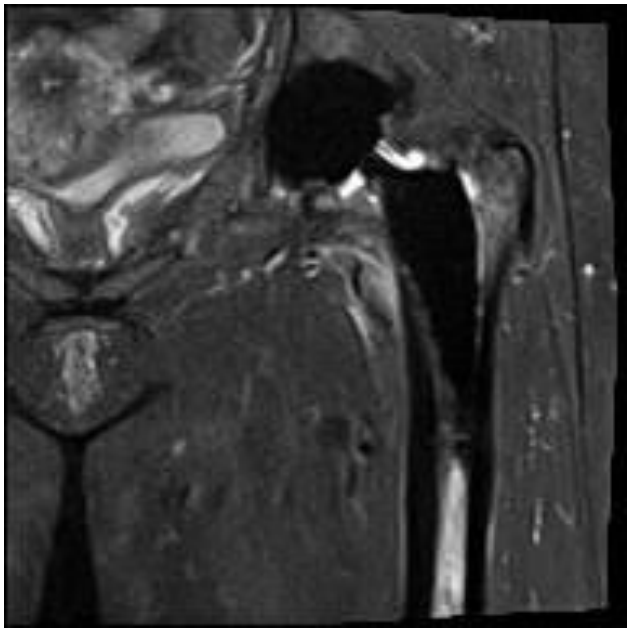
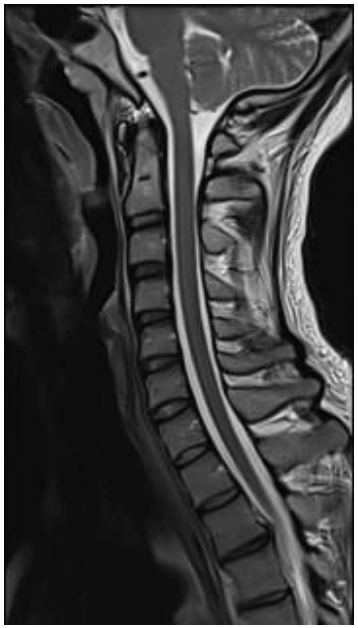
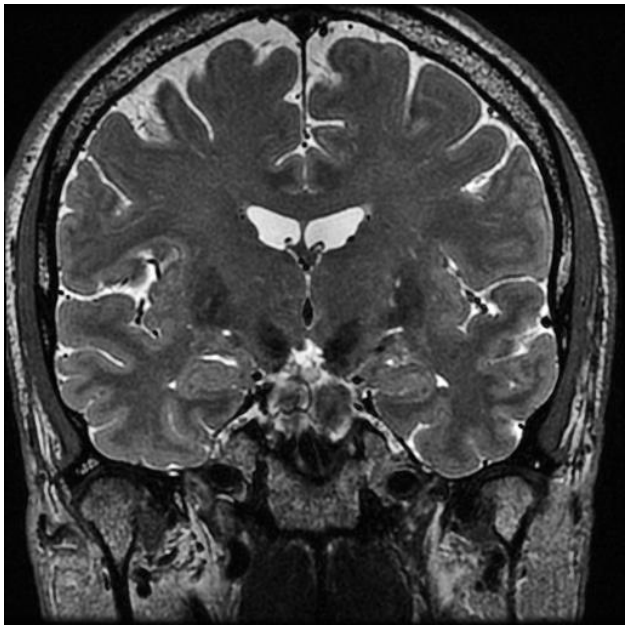
Magnetic Resonance Imaging

F.R.C.R. Physics Lectures

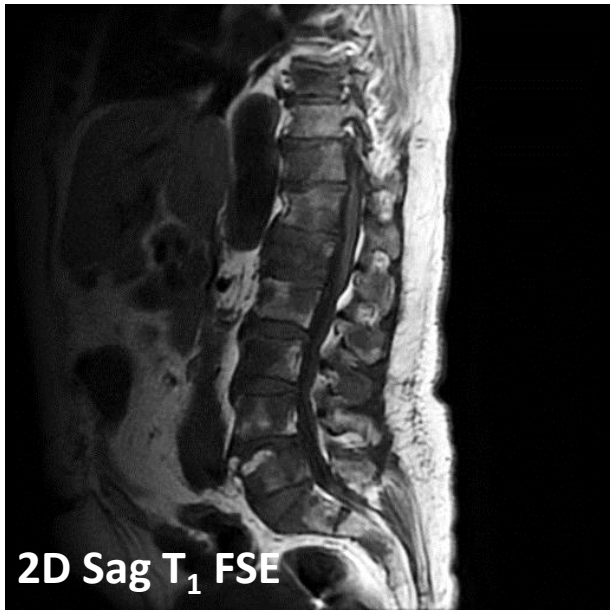
Lawrence Kenning PhD

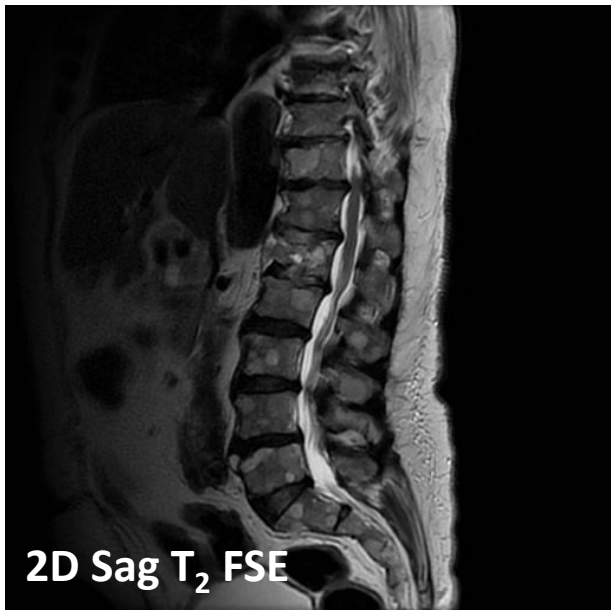
7.5 T_1-dependant techniques

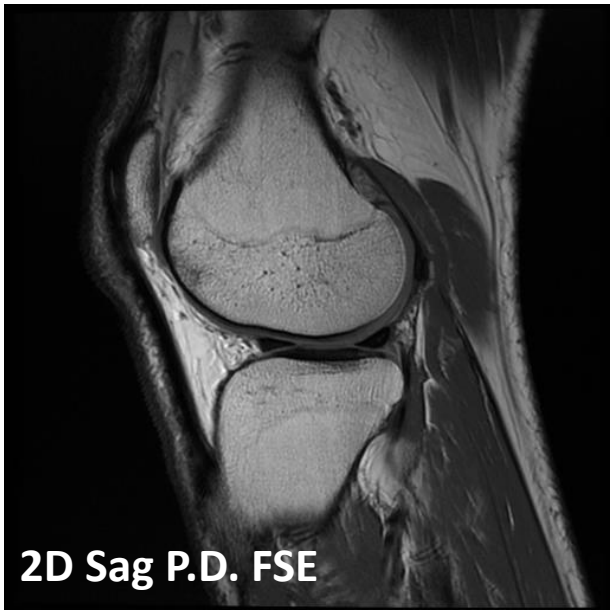
- Inversion recovery (IR)
- Suppression: STIR & FLAIR. The role(s) of TR (and T_1) in determining null point.
- Increase T1-weighting e.g. MPRAGE
- Phase-sensitive IR

Fat issue

- Fat is hyperintense (bright) signal on T_1, T_2 and P.D. weighted imaging
- On T_2 and P.D. weighting imaging fluid is hyperintense
- On T_1 weighed post-contrast imaging, the effect of gadolinium is hyperintense

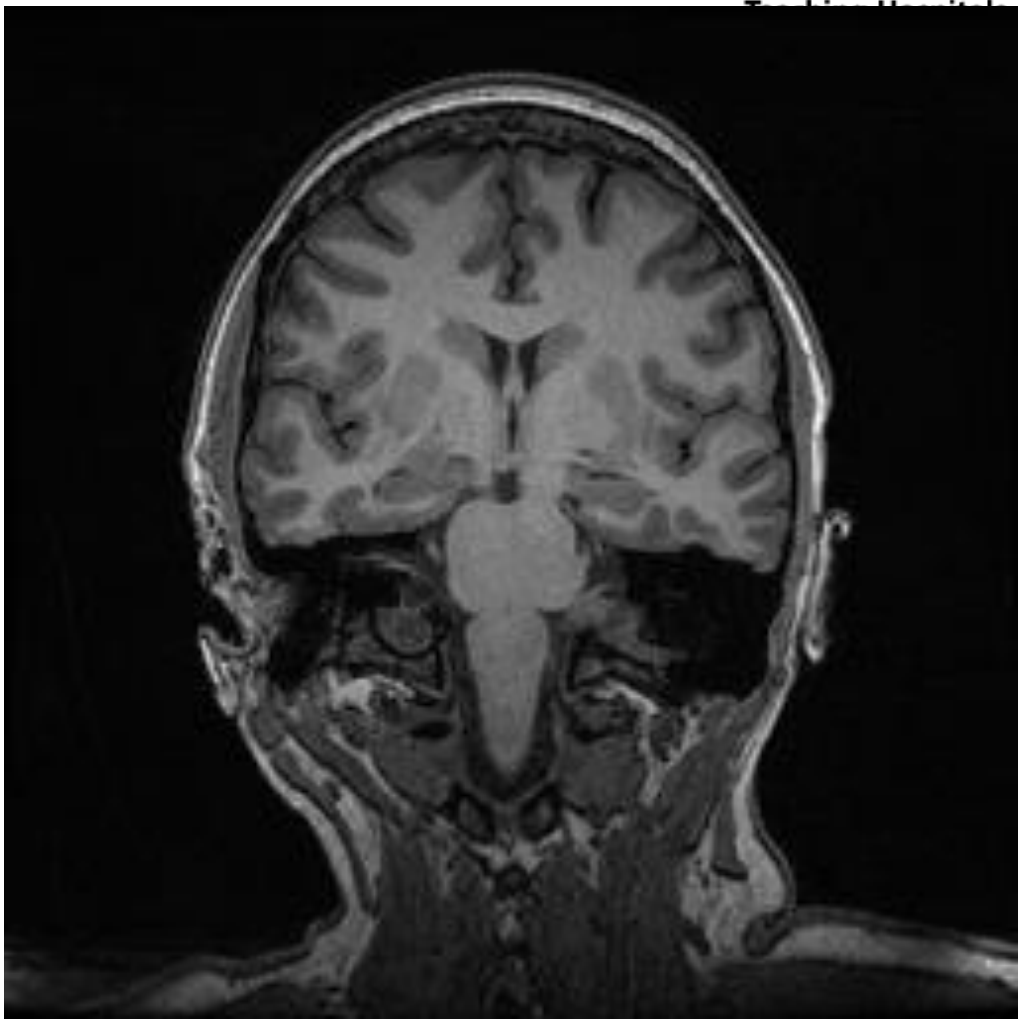




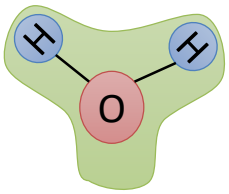


Water and Fat Molecules

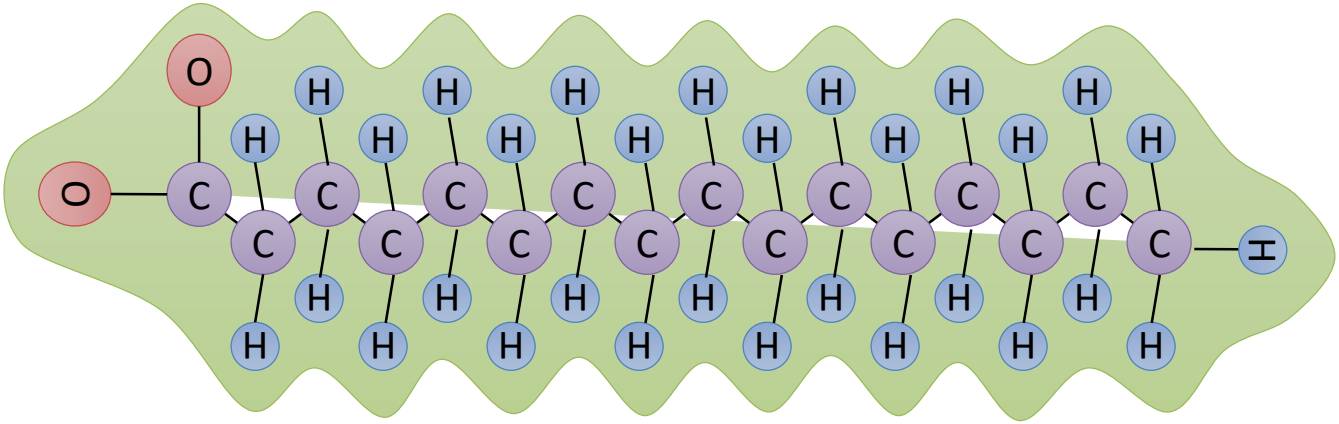
- Signal from water and fat both originate from **protons** but there are two key differences:
- **Relaxation Times**
- **Chemical Shift**



Water molecule H_2O



Fat molecule $–CH_2–$

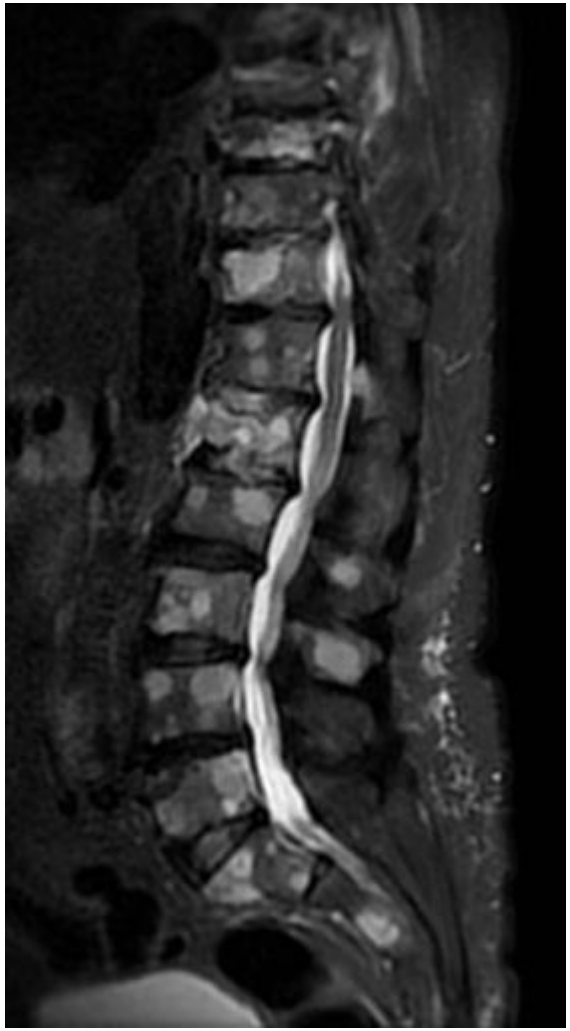


- Fat has a shorter T_1 relaxation rate than water
- The two dominate fat peaks (1.3ppm and 0.9ppm) experience a lower local magnetic field (B_0) due to the electron shielding compared to water (4.7ppm)
- Fat precesses at a lower frequency than water

Tissue	T_1 1.5T (ms)
Fat	260
Liver	500
Grey Matter	900
CSF	2400

Why Suppress?

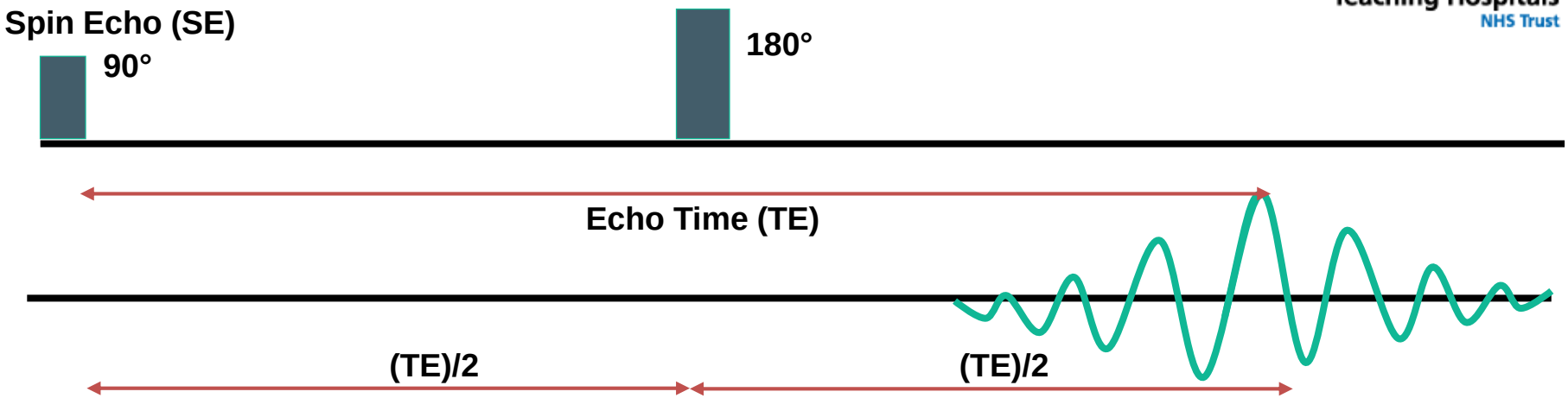
- Remove bright signal from normal tissue
  i.e. fat or CSF
- Improve tissue contrast
  i.e. white-grey matter in the brain
- Improve appearance of contrast agents
- Confirm tissue characterisation
  i.e. fat vs. blood

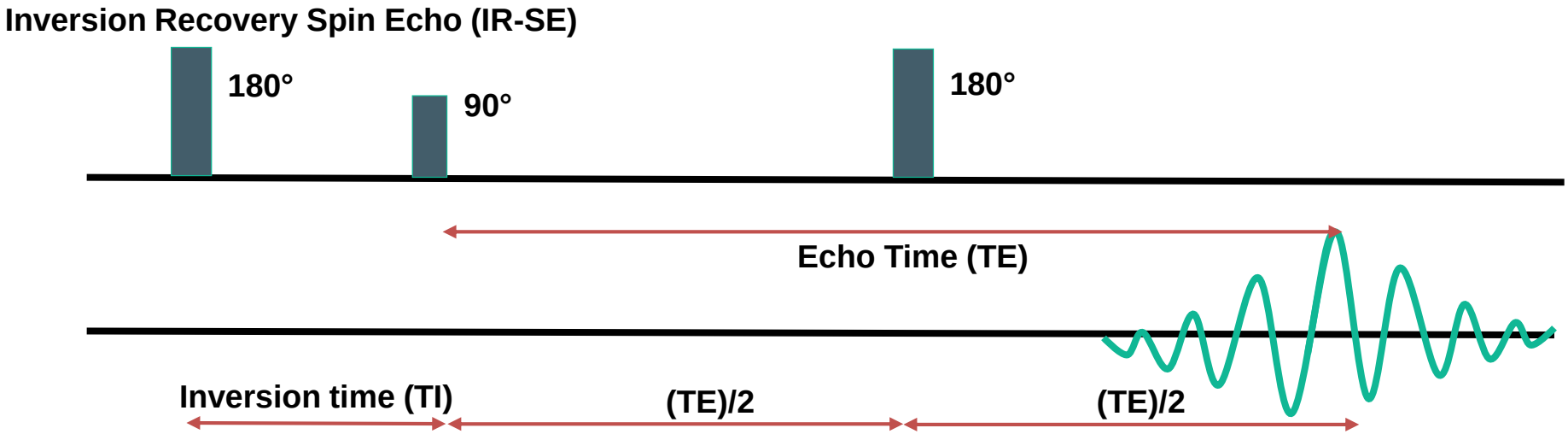


Inversion recovery (IR)

- IR emphasises T_1 relaxation times of the tissues by extending the amplitude of the longitudinal recovery by a factor of 2
- Selection of an appropriate TI can thus suppress tissue signals (e.g., fats/lipids, CSF) depending on their T_1 relaxation times
- Initial 180° RF pulse inverts M_z to $-M_z$
- After a predefined delay, the time of inversion (TI), a 90° RF pulse rotates the recovered fraction of M_z into the transverse (M_{xy}) plane to generate the FID
- A second 180° pulse at TE/2 produces an echo
- The TR for IR is the time between 180° initiation pulses.

Inversion recovery (IR)



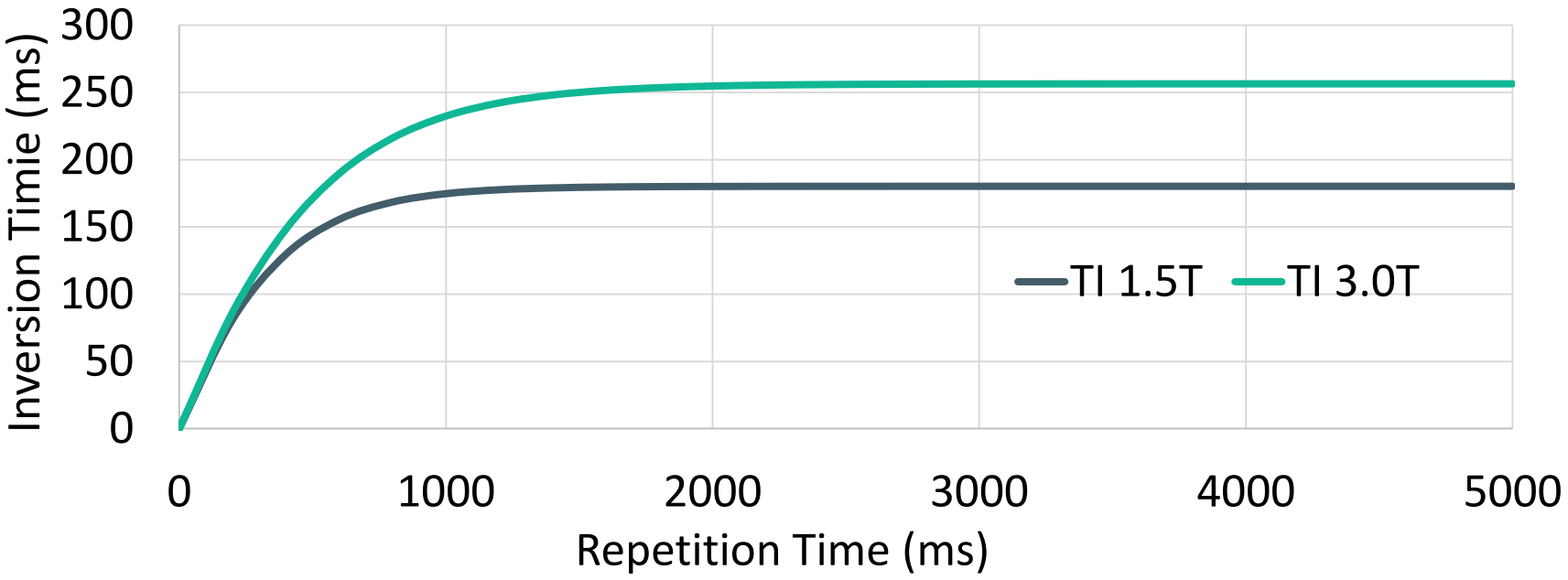


How to calculate TI_{null}

- TI_{null} depends on the ratio TR/T_1. *Fat T1 @ 1.5T = 260ms / 3.0T = 370ms*

- $TI_{null} = T_1 \times (\ln(2) - \ln(1+e^{-TR/T1}))$ for spin echo
- $TI_{null} = T_1 \times (\ln(2) - \ln(1+e^{-TR-TElast/T1}))$ for FSE/TSE

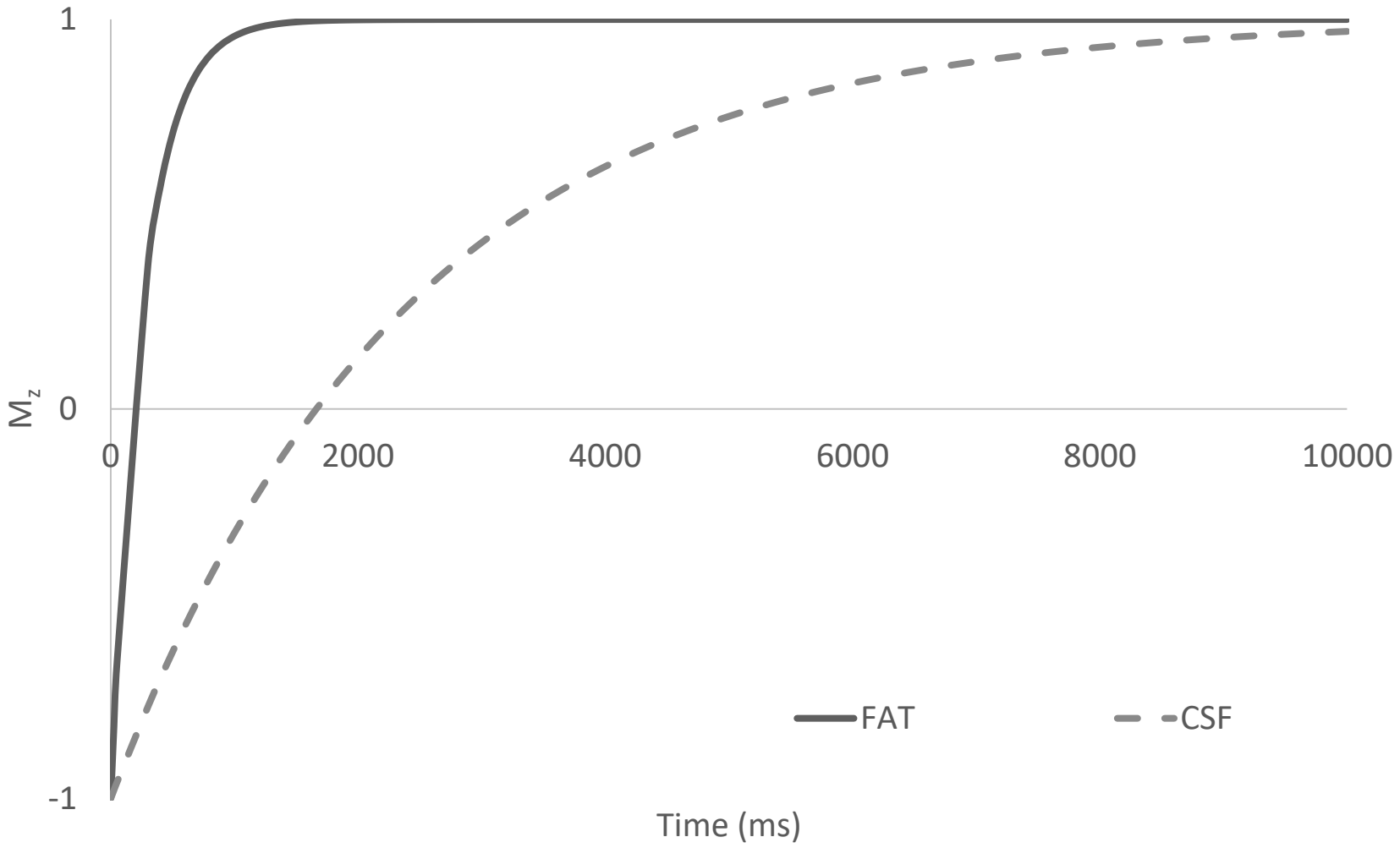
- When $TR >> T_1$ (~5x), the equation can be simplified to $TI_{null} = T_1 \times \ln(2)$



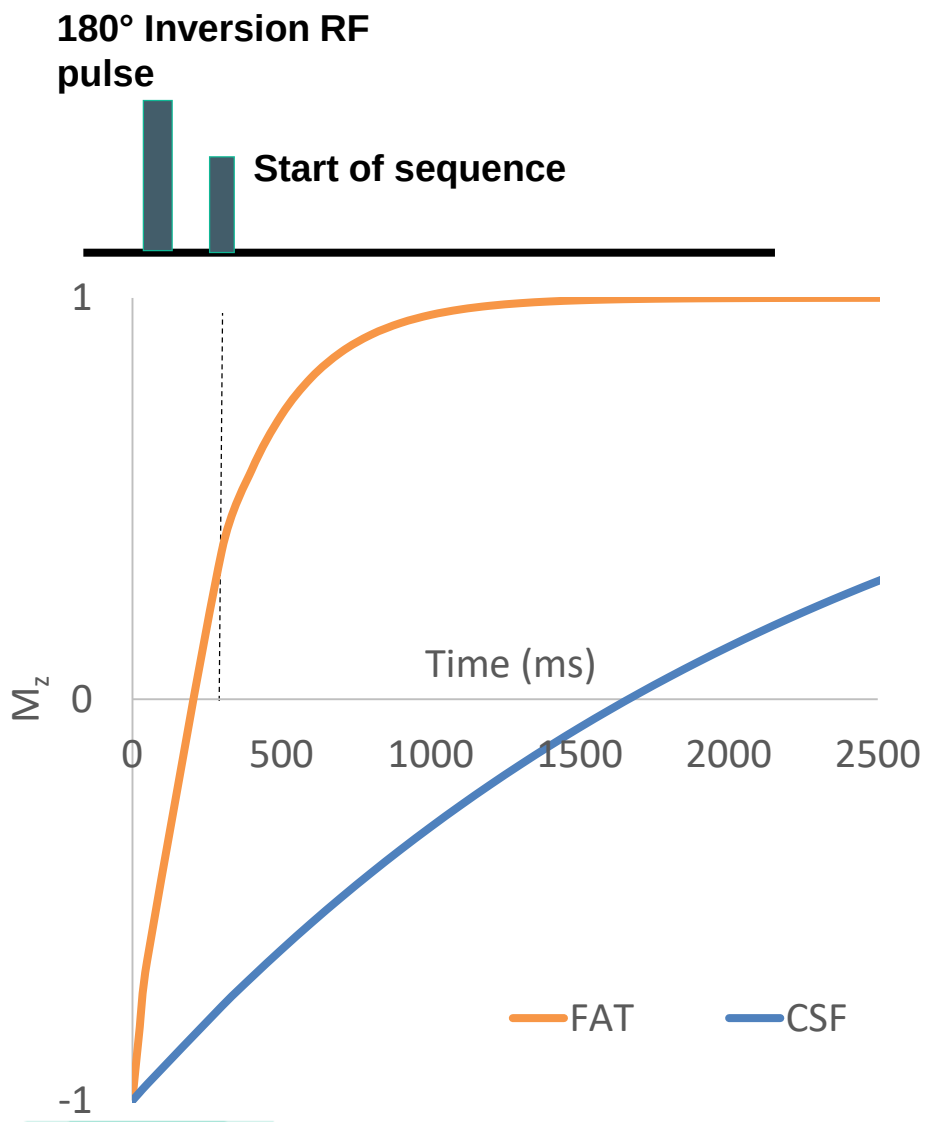
Short Tau Inversion Recovery (STIR)

- Short Tau Inversion Recovery (STIR), is a pulse sequence that uses a very short inversion time (TI) to supress fat
- The signal null (M_z=0) occurs at TI=ln(2)xT_1
- T_1 for fat at 1.5T is approximately 260 ms, therefore TI selected is 0.693 x 260ms = 180 ms
- Typical STIR sequences uses TI of 140-180ms and TR of approximately 2,500ms upwards
- In STIR, T_1 weighting comes from the inversion recovery process
- T_2 weighting generated by using longer TE values
- Generally a higher TI results in darker fat

Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.



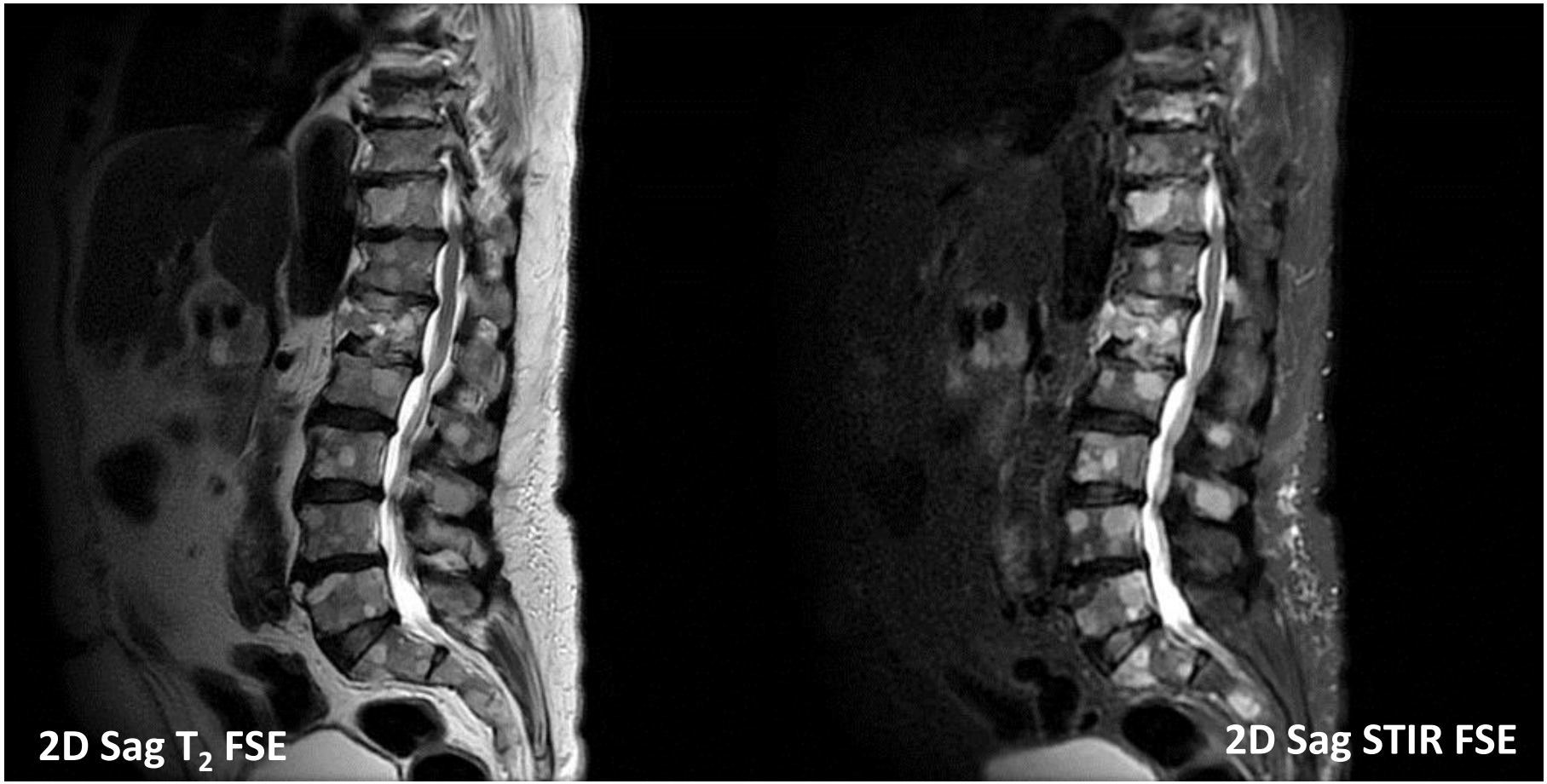
Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.



Short Tau Inversion Recovery (STIR/TIRM)

1. Initial 180° inversion pulse. Fat and water signals begin T_1 relaxation from $-M_z$
2. Fat signal recovers quicker due to shorter T_1
3. Fat passes through the zero line at an inversion time $TI = T_1 \ln(2)$
4. Imaging sequence begins at this point
5. Water still has a longitudinal (reduced) component (negative) but is displayed as a positive value in a magnitude image

Short Tau Inversion Recovery (STIR) / Turbo inversion recovery magnitude (TIRM)



Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.

Short **T**au **I**nversion **R**ecovery (**STIR**) / **T**urbo **I**nversion **R**ecovery **M**agnitude (**TIRM**)

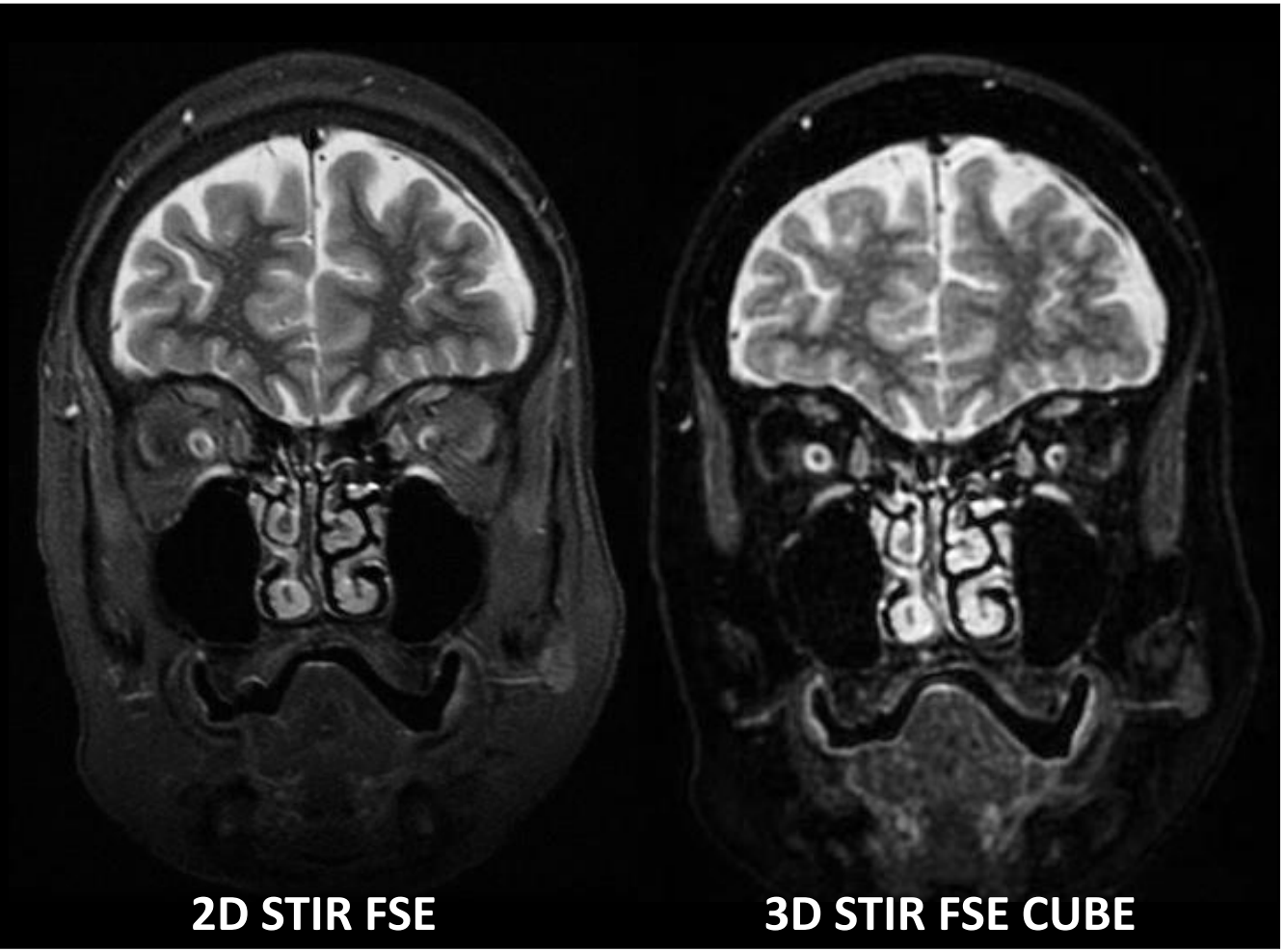
STIR can be implemented as:

2D or 3D TSE/FSE

2D PROPELLER/BLADE

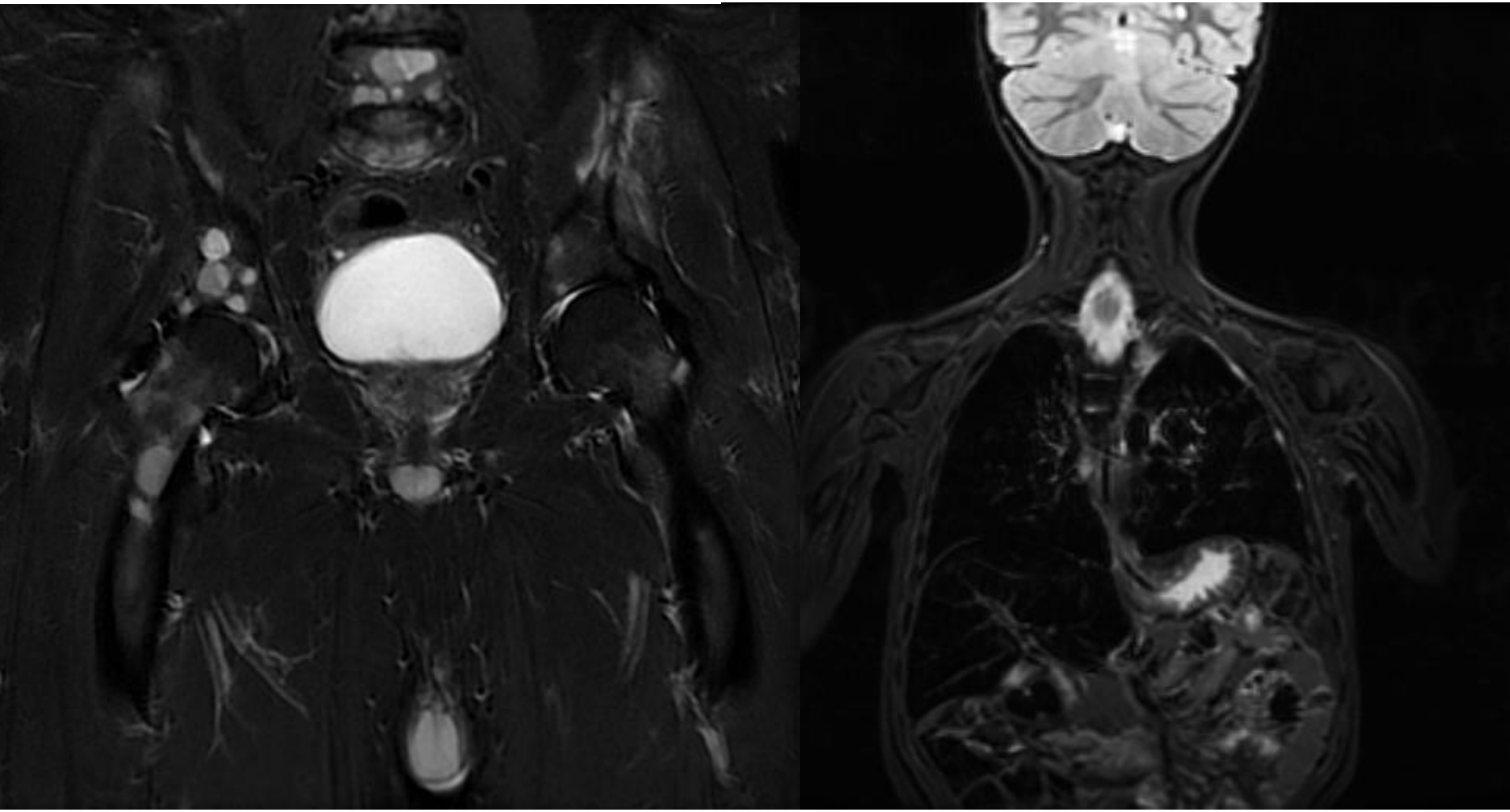
A higher TI can be used for stronger fat suppression

A higher TE can be used for increased T2 weighting



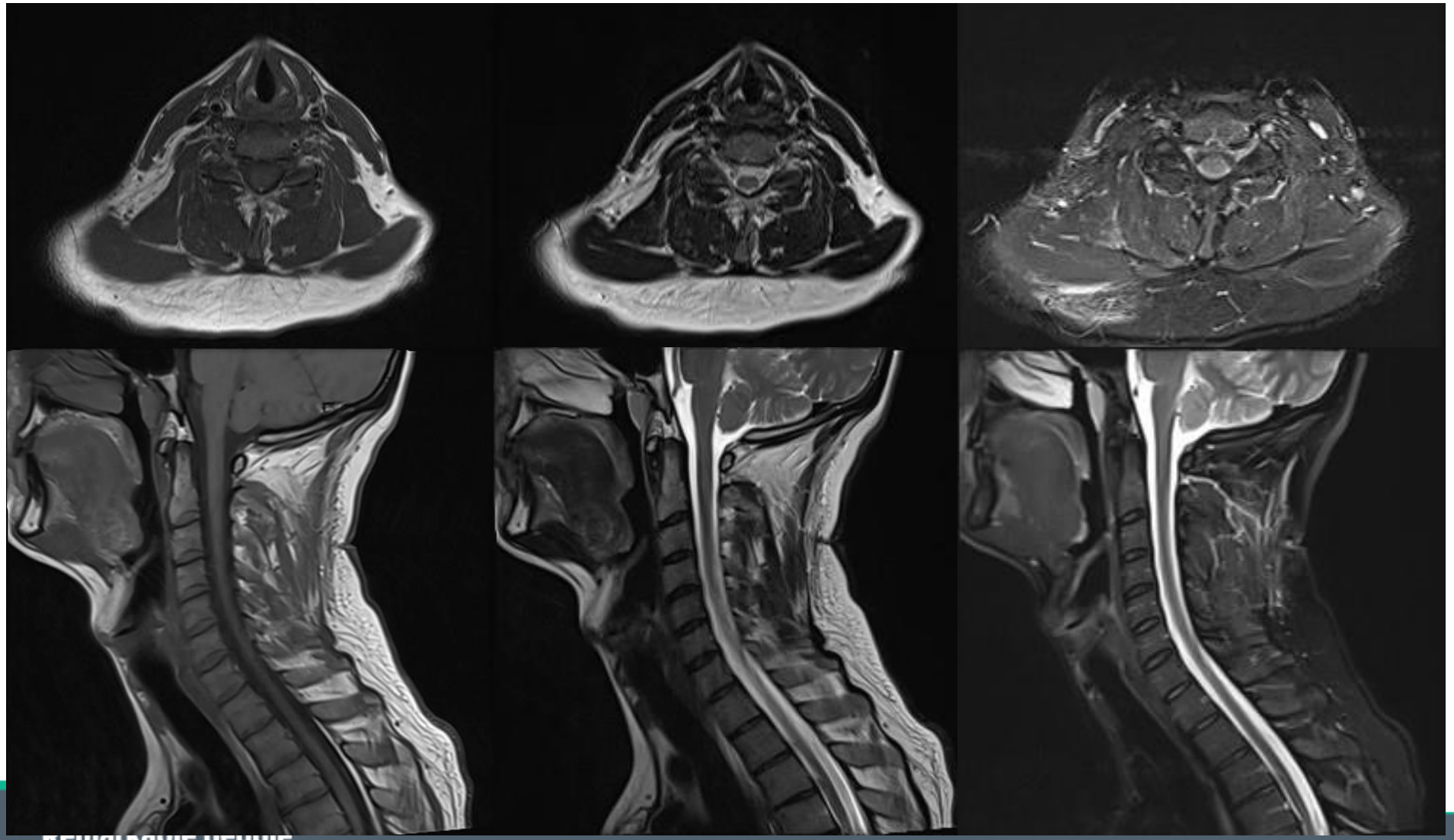
Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.

Short Tau Inversion Recovery (STIR)



Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.

Short Tau Inversion Recovery (STIR)



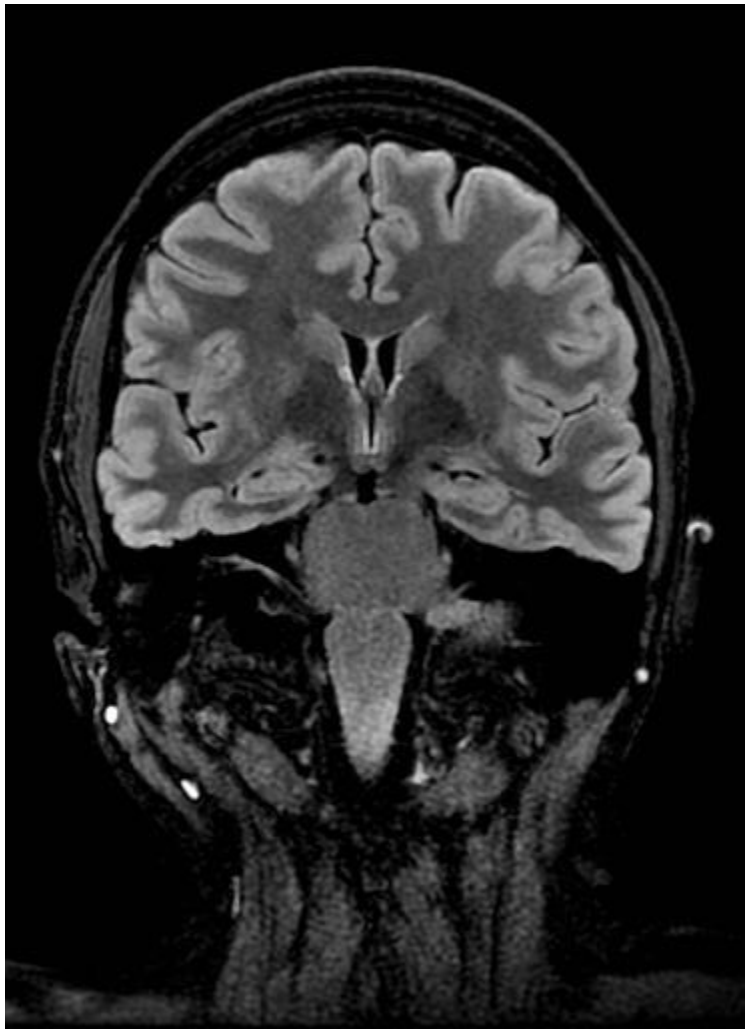
Short Tau Inversion Recovery (STIR)

Advantages	Disadvantages	Suggested applications
Insensitive to B_0 and B_1 inhomogeneities	Inefficient, long scan times	Anywhere CHESS or water excitation fail
Robust fat nulling even over large FOV's	Low SNR Water signal reduced by ~40-50%	Especially good for large FOV, unfavourable anatomy or in the presence of metal
	Suppresses all short T_1 species (Gad)	
	Mixed image weighting (T_1)	
	Limited to T_2 and P.D. (requires long TR)	

T_2 Fluid Attenuated Inversion Recovery (FLAIR)

- Fluid attenuated inversion recovery (FLAIR) sequence, nulls the CSF signal and other water-bound anatomy in the MR image by using a TI selected at or near the bounce point of CSF to permit better evaluation of the surrounding anatomy

- TI is around 1700-2500ms for T_2 FLAIR depending on TR and field strength

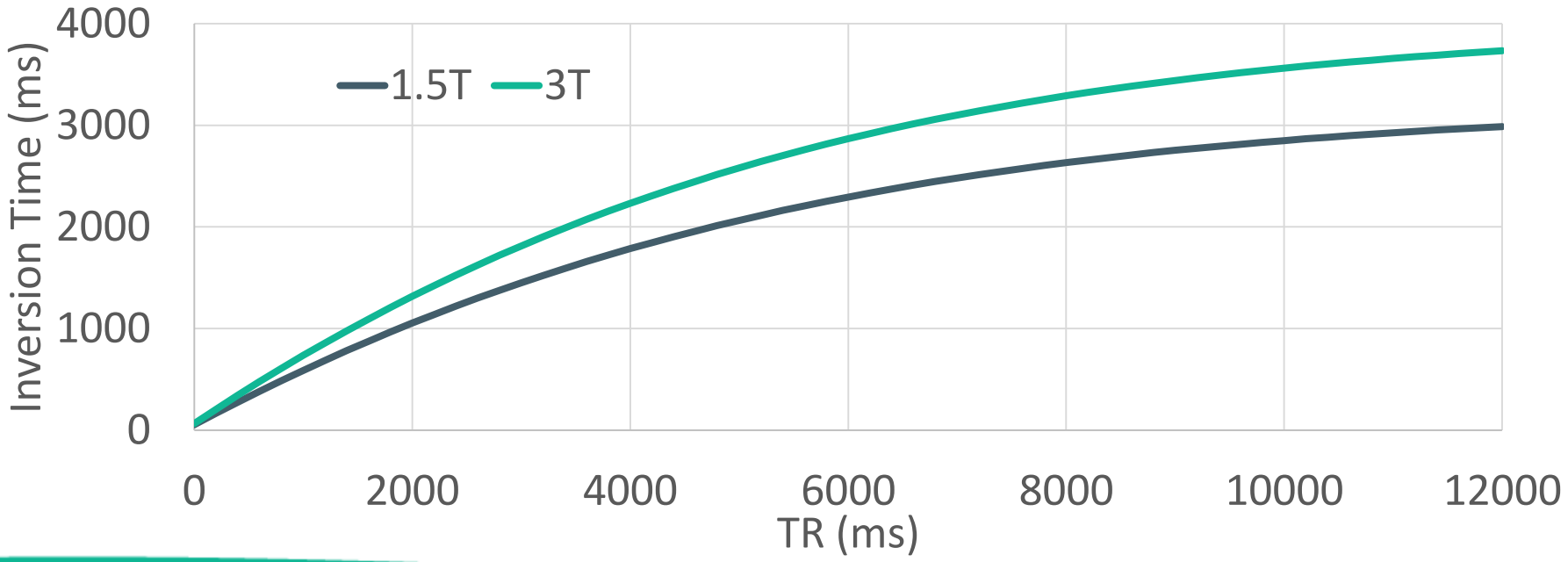


T_2 FLAIR

- TI_{null} depends on the ratio TR/T_1. *CSF T1 @ 1.5T = 4600ms / 3.0T = 5800ms*

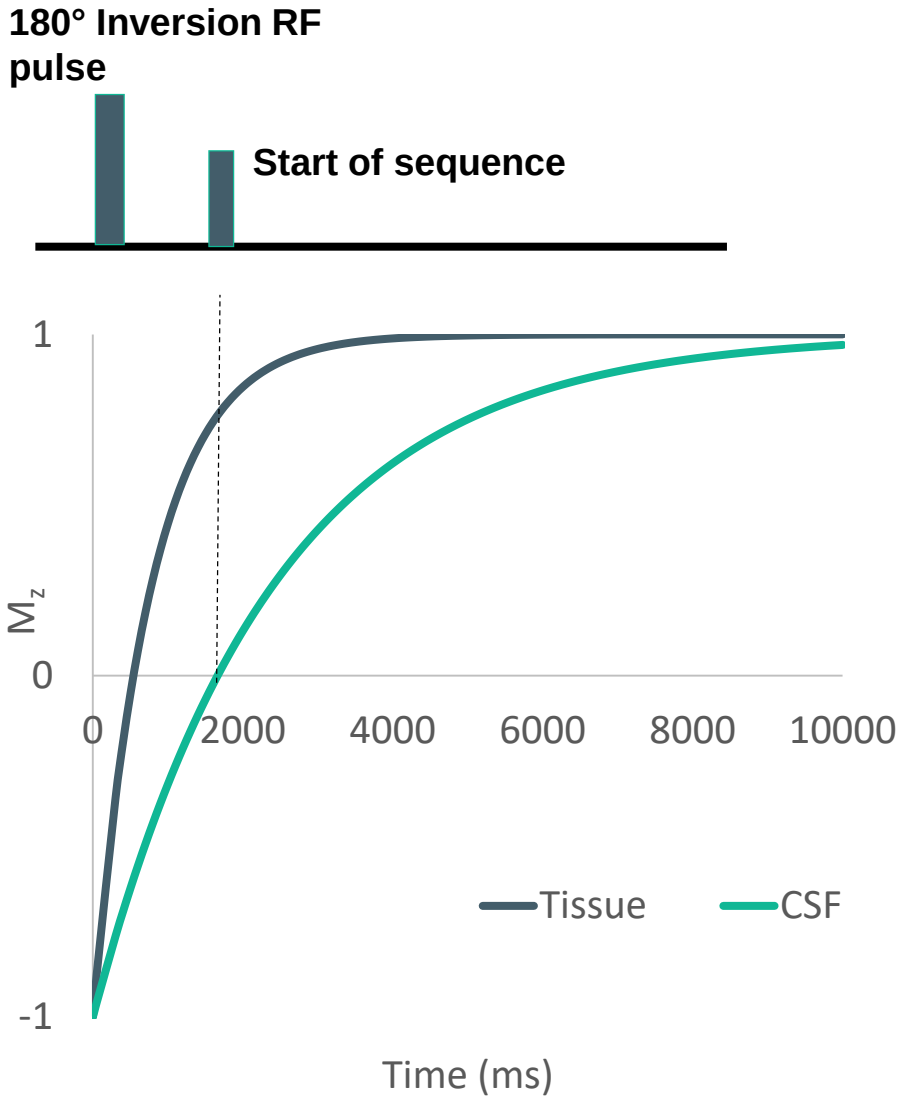
- $TI_{null} = T_1 \times (\ln(2) - \ln(1+e^{-TR/T1}))$ for spin echo
- $TI_{null} = T_1 \times (\ln(2) - \ln(1+e^{-TR-TElast/T1}))$ for FSE/TSE

- When TR is not $>>T_1$ (~5x), the equation cannot be simplified like in STIR.



https://itis.swiss/virtual-population/tissue-properties/database/relaxation-times/

Suppression: STIR & FLAIR. The role(s) of TR (and T1) in determining null point.



T_2 Fluid Attenuated Inversion Recovery (FLAIR)

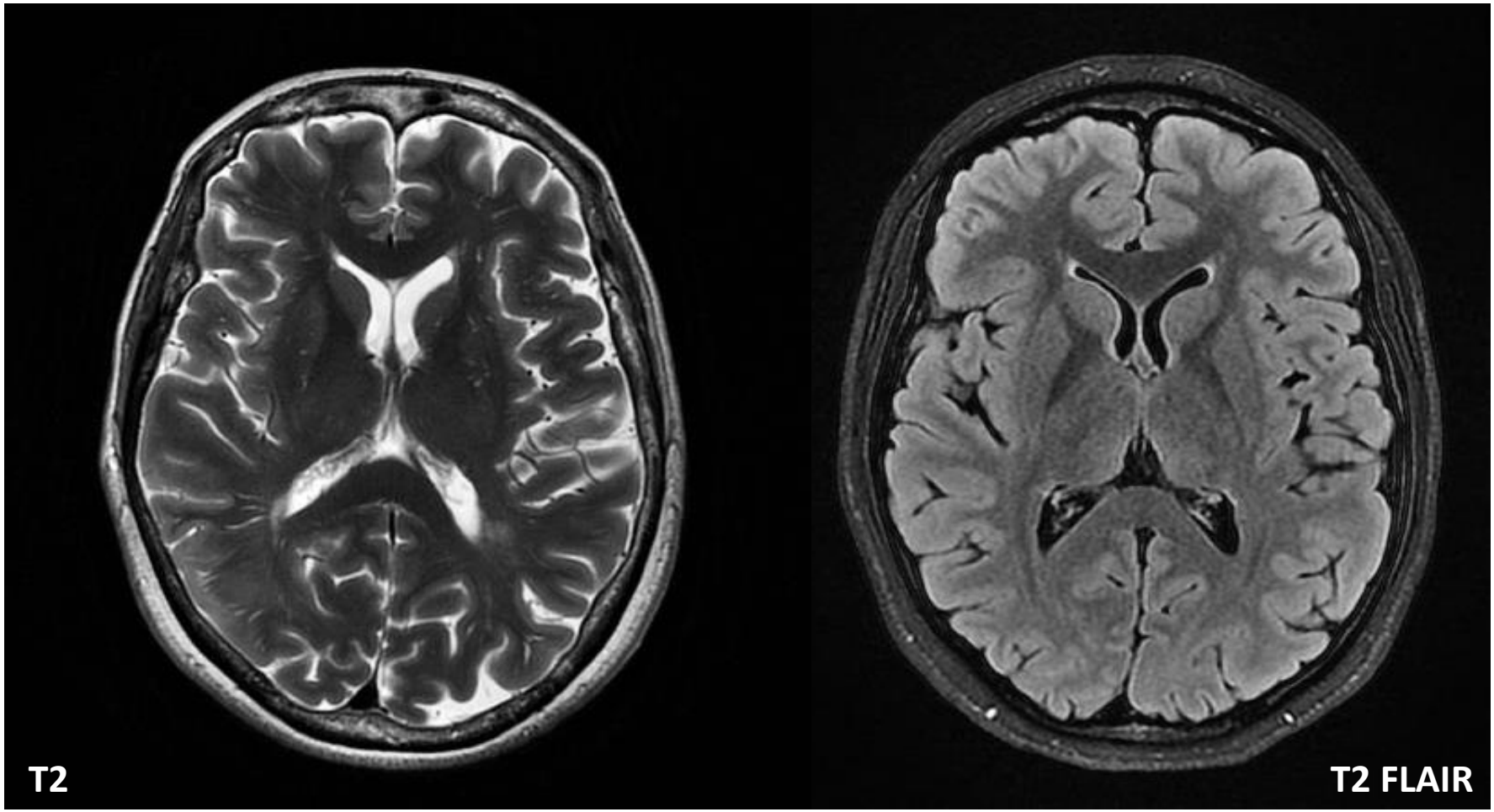
1. Initial 180° inversion pulse. Tissue and CSF signal begin T_1 relaxation from $-M_z$

2. CSF signal recovers slower due to longer T_1 relaxation time

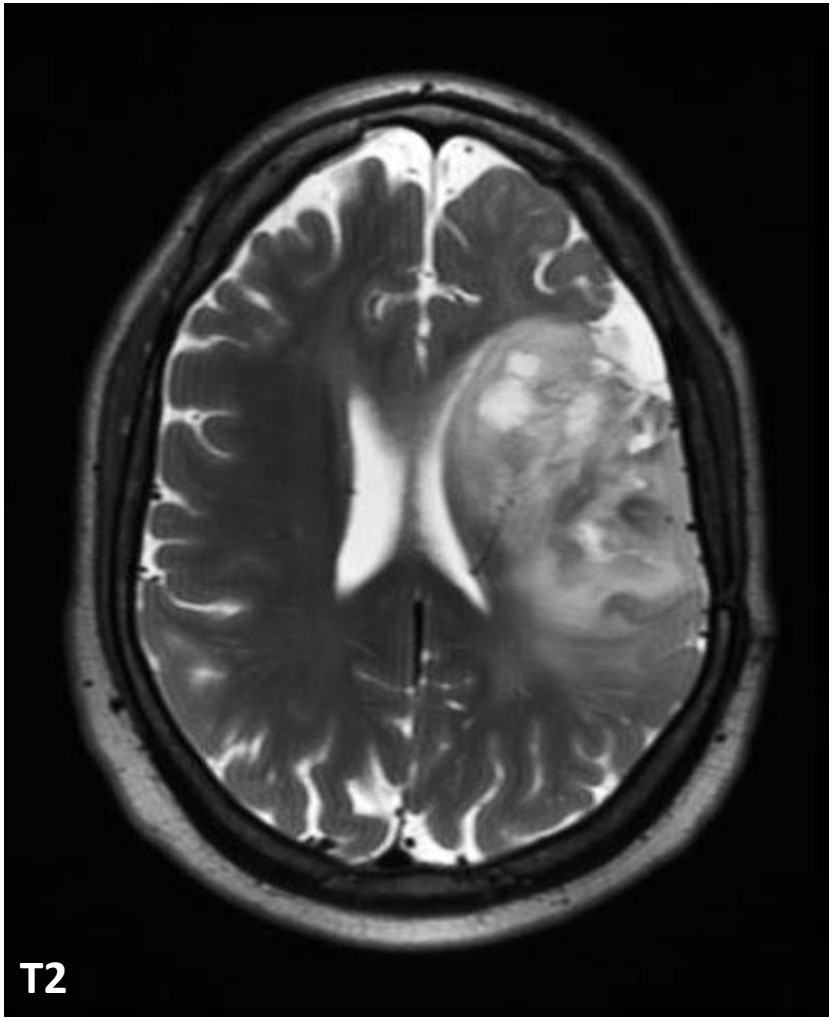
3. CSF passes through the zero M_z at an inversion time
   $TI = T_1 \times (\ln(2) - \ln(1+e^{-TR-TElast/T1}))$ for FSE/TSE

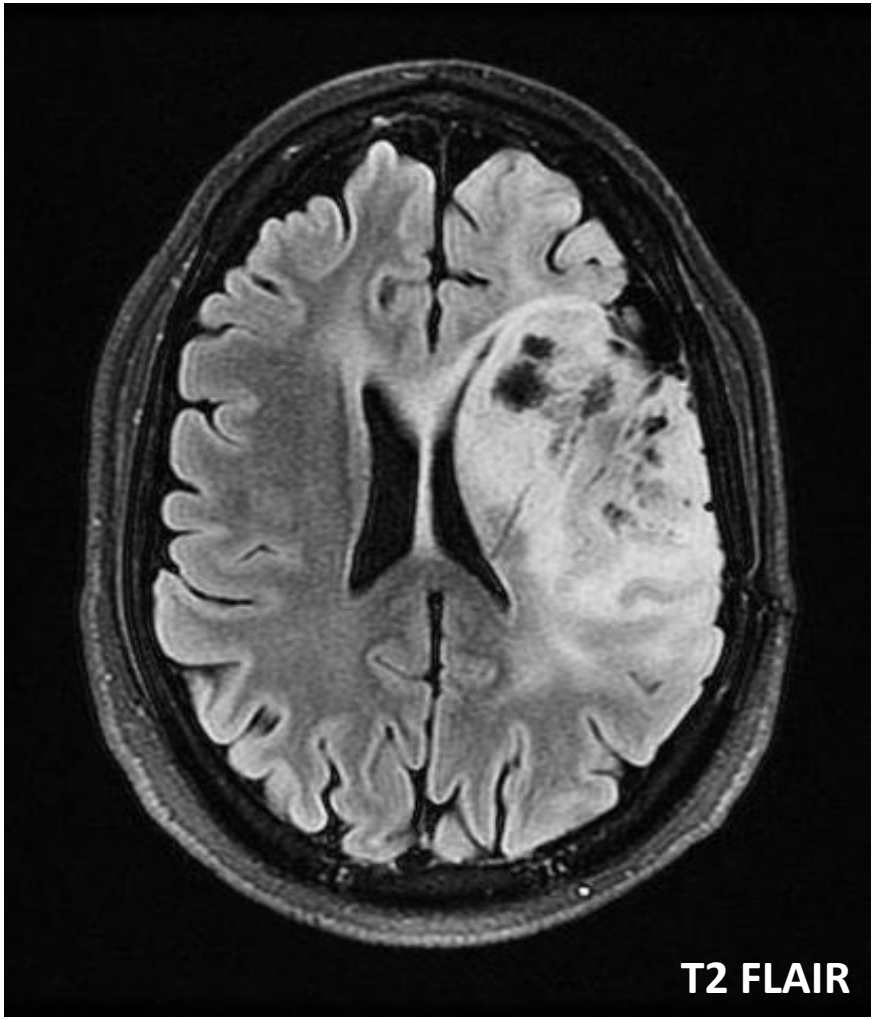
4. Imaging sequence begins at this point

T_2 Fluid Attenuated Inversion Recovery (FLAIR)

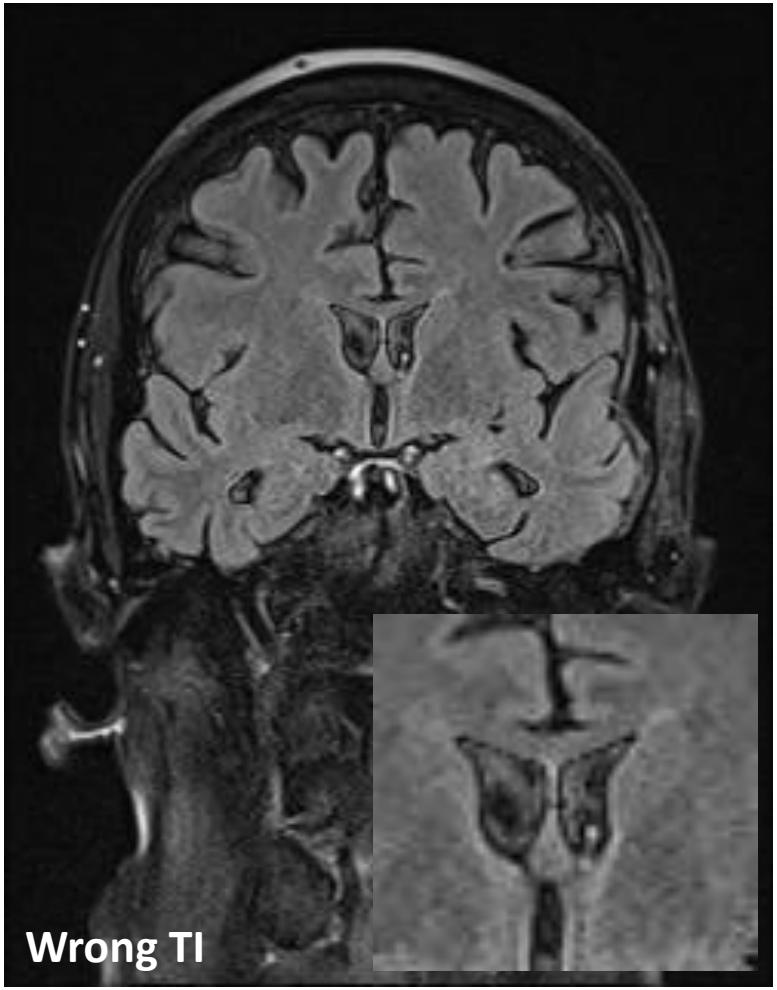


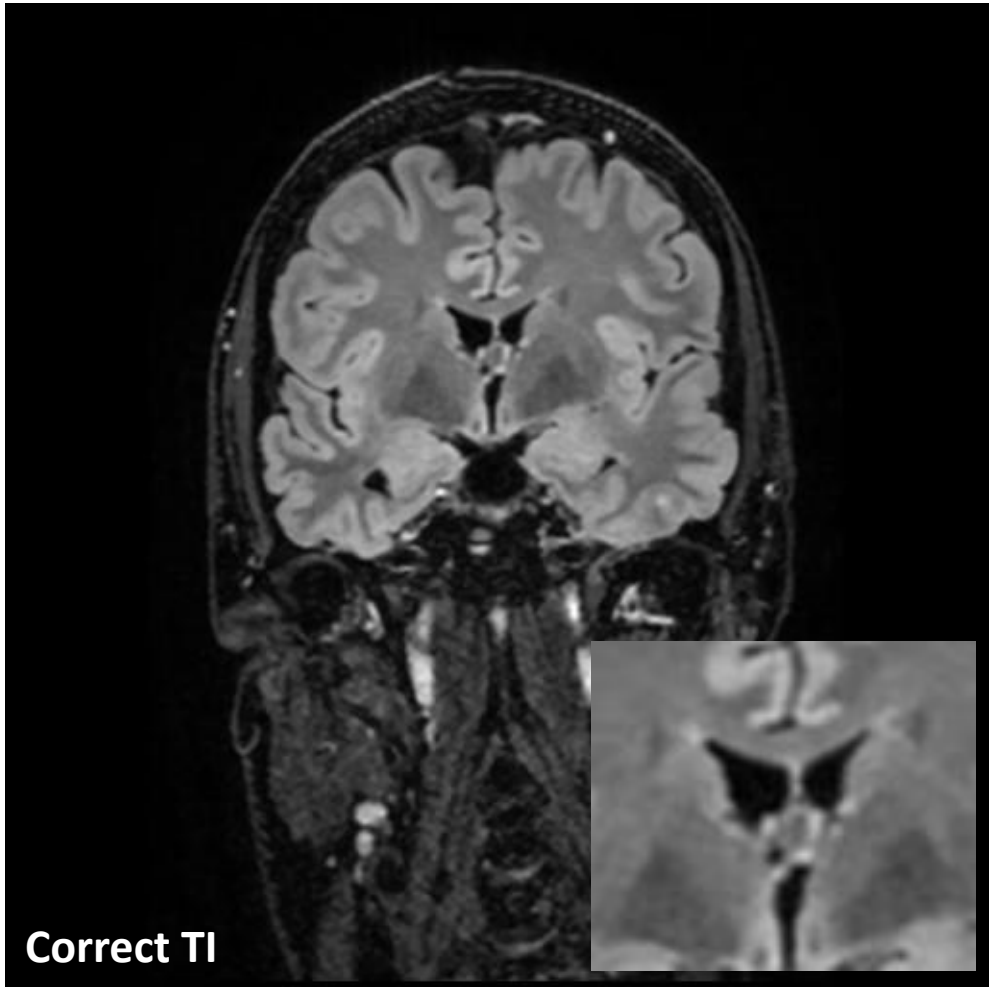
T_2 Fluid Attenuated Inversion Recovery (FLAIR)



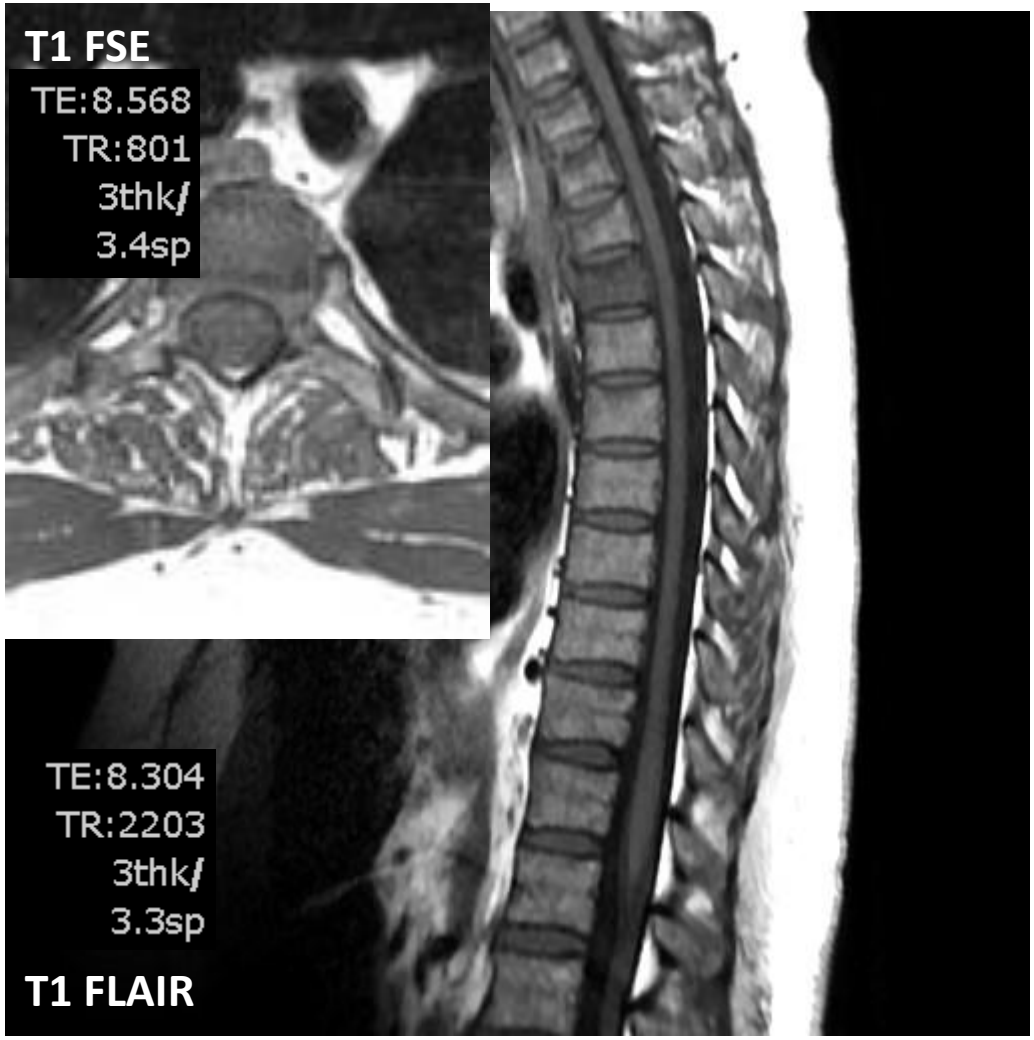


T_2 Fluid Attenuated Inversion Recovery (FLAIR)





T_1 Fluid Attenuated Inversion Recovery (FLAIR)



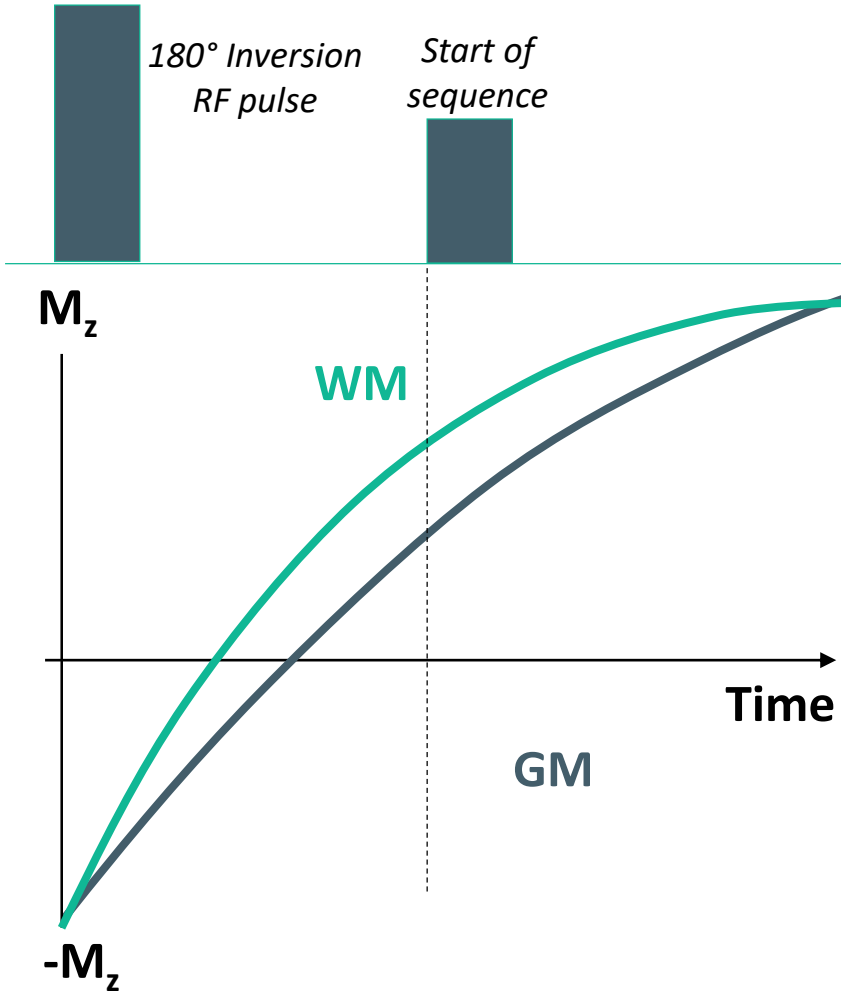
Whilst T_2 weighted FLAIR is the most common variant, T_1W FLAIR is also possible

A T_1W FLAIR will be T_1 weighted with nulling of any tissue with a T1 similar to CSF

At 3.0T the T_1 values of CSF and tissue are longer resulting in poor differentiation of CSF and the cord on spinal imaging

Therefore T_1W FLAIR rather than T_1W imaging is performed for 3T spine imaging

Inversion Recovery Prepped T_1



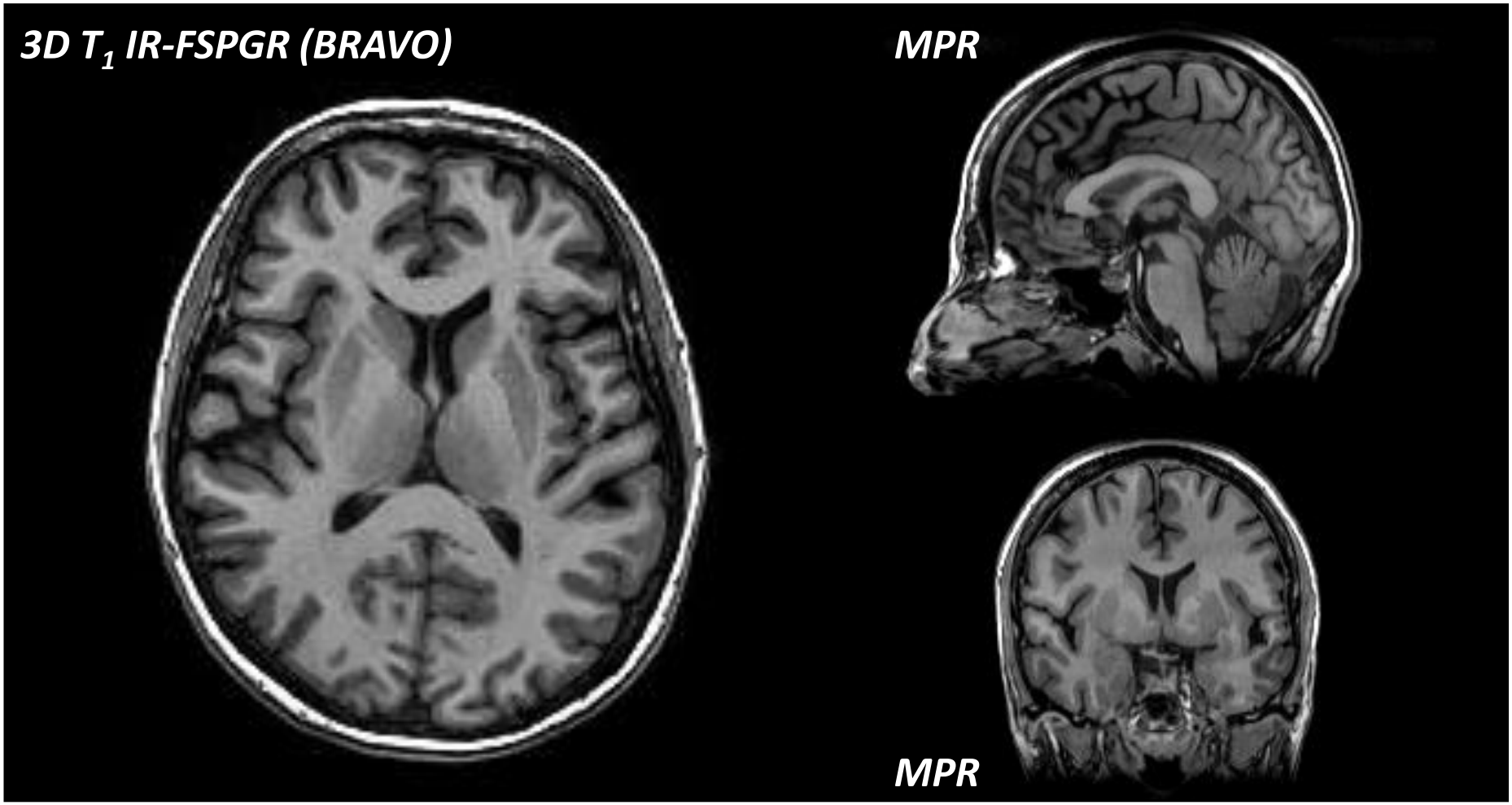
Inversion Recovery Preparation

1. Initial 180° inversion pulse
2. WM and GM signal begin T_1 relaxation from $-M_z$
3. GM signal recovers slower due to longer T_1 relaxation time (900 vs 780ms)
4. When M_z is greatest, imaging sequence begins (TI~450ms)

Tissue	T_1 1.5T (ms)
Fat	260
White Matter	780
Grey Matter	900
Cerebral Spinal Fluid	2400

Increase T_1-weighting e.g. MPRAGE

Inversion Recovery Prepped T_1



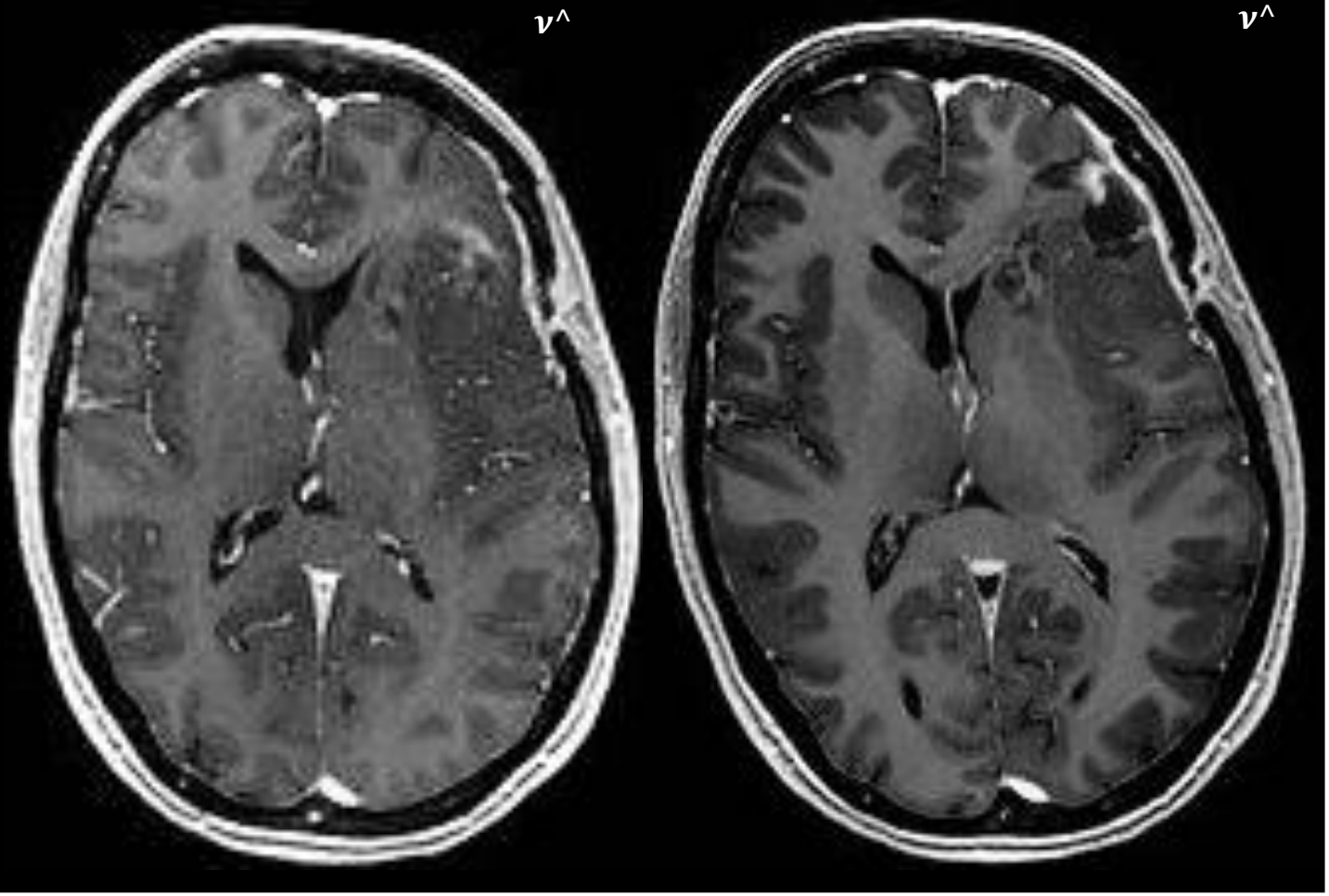
Increase T_1-weighting e.g. MPRAGE

Inversion Recovery Prepped T_1

Hull University Teaching Hospitals NHS Trust

3D T_1 FSPGR +C
FOV = 30cm
Matrix = 256x256
TR/TE = 8.3/3.1ms
SL Thick = 1.3/0mm
Scan Time = 4:17

3D T_1 IR-FSPGR +C
FOV = 25.6cm
Matrix = 256x256
Phase FOV = 0.8
TR/TE = 8.2/3.2ms
TI = 450ms
SL Thick = 1/0mm
ARC = 2 x 1
Scan Time = 3:42



Phase sensitive Inversion Recovery (PSIR)

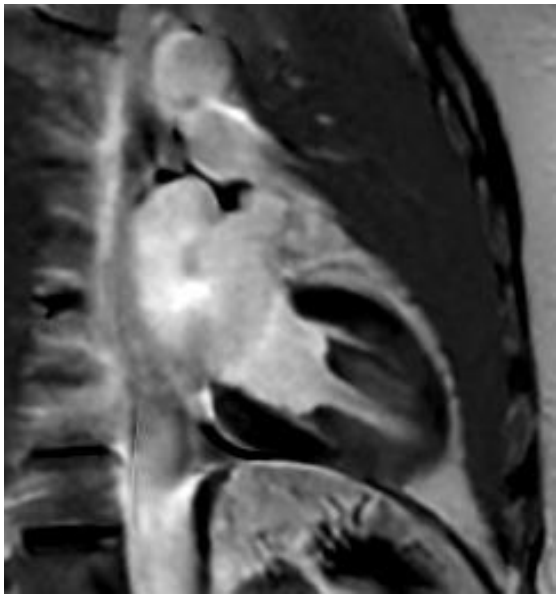
- IR sequences normally use magnitude reconstruction to translate the MR signal into pixel intensity

- Tissue brightness depends only on the magnitude of the longitudinal magnetisation, not its polarity

- Phase-sensitive Inversion Recovery (PSIR) reconstruction preserves the positive and negative polarities of tissues

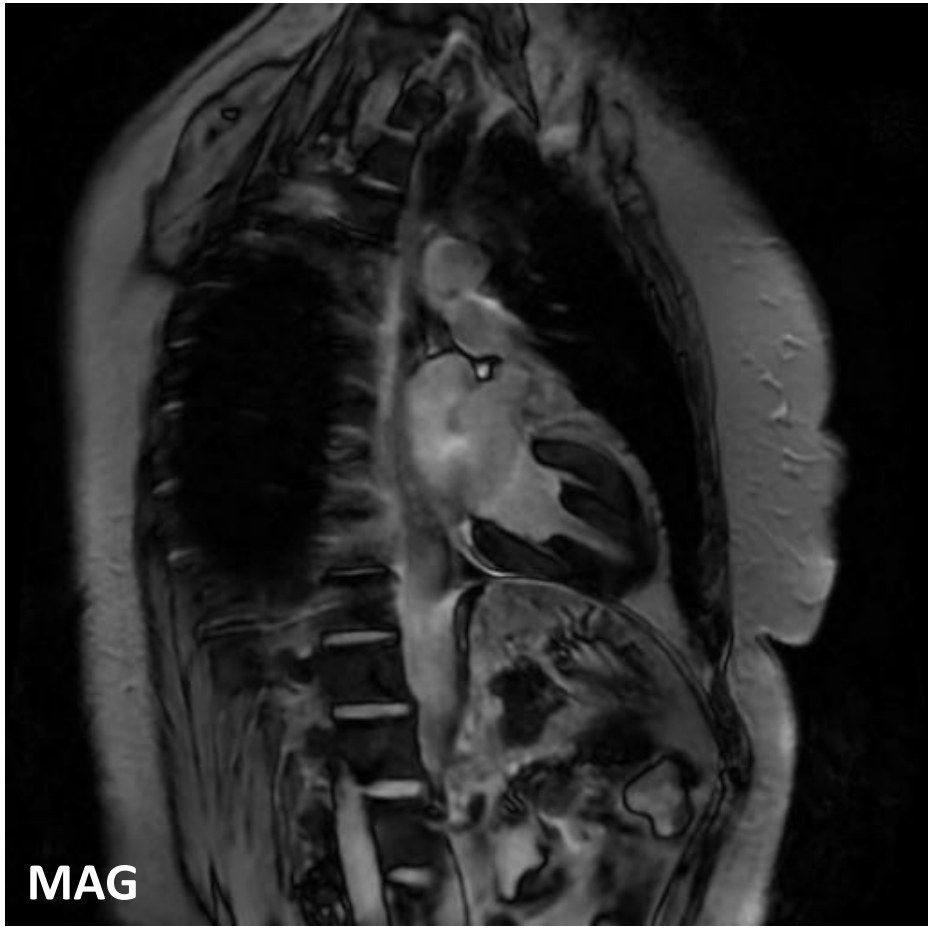
- In PSIR, tissues with more negative longitudinal magnetisation always appear darker than those with more positive magnetisation

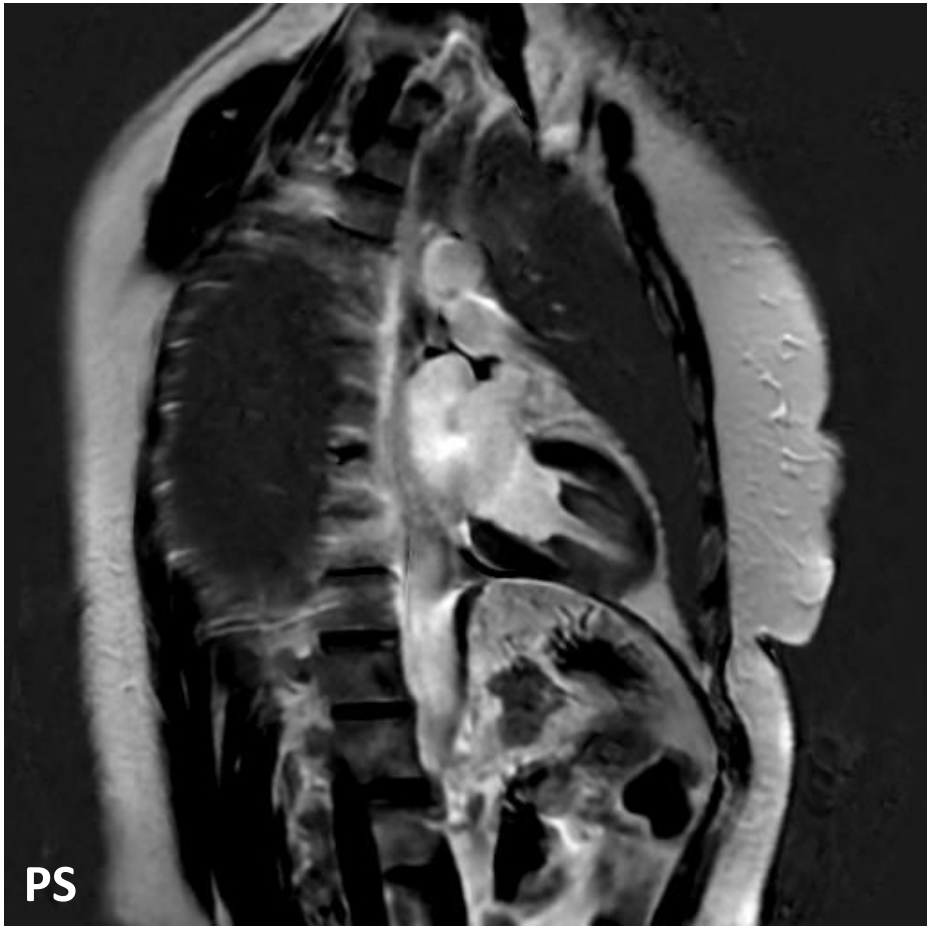
- Contrast-enhancing (infarcted) tissue always has a higher signal than viable myocardium, regardless of the chosen TI



Phase-sensitive IR

PSIR imaging can be further enhanced by optimal estimation of the inversion time using a TI-scouting





7.5 T_1-dependant techniques

- Inversion recovery (IR)
  - *IR emphasises T_1 relaxation times of the tissues by extending the amplitude of the longitudinal recovery by a factor of 2. Uses a 180° pulse to invert spins before beginning pulse sequence.*

- Suppression: STIR & FLAIR. The role(s) of TR (and T_1) in determining null point.
  - *Selection of an appropriate TI can thus suppress tissue signals (e.g., fats/lipids, CSF) depending on their T_1 relaxation times*
  - *STIR. TI = 140-180ms @ 1.5T. Changing TI will effect fat suppression*
  - *T_2 FLAIR. TI = 1700-2500ms @ 1.5T depending on TR*
  - *Both techniques use magnitude reconstruction*

- Increase T_1-weighting e.g. MPRAGE
  - *T_1 values of GM and WM are similar which can result in poor soft tissue contrast. IR-prepped T1 can improve this.*
  - *MPRAGE or BRAVO*

7.5 T_1-dependant techniques

- Phase-sensitive IR
  - *Phase-sensitive Inversion Recovery (PSIR) reconstruction preserves the positive and negative polarities of tissues*
  - *Contrast-enhancing tissue always has a higher signal than viable myocardium, regardless of the chosen TI*

Hull University Teaching Hospitals NHS Trust

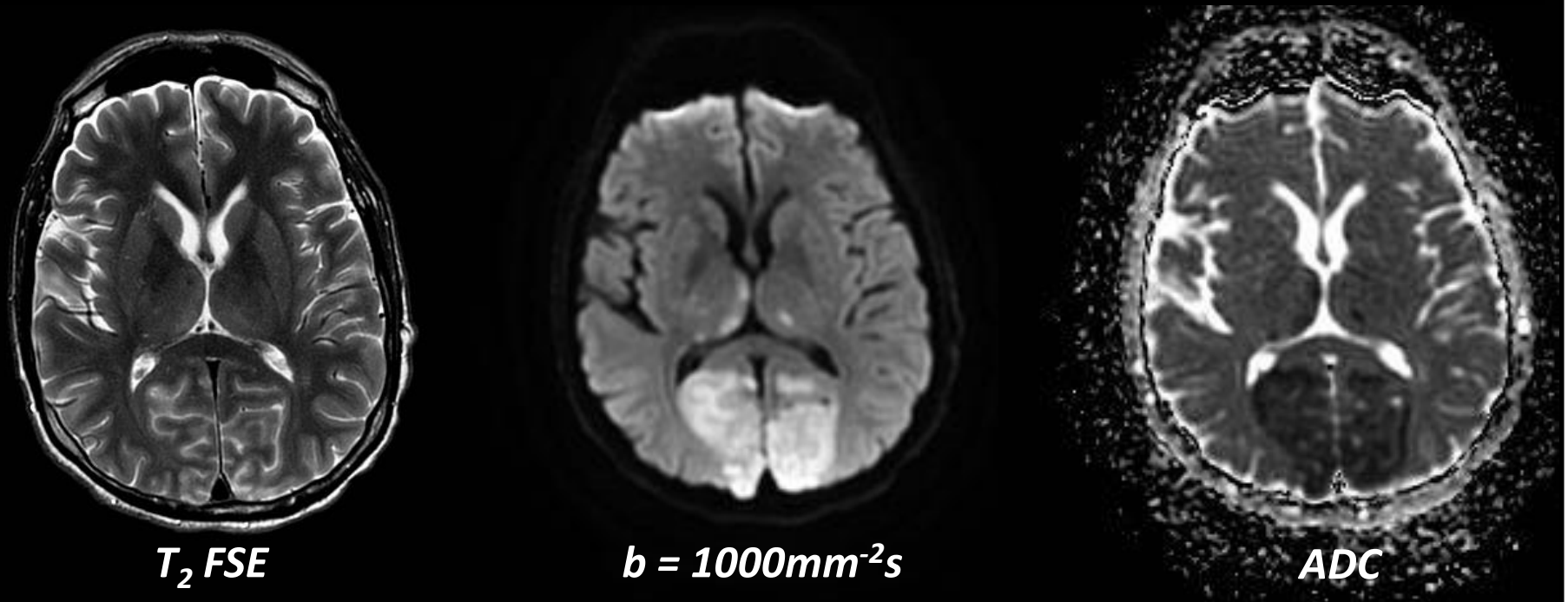
7.6 Diffusion MRI

- Diffusion weighting, relationship with underlying cellularity
- B-values, ADCs and calculated b-values
- Potential perfusion contribution to ADC
- Diffusion anisotropy.

Diffusion weighting, relationship with underlying cellularity

NHS
Hull University Teaching Hospitals
NHS Trust

- DWI can provide real-time information about the extracellular environment



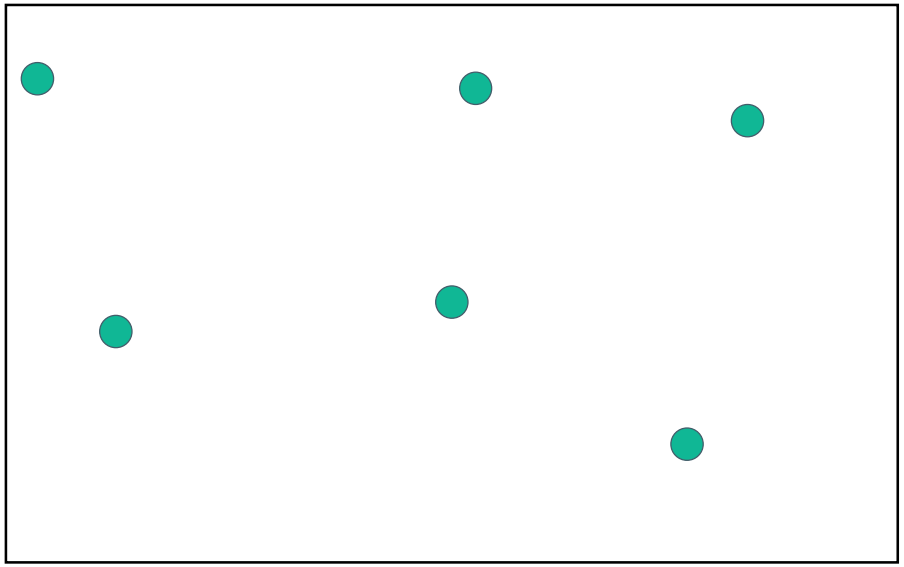
T_2 FSE ***b = 1000mm^{-2}s*** ***ADC***

- Using DWI, alterations in water diffusion can be detected within a few minutes from onset of ischaemia
- Net increase of water detected as an increase of T_2 signal takes 1–4 hours

Thomalla *et al. The Lancet Neurology 2011, pg 978–86*

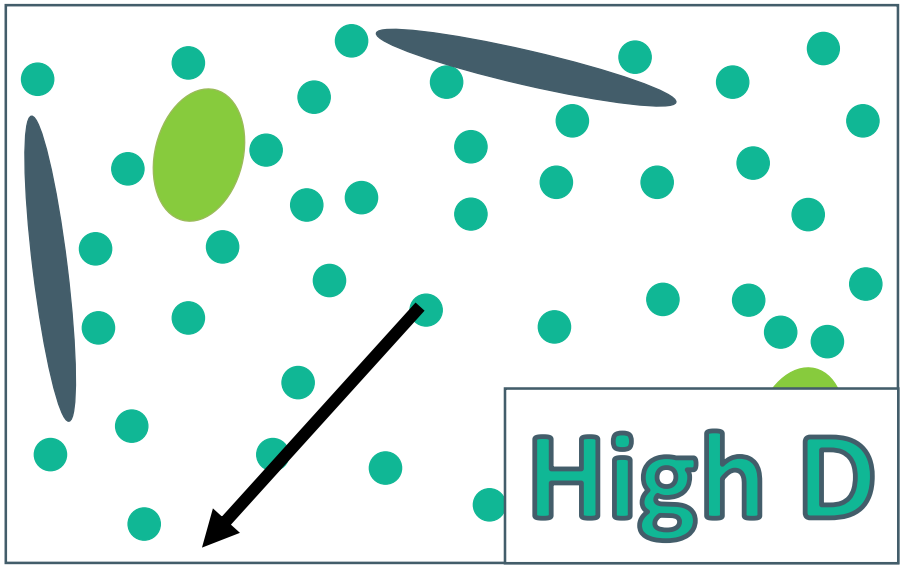
- Random movement ‘walk’ of particles in a fluid or gas is known as: Brownian Motion (Robert Brown, 1827)

- Caused by thermal motion (agitation)



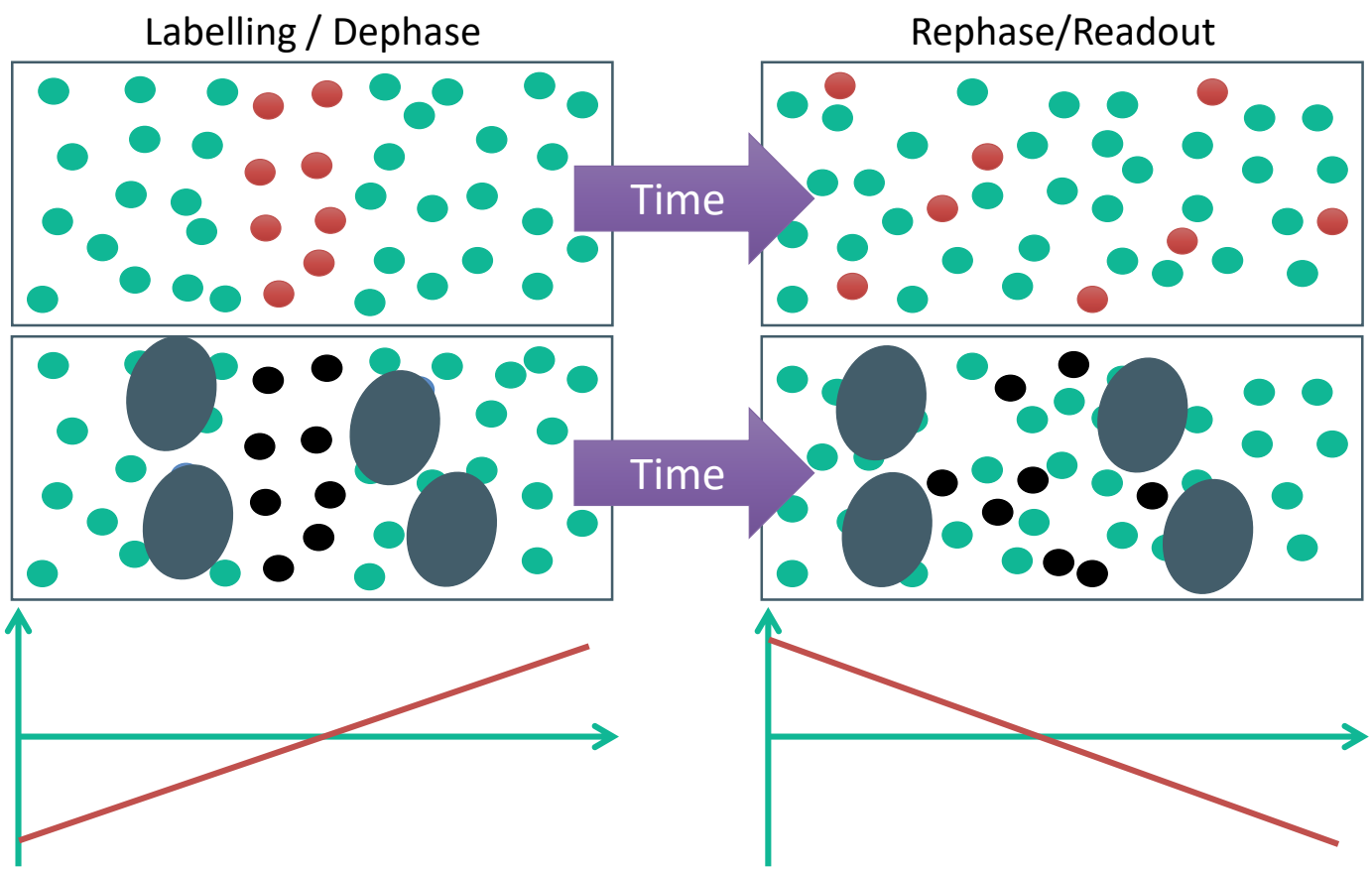
- Free diffusion

Diffusion weighting, relationship with underlying cellularity

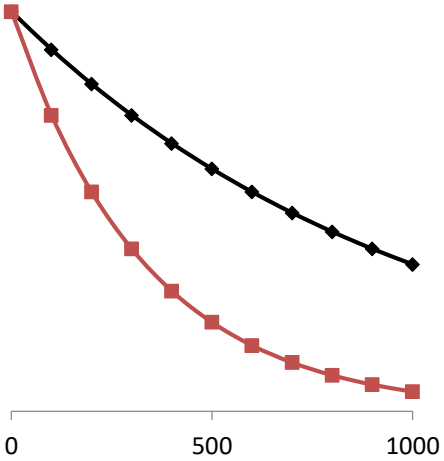


- *in vivo* molecular diffusion in tissue is not free, due to interactions with macromolecules, fibres and membranes
- Reveal microscopic details about tissue architecture
- Einstein's equation for diffusion due to Brownian motion $\langle r^2 \rangle = 6D\tau$
- $\langle r^2 \rangle / 6\tau$ = distance/time = D = self diffusion coefficient

Diffusion Weighting



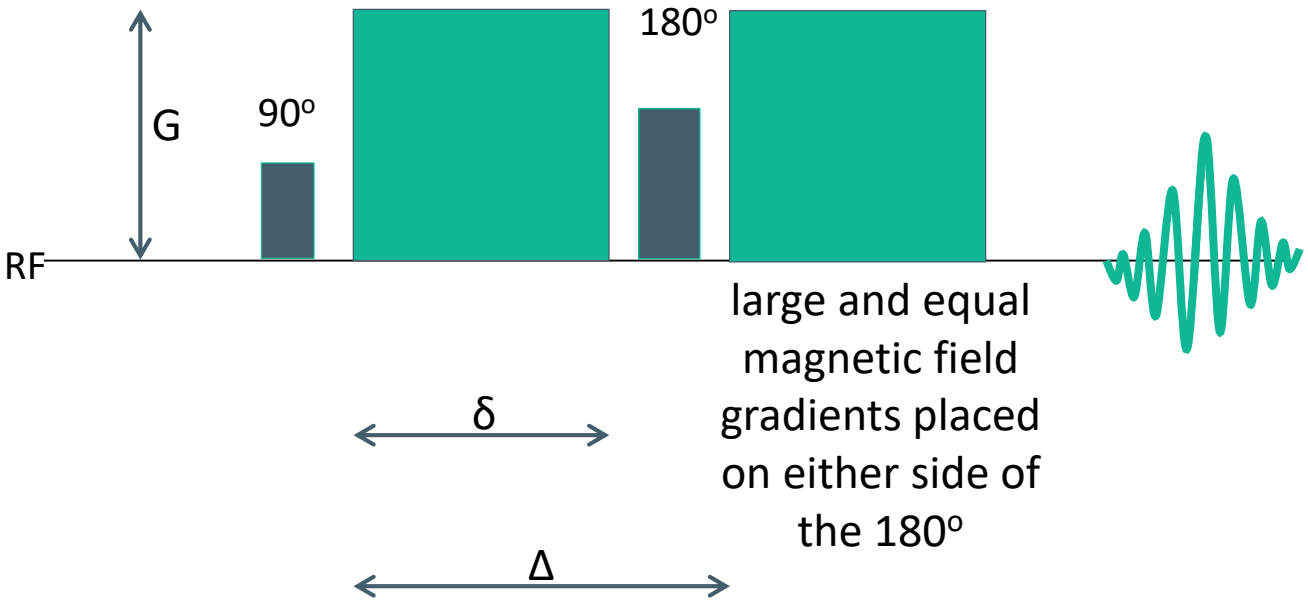
Greater Phase Dispersion
=
Lower Signal



Less Phase Dispersion
=
Higher Signal

Importantly, diffusion can only be measured in a single direction

Pulsed Gradient Spin Echo (and changing b-value)

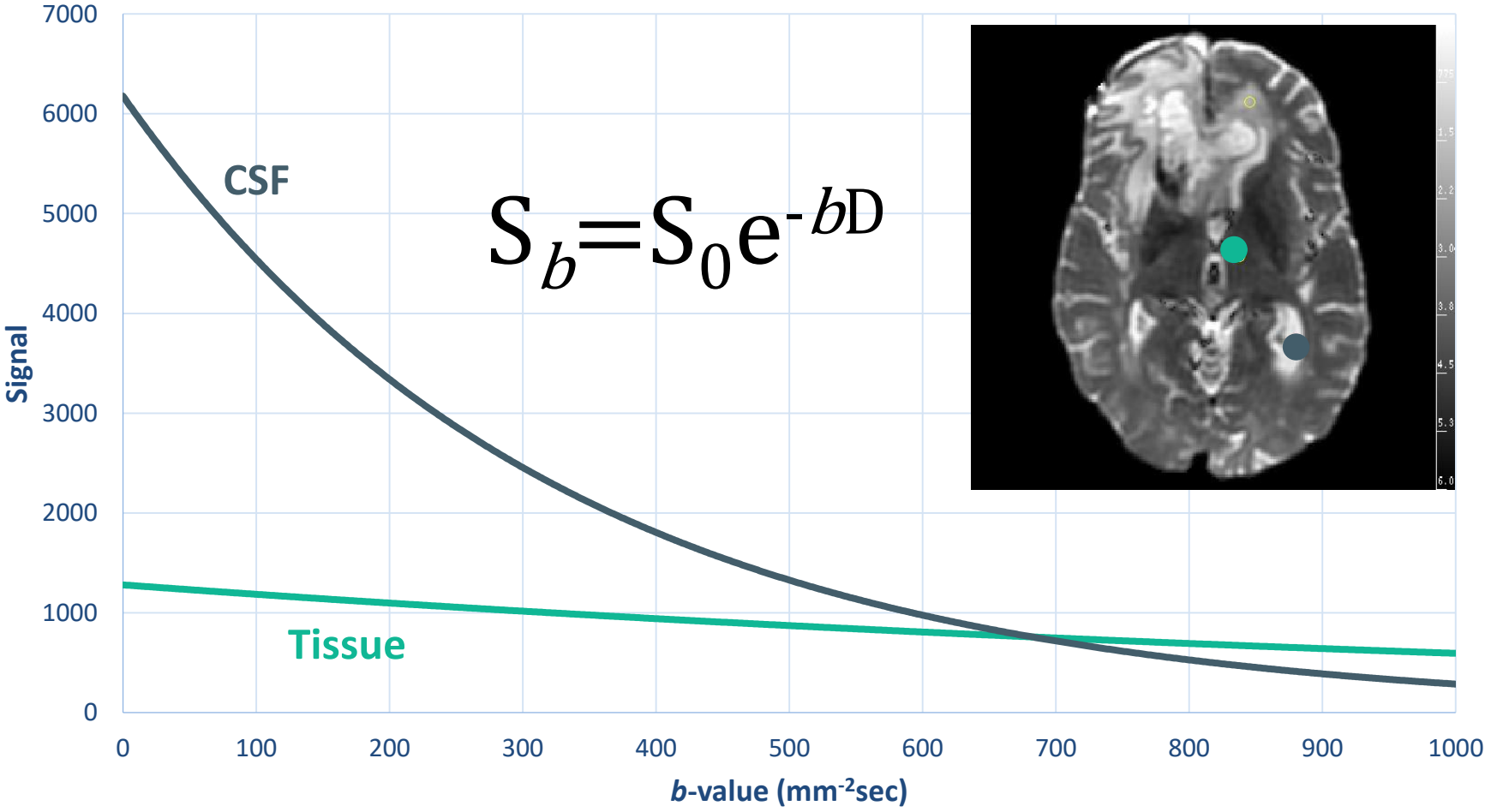


b-value (diffusion sensitivity)	$b=\gamma^2\delta^2G^2T_D$	($sec^{-1}m^2$)
diffusion time (τ)	$T_D=\Delta-\delta/3$	(sec)

γ = gyromagnetic ratio, **G** = strength of the gradient pulse
δ = duration of the pulse, **Δ** time between the two pulses

Stejskal and Tanner. *Spin Diffusion Measurements: Spin Echoes in the Presence of a Time Dependent Field Gradient J. Chem. Phys. 42, 288 (1965)*

Signal Decay – mono-exponential model for extracellular diffusion



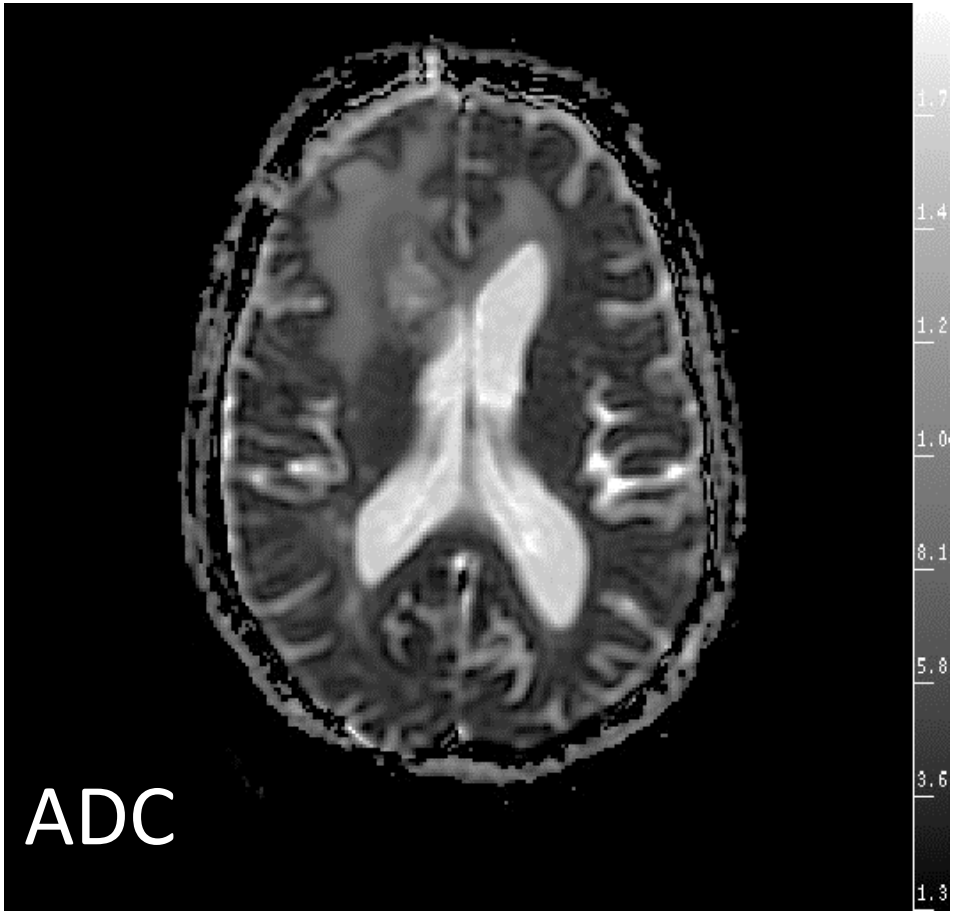
Stejskal and Tanner. *Spin Diffusion Measurements: Spin Echoes in the Presence of a Time Dependent Field Gradient J. Chem. Phys. 42, 288 (1965)*

Apparent Diffusion Coefficient (ADC)

$$S_b = S_0 e^{-bD}$$

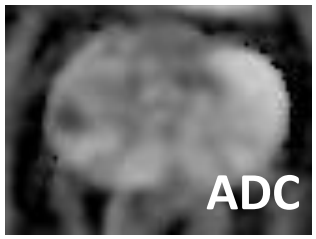
$$ADC = D = \frac{-1}{b} \ln \frac{S_b}{S_0}$$

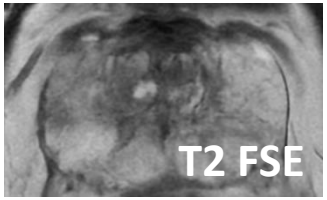
$$ADC = mm^2/sec$$

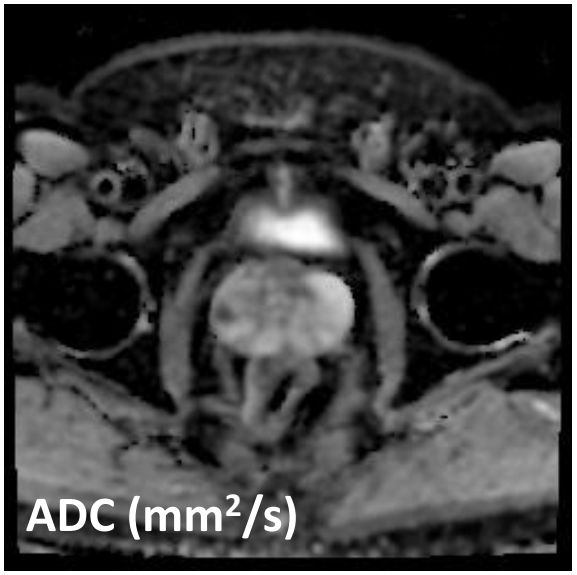


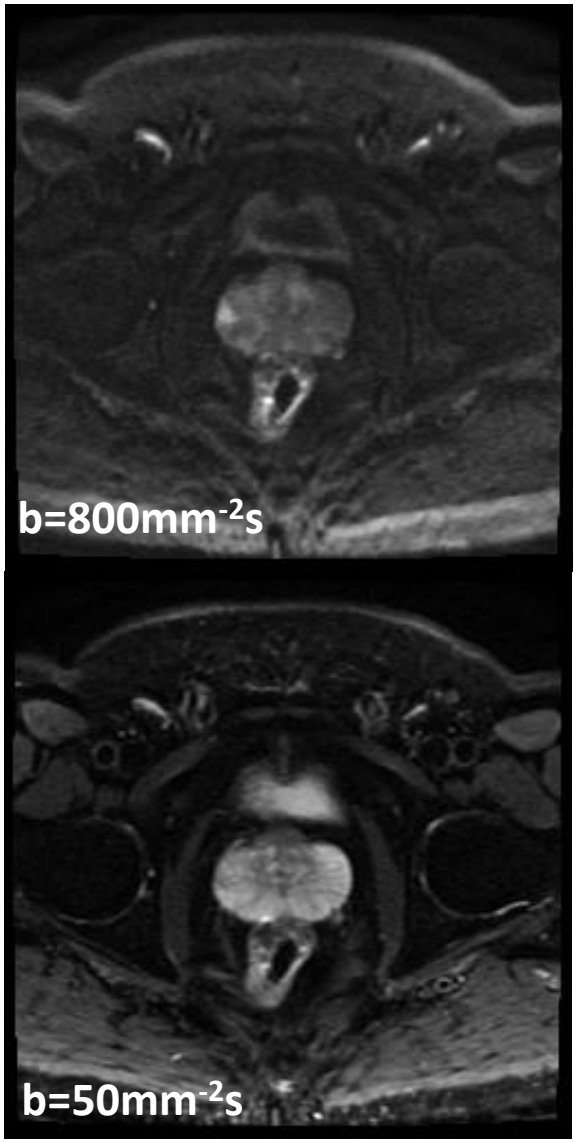
Apparent Diffusion Coefficient (ADC)

$$\mathrm{ADC} = \mathrm{D} = \frac{-1}{b} \ln \frac{S_{high}}{S_{low}}$$



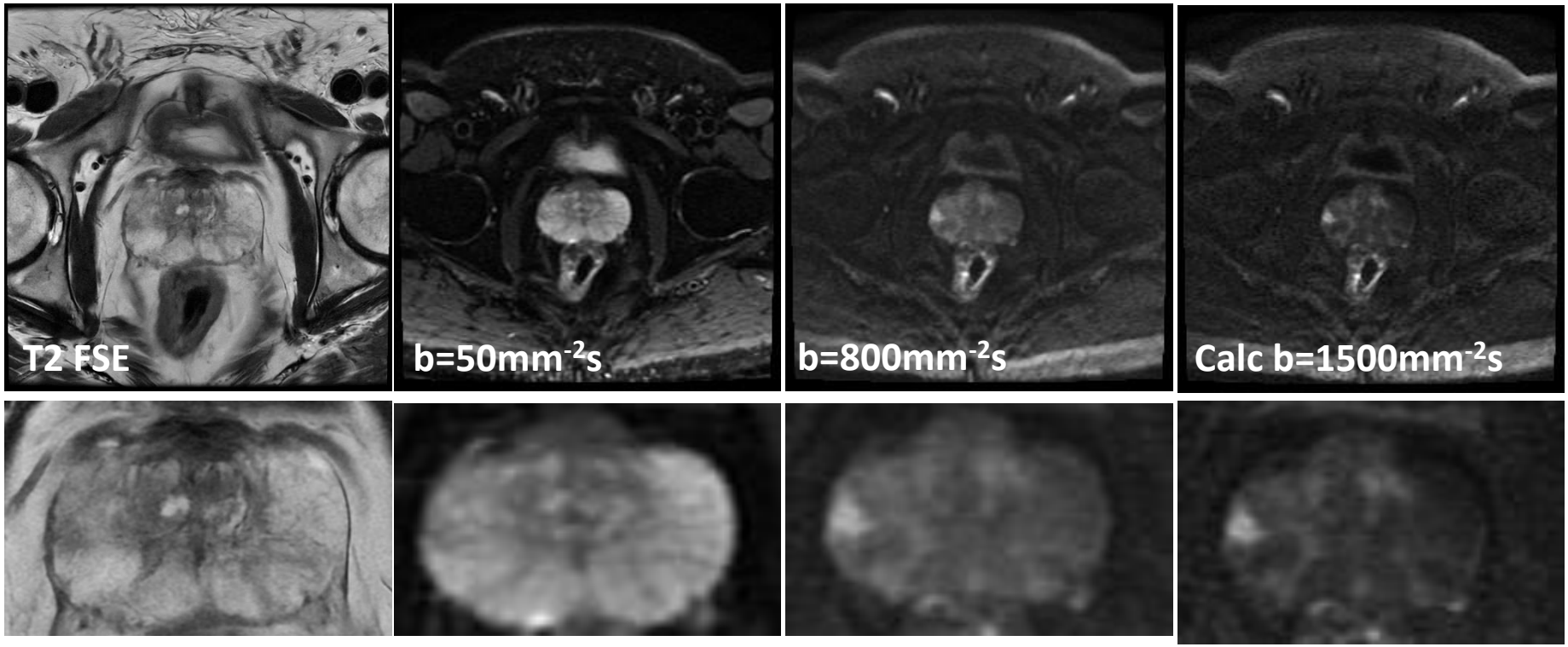






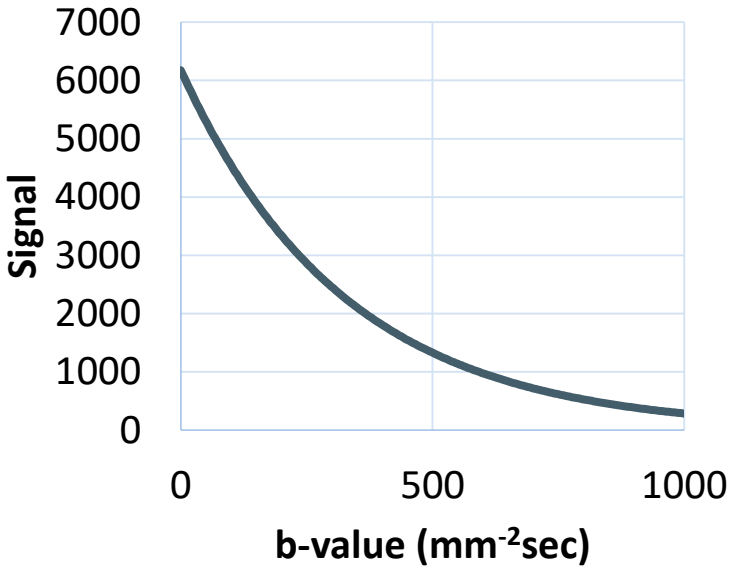
Calculated b-values

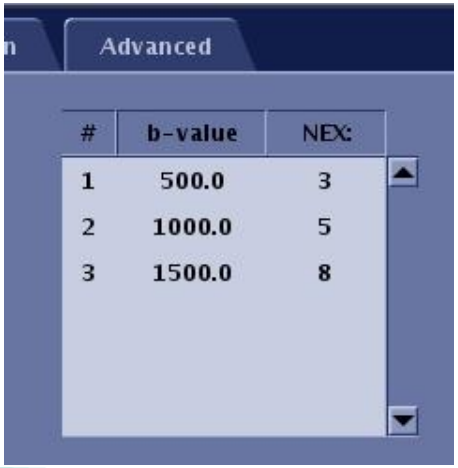
- Once the ADC value for each voxel is calculated from 2 or more b-values, it's possible to extrapolate the fit to generate synthetic/calculated b-value images ($S_b=S_0e^{-b.ADC}$)



Benefits of Calculated b-values

- The MR diffusion weighted signal decays with increasing *b*-value
- Higher b-values have lower signal AND the higher b-value requires a longer TE which in turn also causes signal loss
- Calculated/synthetic diffusion generates the high b-value image contrast without scan time or signal penalties
- The shorter TE also results in less EPI distortion
- Higher b-value images usually increase lesion conspicuity

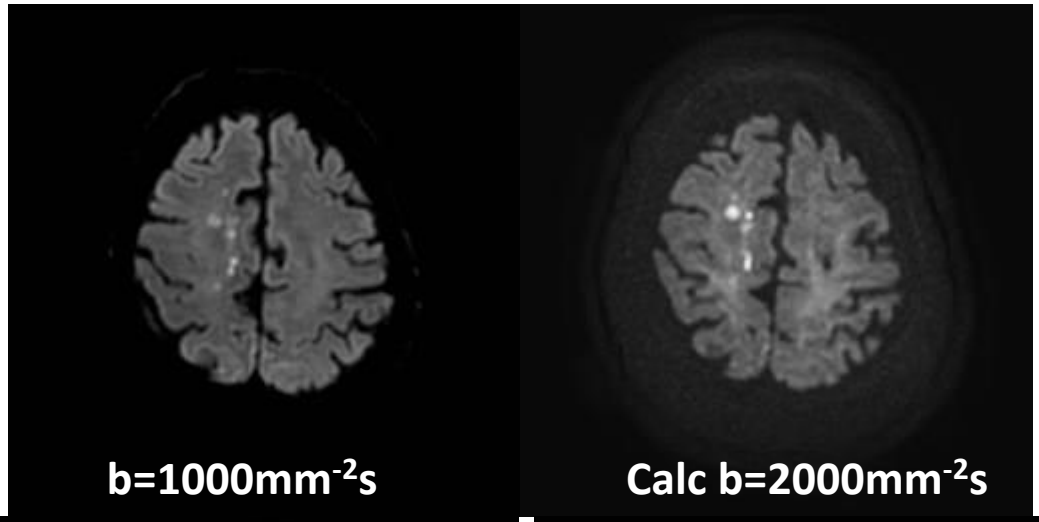


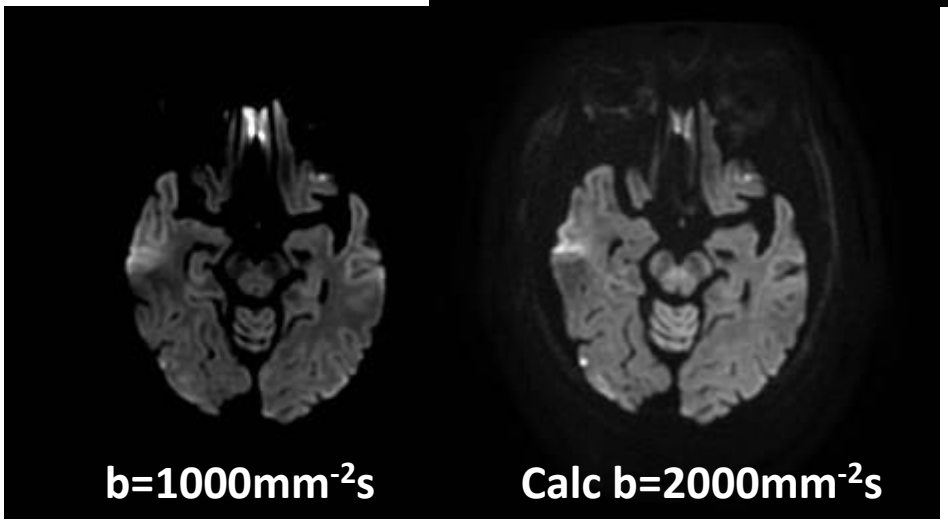


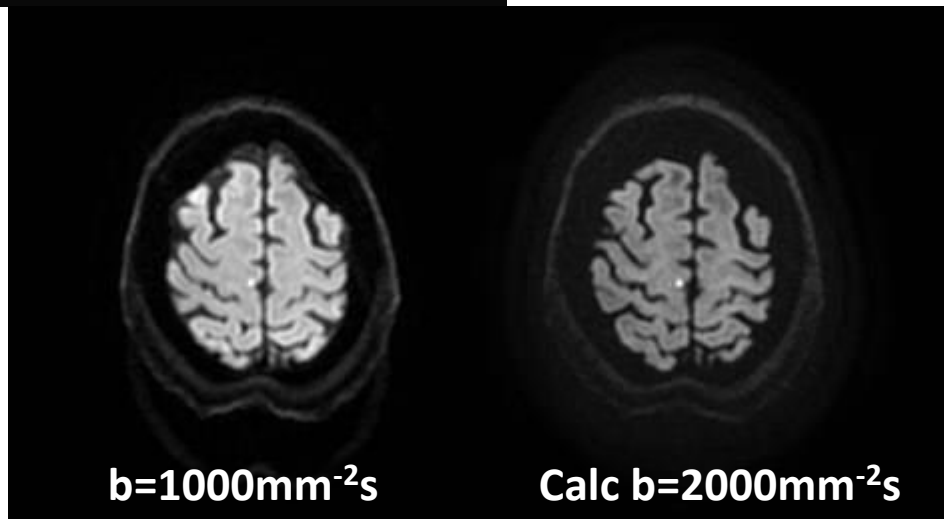
B-values, ADCs and calculated b-values

NHS
Hull University Teaching Hospitals
NHS Trust

Stroke





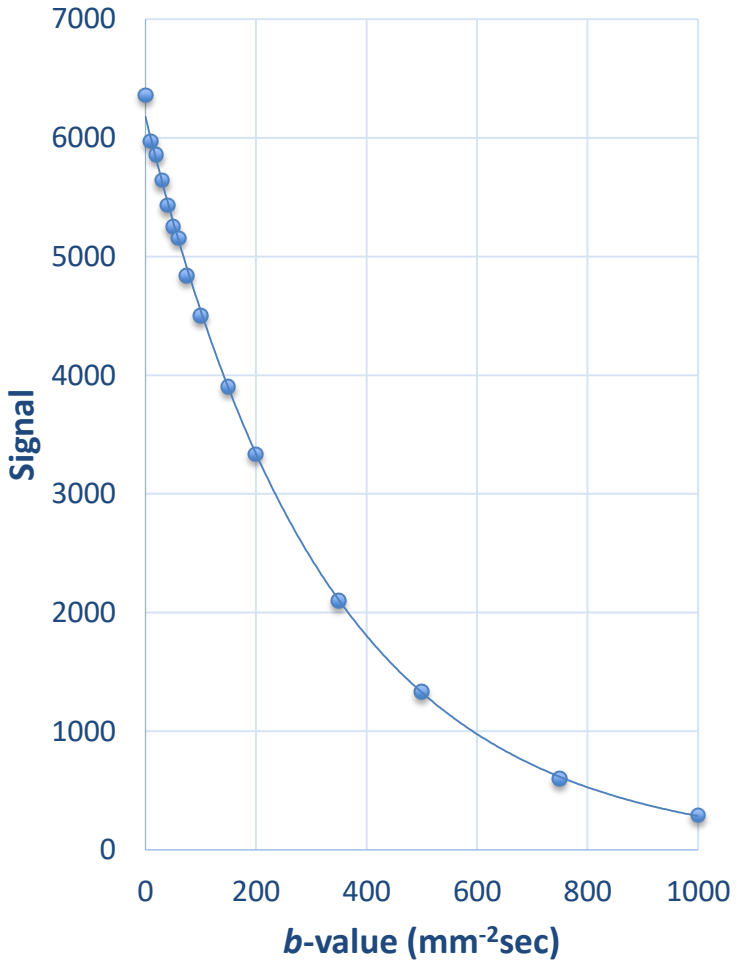


b-value choice

- *b*-values should attenuate the healthy background tissue more than the lesion
- optimal *b*-value is approximately the reciprocal ADC value of normal background tissue WM+GM+CSF
- Brain = ~$1.0x10^{-3}mm^2sec^{-1}$ => $1/1.0x10^{-3} = 1000mm^{-2}sec$
- Trade-off between signal attenuation from diffusion and background noise

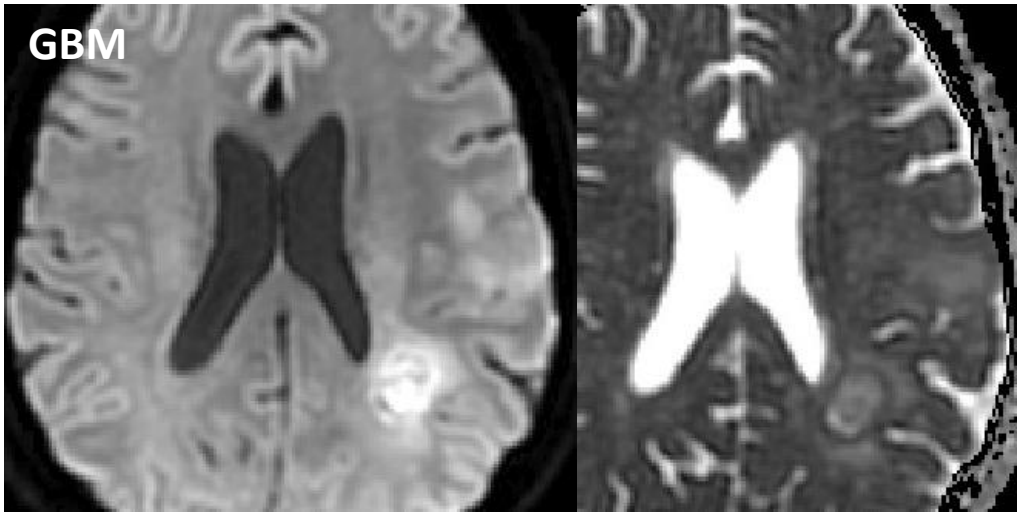
B-values, ADCs and calculated b-values

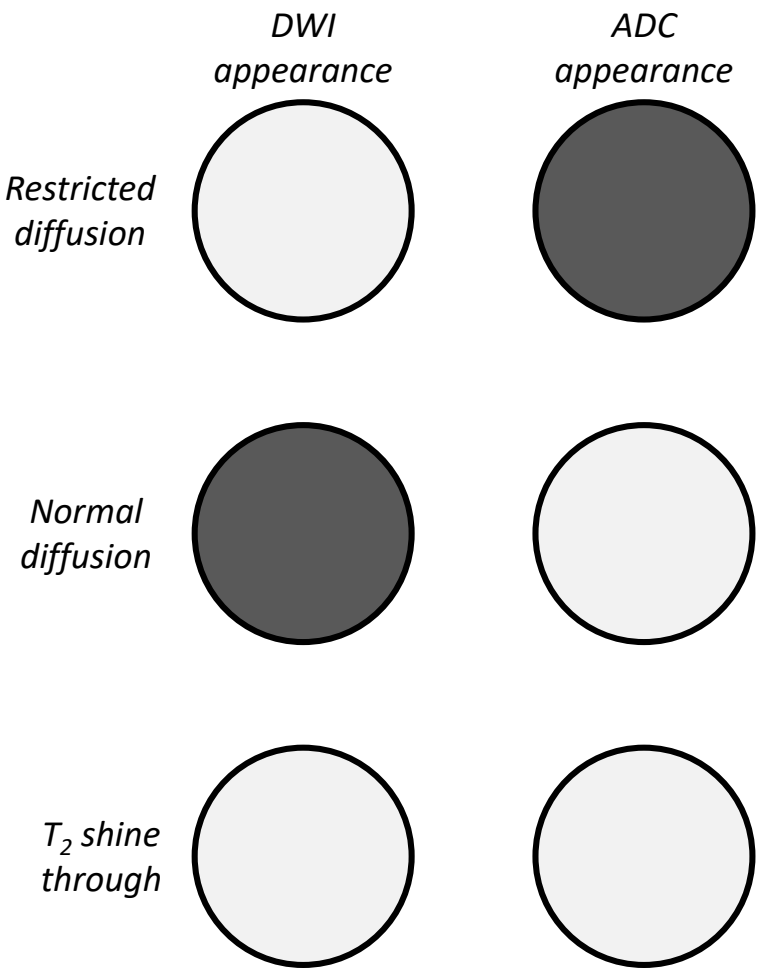
Average/NEX/NSA Assignment



Artefacts - T_2 Shine-Through

- High signal on DW images
  - not due to restricted diffusion
- Caused by long T_2 values in lesions
  - mimicking restricted diffusion
- Seen in many pathologies
  - sub-acute infarctions
  - epidermoid cysts
- ADC provides 'real' answer





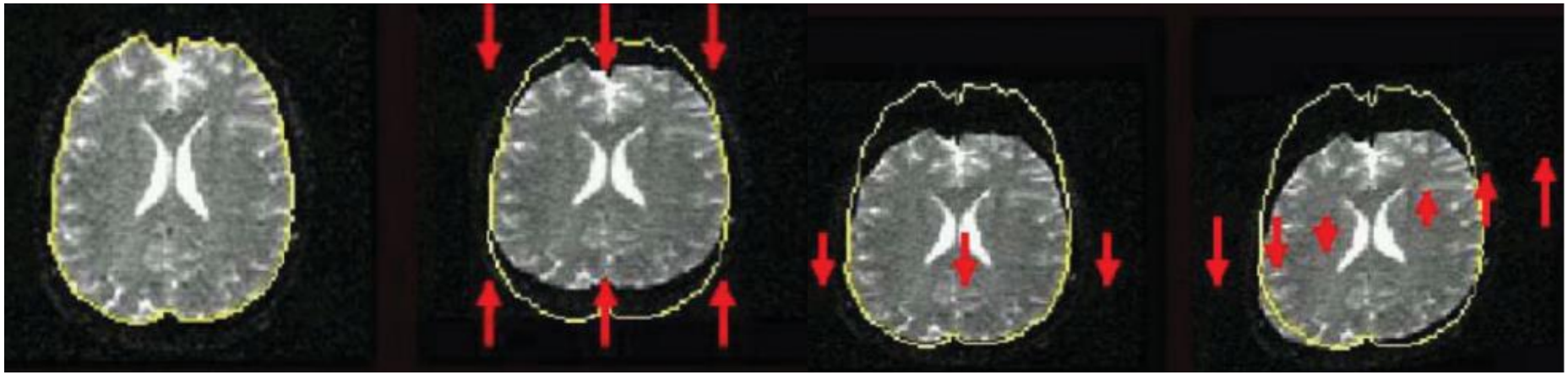
Susceptibility Issues

- Susceptibility Sources
  - Bone
  - Air
  - Hemosiderin
  - Dental work / clips
  - Tissue boundaries

- Solution?
  - Brain DWI PROPELLER / BLADE
    - FSE read-out, to minimize or eliminate distortions.
  - Multishot DWI – MUSE/RESOLVE

Eddy Currents

- large, rapidly switched magnetic field gradients induce eddy currents in the conductive structures of the scanner
- produce additional unwanted magnetic fields
- field gradient at the sample differs from the prescribed field gradient, resulting in a difference between the actual and prescribed b-matrix
- slowly decaying field during readout of the image causes geometrical distortion

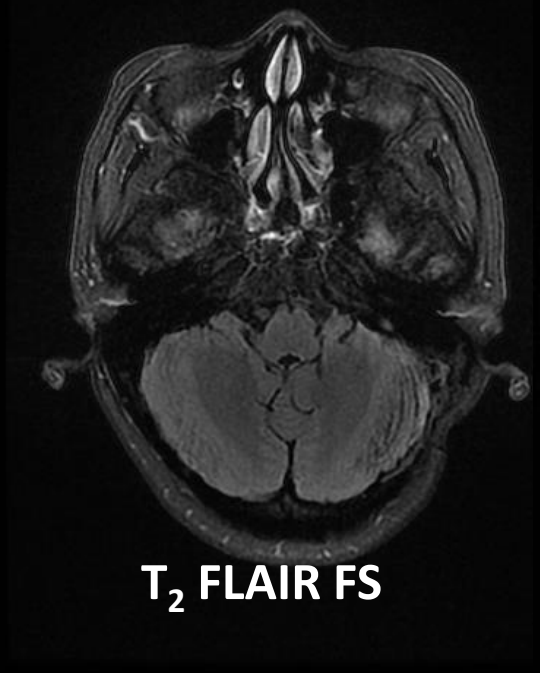


Artifacts and Pitfalls in Diffusion MRI - Le Bihan et al, JMRI 2006

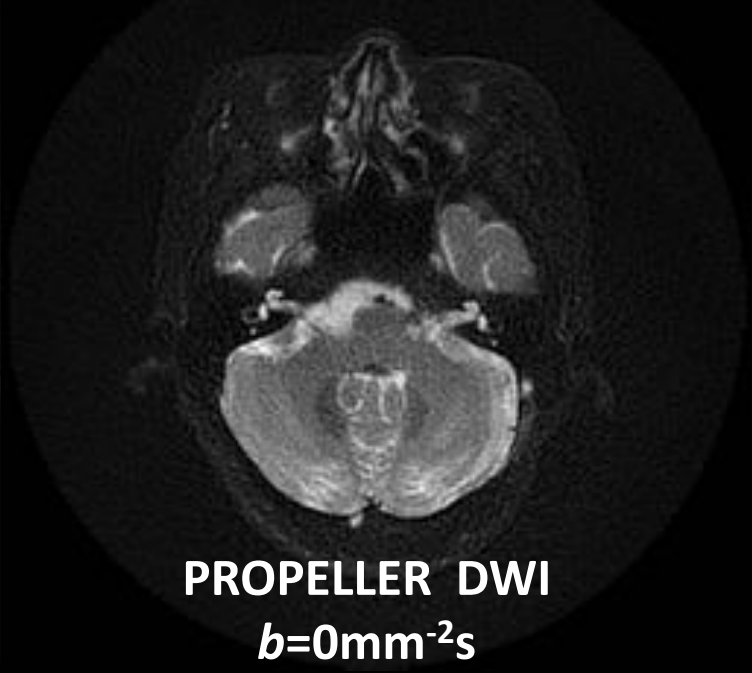
B-values, ADCs and calculated b-values

Hull University Teaching Hospitals NHS Trust

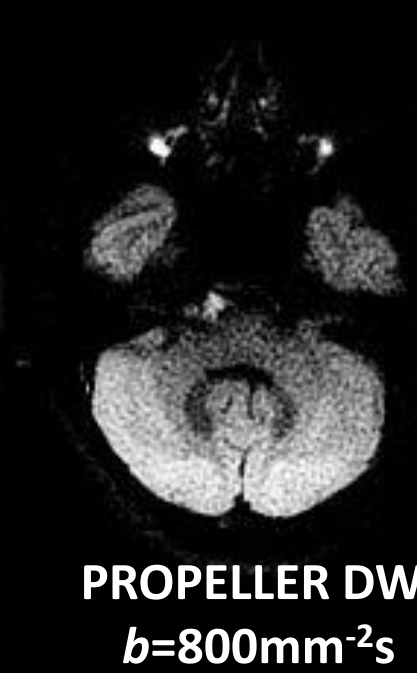
PROPELLER DWI



T_2 FLAIR FS



PROPELLER DWI
***b*=0mm^{-2}s**

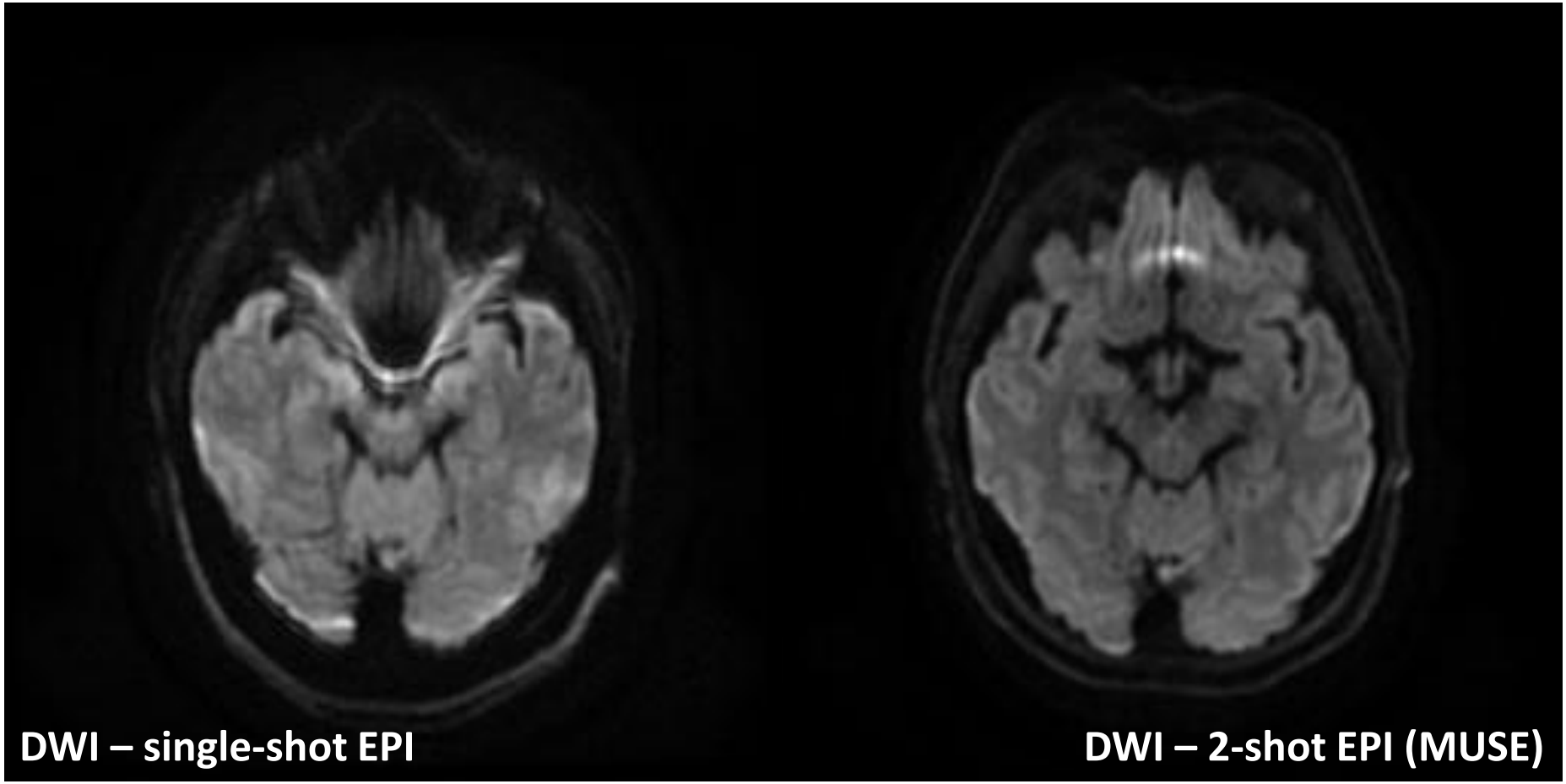


PROPELLER DWI
***b*=800mm^{-2}s**

- Disadvantage of DWI PROPELLER (FSE) is the relatively long acquisition time and low SNR compared to EPI-DWI

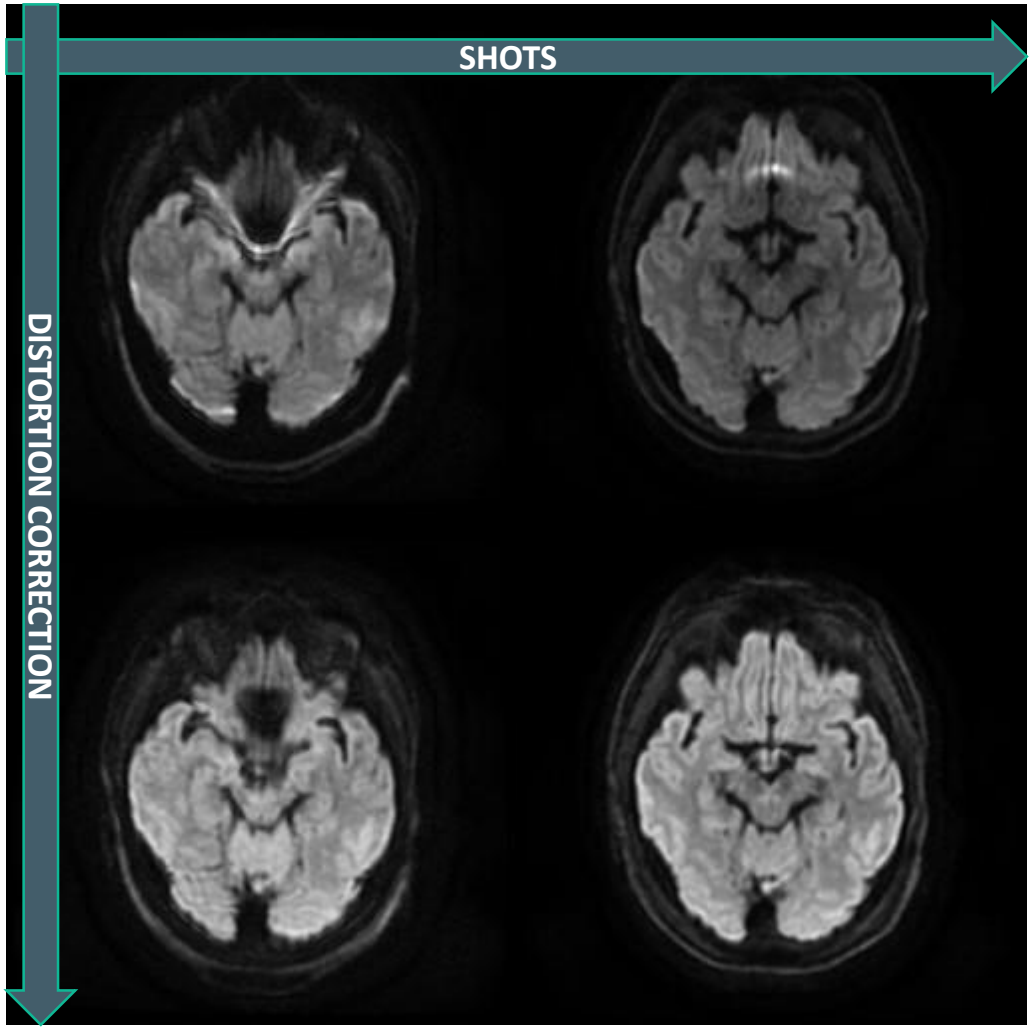
B-values, ADCs and calculated b-values

Single shot versus multi shot



B-values, ADCs and calculated b-values

DWI – single-shot EPI



DWI – 2-shot EPI (MUSE)

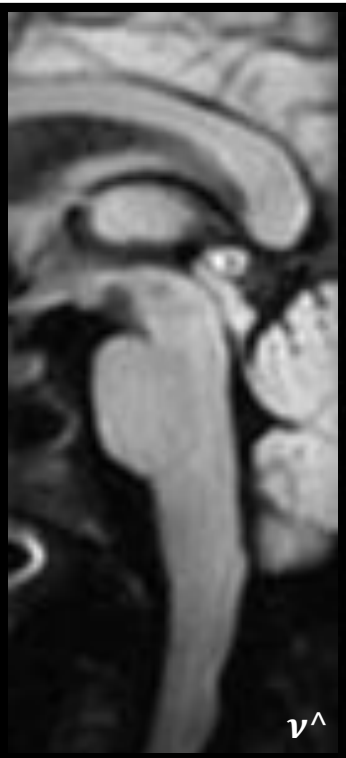
DWI – single-shot EPI With Distortion Correction

DWI – 2-shot EPI (MUSE) With Distortion Correction

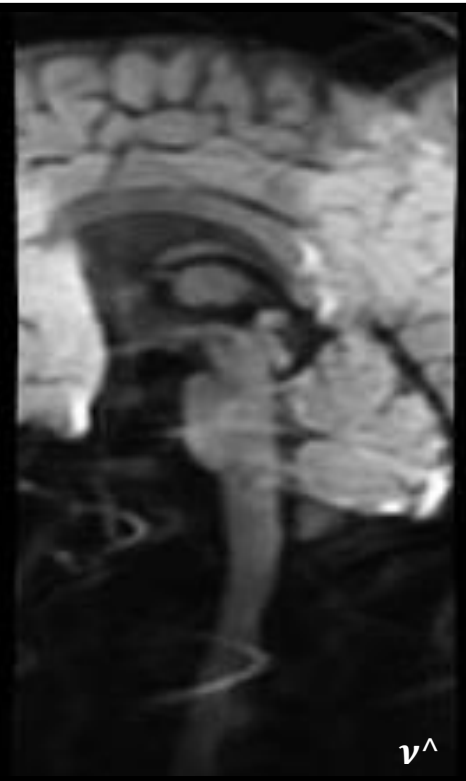
rFOV Diffusion

- FOCUS/ZOOMit/iZOOM
- Small FOV diffusion sequence
- 2D RF excitation pulse that is spatially selective in both the slice select and phase-encoding directions

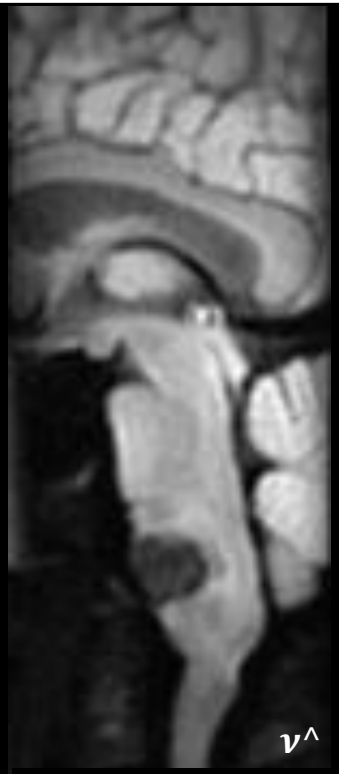
FOV	= 240mm
Phase FOV	= 0.3
Slice thick./spacing	= 3.0/0.0mm
TR/TE	= 2500/52ms
Frequency/phase	= 128/38
Excitation Mode	= FOCUS
Tensor	= 12 directions
b-value	= 600mm^{-2}s
NEX/ per direction	= 12
Shim Volume	



Selective excitation (FOCUS)

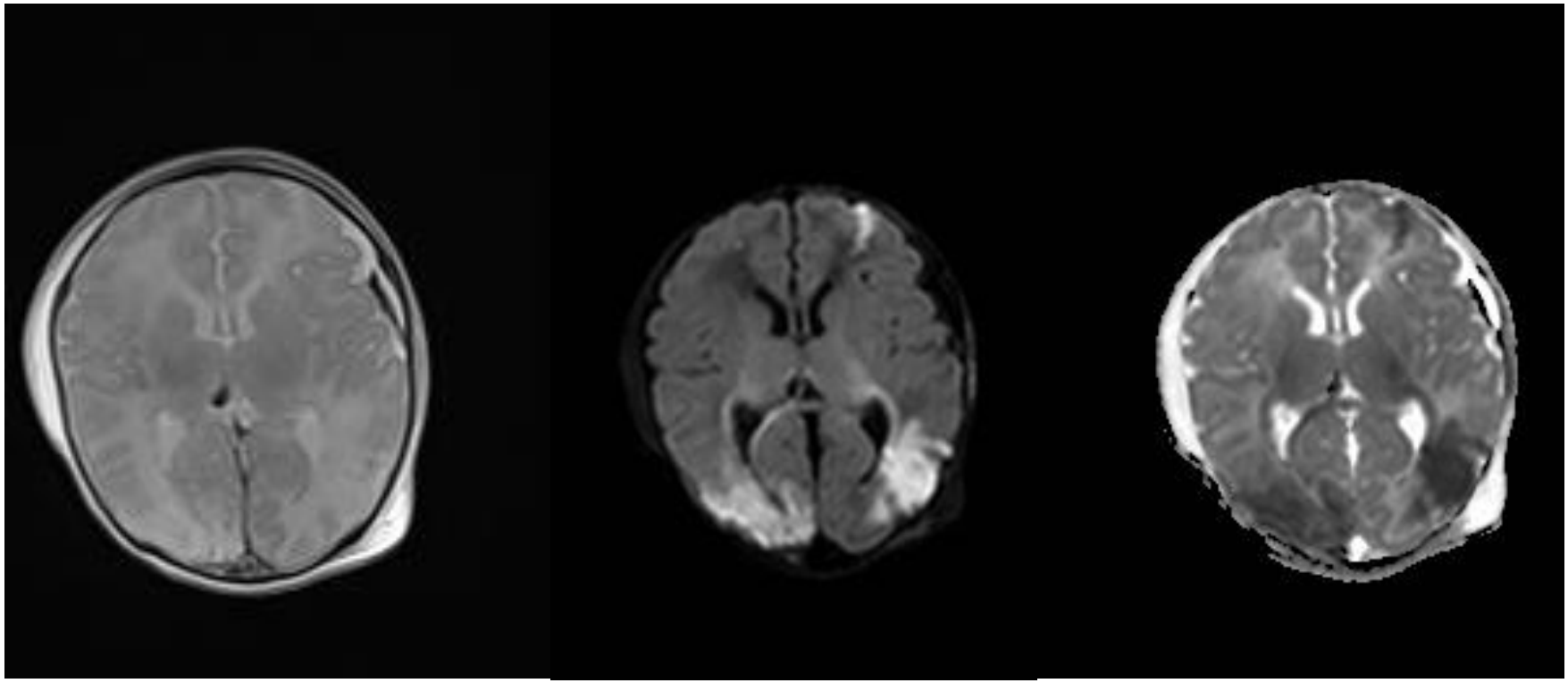


Non-selective excitation



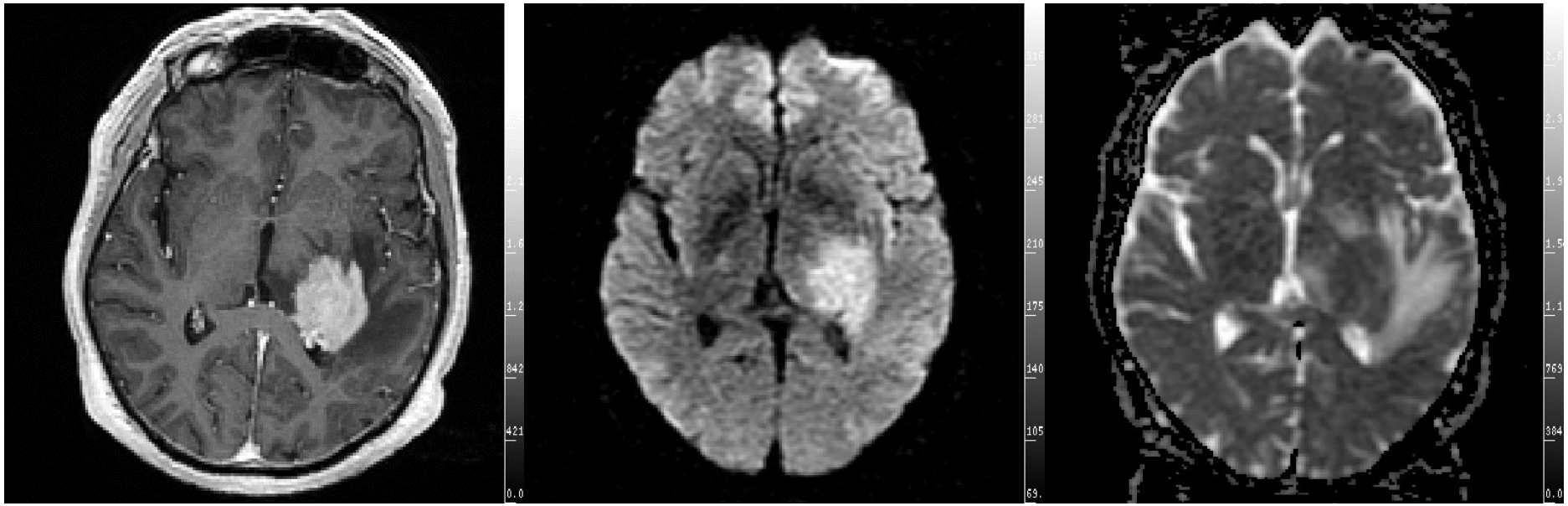
Selective excitation (FOCUS)

Hypoxic Injury

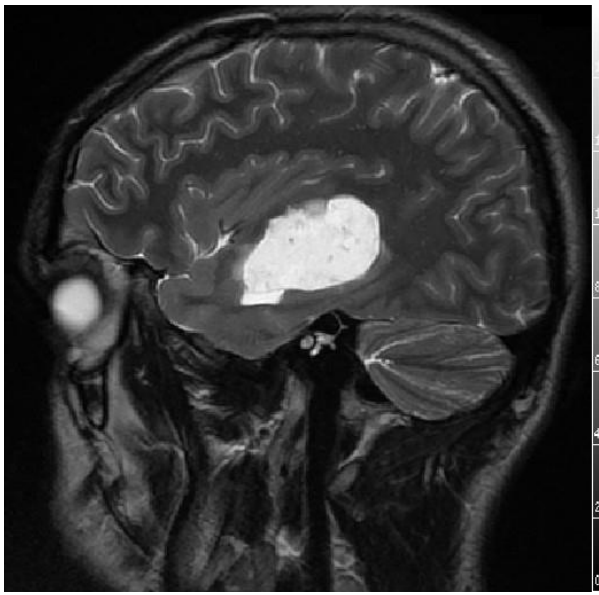


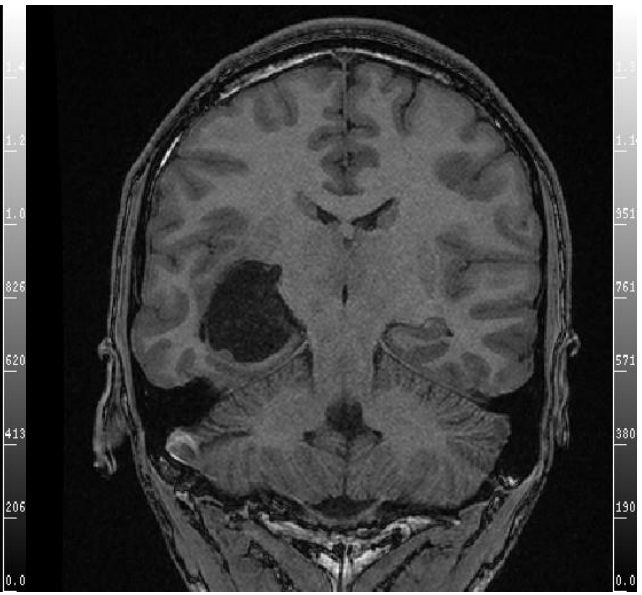
B-values, ADCs and calculated b-values

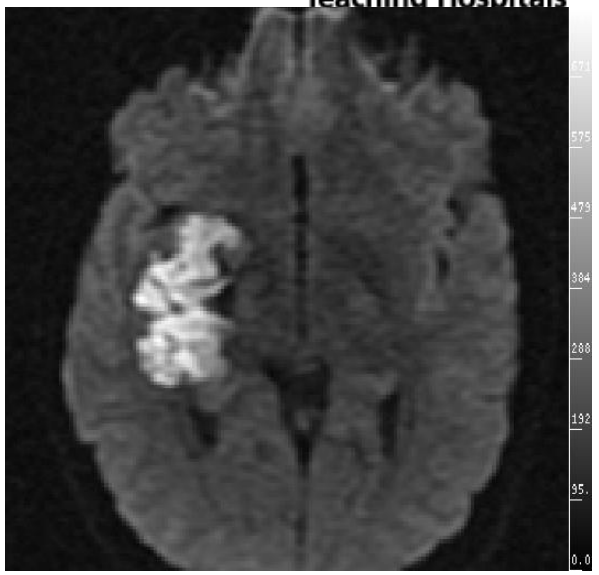
Lymphoma

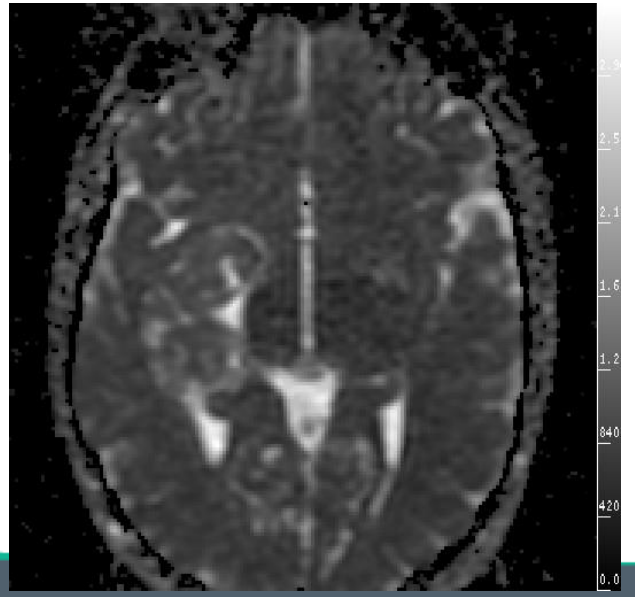


Epidermoid



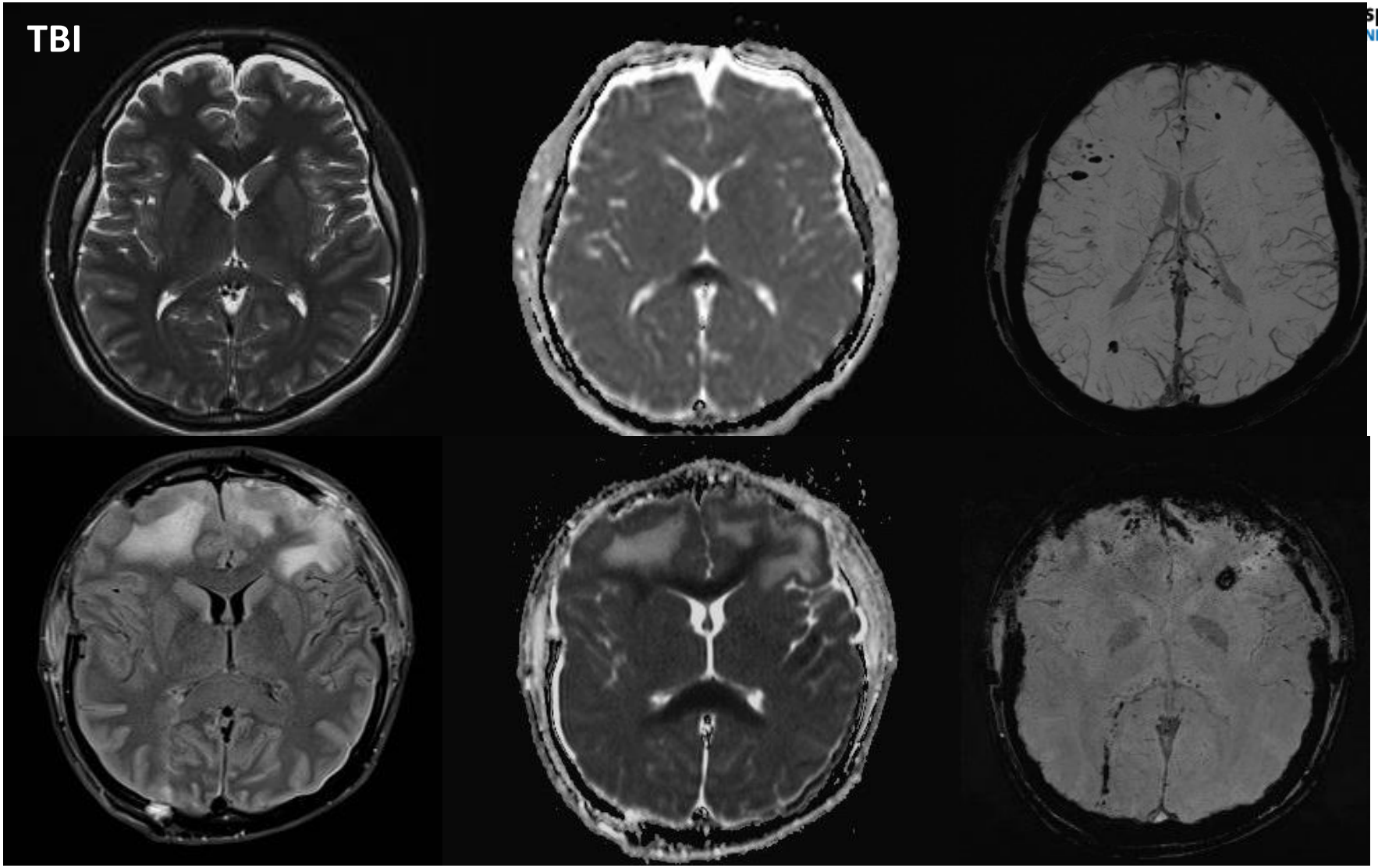




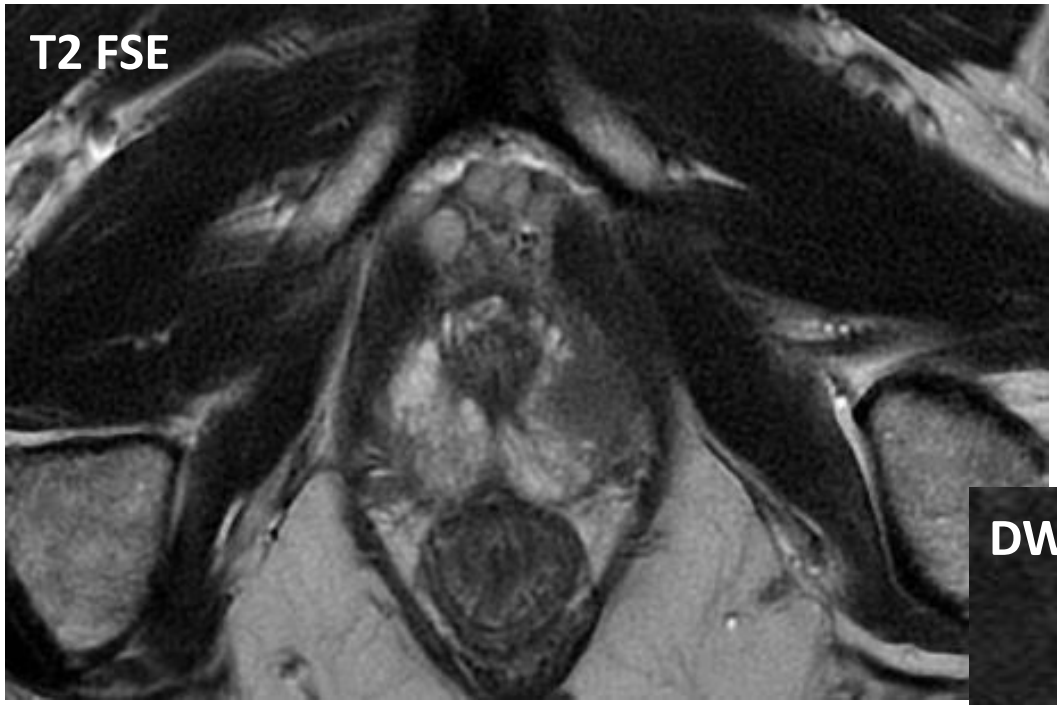


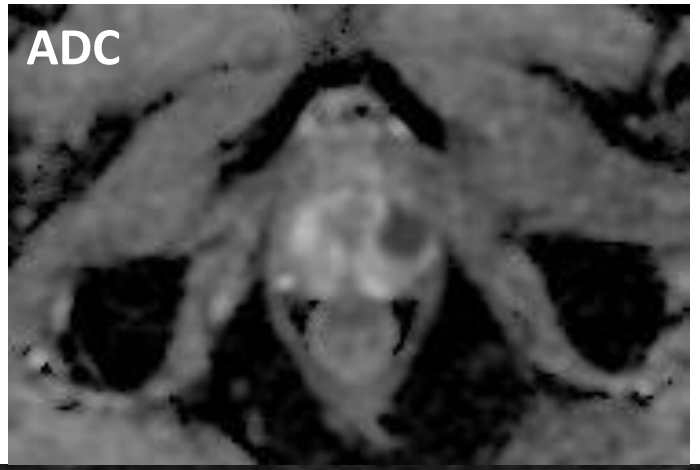
Differential Diagnosis:
Arachnoid cyst vs. Epidermoid

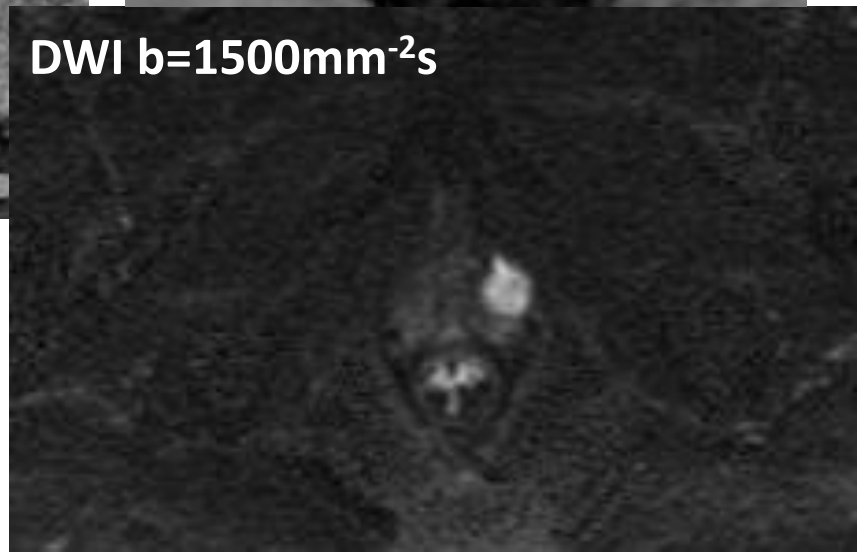
B-values, ADCs and calculated b-values



Prostate DWI

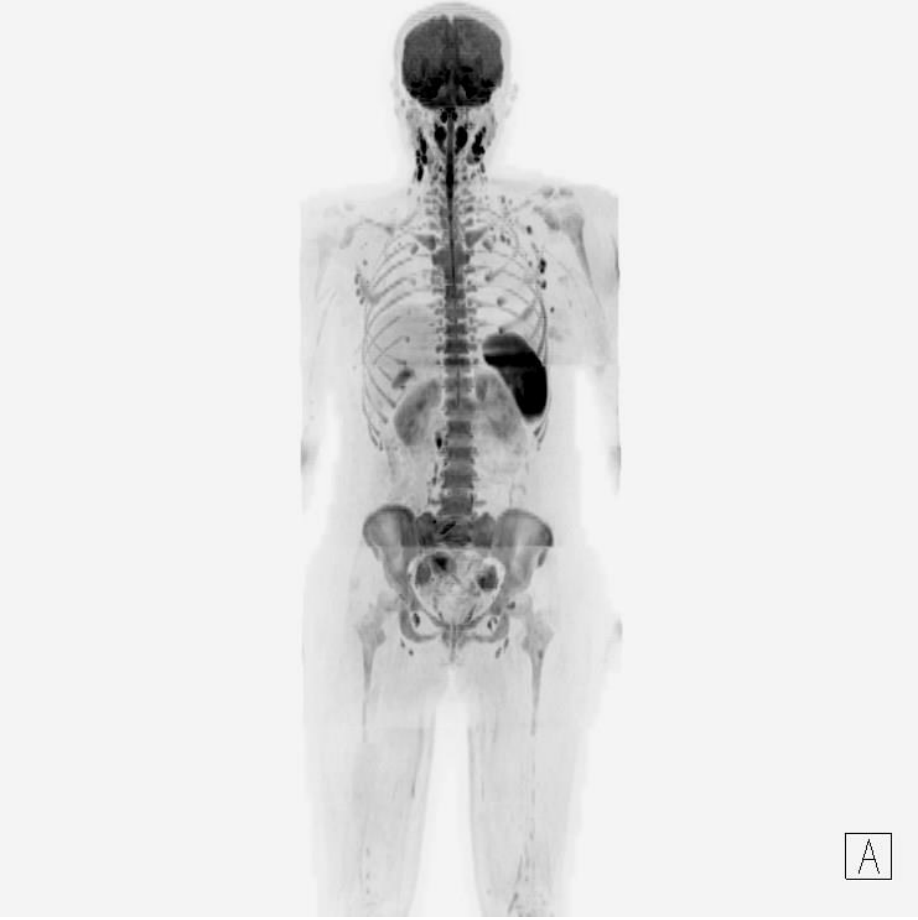






Lesion in the peripheral zone of the gland apex

Whole Body DWI



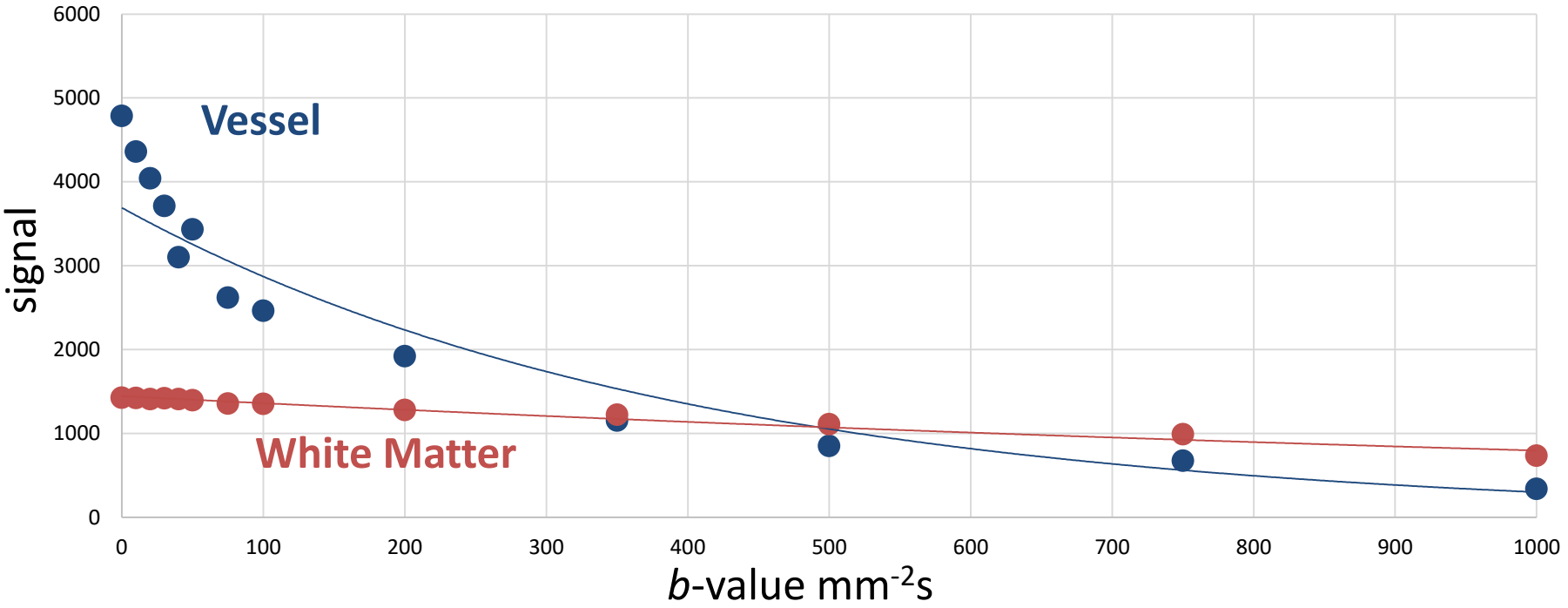
Whole Body DWI MIP – Calculated b=1400mm^{-2}s

Potential perfusion contribution to ADC

Hull University Teaching Hospitals NHS Trust

Conventional DWI ($S_b=S_0e^{-b.ADC}$) uses mono-exponential fitting (straight line in the log domain) to estimate ADC values

Perfusion effects can contribute increased signal at very low b-value which can lead to overestimation of ADC values in highly perfusion organs and/or lesions



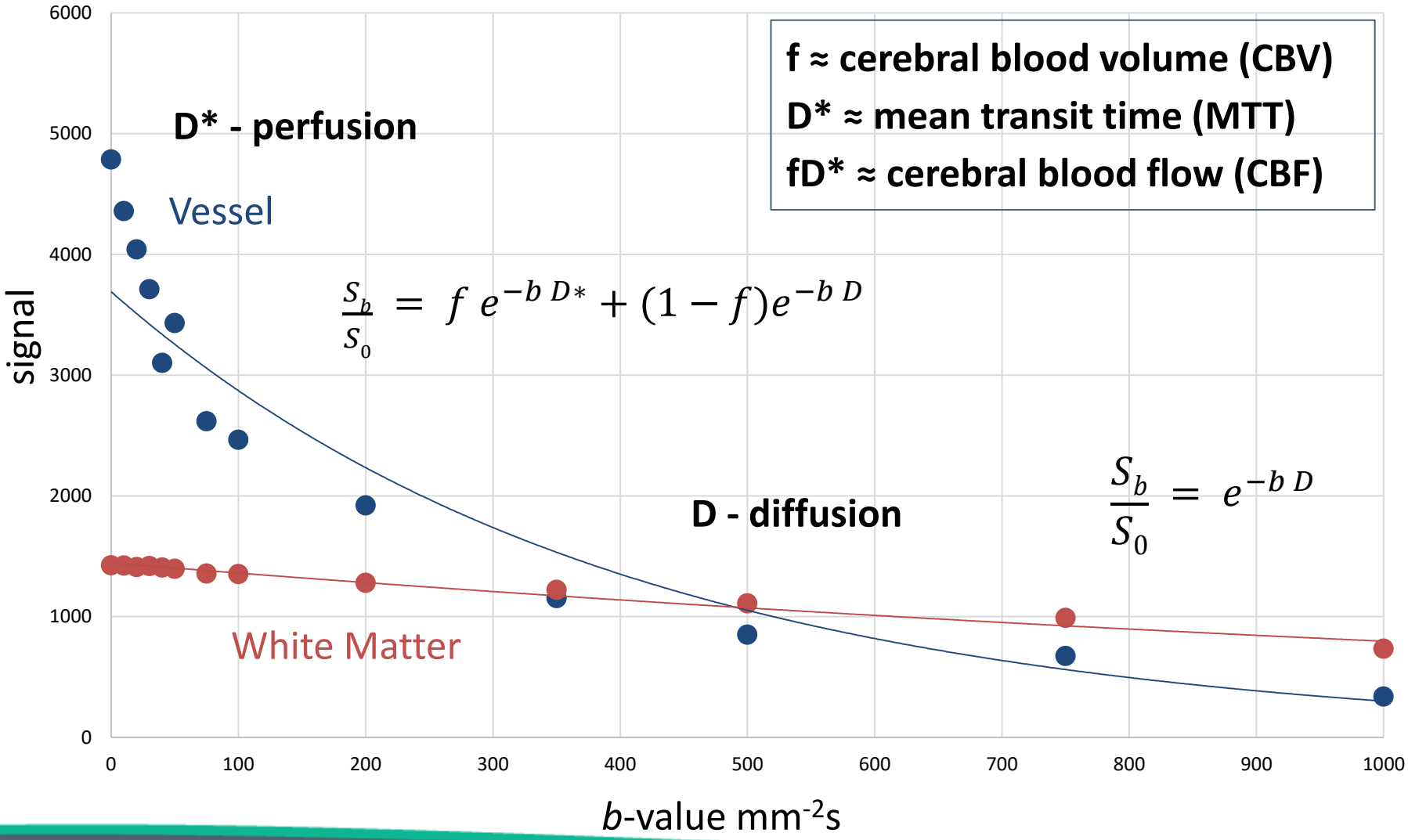
Intravoxel Incoherent Motion (IVIM)

- Water mobility in capillaries has a different nature than that resulting from thermal motion in tissue
- Le Bihan *et al.* (1986, 1988) separated both processes
- The concept of Intra-Voxel Incoherent Motion incorporates both thermally generated water mobility (D) and that resulting from the mobility within the capillary network (D*)
- The perfusion fraction is *f*

$$\frac{S_b}{S_0} = f\, e^{-b\,D*} + (1-f)e^{-b\,D}$$

f ≈ cerebral blood volume (CBV)
D* ≈ mean transit time (MTT)
fD* ≈ cerebral blood flow (CBF)

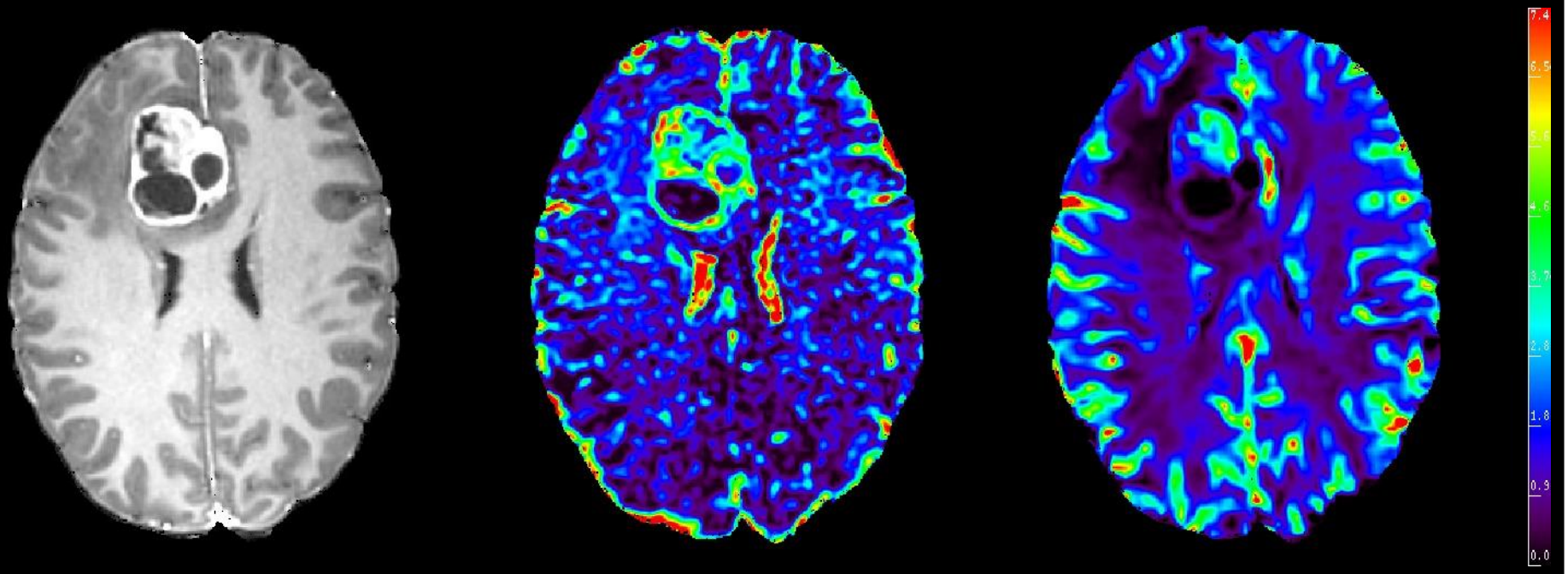
Potential perfusion contribution to ADC



Potential perfusion contribution to ADC

NHS
Hull University Teaching Hospitals
NHS Trust

Anaplastic Meningioma WHO III



3D T_1 IR-FSPGR +C

f ≈ cerebral blood volume
IVIM

rCBV
DSC

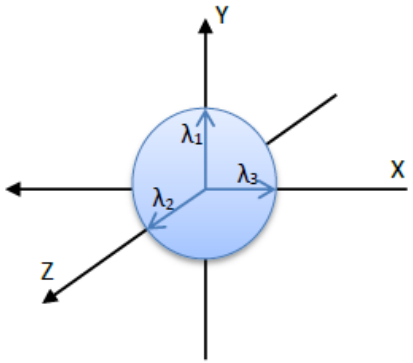
Diffusion anisotropy

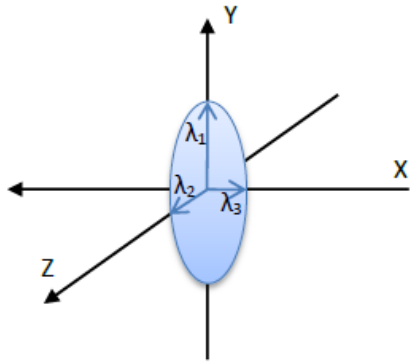
Iso- / Aniso- trophic diffusion

Isotropic Diffusion	Anisotropic Diffusion
A c	A c
A c	A c
A c	A c
Mea /sec	Mea /sec

$1x\ 10^{-3}$ mm^2/sec

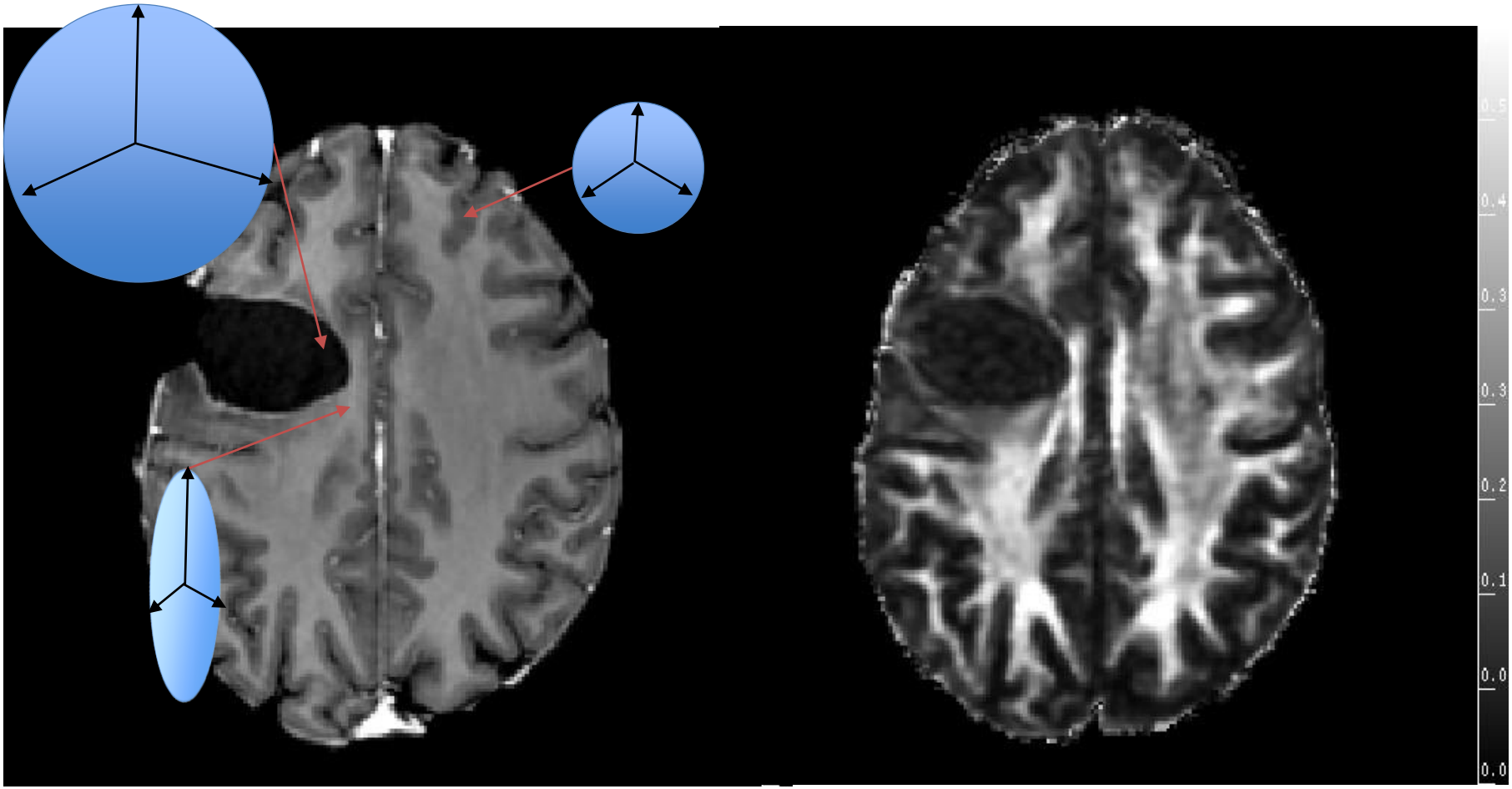
$1x\ 10^{-3}$ mm^2/sec



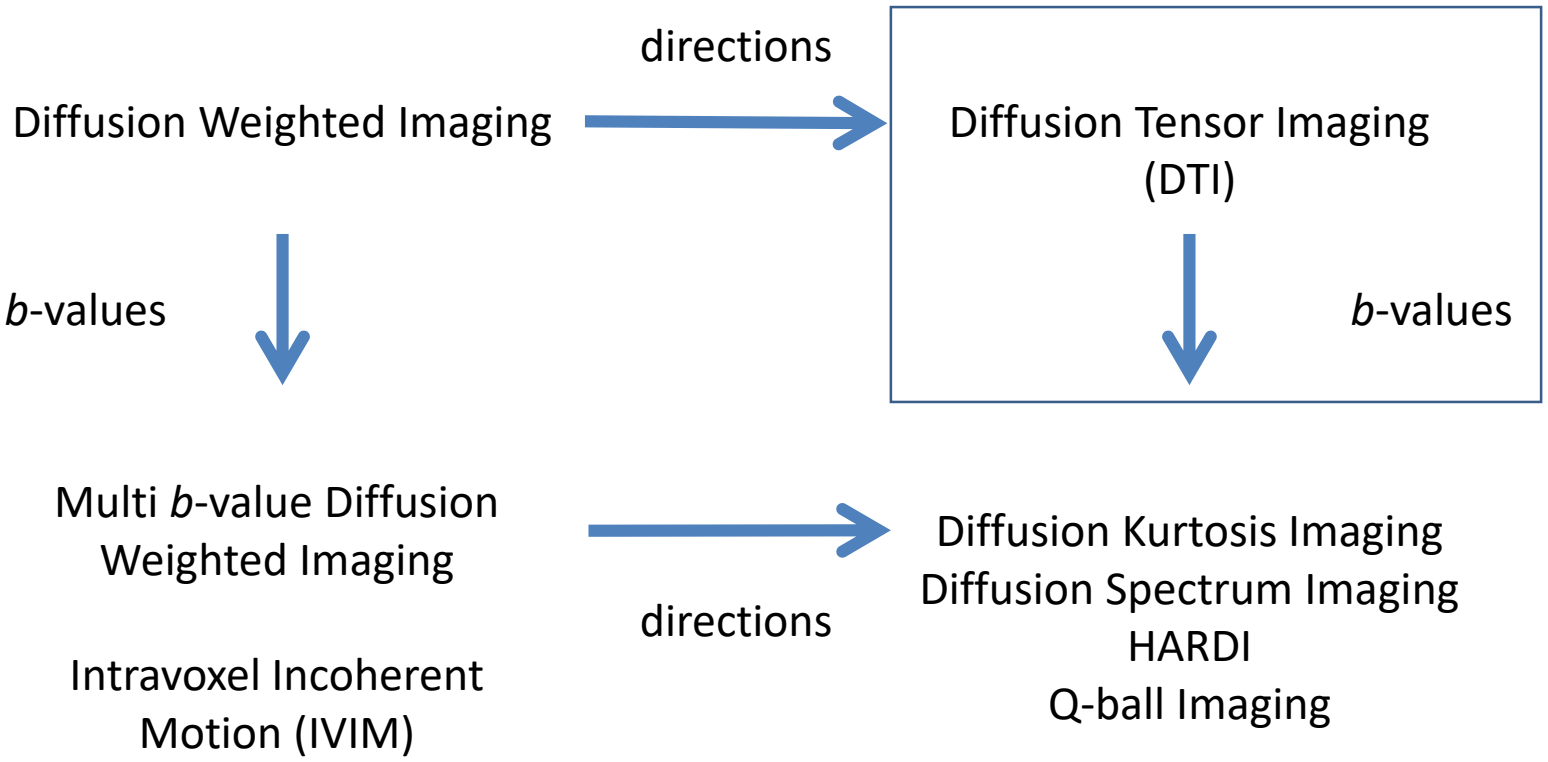


Diffusion anisotropy

Anisotropy Mapping



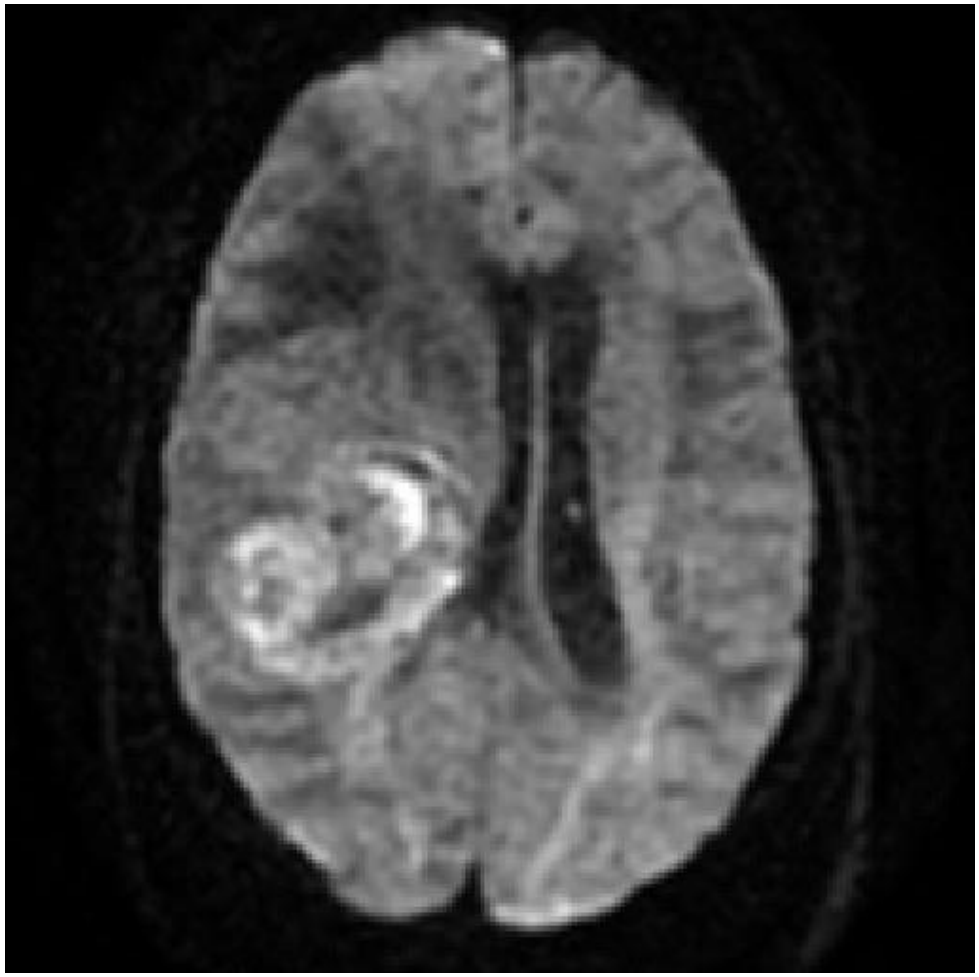
Variations of diffusion imaging



The Tensor

$$\begin{matrix} Dxx & Dxy & Dxz \\ Dxy & Dyy & Dyz \\ Dxz & Dyz & Dzz \end{matrix}$$

- D_{xx}, D_{yy}, D_{zz} are the diffusion in the main axis
- Assume that:
  $D_{xy} = D_{yx}$ $D_{xz} = D_{zx}$ $D_{yz} = D_{zy}$
- Therefore: only 6 elements are needed to create the tensor.
- 6 non-collinear gradient directions

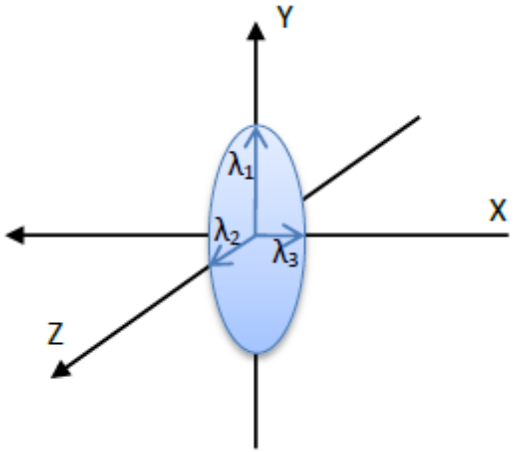


The Tensor

For each voxel:

eigenvalues eigenvectors

$$\begin{matrix} Dxx & Dxy & Dxz \\ Dxy & Dyy & Dyz \\ Dxz & Dyz & Dzz \end{matrix} \xrightarrow{\text{SVD}} \begin{matrix} \lambda_1 & 0 & 0 \\ 0 & \lambda_2 & 0 \\ 0 & 0 & \lambda_3 \end{matrix} \quad [e_1\ e_2\ e_3]$$



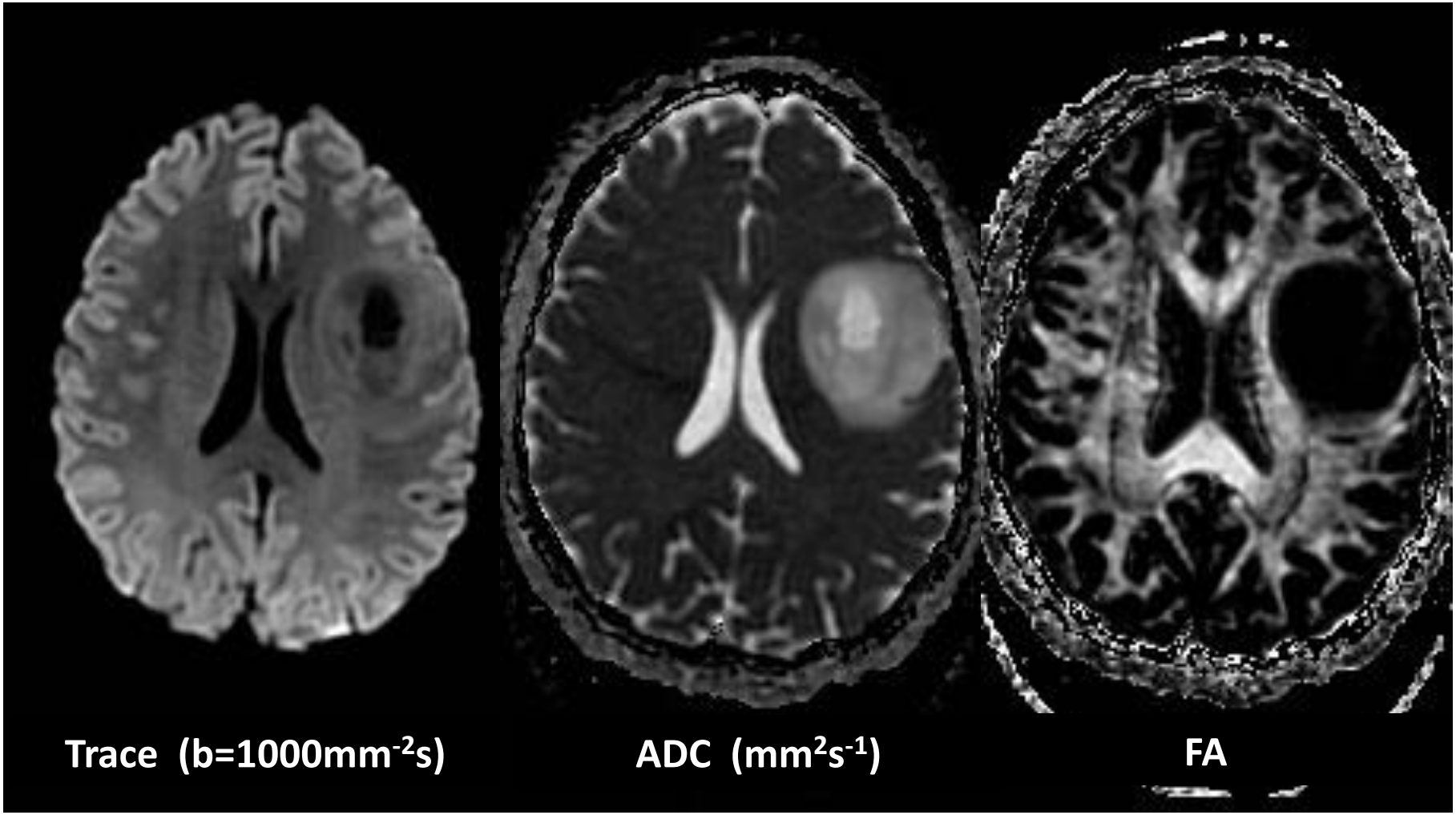
$$ADC = \frac{\lambda_1 + \lambda_2 + \lambda_3}{3}$$

$$FA = \sqrt{\frac{3}{2}} \sqrt{\frac{(\lambda_1 - ADC)^2 + (\lambda_2 - ADC)^2 + (\lambda_3 - ADC)^2}{\lambda_1^{\,2} + \lambda_2^{\,2} + \lambda_3^{\,2}}}$$

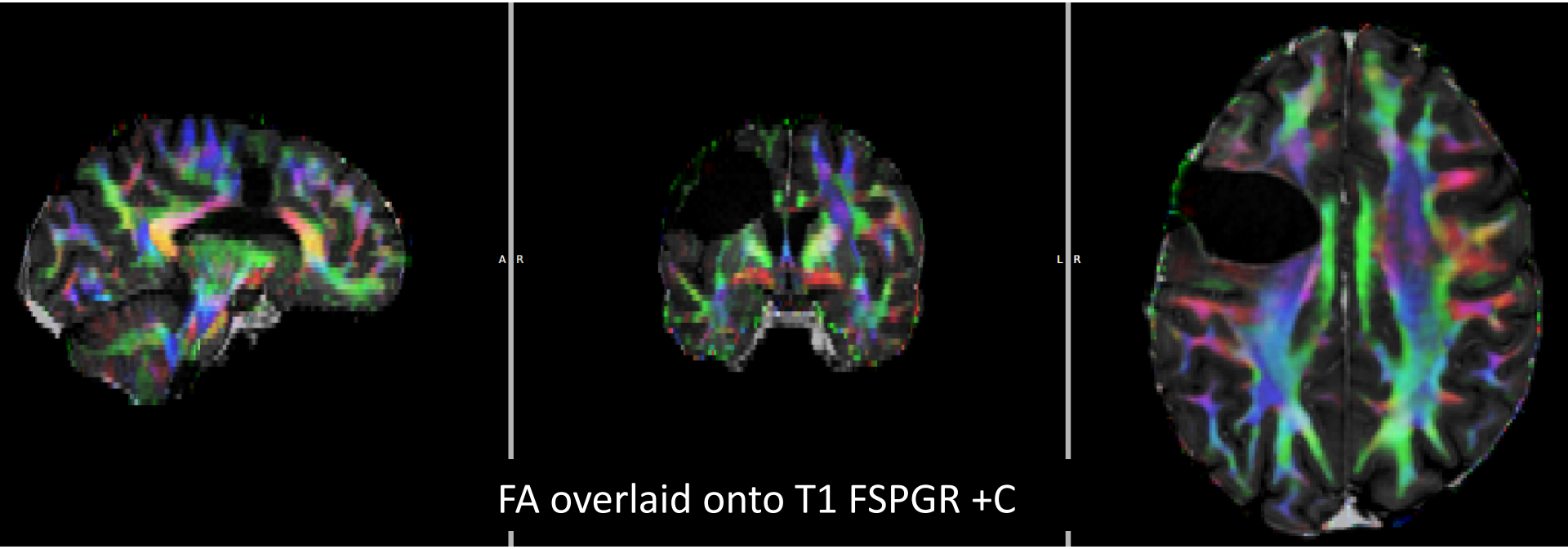
Diffusion anisotropy

NHS
Hull University Teaching Hospitals
NHS Trust

Low Grade Glioma



Diffusion Tensor Imaging

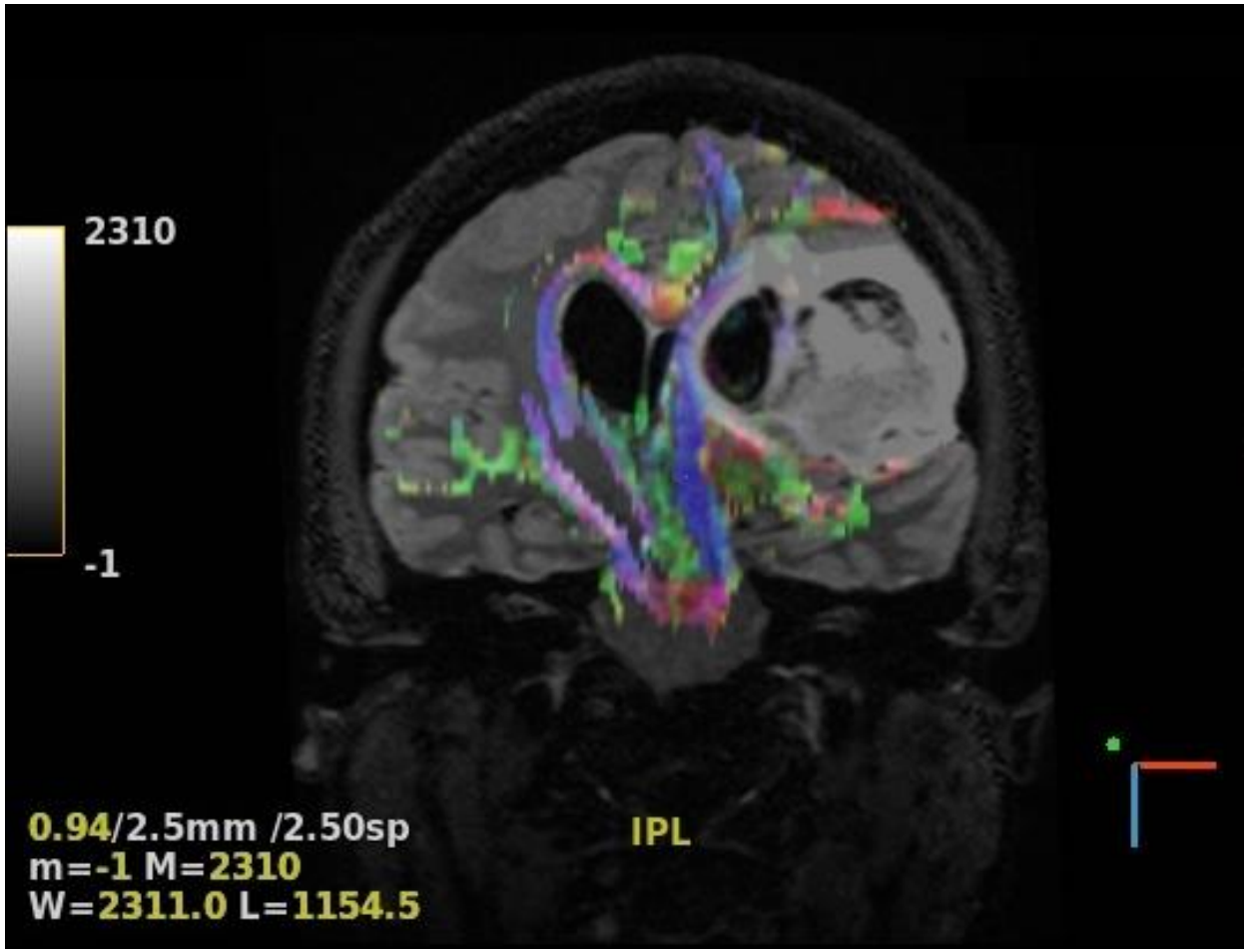


FA overlaid onto T1 FSPGR +C

Red	=	magnitude of eigenvector(1)	Right-Left
Green	=	magnitude of eigenvector(2)	Anterior-Posterior
Blue	=	magnitude of eigenvector(3)	Superior-Inferior

DTI Tractography for Surgical Planning

- Best possible resection
- Preserve eloquent areas and major white matter tracts
- Minimise postoperative morbidity



Diffusion anisotropy

Hull University Teaching Hospitals NHS Trust

	Modeling approach	DWI sampling	DWI weighting	Metrics	Seminal references
ADC	$S = S_0 e^{-bD}$	Few DWIs; one b=0 and 1 DWI	Low	ADC	Eccles et al., 1988
DTI	$S = S_0 e^{-bxD}$	Few DWIs; one shell, minimum six directions	Low	Combinations of λ1, λ2, λ3. e.g. FA, TR, WL, WP	Basser et al., 1994
DKI	$S = S_0 e^{-bD + \frac{1}{6}b^2D^2K}$	Moderate number of DWIs; two shells	Low and moderate only	DTI metrics and mean kurtosis, axial and radial kurtosis and KFA	Jensen et al., 2005; Tabesh et al., 2011; Glenn et al., 2015
MAP	Asymmetric simple harmonic oscillator reconstruction and estimation	Moderate to many DWIs; multishell acquisition	Low, moderate, and high	DTI metrics and non-Gaussianity, zero-displacement probabilities, propagator anisotropy, ODFs	Özarslan et al., 2013; Avram et al., 2016
DSI	Model-free	Many DWIs; Cartesian grid	Low, moderate, and high	ODFs possible to generate zero-displacement probabilities	Tuch et al., 2003
Q-ball	Model-free	Moderate no. of DWIs; single-shell HARDI acquisition	High	ODFs possible to generate zero-displacement probabilities	Tuch, 2004
CHARMED	Intra/extra-axonal compartments modeled by restricted/hindered sheets and cylinders	Multishell acquisition	Low, moderate, and high	Restricted and hindered component fractions; cone of uncertainty	Assaf & Basser, 2005
Axcaliber	Similar to CHARMED, but with additional modeling of axon diameter	Multishell acquisition	Low, moderate, and high flexible	CHARMED metrics and axon diameter	Assaf et al., 2008
NODDI	Watson distributed cylinders and sticks	Moderate no. of DWIs: multishell acquisition	Low, moderate, and high (flexible)	Cellular fractions, orientation dispersion index	Zhang et al., 2012; Tariq et al., 2016
WMTI	Intra/extra-axonal compartments modeled with the Gaussian part of the DKI model	Moderate no. of DWIs; two shells (same as DKI)	Low and moderate only	Axonal water fraction, intra-axonal diffusivity, extra-axonal radial/axial diffusivity, extra-axonal tortuosity	Fieremans et al., 2011

Dif ging

b-va -values

ging

aging

Hutchinson, EB, Schwerin, SC, Avram, AV, Juliano, SL, Pierpaoli, C. Diffusion MRI and the detection of alterations following traumatic brain injury. J Neuro Res. 2018; 96: 612– 625. https://doi.org/10.1002/jnr.24065

7.6 Diffusion MRI

- Diffusion weighting, relationship with underlying cellularity
  - *Diffusion weighted imaging is highly sensitive to tissue environment. Traditionally used to examine extracellular conditions of tissue (DWI)*

- B-values, ADCs and calculated b-values
  - *Select b-value according to ADC of the tissue being scanned. Beware of T_2 shine through. ADC (mm^2/s) is "gold standard"*

- Potential perfusion contribution to ADC
  - *Use b-values that are less sensitive to perfusion (50-100 rather than $0mm^{-2}s$). IVIM.*

- Diffusion anisotropy.
  - *Can be used to interrogate tissue structure (DTI)*