Contrast and Pulse Sequences

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Image contrast

- Contrast determined by:
  - sample properties:
    - Proton density, T_1, T_2, ...
  - pulse sequence type:
    - IR, SE, ...
  - pulse sequence timing:
    - TI, TE, TR, ...
- Main source of contrast between tissues in MRI is differences in T_1 and T_2

The Bloch Equations

- Set of coupled differential equations, describing the behaviour of the magnetisation **M**
- The Bloch equations can be solved to give the x', y', z components of **M** as a function of time.

$$\frac{dM_{x'}}{dt} = (\omega_0 - \omega)\, M_{y'} - \frac{M_{x'}}{T_2}$$

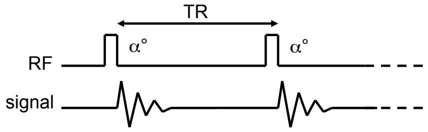
$$\frac{dM_{y'}}{dt} = (\omega_0 - \omega)\, M_{x'} - \frac{M_{y'}}{T_2} \quad + \gamma\, B_1 M_z$$

$$\frac{dM_z}{dt} = \frac{(M_z - M_z^0)}{T_1} \quad - \gamma\, B_1 M_{y'}$$

If B_1 is present along x'

Saturation recovery sequence

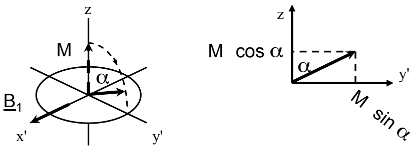
- SR is the simplest pulse sequence
  - repeated excitation pulses (with gradients)



- Contrast depends on
  - T_1 value
  - flip angle of pulses (α)
  - repetition time (TR)

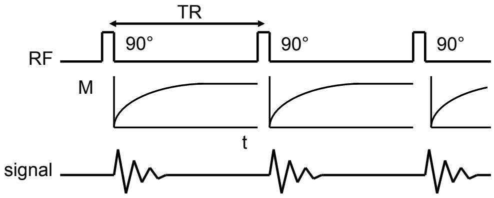
Effect of RF pulse

- Effect of an RF pulse with flip angle α:



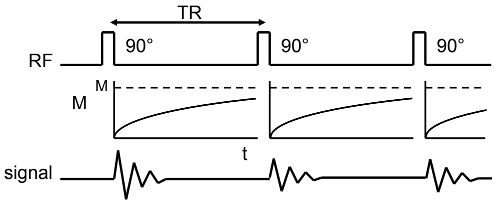
- Signal M which in turn is
  M which existed before the 90° pulse
- A 90° pulse converts all longitudinal magnetisation into transverse magnetisation

Saturation recovery: TR > T_1



- If TR > T_1, magnetisation can relax back fully to M_0 between RF pulses
- Maximum signal strength ($\propto M_0$) is obtained after each pulse

Saturation recovery: TR < T_1



- If TR < T_1, magnetisation cannot recover fully to M_0 between pulses
- Signal strength is less than M_0 (except for the signal following the first 90° pulse)
- This is called partial <u>saturation</u>

SR - expressions for signal

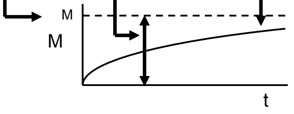
- Need to solve the Bloch equations to obtain an expression for the signal strength as a function of M_0, T_1 and TR
- Use the following facts:
    - 90° pulse converts longitudinal magnetisation into transverse magnetisation
        - so M_z=0 immediately <u>after</u> each 90° pulse (t=0)
    - Transverse magnetisation dephases rapidly after excitation, i.e. $T_2^* << TR$
        - therefore M_{xy}= 0 <u>before</u> each 90° pulse
    - We only need to solve for the behaviour of M_z

SR - expression for signal

- The general solution is given by:

$$M_z(t) = M_0 - \underbrace{(M_0 - M_z(0))}_{\text{"distance" between start and "target" } M_z} \underbrace{\exp(-t/T_1)}_{\text{exponential return towards } M_0}$$

equilibrium magnetisation (at infinite TR)



SR - expressions for signal

- Expression for signal (90° pulses):
  $S(M_0, T_1, TR) = M_0 [1 - \exp(-TR/T_1)]$
- If $TR \geq 5T_1$, $\exp(-TR/T_1) = 0$, so $S = M_0$
  "proton density" image
- General expression for signal,
  no longer assuming 90° pulses:
  (α = flip angle)

$$S(M_0, T_1, TR, \alpha) = M_0 \frac{[1 - \exp(-TR/T_1)] \sin\alpha}{[1 - \exp(-TR/T_1) \cos\alpha]}$$

Ernst Angle

- If $TR < T_1$, a 90° pulse does not give the largest signal, due to partial saturation
- The flip angle which gives the maximum signal is called the Ernst Angle
  - obtained by differentiating general expression for S w.r.t. α, and finding the stationary value
- Ernst angle: $\alpha_E = \cos^{-1} [\exp(-TR/T_1)]$

e.g. $TR = T_1 = 1000\text{ms}$: $\alpha_E = 68°$
$TR = 100\text{ms}$, $T_1 = 1000\text{ms}$: $\alpha_E = 25°$

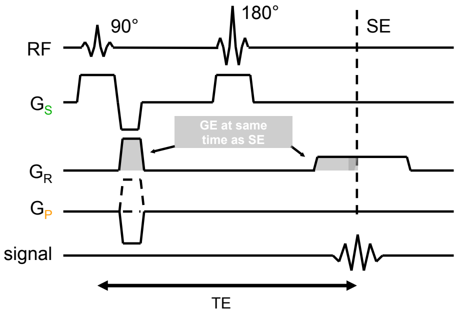
SR - effect of T_2^*

- We should also take account of the effect of transverse relaxation (T_2^*) during the time between the 90° pulse and the centre of the gradient echo
  - to do this, simply multiply the expression for the signal by $\exp(-TE/T_2^*)$:

$$S = M_0 [1 - \exp(-TR/T_1)] \exp(-TE/T_2^*)$$

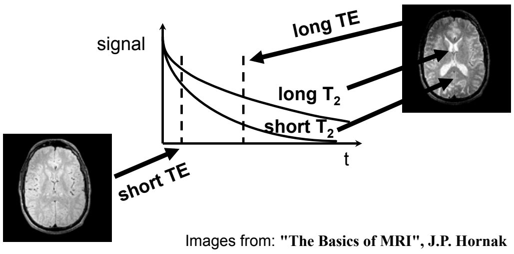
- If TE is short ($TE \ll T_2^*$), the contribution of this term will be very small

Spin echo pulse sequence



Spin Echo image contrast

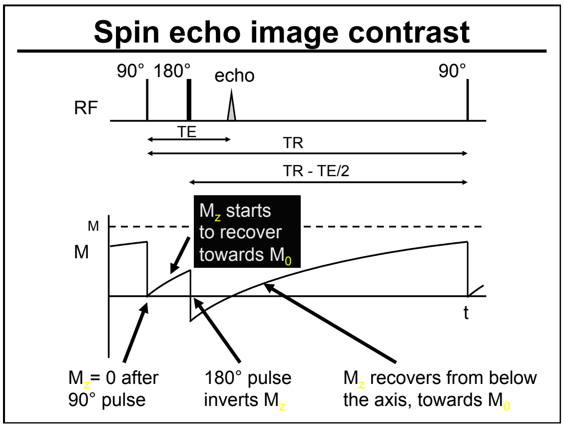
- Assuming TR>>T_1, the signal strength in a spin echo sequence is given by **$S = M_0 \exp(-TE/T_2)$**



Images from: **"The Basics of MRI", J.P. Hornak**
http://www.cis.rit.edu/htbooks/mri/

Spin echo image contrast

- A full analysis should include the possibility of TR ≤ T_1, i.e it should include the effect of partial saturation
- As in the case of SR, in order to determine the effect of saturation, we assume:
  - TE<<TR, so that $M_{xy} = 0$ before each 90° pulse
- Therefore, we only need to consider the behaviour of the longitudinal magnetisation M_z during the pulse sequence



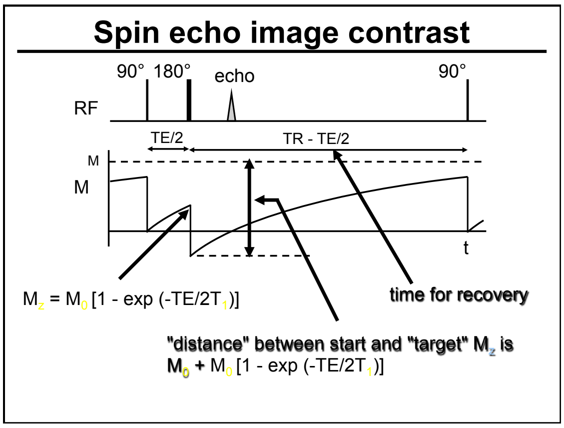
Spin echo image contrast

M_z starts to recover towards M_0

M_z = 0 after 90° pulse

180° pulse inverts M_z

M_z recovers from below the axis, towards M_0

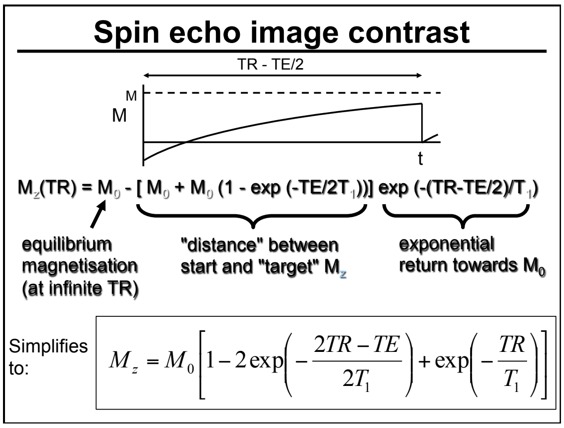


Spin echo image contrast

$M_z = M_0 [1 - \exp(-TE/2T_1)]$

time for recovery

"distance" between start and "target" M_z is $M_0 + M_0 [1 - \exp(-TE/2T_1)]$



Spin echo image contrast

$$M_z(TR) = \underset{\substack{\uparrow \\ \text{equilibrium magnetisation (at infinite TR)}}}{M_0} - \underbrace{[\, M_0 + M_0 (1 - \exp(-TE/2T_1))]}_{\text{"distance" between start and "target" } M_z} \underbrace{\exp(-(TR-TE/2)/T_1)}_{\text{exponential return towards } M_0}$$

Simplifies to:

$$M_z = M_0\left[1 - 2\exp\left(-\frac{2TR - TE}{2T_1}\right) + \exp\left(-\frac{TR}{T_1}\right)\right]$$

Spin echo image contrast

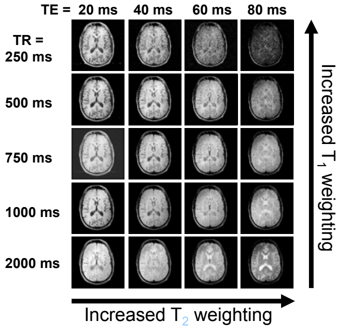
$$M_z = M_0\left[1-2\exp\left(-\frac{2TR-TE}{2T_1}\right)+\exp\left(-\frac{TR}{T_1}\right)\right]$$

- 90° pulse converts M_z to M_{xy}, so signal strength is equal to above expression multiplied by spin-spin relaxation factor:

$$S = M_0\left[1-2\exp\left(-\frac{2TR-TE}{2T_1}\right)+\exp\left(-\frac{TR}{T_1}\right)\right]\exp\left(-\frac{TE}{T_2}\right)$$

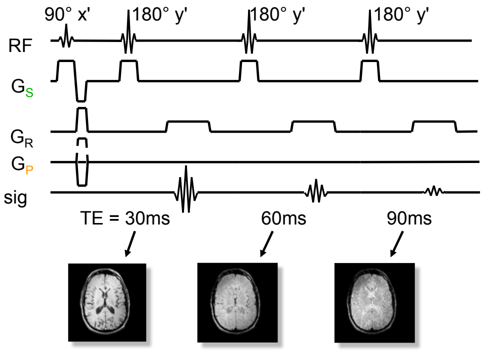
- The above is the full expression for the signal strength in a spin echo experiment, taking into account partial saturation and spin-spin relaxation

Spin Echo - T_1 and T_2 contrast



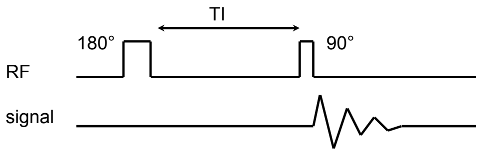
Images from: **"The Basics of MRI", J.P. Hornak:** www.cis.rit.edu/htbooks/mri/

Multi-echo pulse sequence (CPMG)

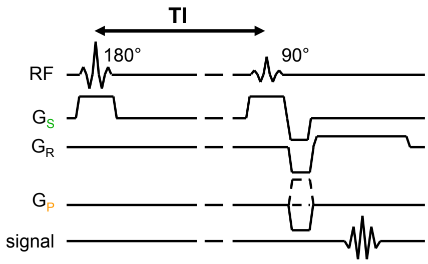


T_1 contrast: Inversion recovery

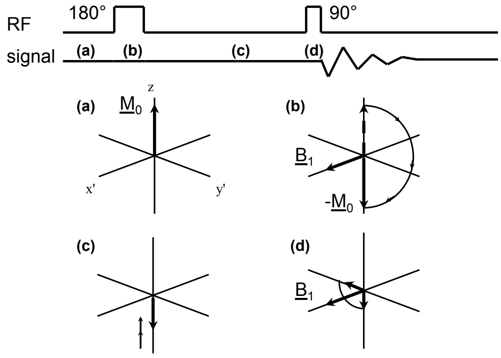
- 180° inversion pulse followed, after delay TI, by 90° pulse



Inversion recovery pulse sequence

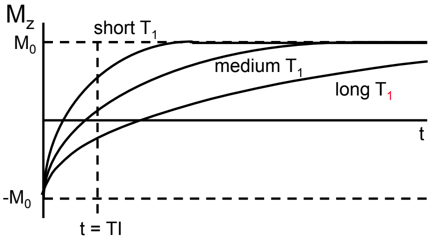


Inversion recovery



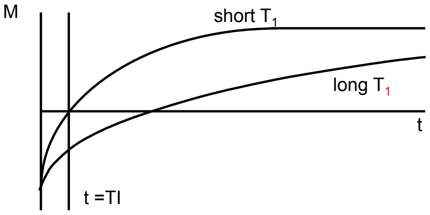
Inversion recovery

$$M_z(t) = M_0[1 - 2\exp(-t / T_1)]$$



NMR signal: S(TI) = M (TI) = M [1 - 2exp(-TI / T)]

STIR (<u>S</u>hort <u>TI</u> <u>I</u>nversion <u>R</u>ecovery)



- Zero signal from short-T sample
- Large signal from long-T sample
- Used to "null out" fat from image
  (fat has short T compared with muscle etc.)

STIR sequence – example of use

From: **MRI in Practice** By Catherine Westbrook, Carolyn Kaut Roth, John Talbot

Short TI (tau)	150–175 ms (to suppress fat depending on field strength)
Long TE	50 ms+ (to enhance signal from pathology)
Long TR	4000 ms+ (to allow full recovery)
Long turbo factor	16–20 (to enhance signal from pathology)
Average scan time	5–15 min

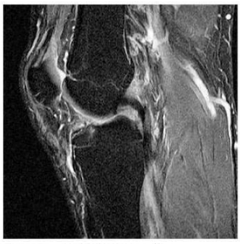


Figure 5.18 Sagittal STIR sequence of the knee. Normal bone marrow has been nulled. Synovial fluid in the joint has a high signal as the TE is long and the image is therefore T2 weighted.

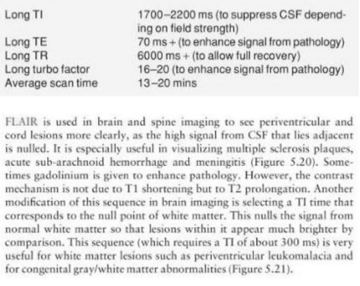
STIR is an extremely important sequence in musculoskeletal imaging as normal bone, which contains fatty marrow, is suppressed and lesions within bone such as bone bruising and tumors are seen more clearly (Figure 5.18 and 5.19). It is also a very useful sequence for suppressing fat in general MR imaging (*see* Chapter 6).

FLAIR (fluid attenuated inversion recovery)

- Similar concept to STIR, but used to "null out" signal from long-T_1 fluid, especially cerebro-spinal fluid (CSF)
- Used with spin-echo sequence to attenuate signal from ventricles in brain, allowing surrounding structures and pathology to be seen
- $TI_{null} = \ln(2) \times T_1$ = 2000ms (approx) for CSF

FLAIR sequence – example of use

From: **MRI in Practice** By Catherine Westbrook, Carolyn Kaut Roth, John Talbot



Long TI	1700–2200 ms (to suppress CSF depending on field strength)
Long TE	70 ms + (to enhance signal from pathology)
Long TR	6000 ms + (to allow full recovery)
Long turbo factor	16–20 (to enhance signal from pathology)
Average scan time	13–20 mins

FLAIR is used in brain and spine imaging to see periventricular and cord lesions more clearly, as the high signal from CSF that lies adjacent is nulled. It is especially useful in visualizing multiple sclerosis plaques, acute sub-arachnoid hemorrhage and meningitis (Figure 5.20). Sometimes gadolinium is given to enhance pathology. However, the contrast mechanism is not due to T1 shortening but to T2 prolongation. Another modification of this sequence in brain imaging is selecting a TI time that corresponds to the null point of white matter. This nulls the signal from normal white matter so that lesions within it appear much brighter by comparison. This sequence (which requires a TI of about 300 ms) is very useful for white matter lesions such as periventricular leukomalacia and for congenital gray/white matter abnormalities (Figure 5.21).

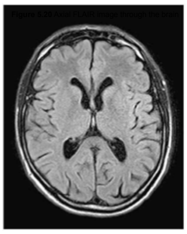


Figure 5.20 Axial FLAIR image through the brain