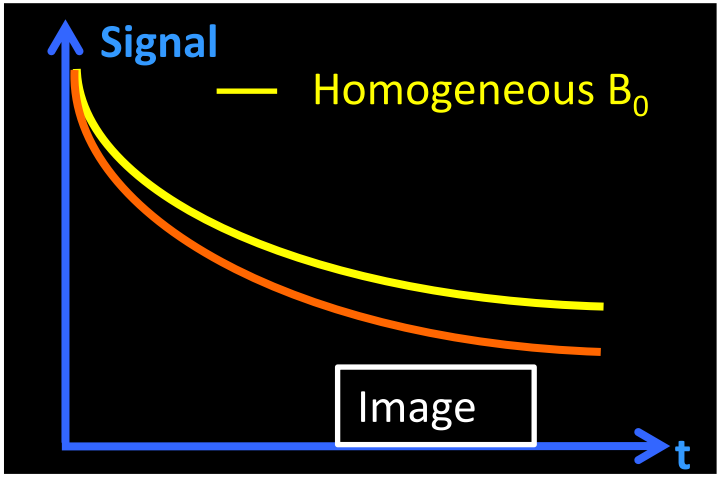
Strong B_0 inhomogeneities



Full signal decay before image acquisition

Signal dropout

Moderate B_0 inhomogeneities

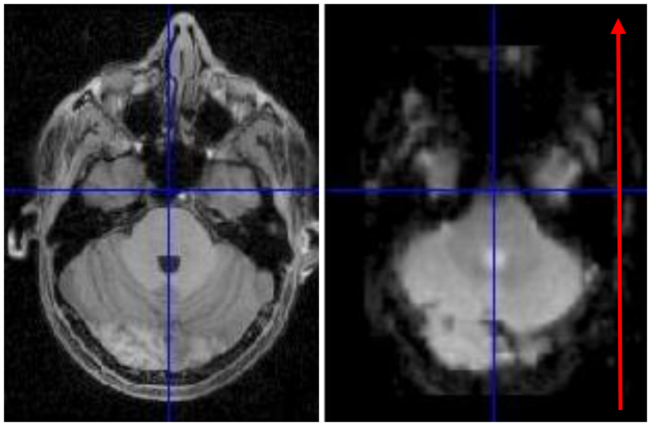


Increased signal decay during image acquisition

Image distortions

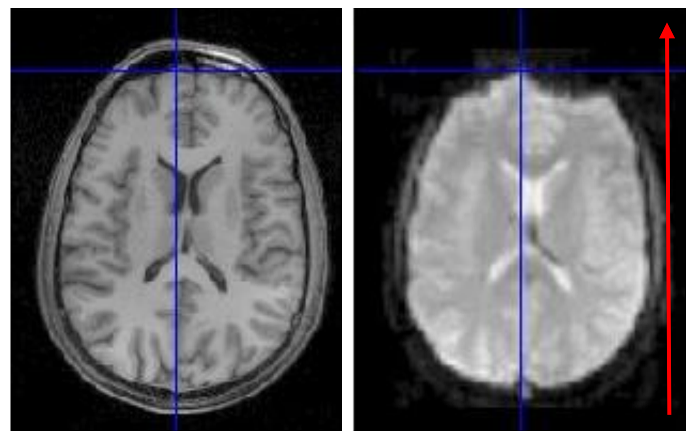
Susceptibility effects in EPI: distortion and dropout

Distortion



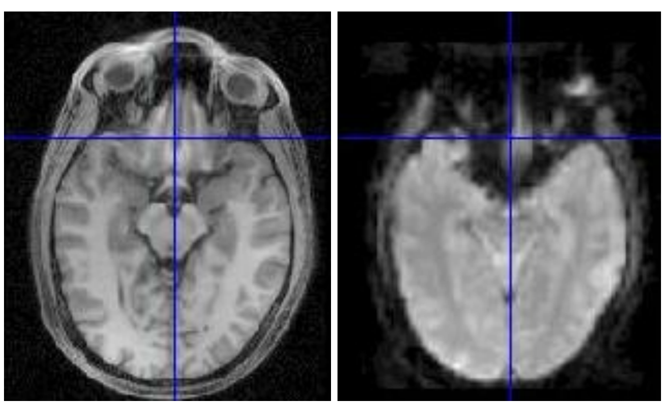
Phase-encode direction

Dropout and distortion



Phase-encode direction

Dropout



EPI distortion correction with field map

Fieldmap toolbox

Mapping of B0 inhomogeneities calculated from 'fielmap data'

B0 field

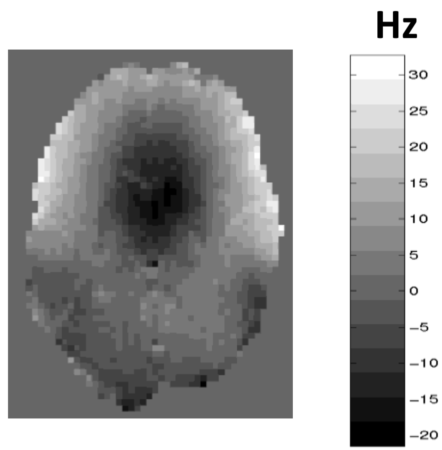
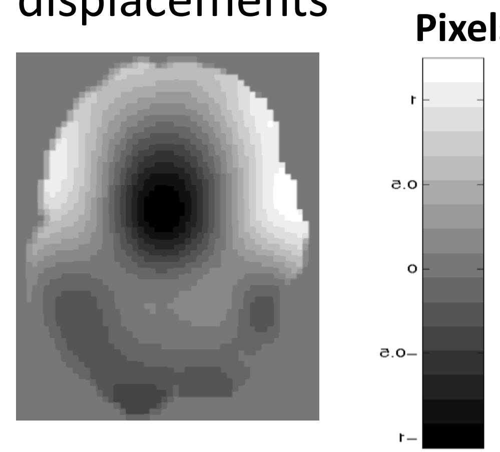


Image processing ➔

Map of voxel displacements



Jezzard and Balaban, 1995, MRM, 34(1);65-73; Hutton et al, 2002, Neuroimage, 16(1);217-240

EPI distortion correction with field map

Fieldmap toolbox

Mapping of B0 inhomogeneities calculated from ‘fielmap data’

B0 field

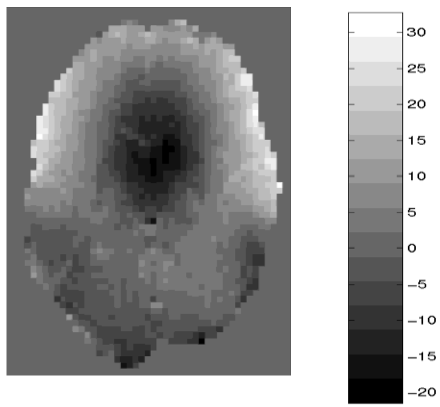
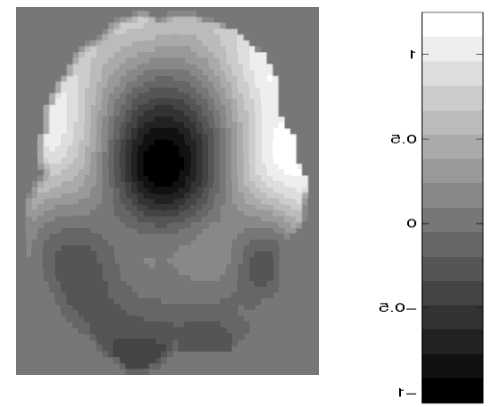


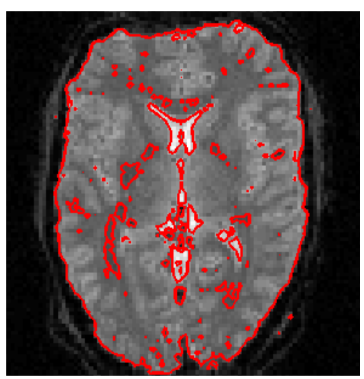
Image processing

➔

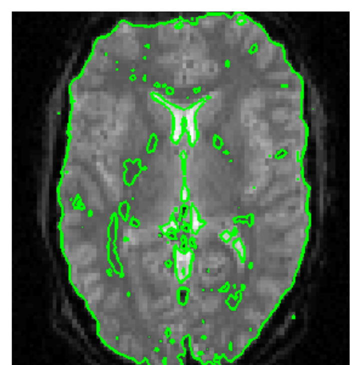
Map of voxel displacements

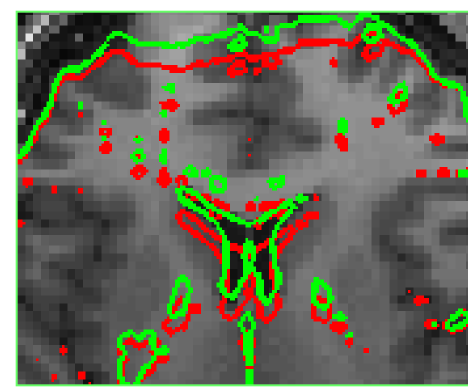


Use pixel shift map to unwarp image ➔



➔

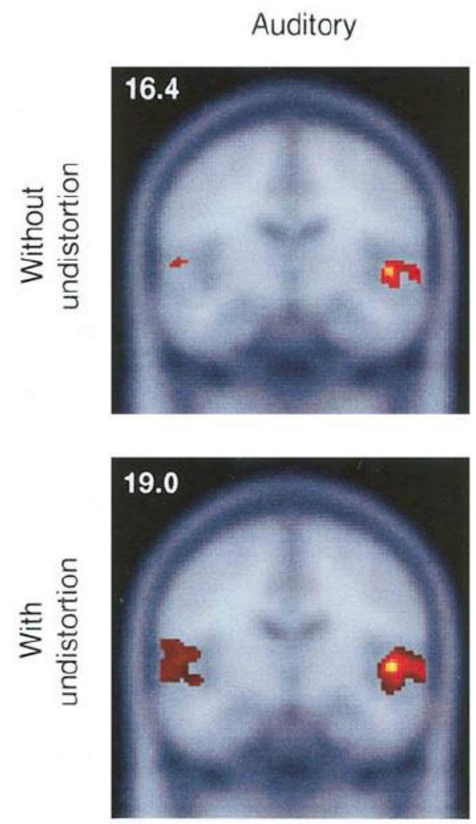




Jezzard and Balaban, 1995, MRM, 34(1);65-73; Hutton et al, 2002, Neuroimage, 16(1);217-240

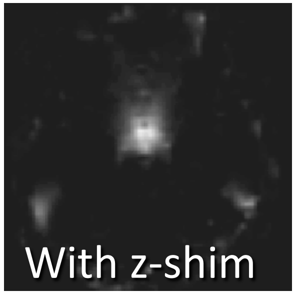
Susceptibility effects in EPI: distortion

- **Distortion**
  - Pixel displacement in **phase-encoding direction**
  - Problem for spatial localisation of activations.
  - Inaccurate coregistration reduces sensitivity of group studies.
- **Reduce distortion**
  - Shorter acquisition times, use parallel imaging
- **Distortion correction**
  - Post-processing using field maps



Cusack et al., Neuroimage 2003

Dropout compensation: z-shimming



- **Use of preparation gradient pulses (*zshim gradients*) to compensate local dropouts**
- ***But:* Reduces signal in normal areas**

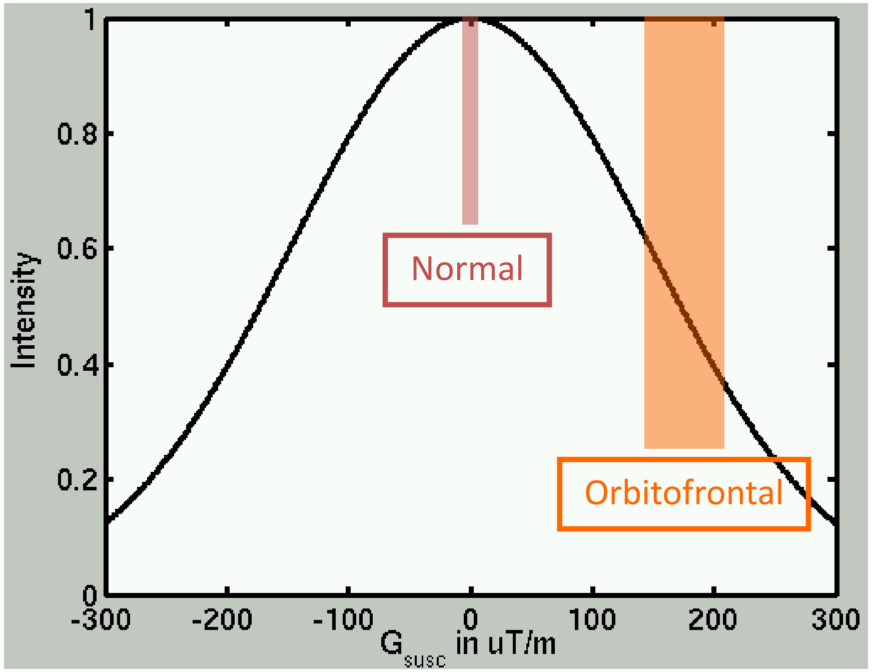
Acquisition of several images with different z-shimming reduces temporal resolution

⇒ Optimal compromise: moderate zshimming

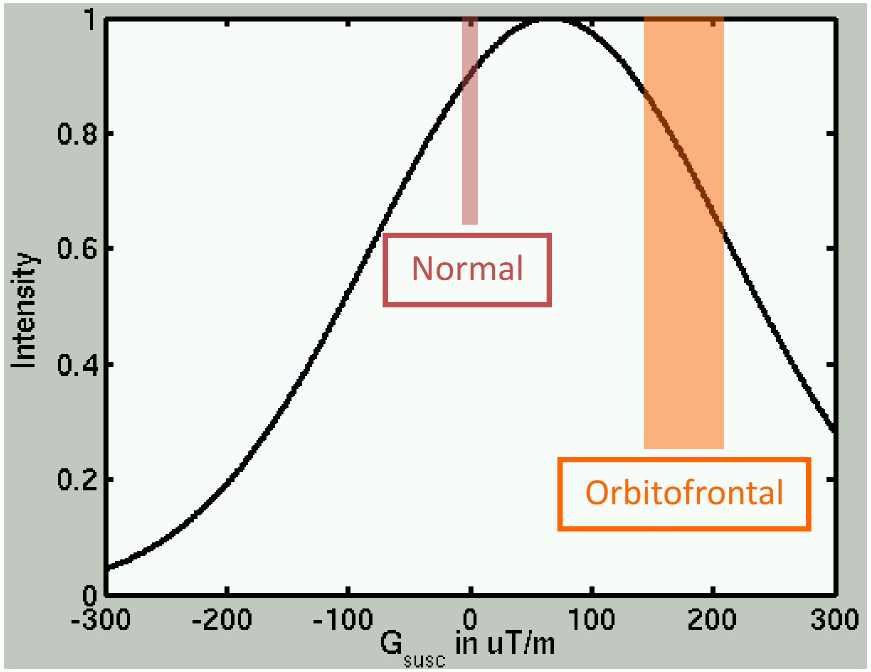
Frahm et al., MRM 1988; Ordidge et al., MRM 1994

Moderate z-shimming: trade-off

(Simulation for slice thickness of 2 mm)



No z-shimming



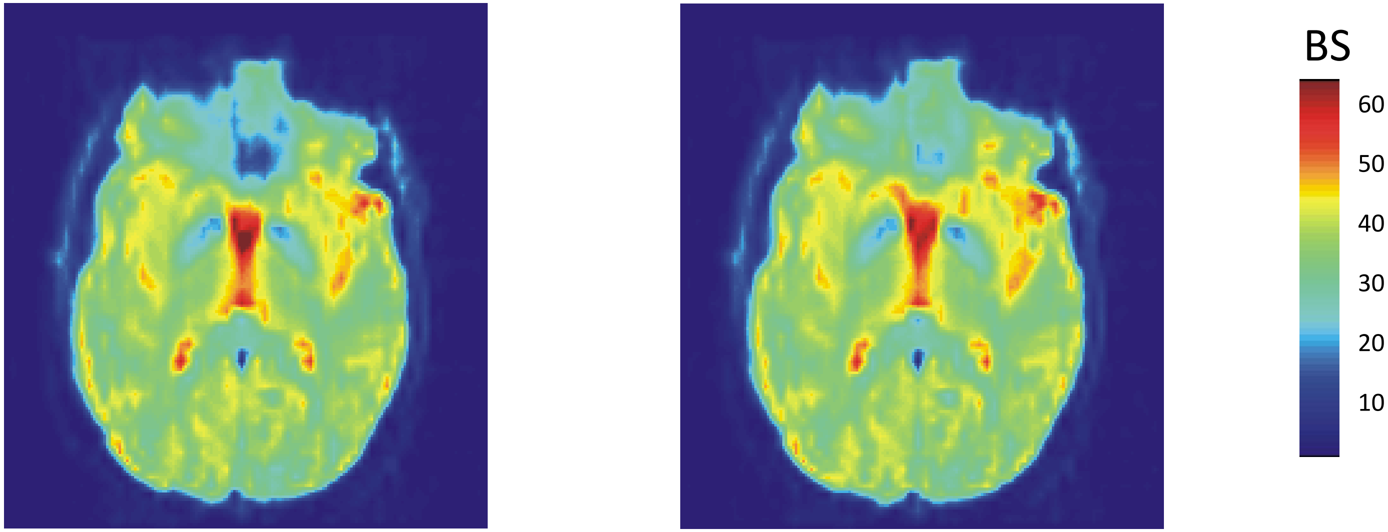
z-shimming with -2 mT/m*ms

Deichmann et al., Neuroimage 2003

Moderate z-shimming: example

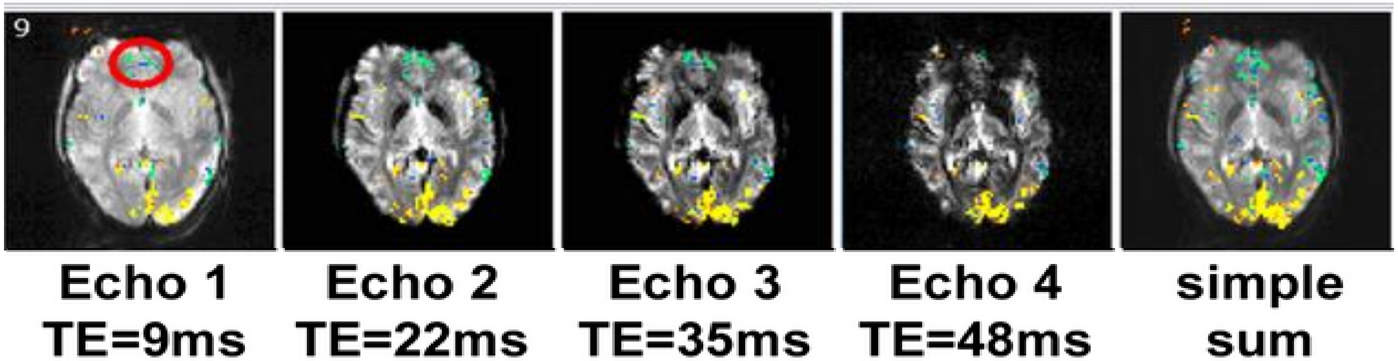
Standard EPI

EPI + *z-shim*



Dropout compensation - multi-echo EPI

- Acquire multiple EPI readouts (=images) after a single RF excitation pulse
- Short TE images recover dropouts



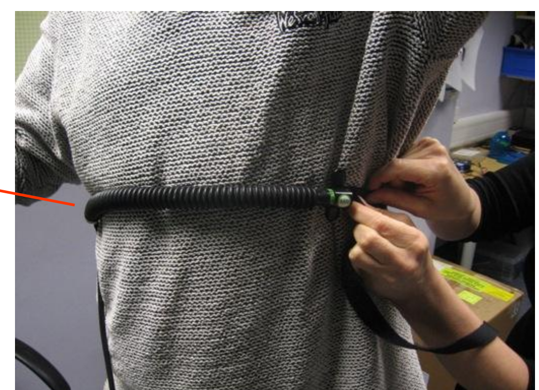
Poser et al., Neuroimage 2009

- Enhanced BOLD sensitivity over the whole brain
- Pitfall: increased acquisition time

Measuring cardiac and respiratory effects

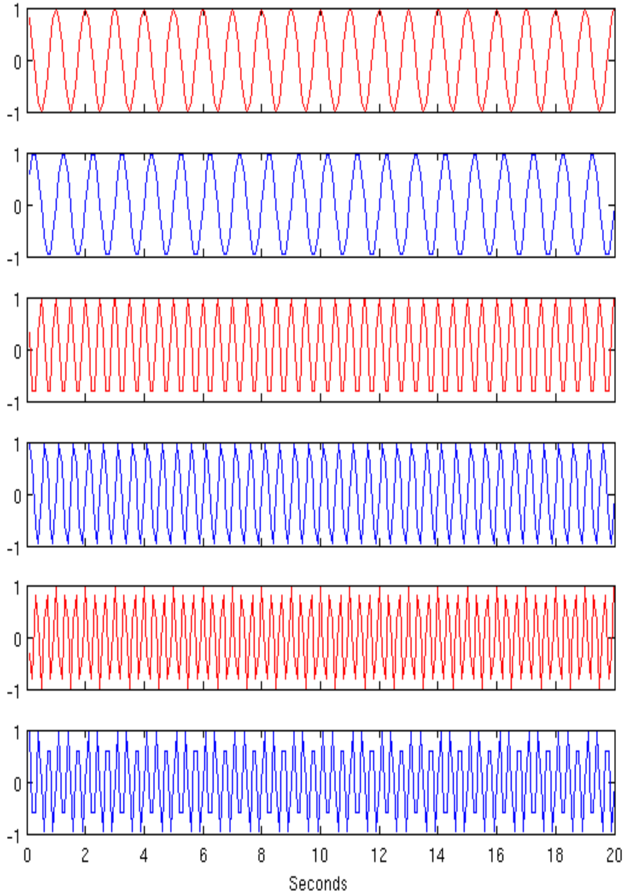
- Model based on peripheral measurements:

Pulse oximeter



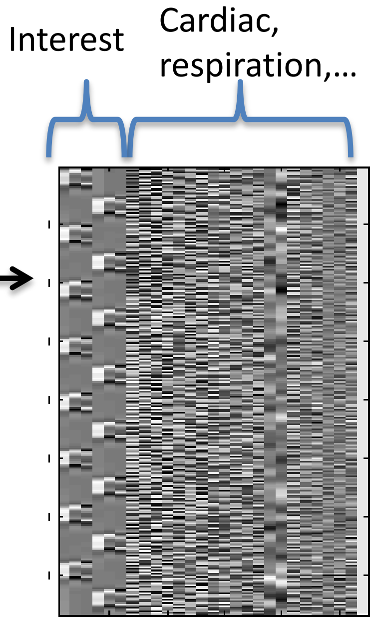
Respiration belt

Modelling and correcting for cardiac and respiratory effects



- Measured cardiac and respiratory phase can be modelled using a sum of periodic functions e.g. sines and cosine of increasing frequency (Fourier set)
- Modelled effects can be
  - removed from original fMRI signal

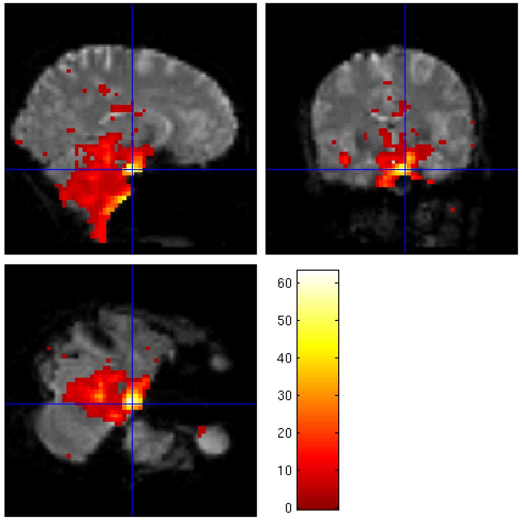
    or
  - included in fMRI statistical model



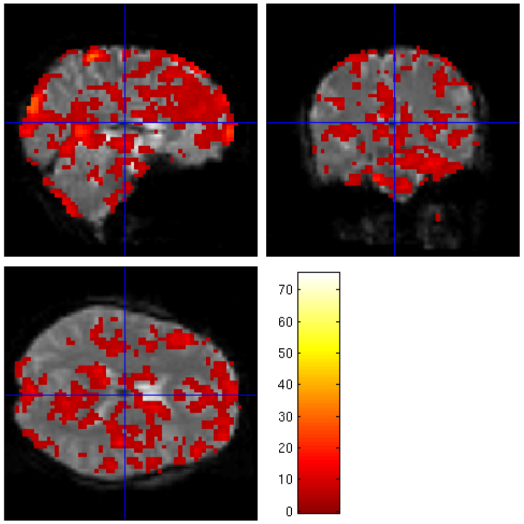
Glover G.H. Et al. MRM 2000; Hutton et al., Neuroimage 2011

Physiological effects in BOLD

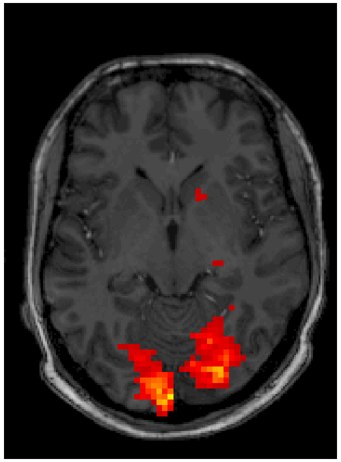
Cardiac effects - vessels



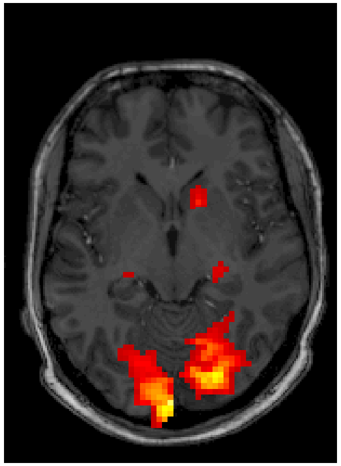
Respiratory effects - global



standard



corrected



Activation in visual cortex and LGN with and w/o physiological noise correction

Physiological correction enhances BOLD sensitivity

Hutton et al., Neuroimage 2011

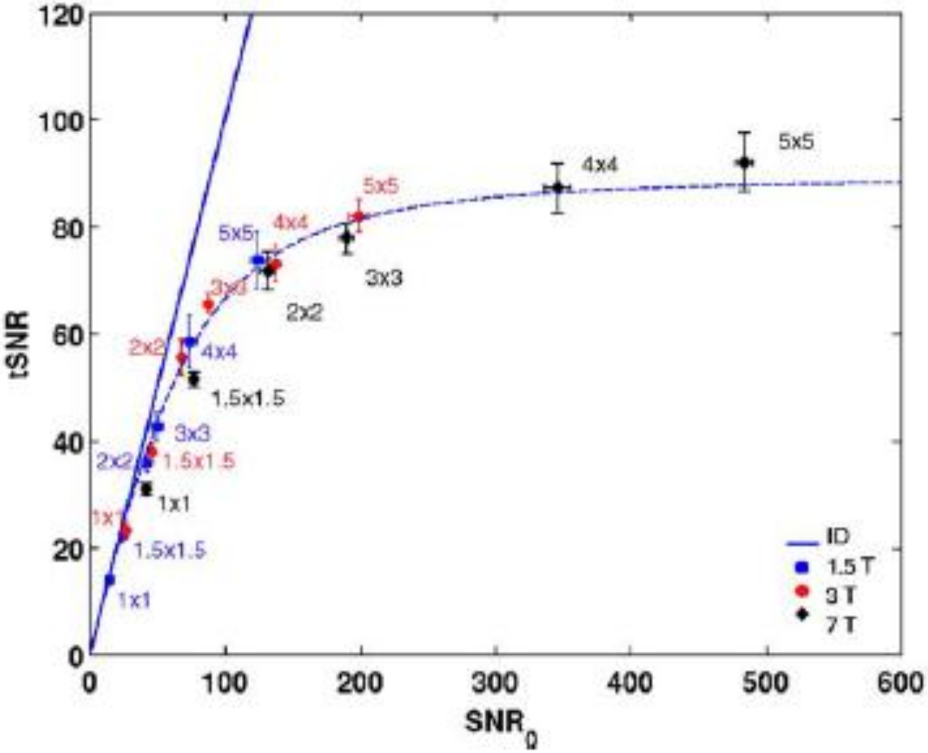
3D EPI acquisitions for fMRI

3D EPI yields higher image signal-to-noise (SNR_0)

Temporal stability (tSNR) is an indicator of BOLD sensitivity

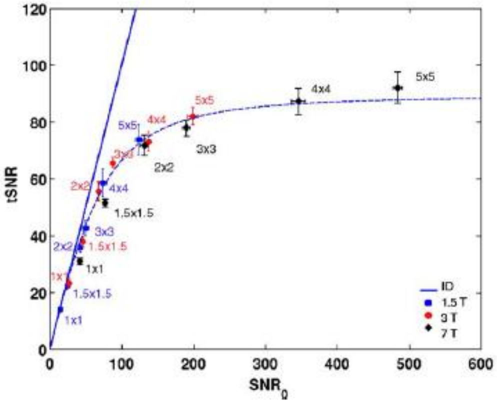
tSNR vs SNR_0

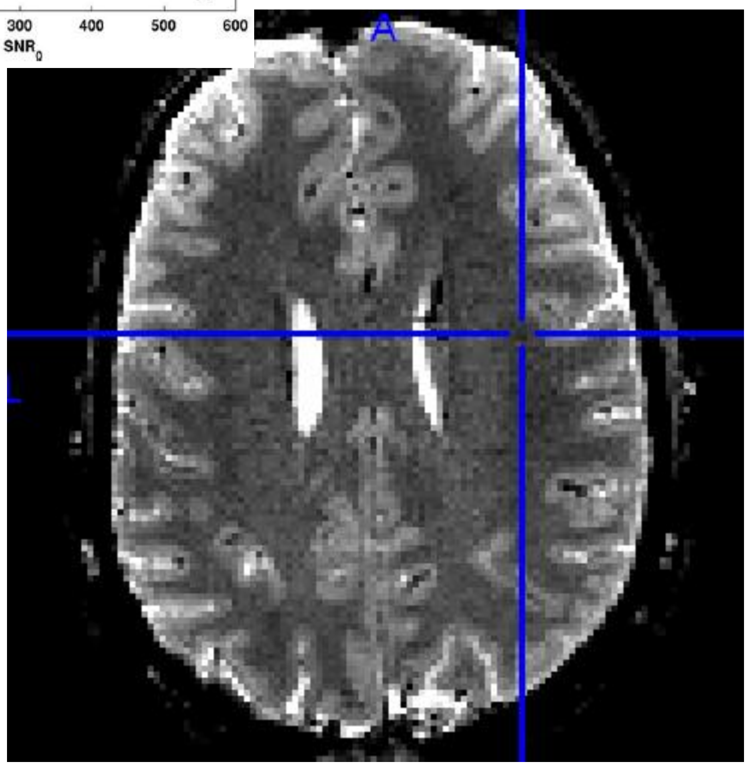
$$tSNR = \frac{SNR_0}{\sqrt{1 + \lambda^2 \times SNR_0^2}},$$



Krueger, G., Glover, G.H. MRM 2001, Triantafyllou, C. et al Neuroimage 2005

High-resolution EPI: 1.5mm 2D/3D EPI at 3T

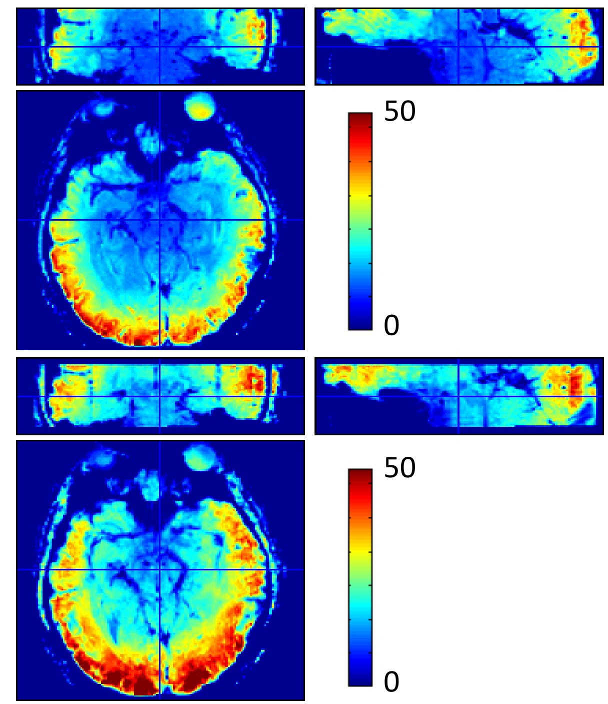




Temporal SNR

2D

3D



$tSNR_{3D}$

- 128% $tSNR_{2D}$ in VC
- 164% $tSNR_{2D}$ in LGN

Lutti et al., Magn Reson Med 2013

3D EPI acquisitions for fMRI

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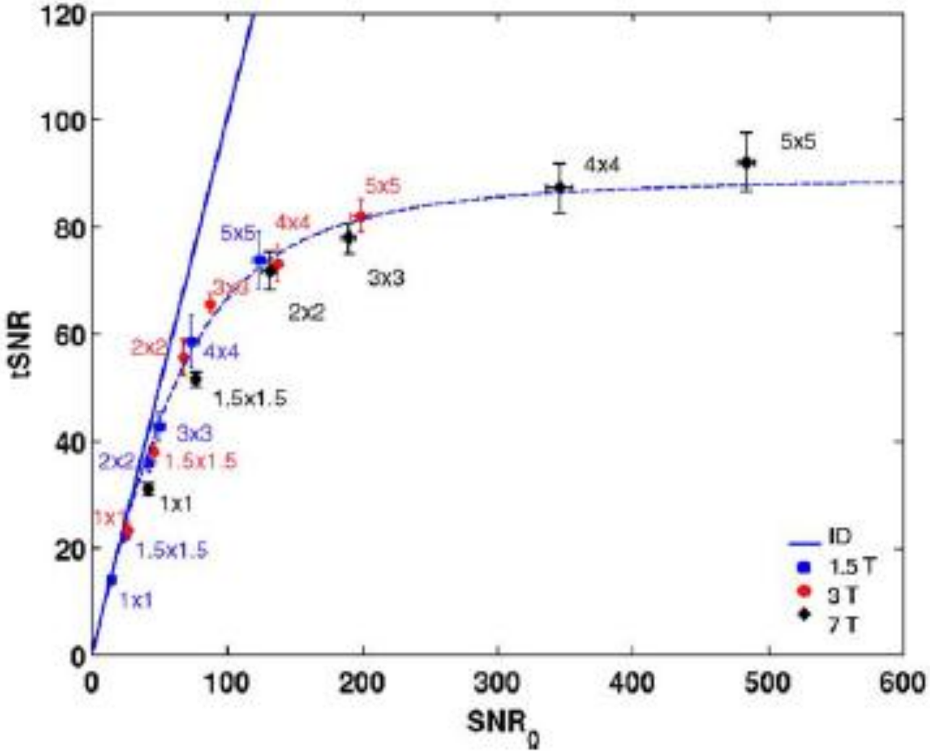
tSNR vs SNR_0

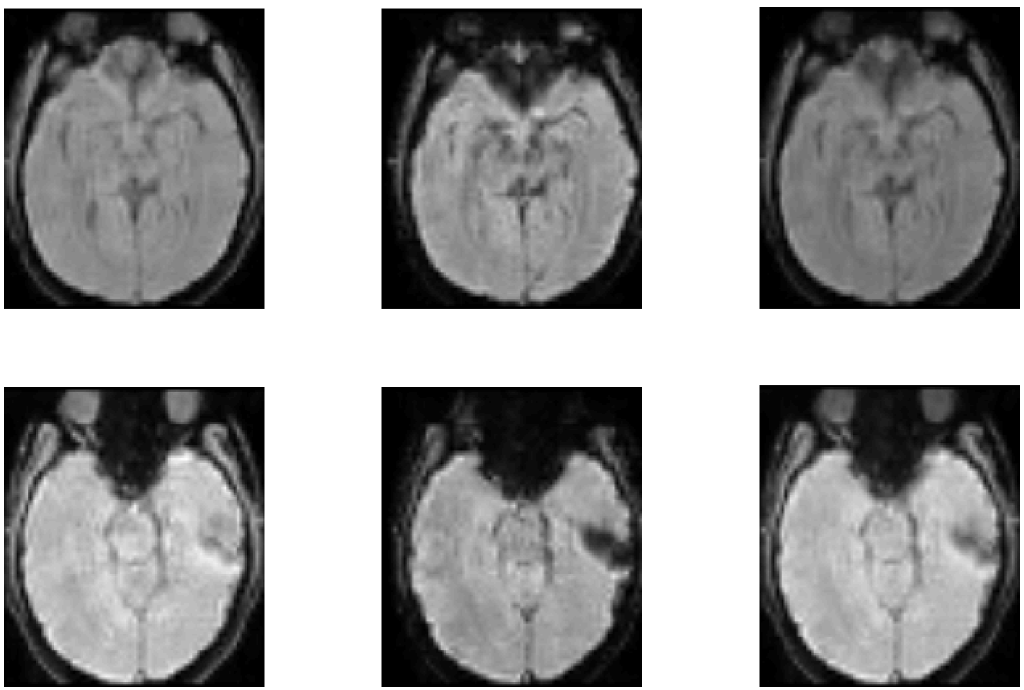
$$\mathrm{tSNR} = \frac{\mathrm{SNR}_0}{\sqrt{1 + \lambda^2 \times \mathrm{SNR}_0^2}},$$



Krueger, G., Glover, G.H. MRM 2001, Triantafyllou, C. et al Neuroimage 2005

Ultra-fast fMRI - 3mm^3 resolution

Echo1 TE=15.85ms | Echo2 TE=34.39ms | Echo1 +Echo2



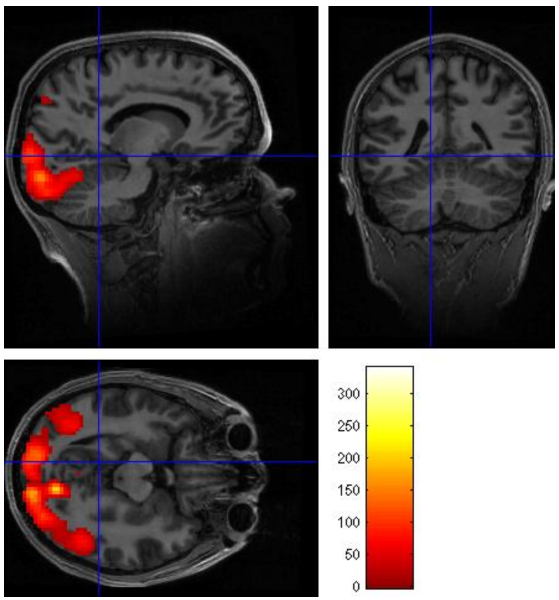
- Dual-echo whole-brain EPI acquisition
- Matrix size: 72x64x60 (PExROxPA)
- Acceleration factor 2 and 3 along the phase and partition directions
- **TR = 1s**

Poser B.A., Norris D.G. Neuroimage 2009;

Ultra-fast fMRI - $3mm^3$ resolution

Visual stimulus left-rest-right-rest flickering checkerboard.

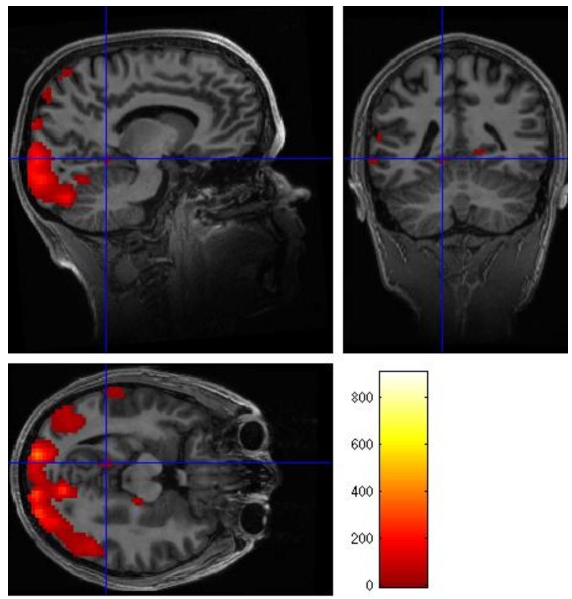
2D EPI



50 transverse slices

TR = 3s

3D EPI

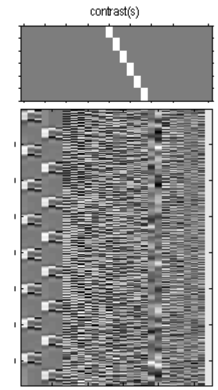


60 sagittal slices

TR = 1s

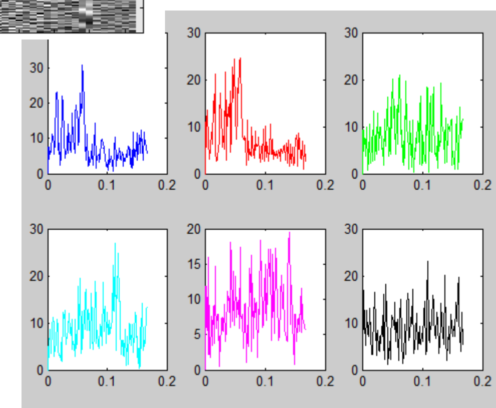
Mean F-value for visual excitation: 2D EPI: 36;3D EPI: 50

Mean T-value for visual excitation: 2D EPI: 4.5;3D EPI: 6

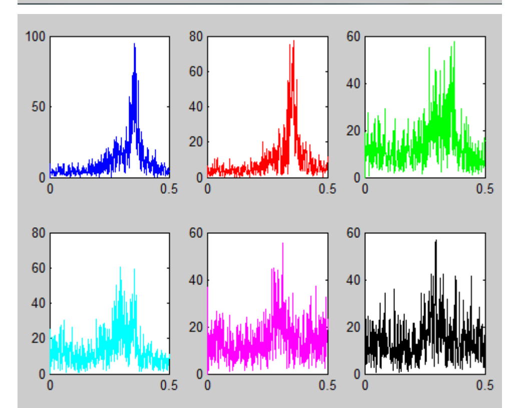


respiration

2D EPI



3D EPI



Functional imaging - summary

- fMRI: brain activation detected via increased metabolim (‘*BOLD effect*’)
- EPI acquisitions allow optimal sampling of BOLD response
- EPI images/time-series:
  - Distortions – corrected **at post-processing**
  - Signal dropouts –minimized **at run time**
  - Physiological instabilities - **online monitoring + offline processing**

  Correction yields optimal BOLD sensitivity
- Advanced acquisitions:
  - Enhanced BOLD sensitivity – high resolution
  - Rapid acquisitions – higher efficiency