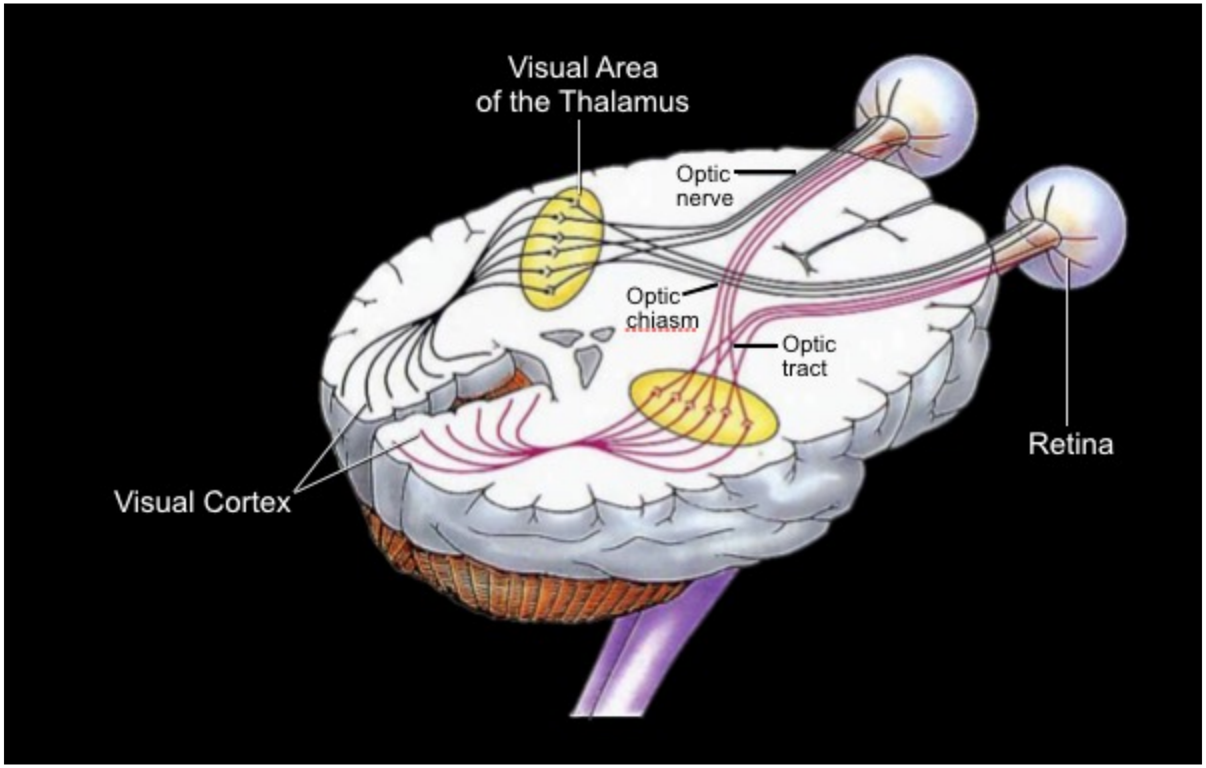
Basic Pulse Sequences

Saturation and Inversion Recovery

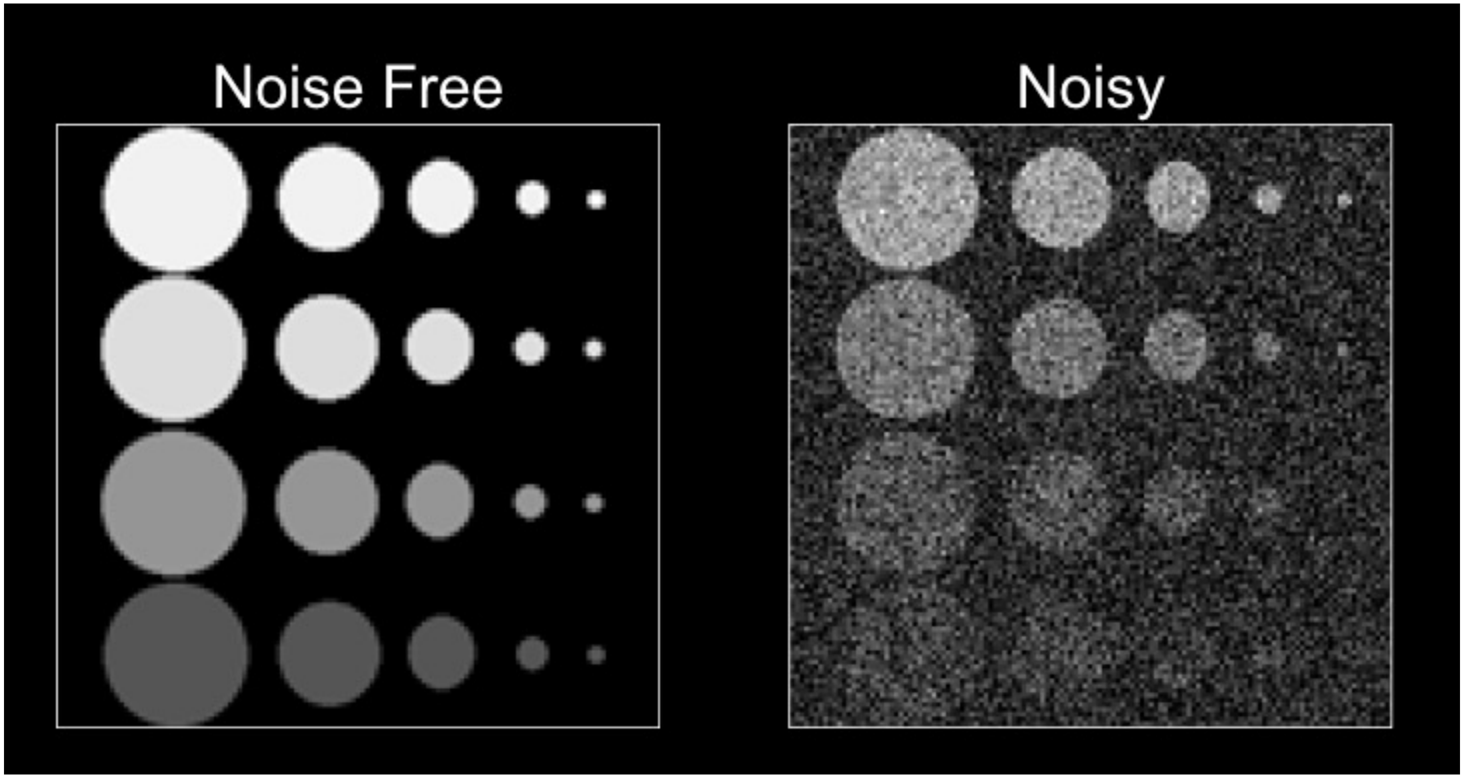
Image Contrast

Image Contrast



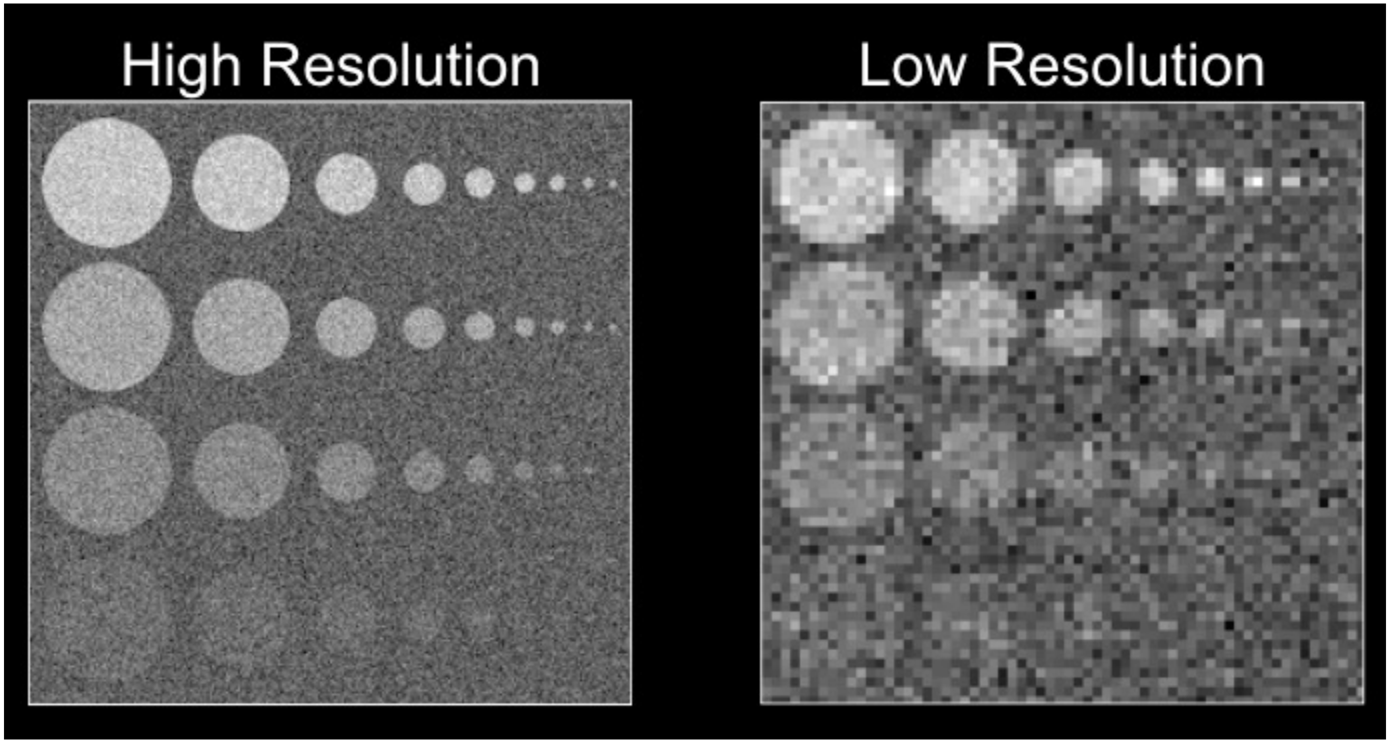
- **Human visual system is more sensitive to *contrast* than *absolute* luminance**

CNR, Object Size, and Noise



- **Large, high-contrast objects are easier to see in the presence of noise**

CNR, Object Size, and Noise



- **Small, low-contrast objects are easier to see with higher resolution**

Image Contrast

$$\mathcal{C}_{AB} = \frac{|I_A - I_B|}{I_{ref}}$$

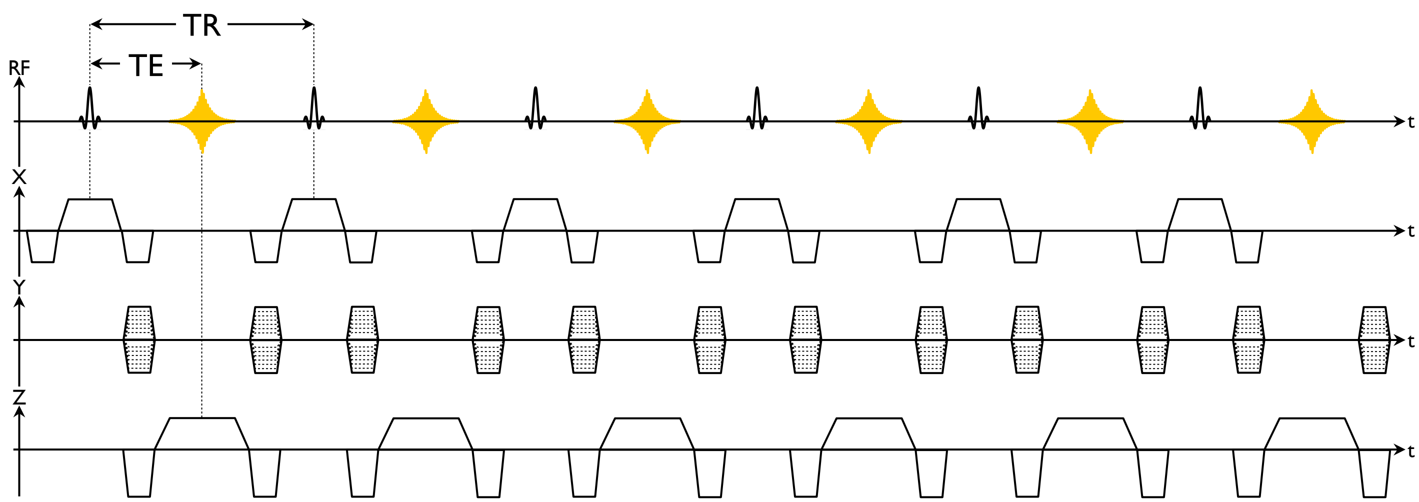
$$\mathcal{C}_{AB} = f(\rho, T_1, T_2, T_2^*, D, \ldots)$$

$$\mathcal{C}_{AB} \approx f(T_1) \qquad \mathcal{C}_{AB} \approx f(T_2)$$

Central goal in MRI is to limit image contrast to a single mechanism

Pulse Sequences

Sheet music is a <u>timing diagram</u> for playing the piano



A pulse sequence is a <u>timing diagram</u> for the MRI scanner…

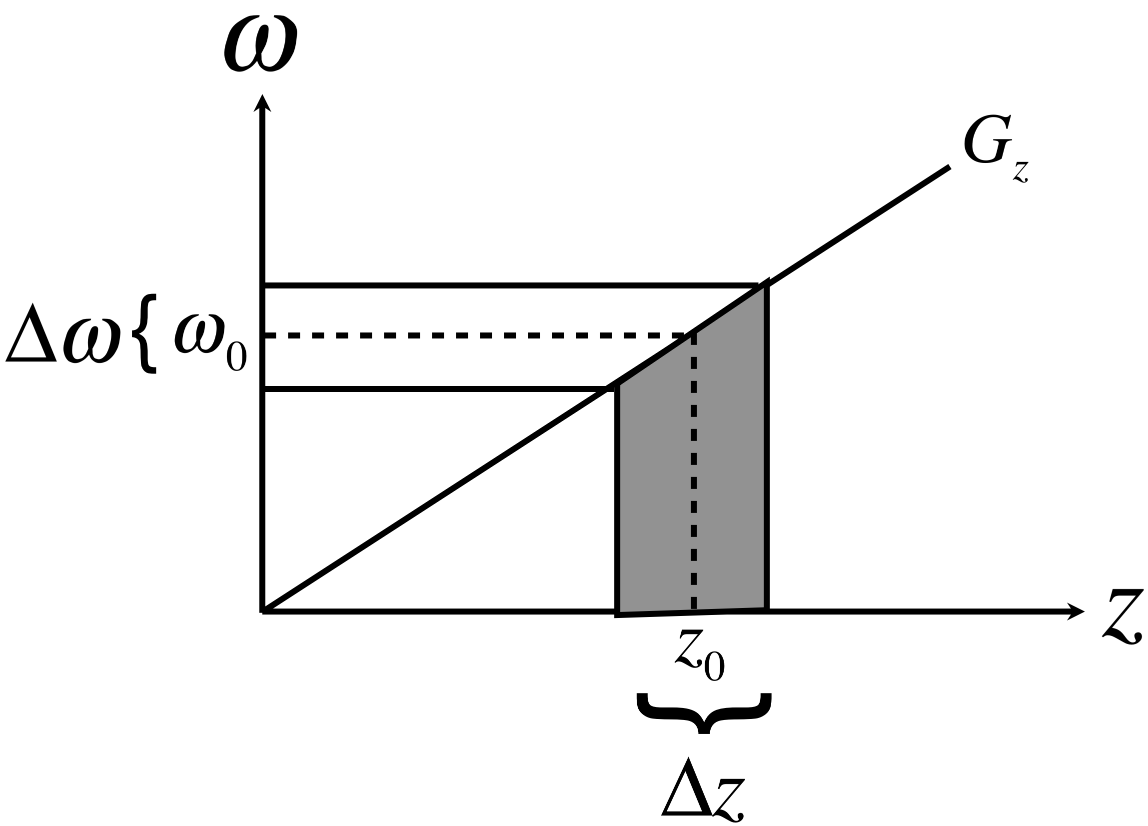
MR Image Formation (review)

- 2D MR images are formed by
  - Slice selection/excitation
  - Phase Encoding
  - Frequency Encoding

- 3D MR images are formed by
  - Slab selection/excitation
  - Phase encode in z-direction
  - Phase encode in y-direction
  - Frequency encode in x-direction

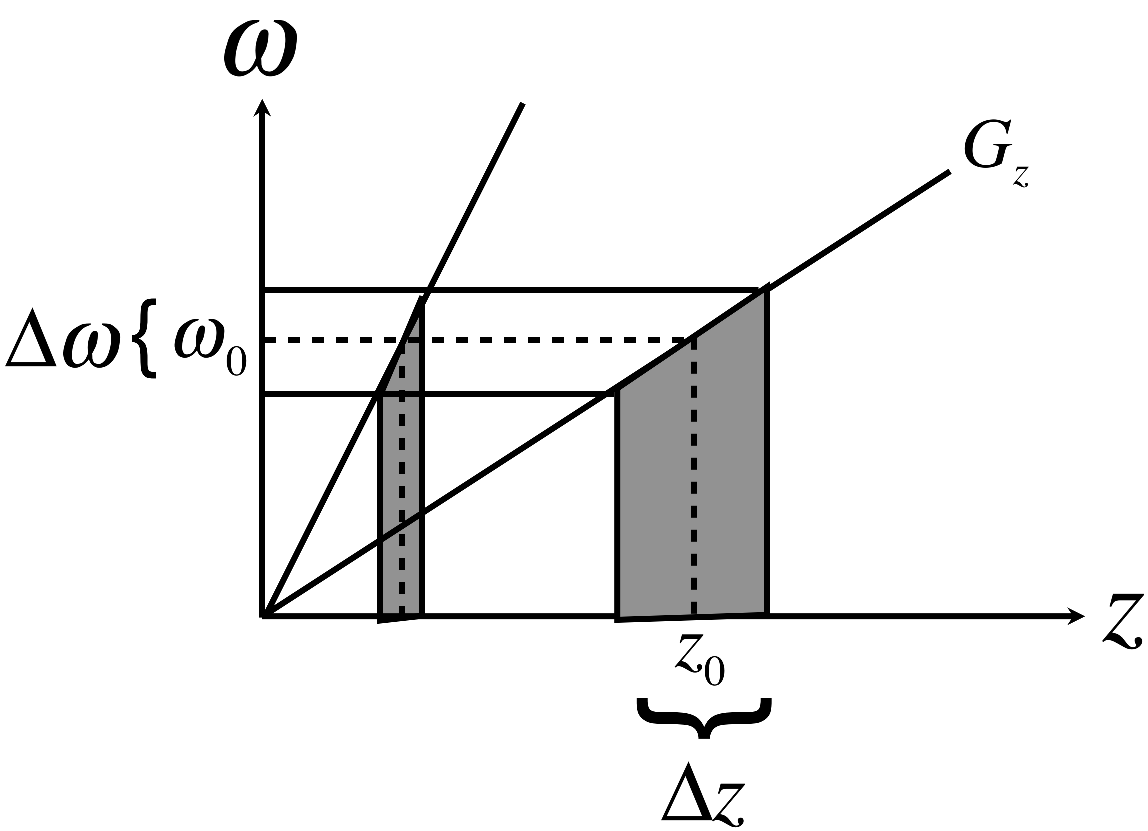
Slice Selection

- Gradient is applied in z-direction (or any direction) during RF excitation
- Only spins within the RF pulse "bandwidth" Δω will be excited



Slice Selection

- Gradient is applied in z-direction (or any direction) during RF excitation
- Only spins within the RF pulse "bandwidth" Δω will be excited
- Larger gradient amplitude + same bandwidth → thinner slice



Slice Selection

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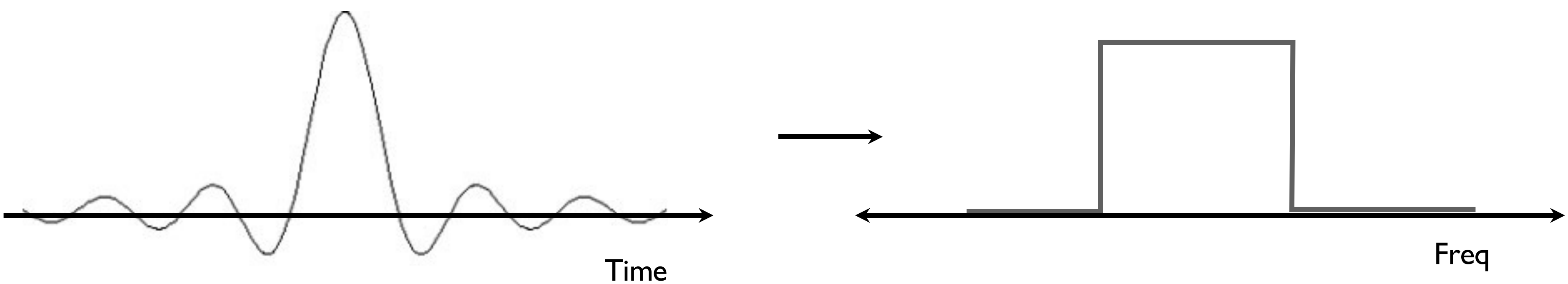
- A *sinc(t)* function is often used to obtain a *rectangular* frequency profile



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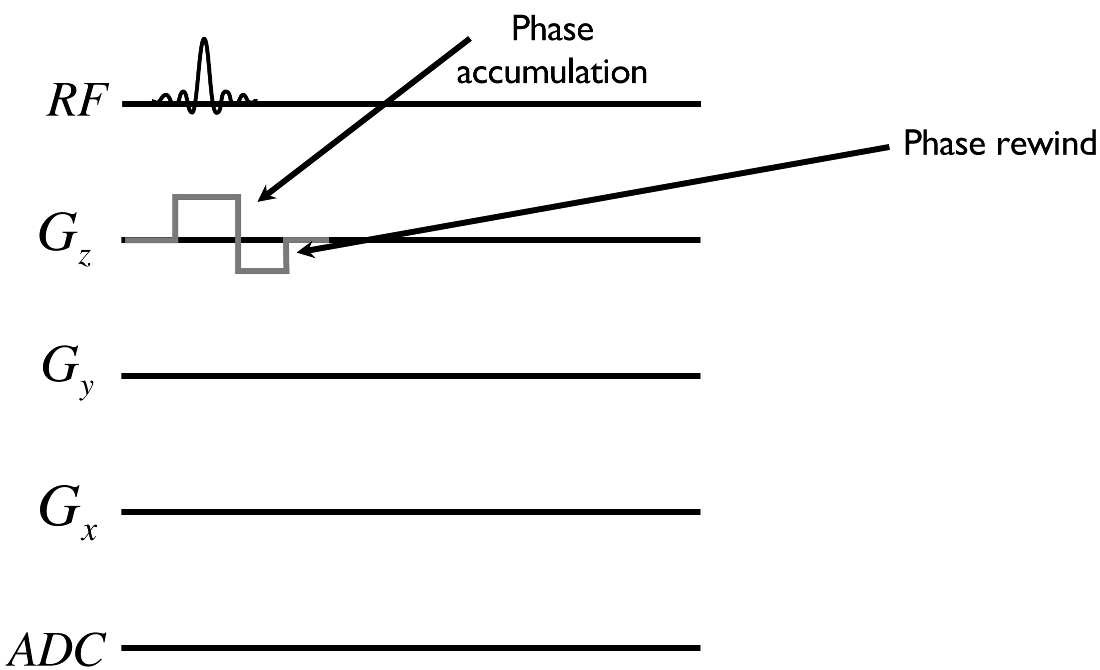


Phase Encode

- Pulsing a field gradient for a short period of time results in a phase offset along the direction of the field gradient.

$$\vec{\mathbf{M}}(\vec{\mathbf{r}},t)=\begin{pmatrix} e^{-t/T_2(\vec{\mathbf{r}})} & 0 & 0\\ 0 & e^{-t/T_2(\vec{\mathbf{r}})} & 0 \\ 0 & 0 & e^{-t/T_1(\vec{\mathbf{r}})}\end{pmatrix}\begin{pmatrix}\cos\left(\omega_0 t+\gamma\int_0^t \vec{\mathbf{G}}(\tau)\bullet\vec{\mathbf{r}}\,d\tau\right) & \sin\left(\omega_0 t+\gamma\int_0^t \vec{\mathbf{G}}(\tau)\bullet\vec{\mathbf{r}}\,d\tau\right) & 0\\ -\sin\left(\omega_0 t+\gamma\int_0^t \vec{\mathbf{G}}(\tau)\bullet\vec{\mathbf{r}}\,d\tau\right) & \cos\left(\omega_0 t+\gamma\int_0^t \vec{\mathbf{G}}(\tau)\bullet\vec{\mathbf{r}}\,d\tau\right) & 0\\ 0 & 0 & 1\end{pmatrix}\vec{\mathbf{M}}^0(\vec{\mathbf{r}},t)+\begin{pmatrix}0\\0\\M_0\left(1-e^{-t/T_1(\vec{\mathbf{r}})}\right)\end{pmatrix}$$

- Consider only the *transverse* magnetization *Mxy(t)* = *Mx(t)* + *j My(t)* for a phase encode gradient in the *y*-orientation:

$$M_{xy}(\vec{\mathbf{r}},t)=\left|\vec{\mathbf{M}}^0(\vec{\mathbf{r}},t)\right|e^{-t/T_2(\vec{\mathbf{r}})}e^{-j\omega_0 t}e^{-j\gamma y\int_0^t G_y(\tau)\,d\tau}$$

Transverse Magnetization

Initial (or steady state) magnetization

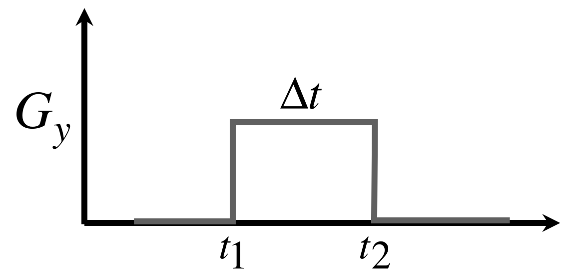
T2 decay

Precession

Contribution from G_y

Phase Encode

- Consider a square-wave gradient with magnitude *Gy* pulsed from t_1 to t_2



$$M_{xy}(\vec{\mathbf{r}},t) = \left|\vec{\mathbf{M}}^0(\vec{\mathbf{r}},t)\right| e^{-t/T_2(\vec{\mathbf{r}})} e^{-j\omega_0 t} e^{-j\gamma y \int_0^t G_y(\tau)\,d\tau}$$

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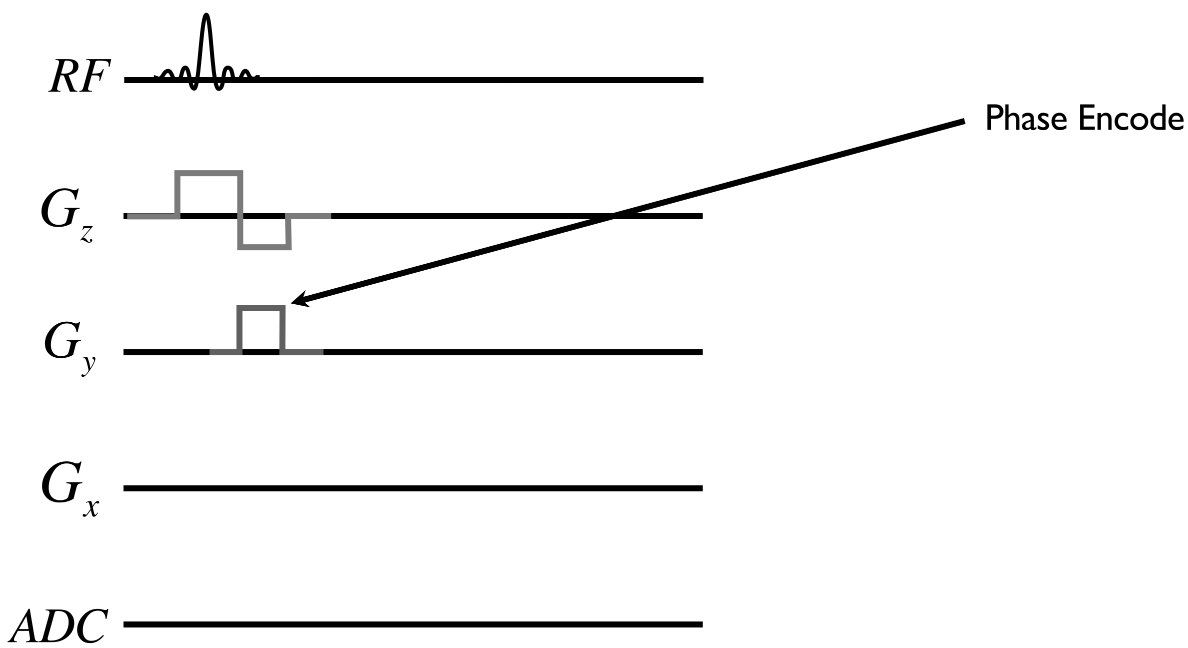
$$M_{xy}(\vec{\mathbf{r}},t) = \left|\vec{\mathbf{M}}^0(\vec{\mathbf{r}},t)\right| e^{-t/T_2(\vec{\mathbf{r}})} e^{-j\omega_0 t} e^{-j\gamma y G_y(t_2 - t_1)}$$

Linear Phase offset

$$\theta(y) = y \cdot (\gamma G_y \Delta t)$$

Phase Encode

- Consider a square-wave gradient with magnitude *Gy* pulsed from t_1 to t_2



Frequency Encode

- Using the same principle, we can apply another gradient in the *x*-direction, but read out the FID during this time
- Position is then encoded by *frequency*

$$M_{xy}(\vec{\mathbf{r}},t) = \left|\vec{\mathbf{M}}^0(\vec{\mathbf{r}},t)\right| e^{-t/T_2(\vec{\mathbf{r}})} e^{-j\omega_0 t} e^{-j\gamma x \int_0^t G_x(\tau)\,d\tau}$$

$$\downarrow$$

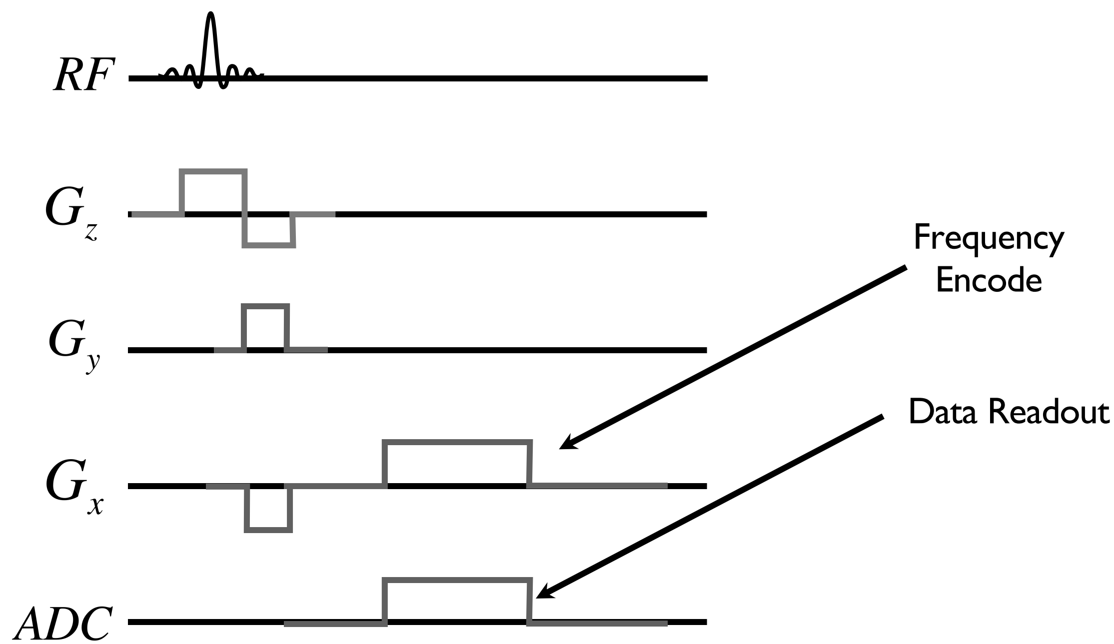
$$e^{-j\gamma x G_x t} = e^{-j\omega t}$$

$$\downarrow$$

$$\omega(x) = x \cdot (\gamma G_x)$$

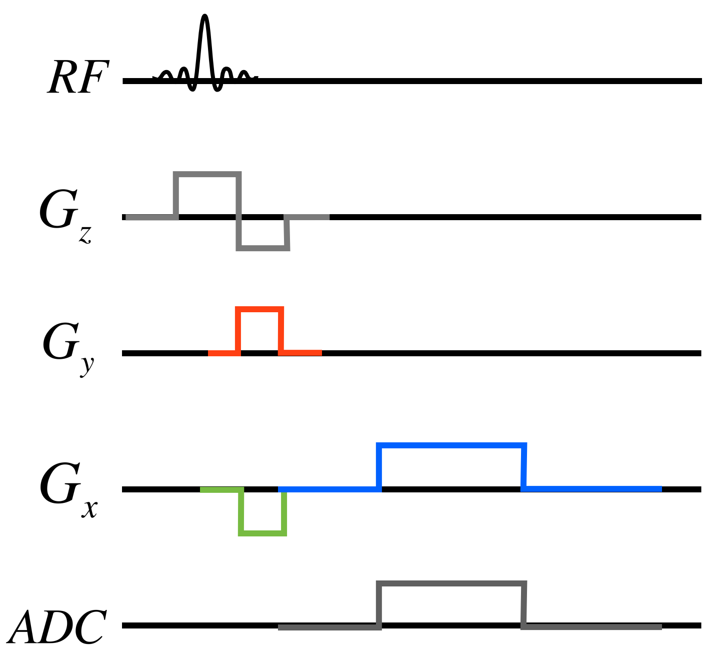
Frequency Encode

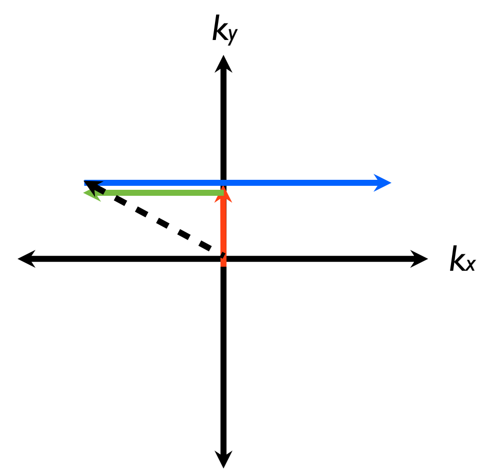
- Using the same principle, we can apply another gradient in the *x*-direction, but read out the FID during this time
- Position is then encoded by *frequency*
- Data is acquired *during* application of the frequency encode gradient



k-Space

- Because all image data is encoded with respect to *spatial frequency* using field gradients, a 2D (or 3D) Fourier transform can be applied to acquired data in order to obtain the original image
- Spatial frequency space = *k*-space
- Application of the gradients at different times results in traversing through *k*-space, while data is only "stored" in *k*-space during data readout.





$$\vec{\mathbf{k}}(t) = \gamma \int_0^t \vec{\mathbf{G}}(\tau) d\tau$$

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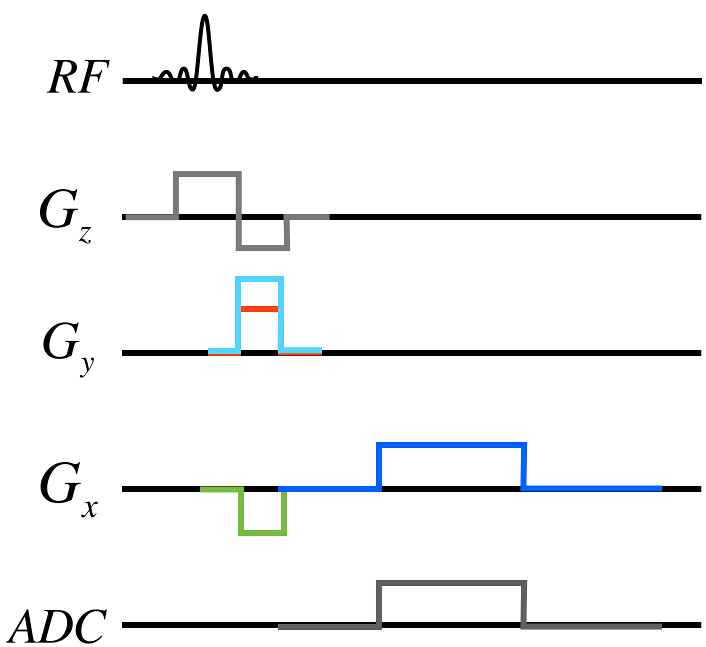
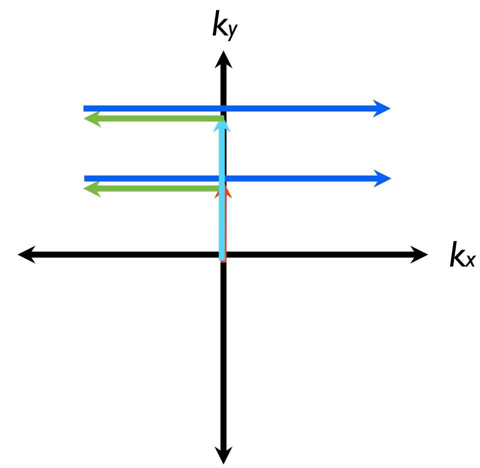


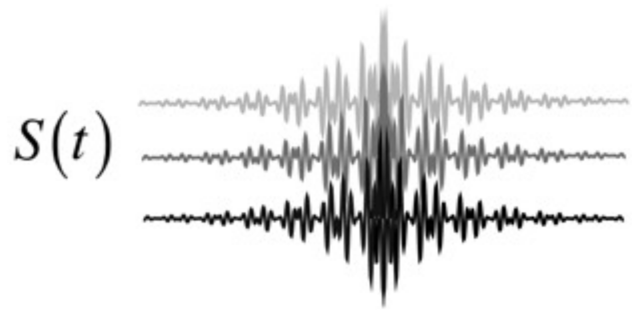


$$\vec{\mathbf{k}}(t) = \gamma \int_0^t \vec{\mathbf{G}}(\tau) d\tau$$

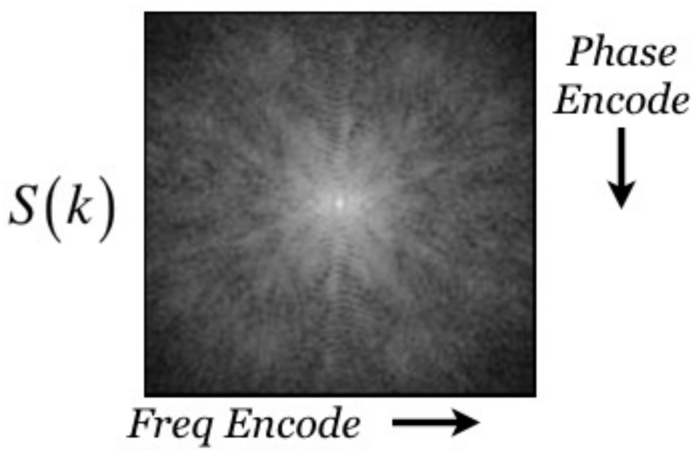
MRI: Dipoles to Images

FID: 2 Frequencies

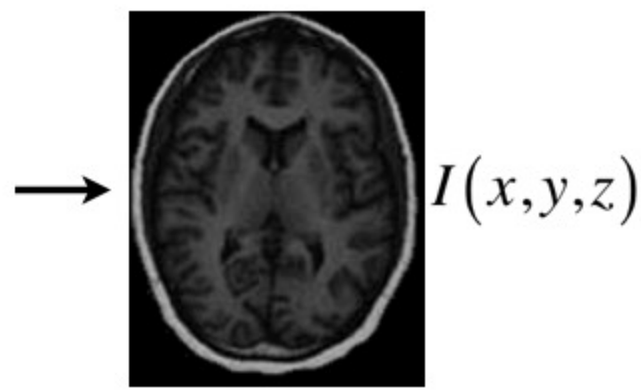
Multiple FIDs: Separated by Phase



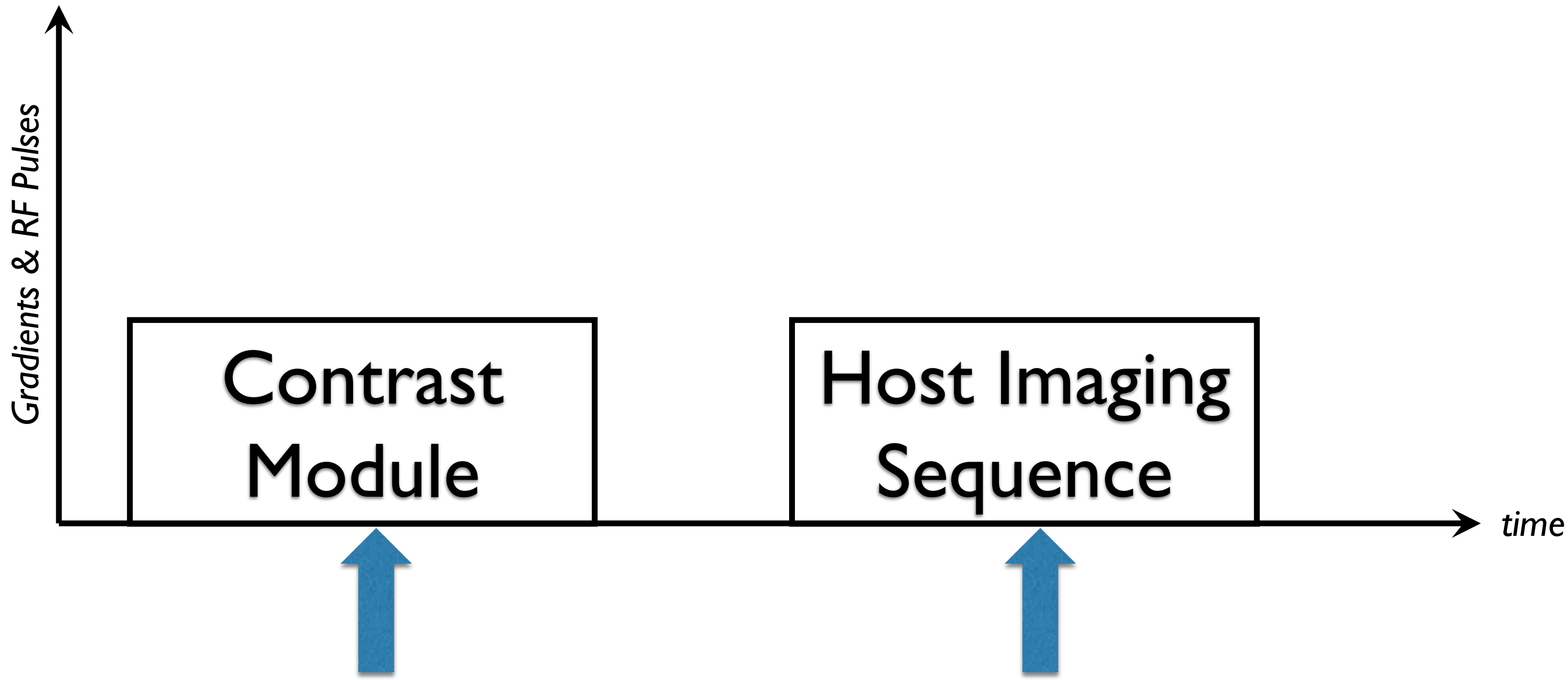
K-Space: 2-D Frequency Domain



MRI Image



Basic Pulse Sequences
(Contrast Module → Host Sequence)



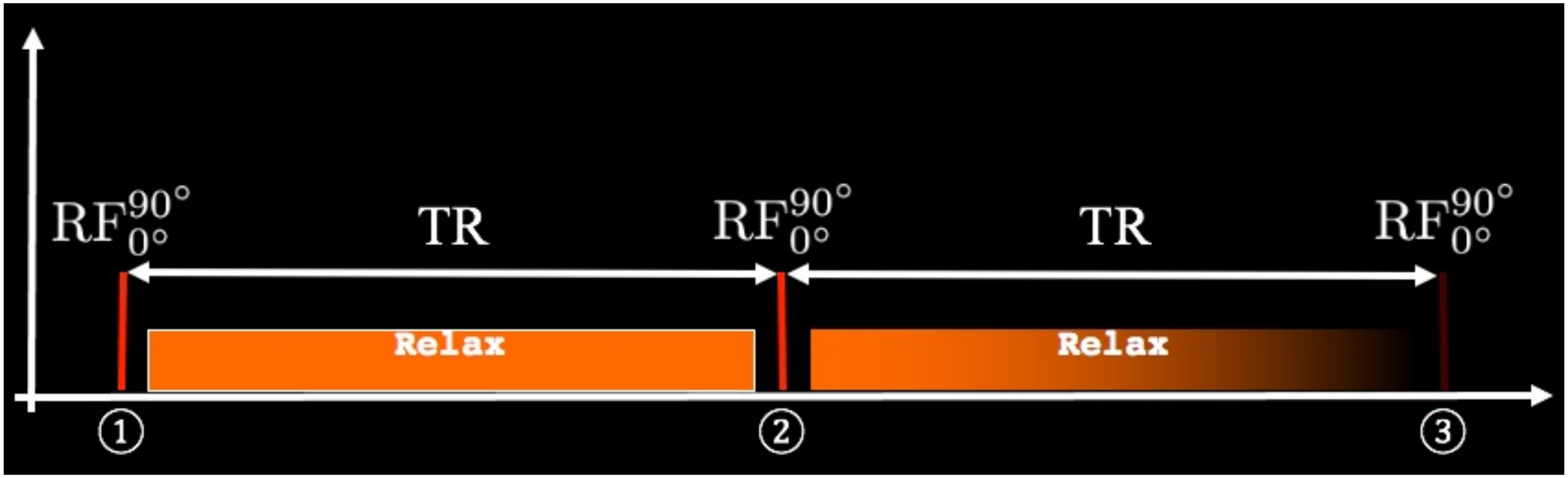
Inversion Recovery
T2-preparation
Fat Saturation
Tagging

Spoiled Gradient Echo
SPGR, FLASH, T1 FFE
bSSFP
FIESTA, Balanced FFE, True FISP
Spin Echo
Fast or Turbo Spin Echo

Pulse Sequence Definitions

$M_z^{(n)}(0_-)$	Longitudinal magnetization ***before*** the n^{th} event.
$M_z^{(n)}(0_+)$	Longitudinal magnetization ***after*** the n^{th} event.
$M_{xy}^{(n)}(0_-)$	Transverse magnetization ***before*** the n^{th} event.
$M_{xy}^{(n)}(0_+)$	Transverse magnetization ***after*** the n^{th} event.

Pulse Sequence Definitions



TR - Repetition Time

Duration of basic pulse sequence repeating block

At least one echo acquired per TR

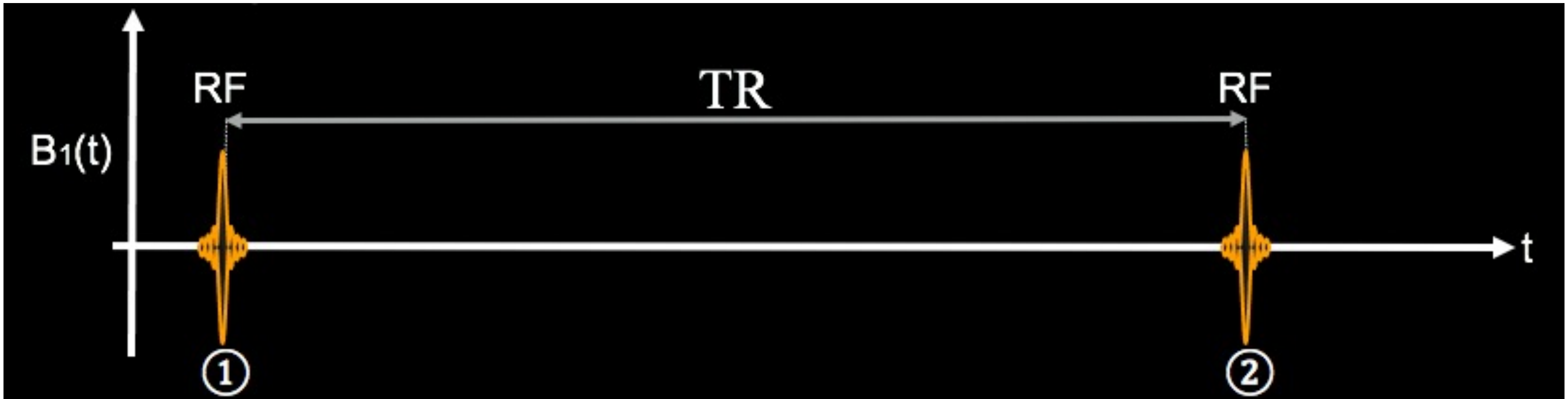
TE - Echo Time

Time from excitation to the maximum of the echo

Data is recorded at time TE to form an image

Echo can occur as a result of a *gradient-echo* or *spin-echo*

Typical Pulse Sequence



$$\vec{M}^{(1)}(0_-) = \vec{M}_0 = \begin{bmatrix} 0 \\ 0 \\ M_0 \end{bmatrix}$$ Equilibrium Magnetization

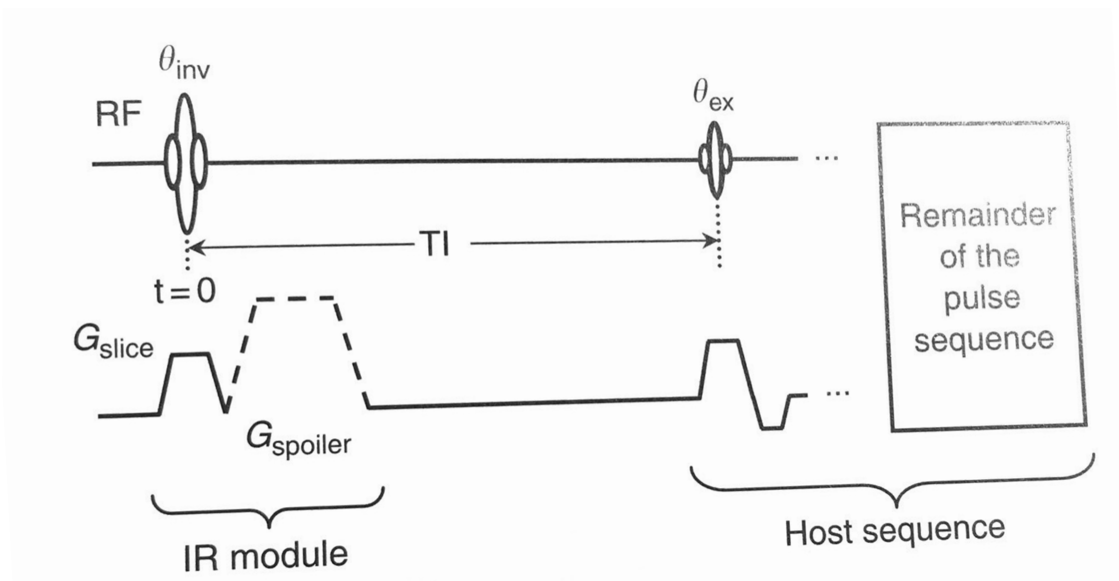
$$\vec{M}^{(1)}(0_+) = RF_\theta^\alpha \vec{M}^{(1)}(0_-)$$ During RF Excitation (1)

$$\vec{M}^{(2)}(0_-) = \mathrm{E}(T_1, T_2, TR)\vec{M}^{(1)}(0_+)$$ Relaxation

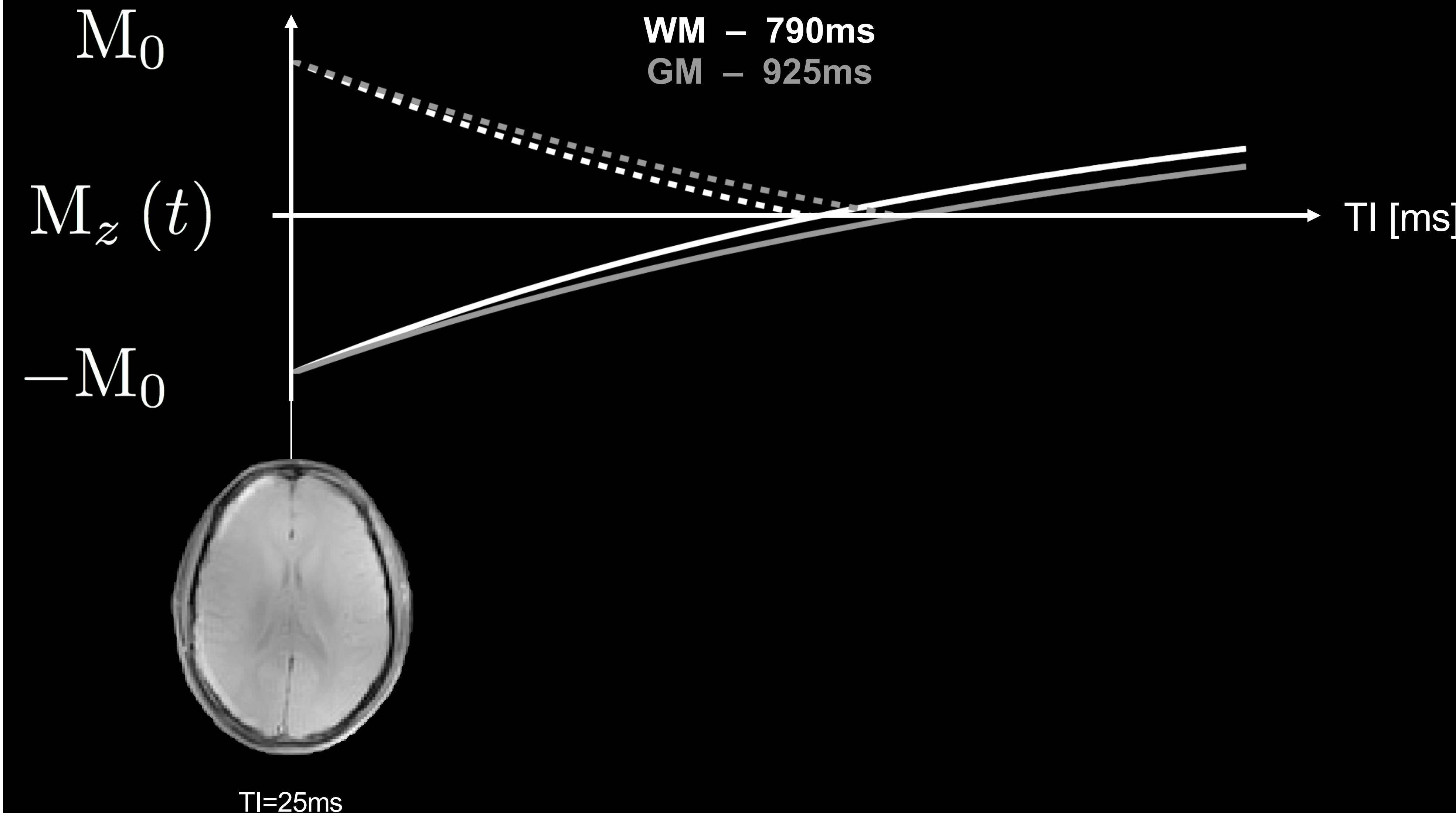
RF Excitation (2) $$\vec{M}^{(2)}(0_+) = RF_\theta^\alpha \vec{M}^{(2)}(0_-)$$

Basic Sequences - Inversion Recovery (IR)

- **Pulse sequences with an inversion pulse followed by a time delay prior to RF excitation are *inversion recovery* pulses**
- **Allow for T1-weighting or T1-weighted magnetization preparation**
- **Delay between inversion and excitation pulses is known as the *inversion time (TI)***
- **IR module followed by Host sequence (e.g. RARE, EPI, etc)**

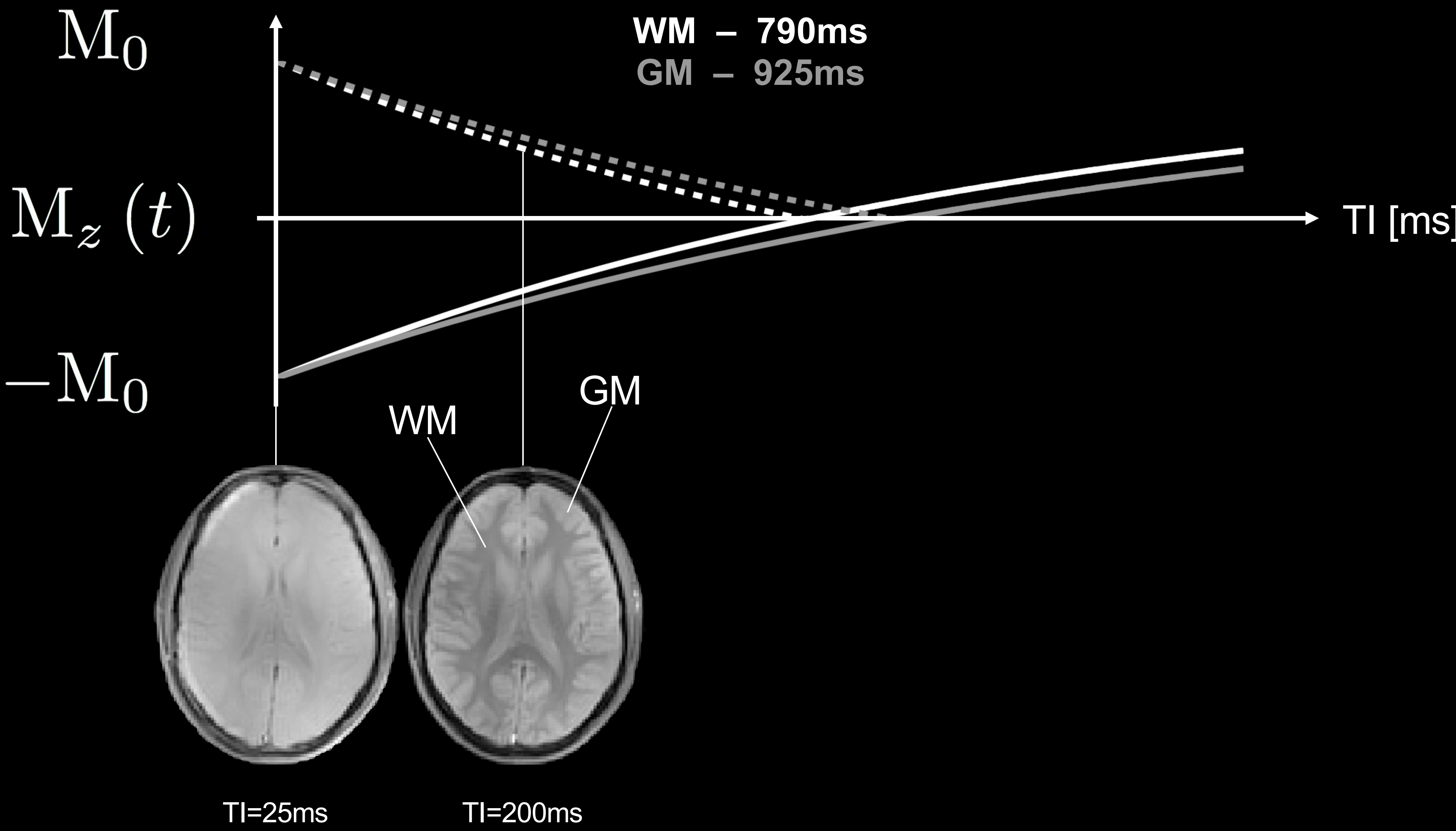


Spin Echo Inversion Recovery



Spin Echo with TE=12ms, TR=2000ms

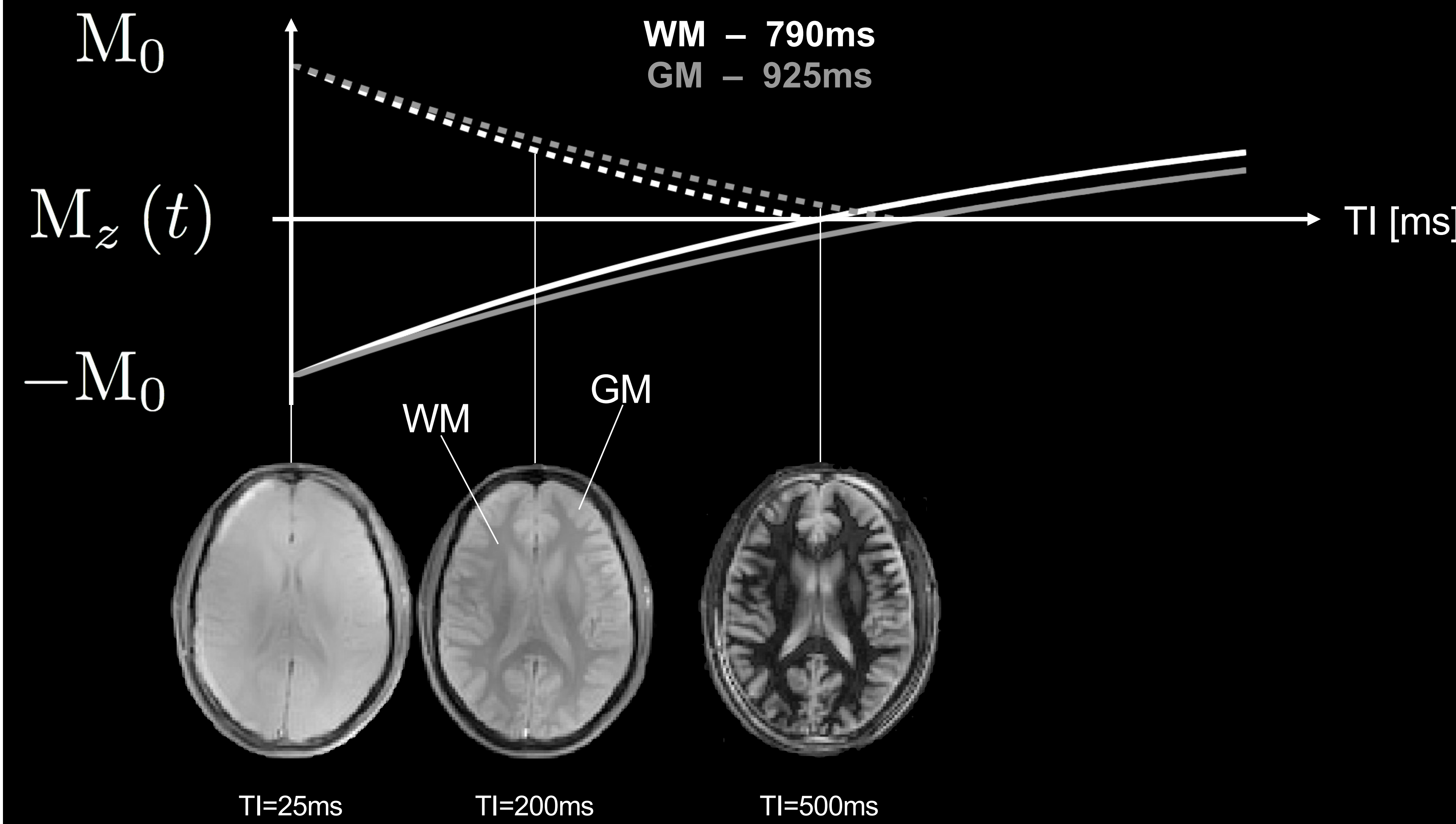
Spin Echo Inversion Recovery



Spin Echo with TE=12ms, TR=2000ms

David Geffen School of Medicine

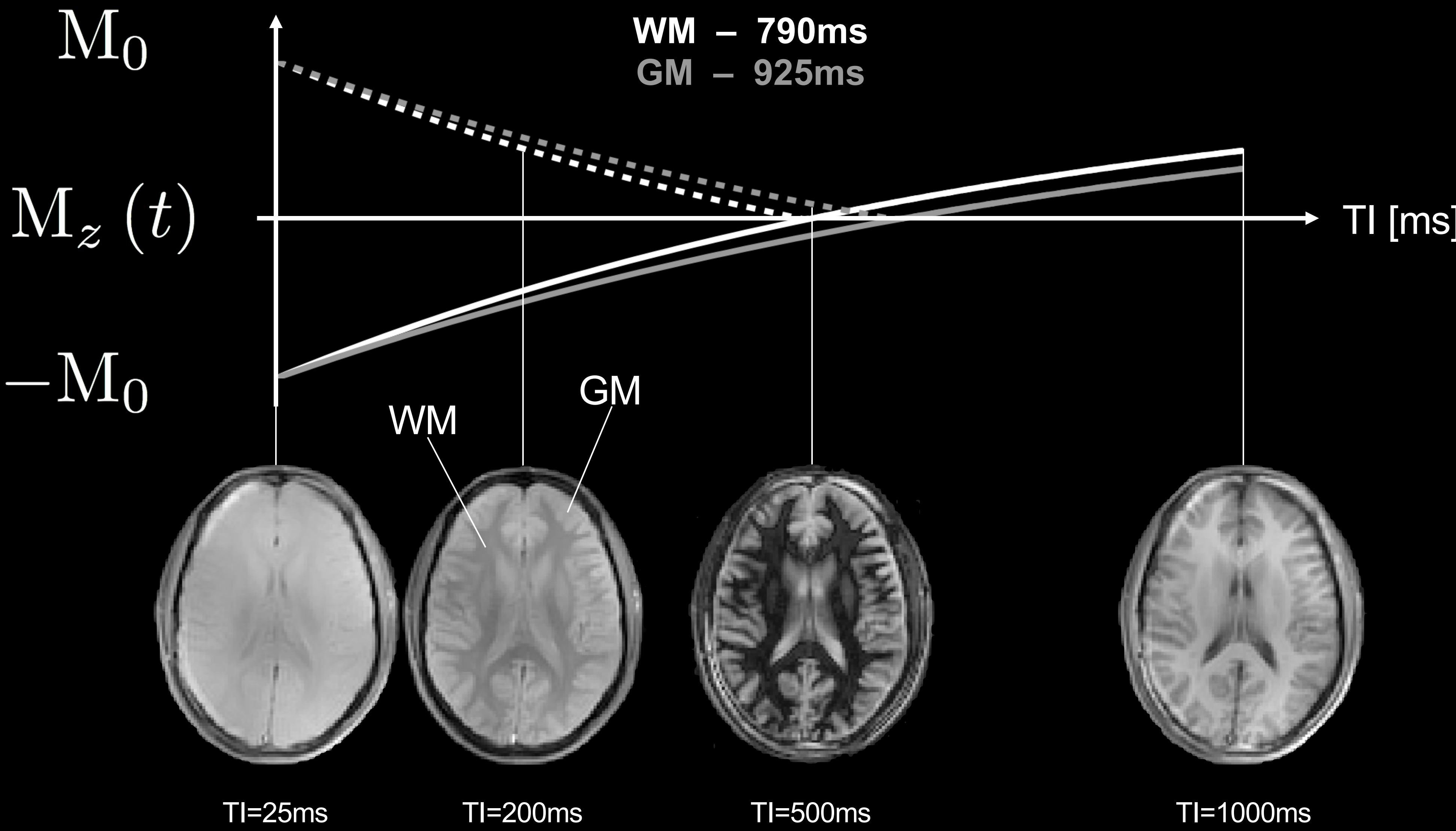
Spin Echo Inversion Recovery



David Geffen School of Medicine

Spin Echo with TE=12ms, TR=2000ms

Spin Echo Inversion Recovery



Spin Echo with TE=12ms, TR=2000ms

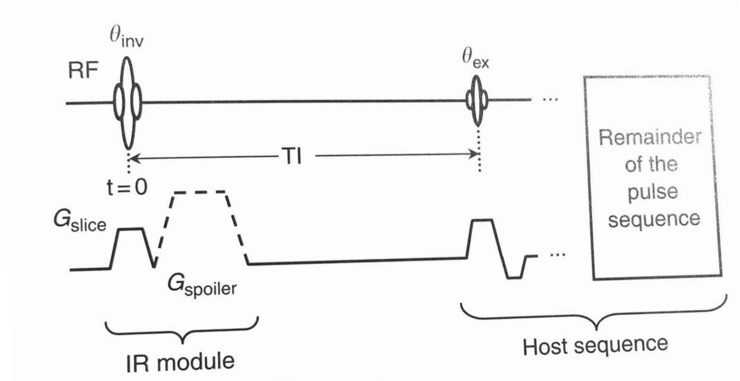
Spin Echo Inversion Recovery

TI=25ms TI=200ms TI=500ms TI=1000ms

TE=12ms, TR=2000ms

David Geffen
School of Medicine

Basic Sequences - Inversion Recovery (IR)

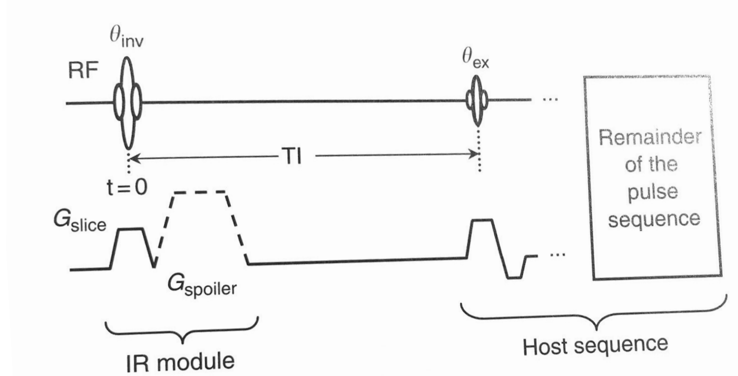


- **Signal Equation for IR:**
  - **Defining the transverse and longitudinal magnetization as:**

$$M_{xy} = M_0 \sin\theta_{inv}$$

$$M_z = M_0 \cos\theta_{inv}$$

Basic Sequences - Inversion Recovery (IR)



- **By applying a spoiler gradient after the inversion pulse, Mxy = 0 (all that is left is longitudinal magnetization)**

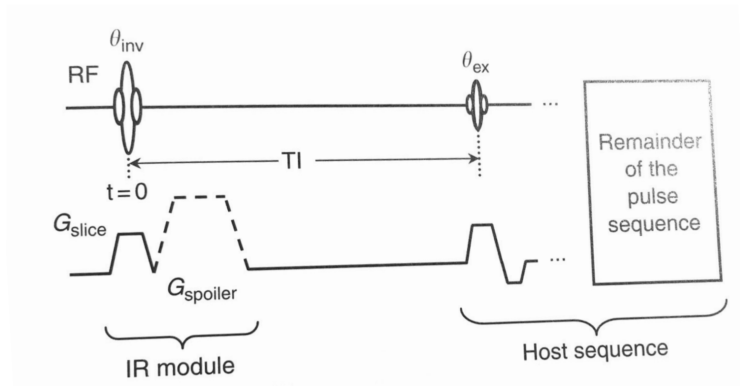
- **Bloch equation:**

$$\frac{dM_z}{dt} = \frac{M_0 - M_z}{T_1}$$

- **With the solution:**

$$M_z(t) = M_0\left[1 - \left(1 - \cos\theta_{inv}\right)e^{-t/T_1}\right]$$

Basic Sequences - Inversion Recovery (IR)



- **If TR is not infinitely long, then the equations for the spin echo and turbo spin echo are:**
- **Spin Echo (SE)**

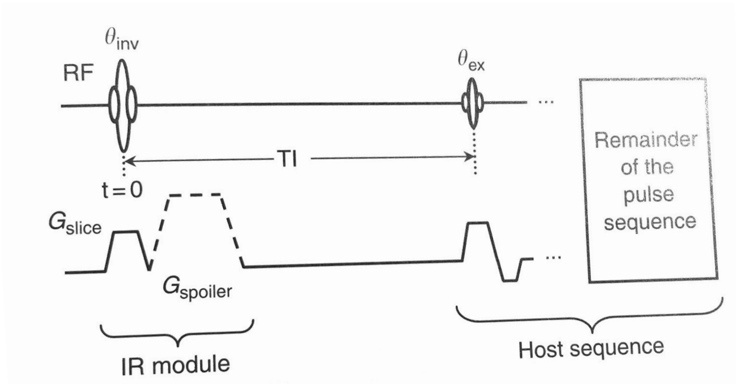
$$M_z(t) = M_0\left[1-(1-\cos\theta_{inv})e^{-t/T_1} + e^{-TR/T_1}\right]$$

- **Turbo spin echo (TSE) or RARE**

$$M_z(t) = M_0\left[1-(1-\cos\theta_{inv})e^{-t/T_1} + e^{-(TR-TE_{last})/T_1}\right]$$

TE_{last} = Last echo in echo train

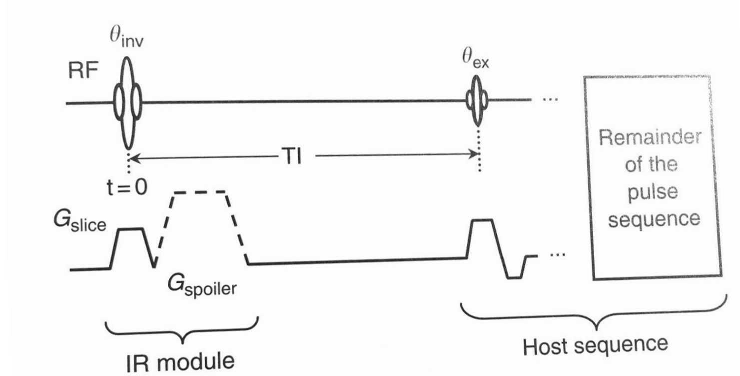
Basic Sequences - Inversion Recovery (IR)



- **Following the IR pulse, the longitudinal magnetization recovers along the z-axis until being nutated by the excitation pulse.**
- **The available magnetization is thus:**

$$M_z(TI) = M_0\left[1 - 2e^{-TI/T_1}\right]$$

Basic Sequences - Inversion Recovery (IR)



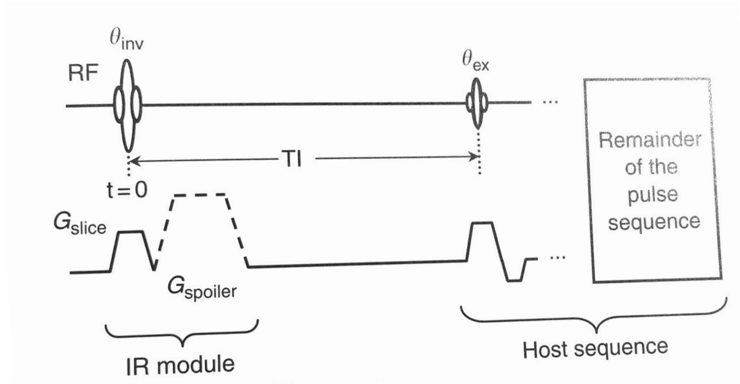
- **Magnetization becomes zero (nulled) when (for infinitely long TR)**

$$TI_{null} = T_1 \ln 2$$

$$TI_{null} = T_1 \left[\ln 2 - \ln\left(1 + e^{-(TR)/T_1} \right) \right]$$ *Spin Echo*

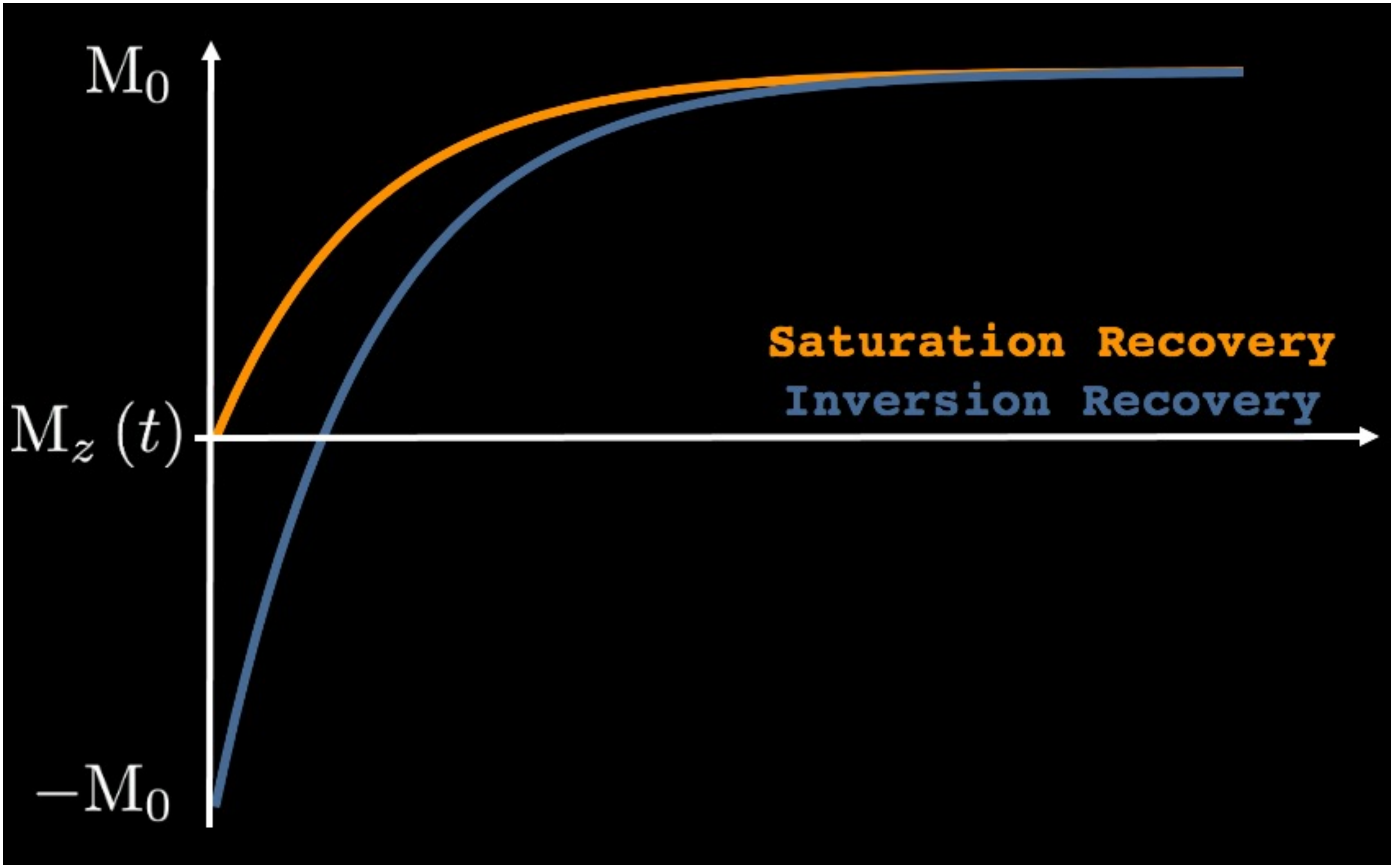
$$TI_{null} = T_1 \left[\ln 2 - \ln\left(1 + e^{-(TR - TE_{last})/T_1} \right) \right]$$ *Turbo Spin Echo*

Basic Sequences - Inversion Recovery (IR)



- **If the inversion pulse is 180 degree, this is an *inversion recovery* sequence**
- **If the inversion pulse is 90 degrees, this is a *saturation recovery* sequence**

Saturation Recovery (SR) vs. Inversion Recovery (IR)



Saturation Condition

- **The *__Saturation Condition__* states:**

$$M_z^{(n)}(0_+) = 0, n \geq 1$$

M_z is ZERO after the event (RF pulse).

This is true if the M_{xy} is "gone" before the next 90° RF-pulse is applied:

No M_{xy} to convert to M_z

How? $TR >> T_2$

What if $TR <\sim 3T_2$?

M_{xy} can be converted back to M_z

Corrupts/complicates image contrast

Solution? Spoiler gradients to disperse M_{xy}

Steady-state solution arises if the saturation conditions are met/enforced

Saturation Recovery Contrast Optimization

$$I(\vec{r})_{TR \rightarrow TR_{opt}} \propto \text{Maximum } T_1 \text{ contrast}$$

$$TR_{opt} = \frac{\ln\left(\frac{T_{1,A}}{T_{1,B}}\right)}{\frac{1}{T_{1,B}} - \frac{1}{T_{1,A}}}$$

Inversion Recovery Contrast Optimization

$$I(\vec{r}) \propto \rho(\vec{r}) \left(1 - 2e^{-TI/T_1(\vec{r})} + e^{-TR/T_1(\vec{r})}\right)$$

Image contrast is controlled by TI and TR

Maximum contrast if $\mathrm{TR} \gg T_1$ and,

$$TI_{opt} = \frac{\ln\left(\frac{T_{1,A}}{T_{1,B}}\right)}{T_{1,A} - T_{1,B}} T_{1,A} T_{1,B}$$

Inversion Recovery

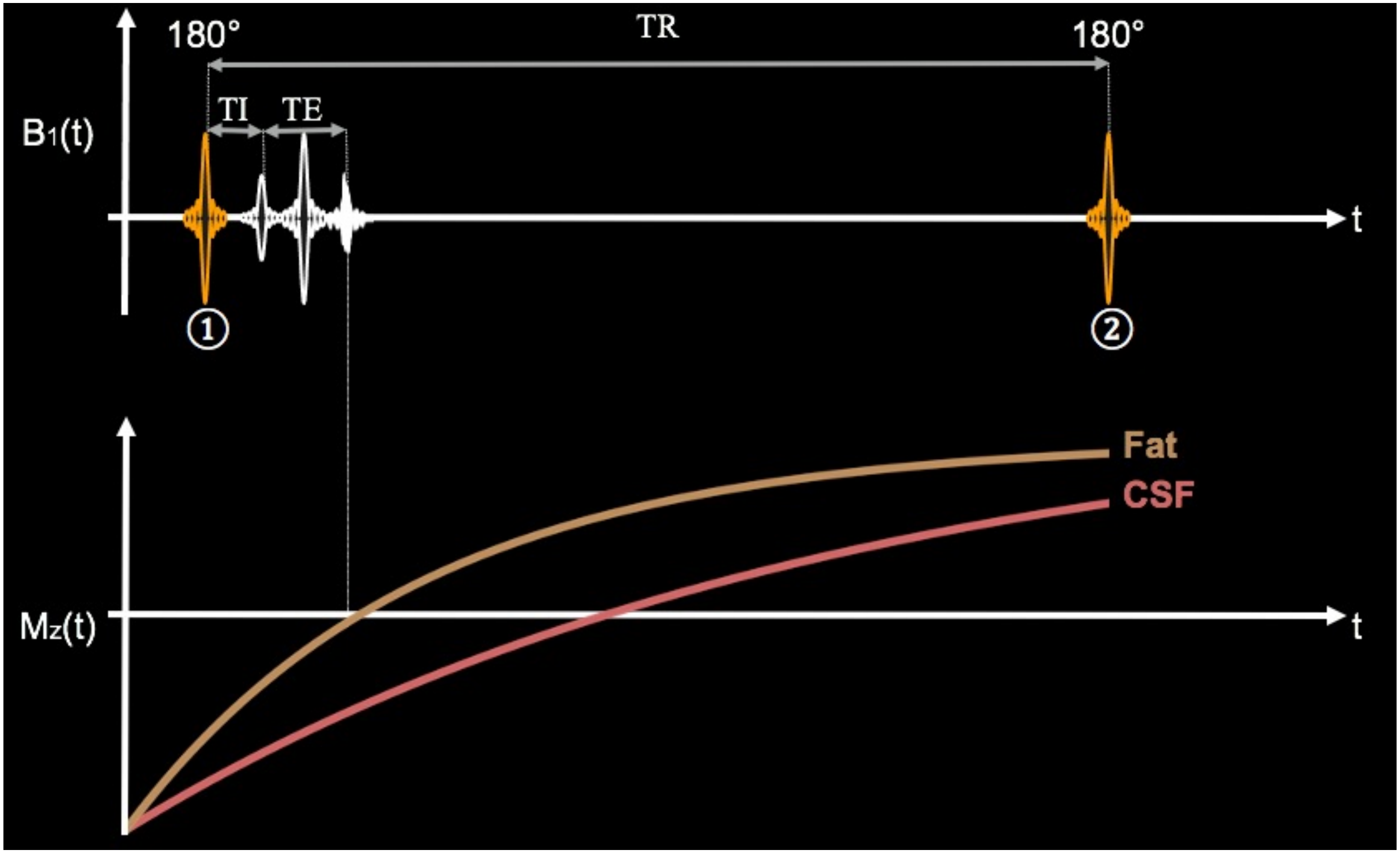
- **Greater T_1 contrast than SR**

- **T_1 species nulling/attenuation**
  - FLAIR (Fluid Attenuated Inversion Recovery)
  - STIR (Short Tau Inversion Recovery)

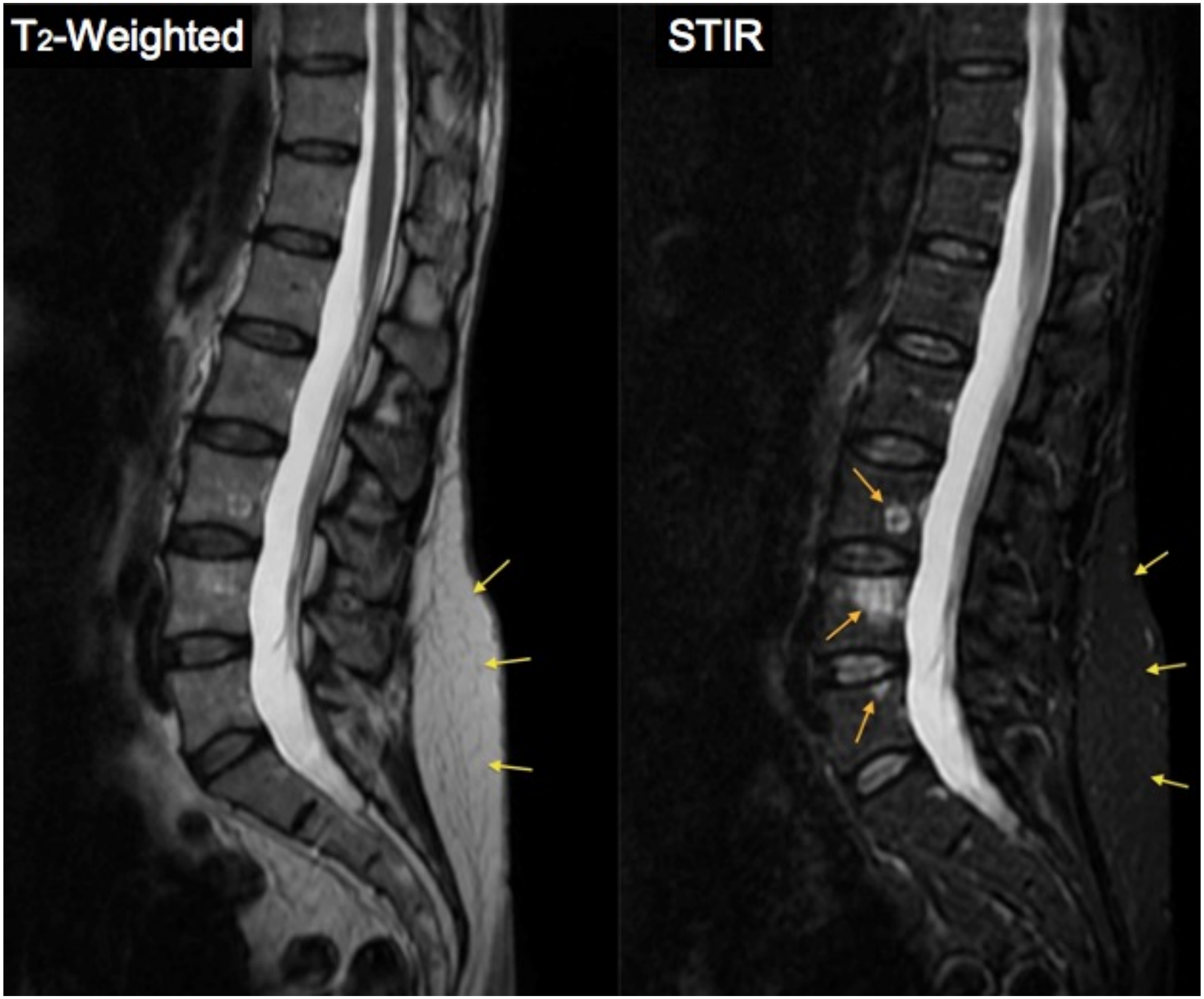
- **IR is better than SR for generating contrast when:**
  - $\rho(A)=\rho(B)$ and $T_2(A)=T_2(B)$
  - AND
  - $T_1(A)$ and $T_1(B)$ are slightly different

- **Quantitative T_1 mapping**

Short Tau Inversion Recovery (STIR)

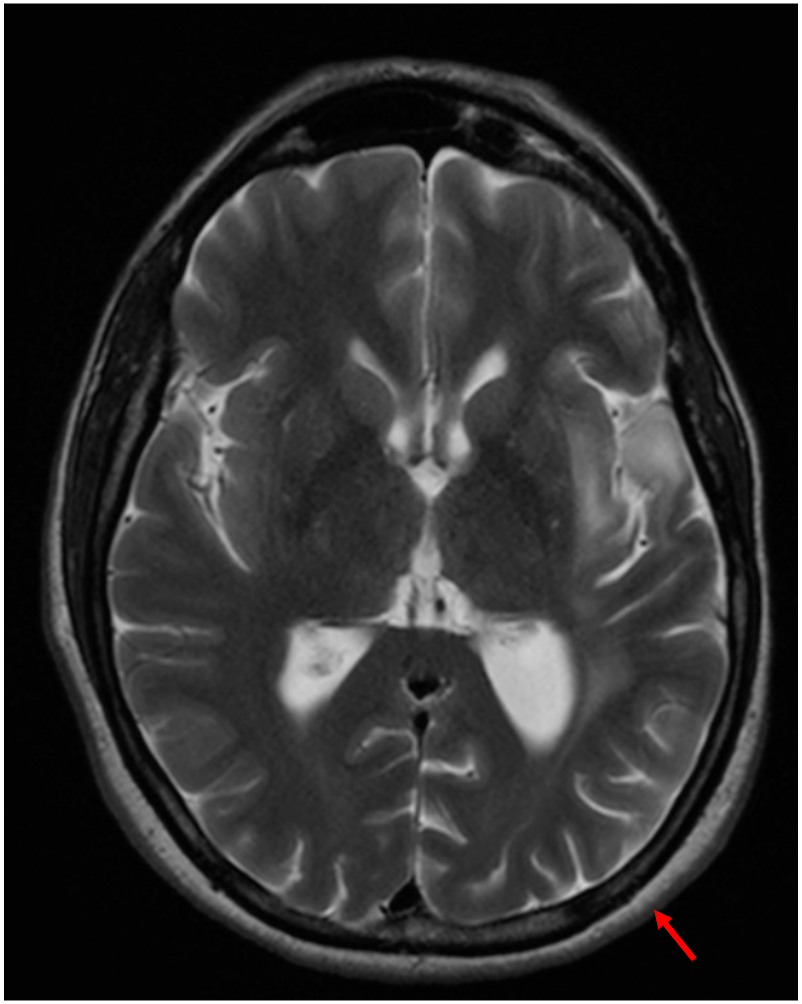


Short Tau Inversion Recovery (STIR)

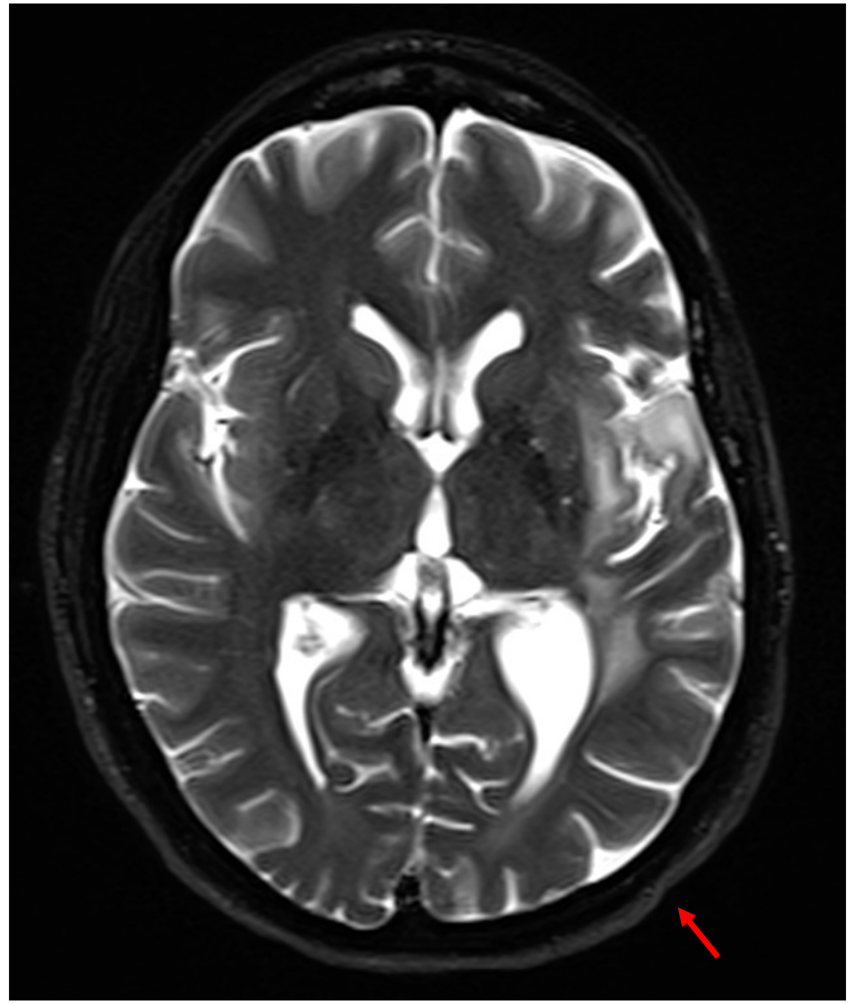


Short Tau Inversion Recovery (STIR)

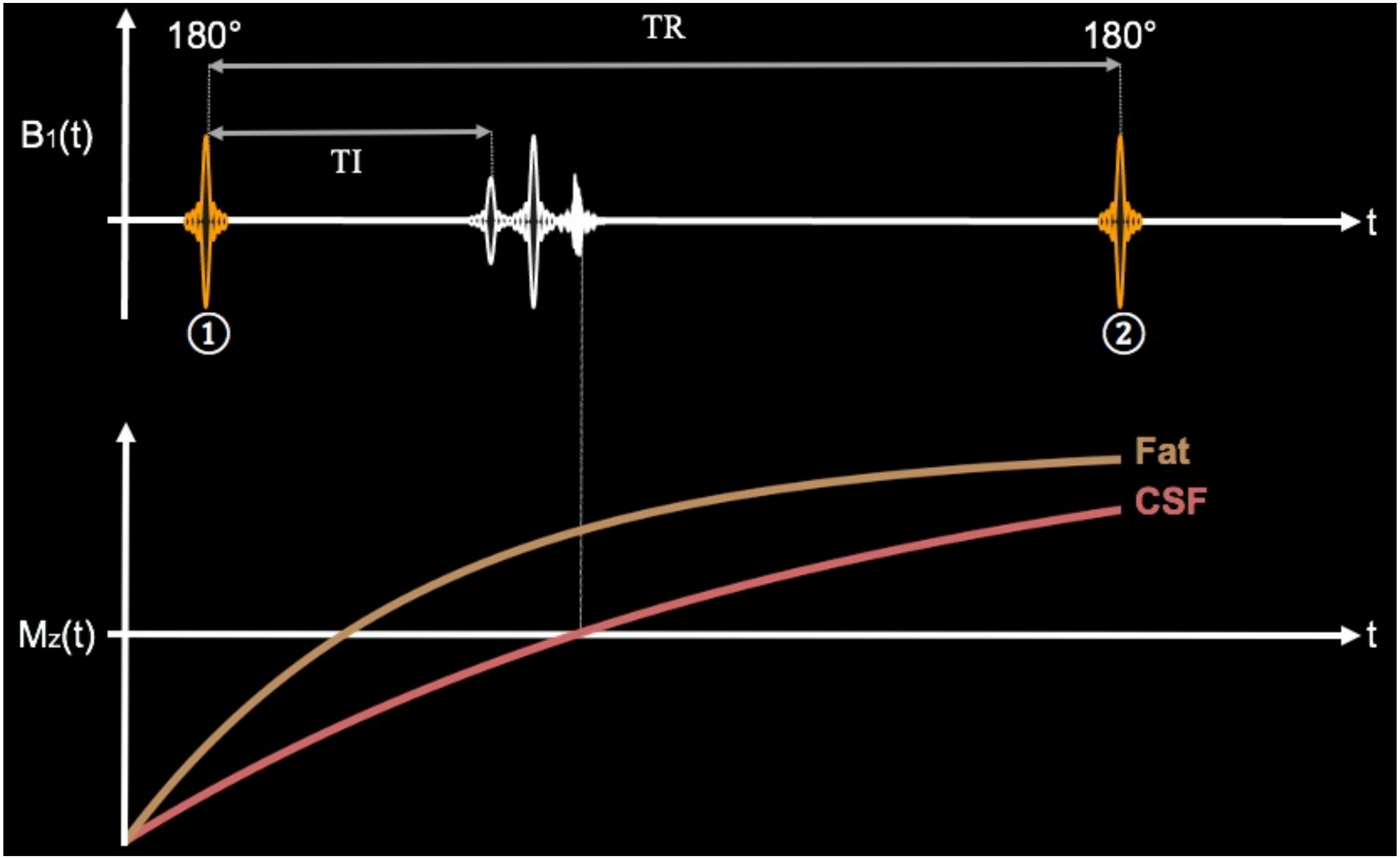
T2-Weighted TSE



T2-Weighted **STIR** TSE

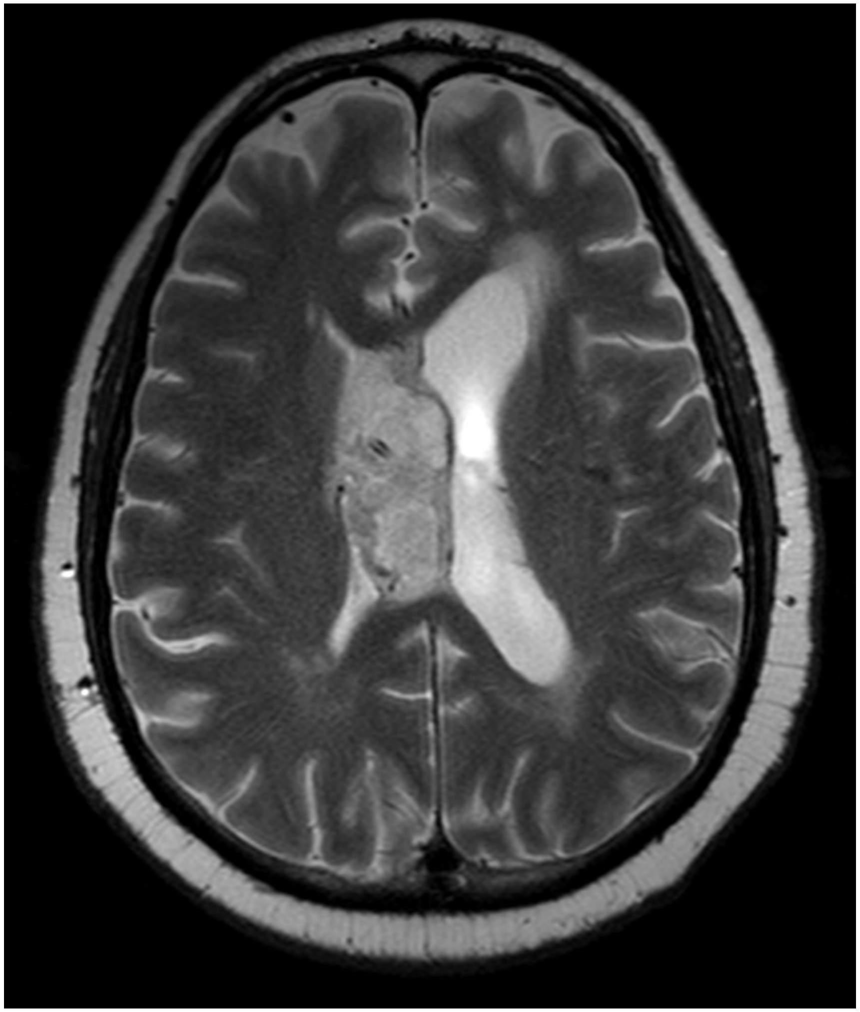


FLuid Attenuated Inversion Recovery (FLAIR)

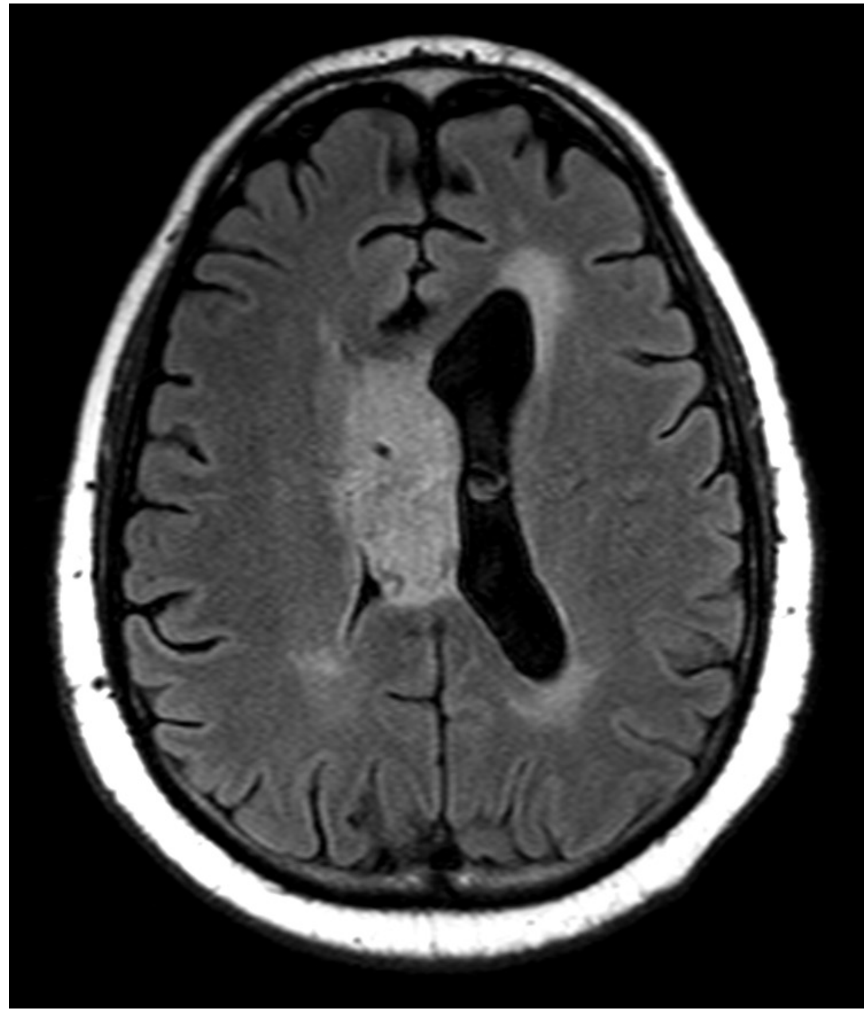


FLuid Attenuated Inversion Recovery (FLAIR)

T2-Weighted TSE

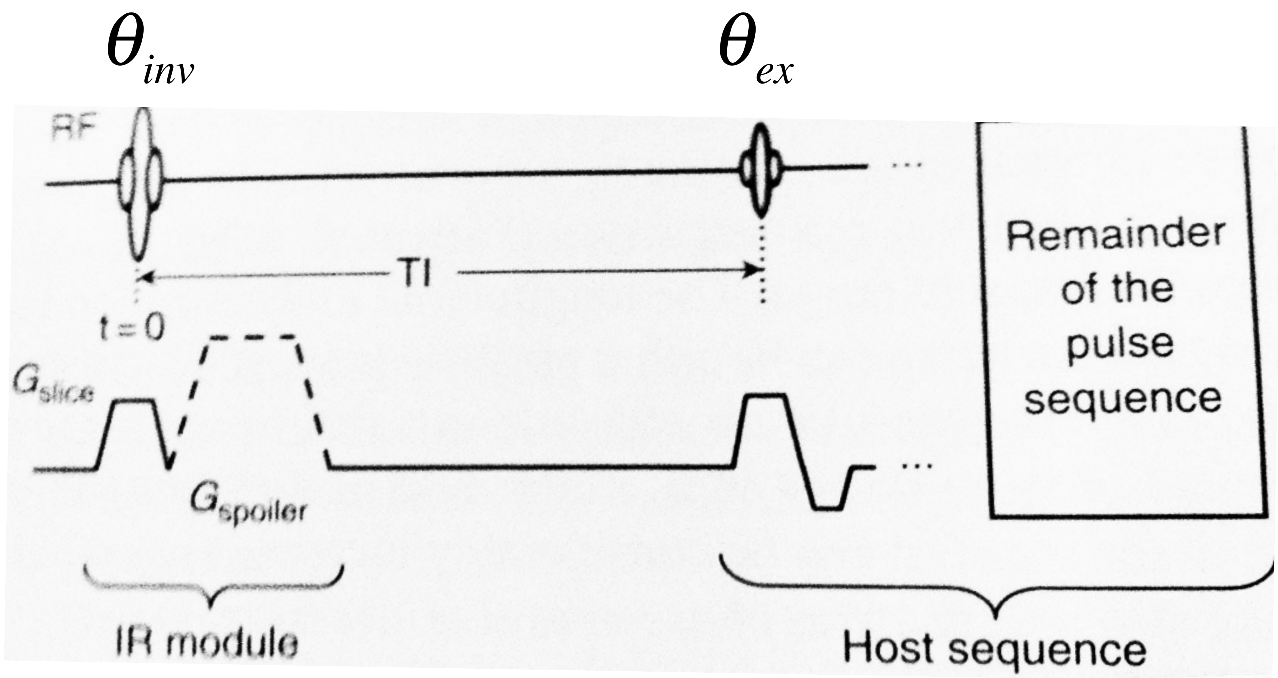


T2-Weighted **FLAIR**



T_1 Measurement with Inversion Recovery (IR)

- The most common method for estimating tissue T1 is through the use of an inversion recovery sequence
- Involves an IR module or "preparation" prior to a "host sequence"



T_1 Measurement with Inversion Recovery (IR)

- For this sequence, the time-dependent longitudinal magnetization is:

$$M_z(t) = M_0\left[1-\left(1-\cos\theta_{inv}\right)e^{-t/T1}\right]$$

- which assumes an infinitely long TR. With finite TR, $M_z(t)$ depends on details of the host sequence.
- For Spin Echo and Turbo Spin Echo (TSE)/RARE sequences:

$$M_z(t) = \begin{cases} M_0\left[1-\left(1-\cos\theta_{inv}\right)e^{-t/T1}+e^{-TR/T1}\right] & \textit{Spin Echo} \\ M_0\left[1-\left(1-\cos\theta_{inv}\right)e^{-t/T1}+e^{-\left(TR-TE_{last}\right)/T1}\right] & \textit{TSE / RARE} \end{cases}$$

Last Echo in the Echo SE Train

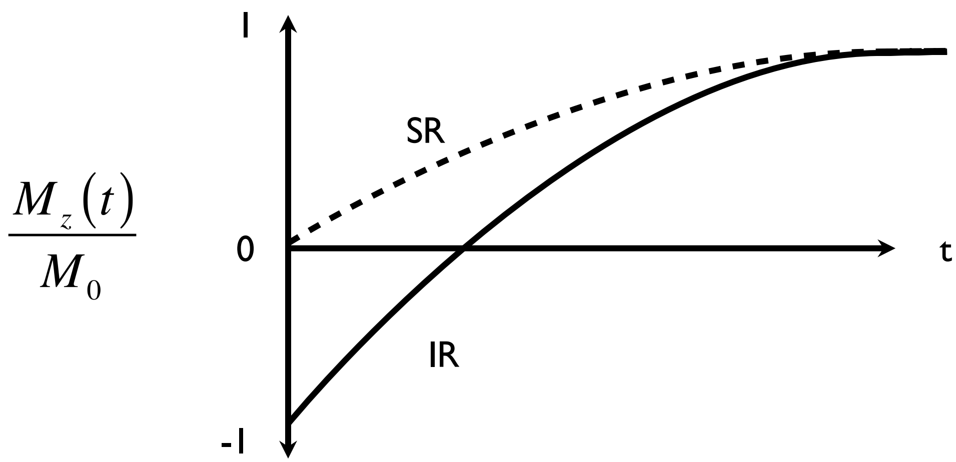
T_1 Measurement with Inversion Recovery (IR)

- For $\theta_{inv} = \pi$ this is complete inversion and the longitudinal magnetization becomes:

$$M_z(t) = M_0\left[1 - 2e^{-t/T1}\right]$$

- For $\theta_{inv} = \frac{\pi}{2}$ this results in *saturation recovery* (SR)

$$M_z(t) = M_0\left[1 - e^{-t/T1}\right]$$



T_1 Measurement with Inversion Recovery (IR)

- To quantify T_1, a series of IR images are acquired from the same location, each with a different TI while keeping all parameters identical.
- To avoid signal saturation, a long TR must be used ($TR > 4T_{1max}$)
- Note that a "phase sensitive" IR sequence needs to be employed to discriminate negative magnetization (particularly around the null point).

$$M_z(t) = \begin{cases} M_0\left[1-\left(1-\cos\theta_{inv}\right)e^{-t/T1} + e^{-TR/T1}\right] & \textit{Spin Echo} \\ M_0\left[1-\left(1-\cos\theta_{inv}\right)e^{-t/T1} + e^{-\left(TR-TE_{last}\right)/T1}\right] & \textit{TSE / RARE} \end{cases}$$

T_1 Measurement with Inversion Recovery (IR)

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$$R_1 = \frac{1}{T_1} = -\frac{1}{TI}\ln\left(\frac{1-\frac{M_z}{M_0}}{\left(1-\cos\theta_{inv}\right)}\right)$$

$$R_1 = \frac{1}{T_1} = -\frac{1}{TI}\ln\left(\frac{1}{2}\left(1-\frac{M_z}{M_0}\right)\right) \quad \text{for } \theta = \pi$$

T_1 Measurement with Inversion Recovery (IR)

- Note that either the inversion time (TI) or the inversion flip angle can be changed and **nonlinear regression** can be used to fit R1 or T1

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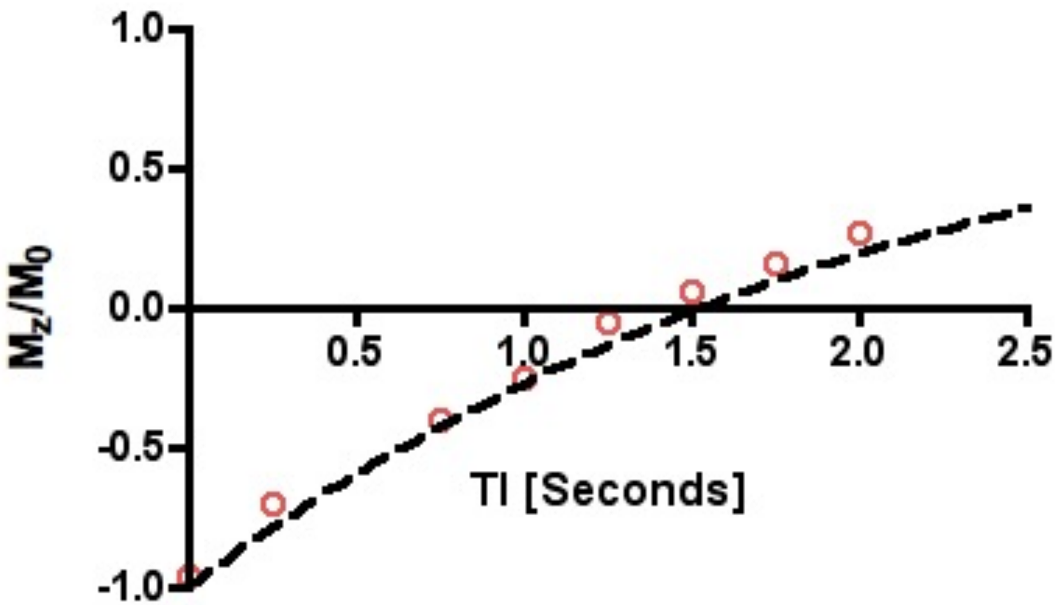
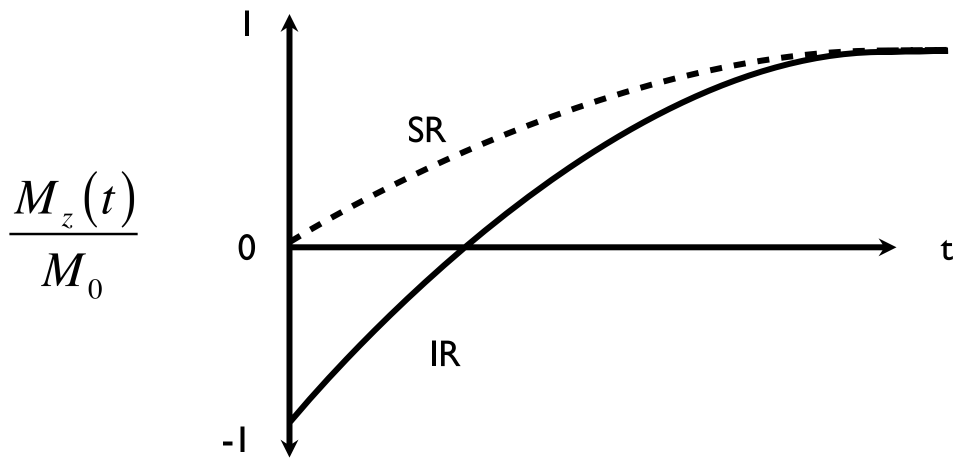


T_1 Measurement with Saturation Recovery (SR)

- For an SR sequence, the IR prep is obtained by using ϑ_{inv} = ***π/2*** , followed by a normal excitation ϑ_{ex} = ***π/2***.
- This is considered a **saturation recovery** sequence.
- Different "T1" values can then be plotted and a similar fit can be used to estimate T1

$$M_z(t) = M_0\left[1 - e^{-t/T1}\right]$$



Summary

- SR and IR are "modules" that can be applied to (most) host sequences
- SR/IR provides controllable T1 contrast/weighting
- SR/IR can be used to "null" or zero out tissues of interest to increase anatomical conspicuity
- Can also be used to quantify T1