Advanced MRI and fMRI Acquisition Methods

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National Institutes of Health

National Institute of Mental Health
Functional MRI Facility

Outline

- NMR: Review of physics basics
- MR Imaging: tools and techniques
- K-space trajectories
- Image acceleration
- Controlling the image contrast

NMR: Classical view

NMR: Nuclear Magnetic Resonance

- Effect is due to intrinsic spin of positively charged atomic nuclei of atoms.
- In the presence of an external magnetic field the nuclei absorb and re-emit electromagnetic radiation
- The radiation at a specific resonance frequency

NMR: Classical view

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- In the presence of an external magnetic field the nuclei absorb and re-emit electromagnetic radiation
- The radiation at a specific resonance frequency

$$\omega = \gamma B$$

- ω : angular frequency. $\omega = 2\pi \mathbf{v}$
- γ : gyromagnetic ratio
- B : strength of the external magnetic field

NMR: Classical view

NMR: Nuclear Magnetic Resonance

- $\omega = \gamma B$
- For ^{1}H (aka protons): $\bar{\gamma}$ = 42.58 MHz / T
  where $\bar{\gamma} = \gamma / 2\pi$
- Magnetization is a vector:

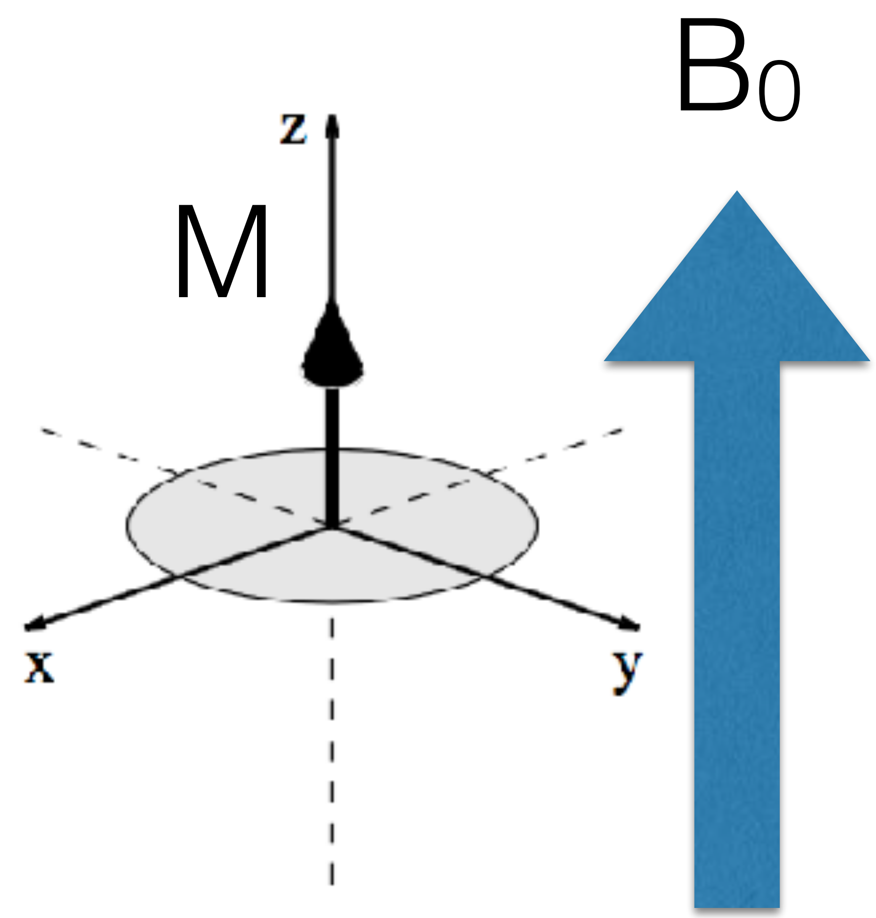
$$\mathbf{M} = (M_x, M_y, M_z)^T$$

- At equilibrium:

$$\mathbf{M} = (0, 0, M_0)^T$$

where

$$\frac{M_0 = N\gamma\hbar^2 I_z(I_z+1)B_0}{3kT}$$



NMR: Classical view

NMR: Nuclear Magnetic Resonance

1.5T: 64MHz
3.0T: 128MHz
7.0T: 298MHz

- $\omega = \gamma B$
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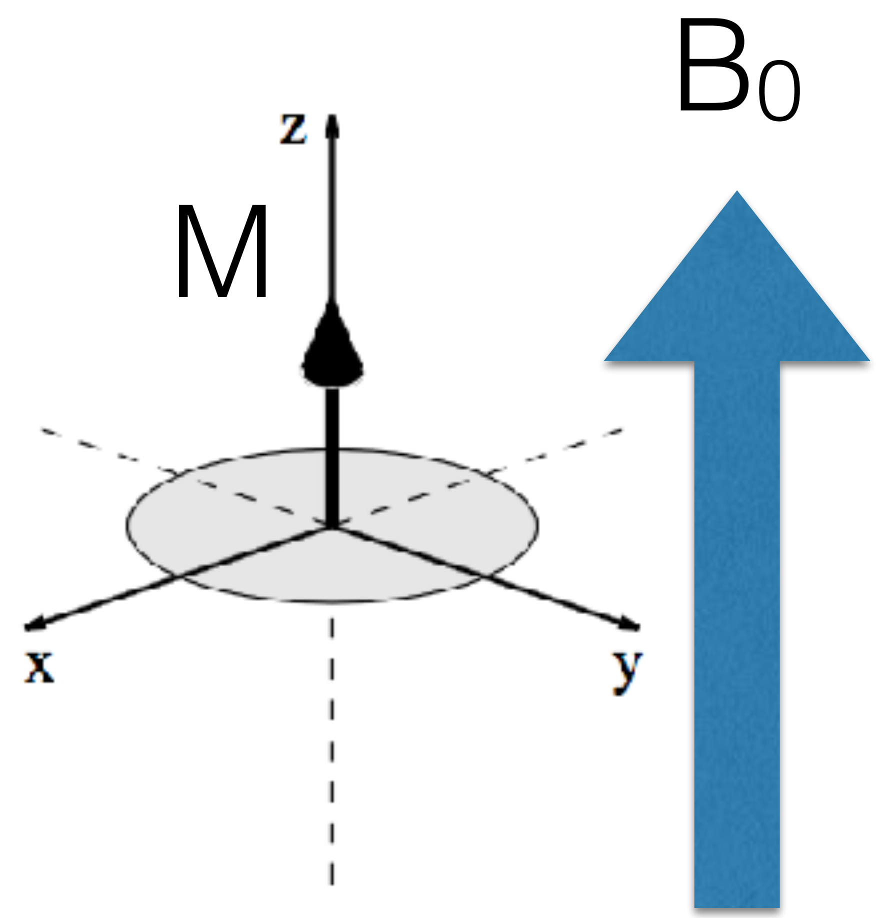
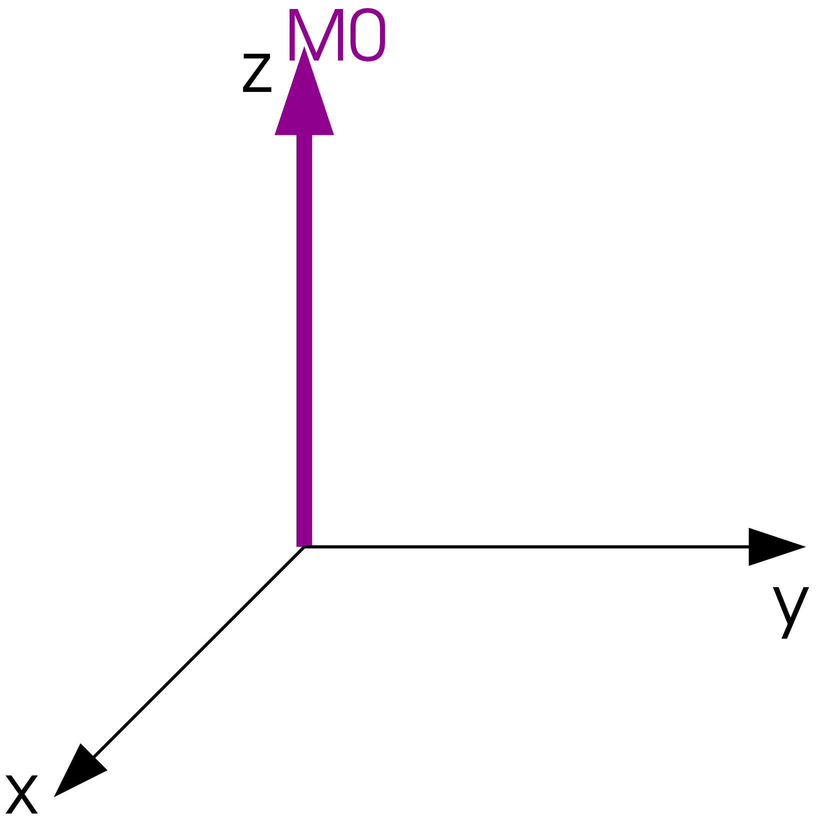


NMR: Classical view

Excitation, Precession and the Rotating Frame

- In lab frame at equilibrium

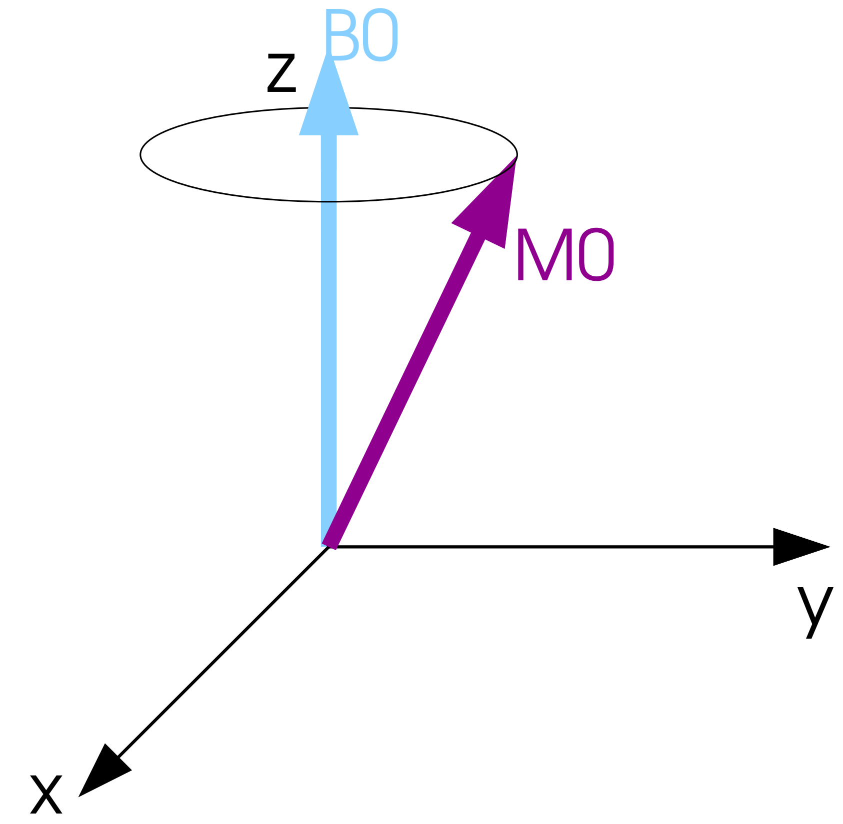


NMR: Classical view

Excitation, Precession and the Rotating Frame

- Excitation is the process of tipping the magnetization away from the direction of the main magnetic field.
- Once excited, the magnetization precesses around the magnetic field with angular frequency

$$\omega = \gamma B$$



NMR: Classical view

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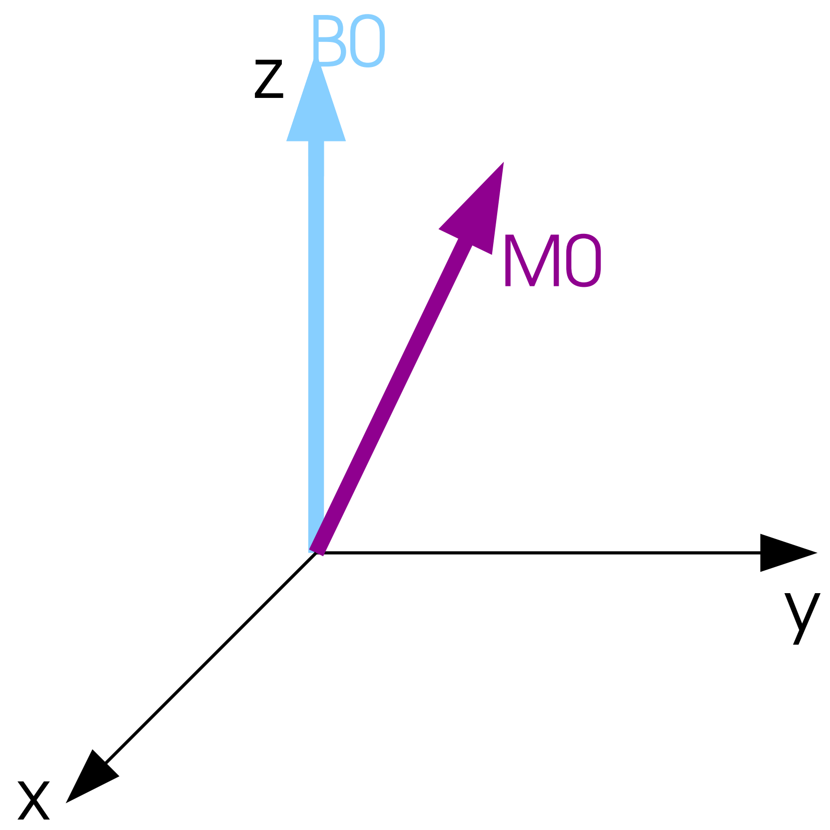
$$\omega = \gamma B$$

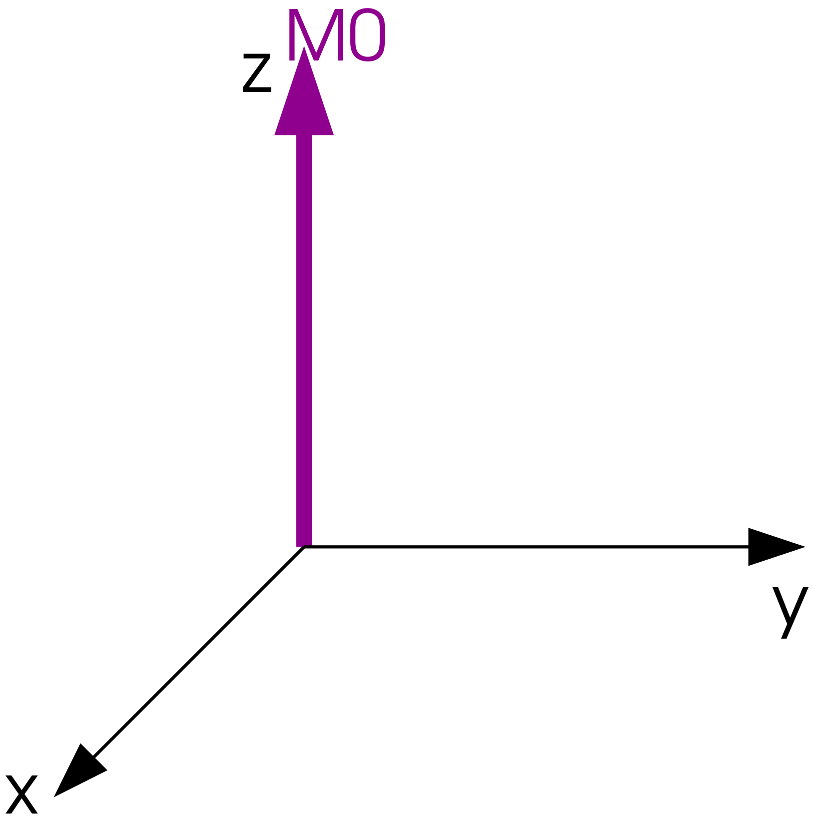
- It is convenient to work in a frame of reference rotating at $\omega = \gamma B$



NMR: Classical view

Excitation, Precession and the Rotating Frame

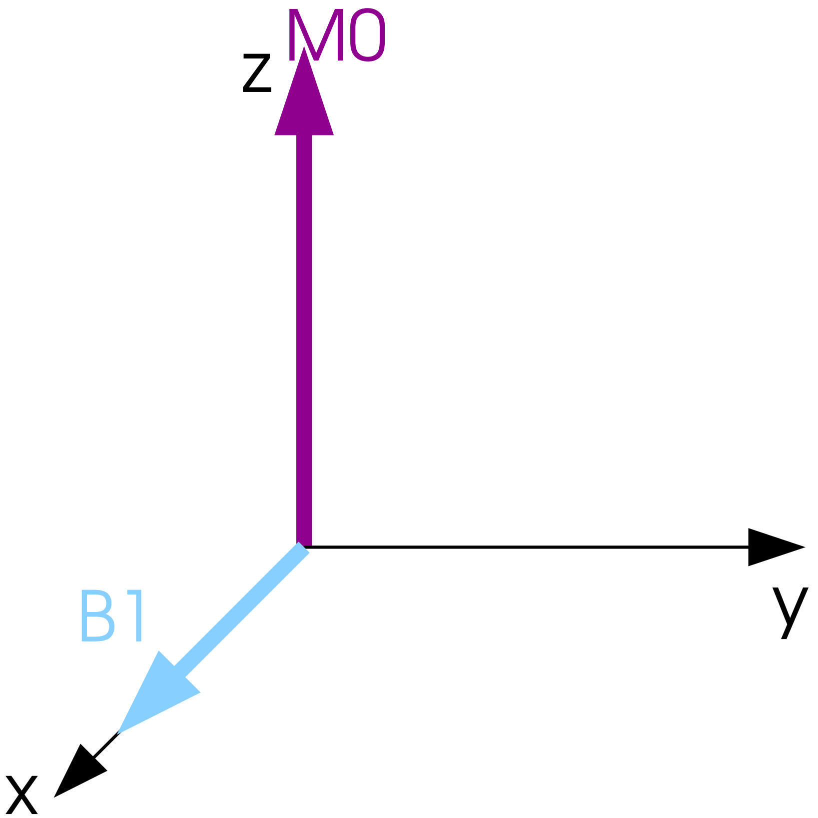
- In rotating frame at equilibrium



NMR: Classical view

Excitation, Precession and the Rotating Frame

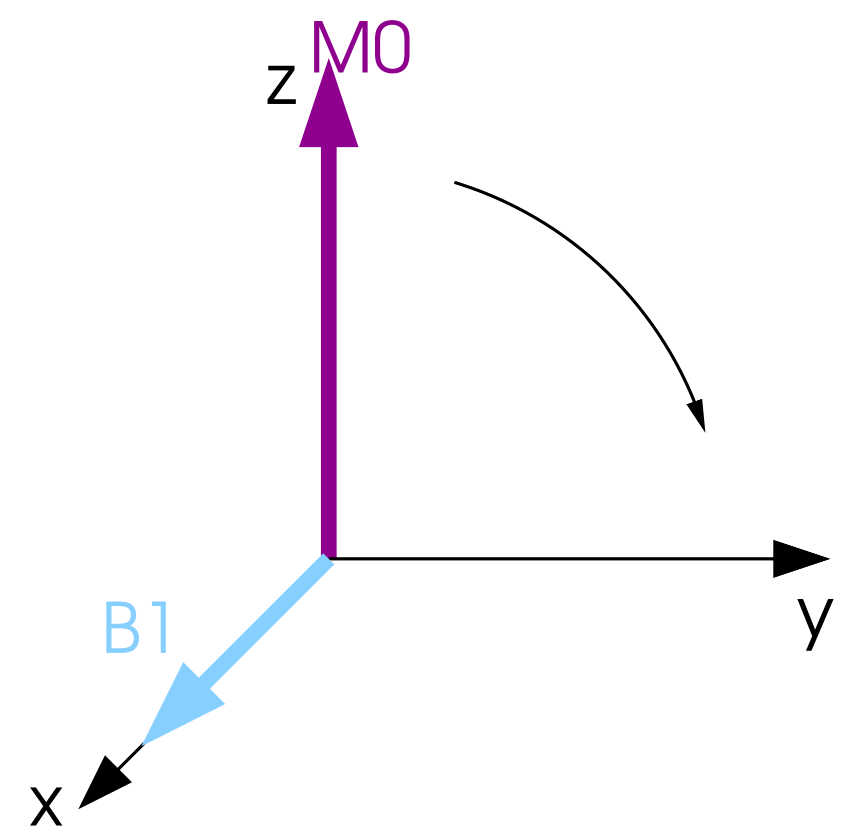
- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating frame) x-axis



NMR: Classical view

Excitation, Precession and the Rotating Frame

- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating-frame) x-axis
- $\omega_1 = \gamma B_1$
- Magnetization rotates towards (rotating-frame) y-axis



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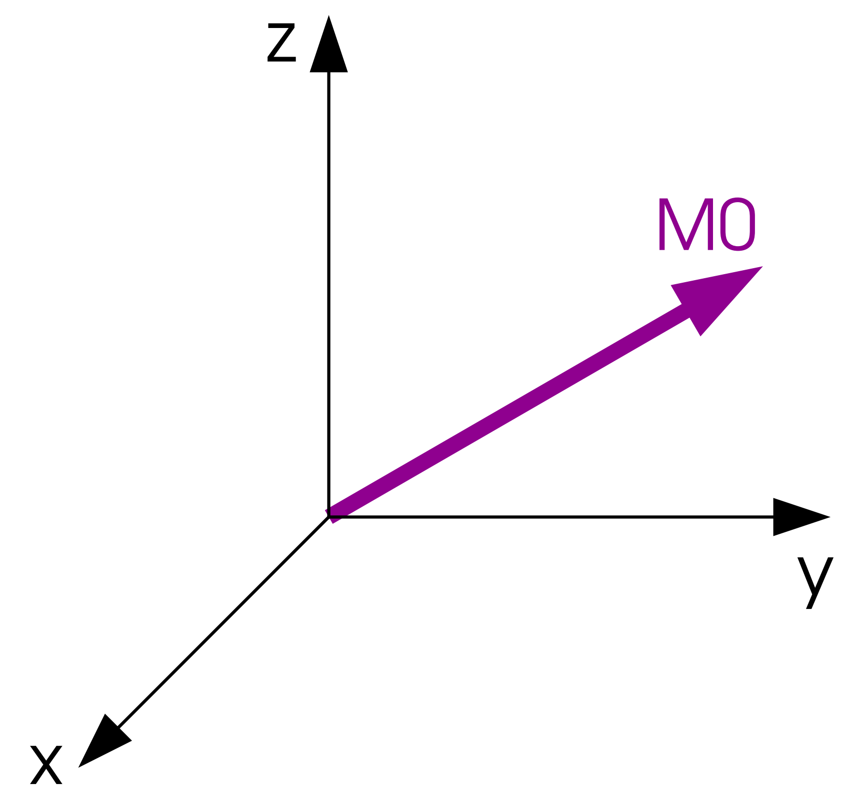
- Turn off B1 field when magnetization reaches the appropriate flip angle with respect to the z-axis



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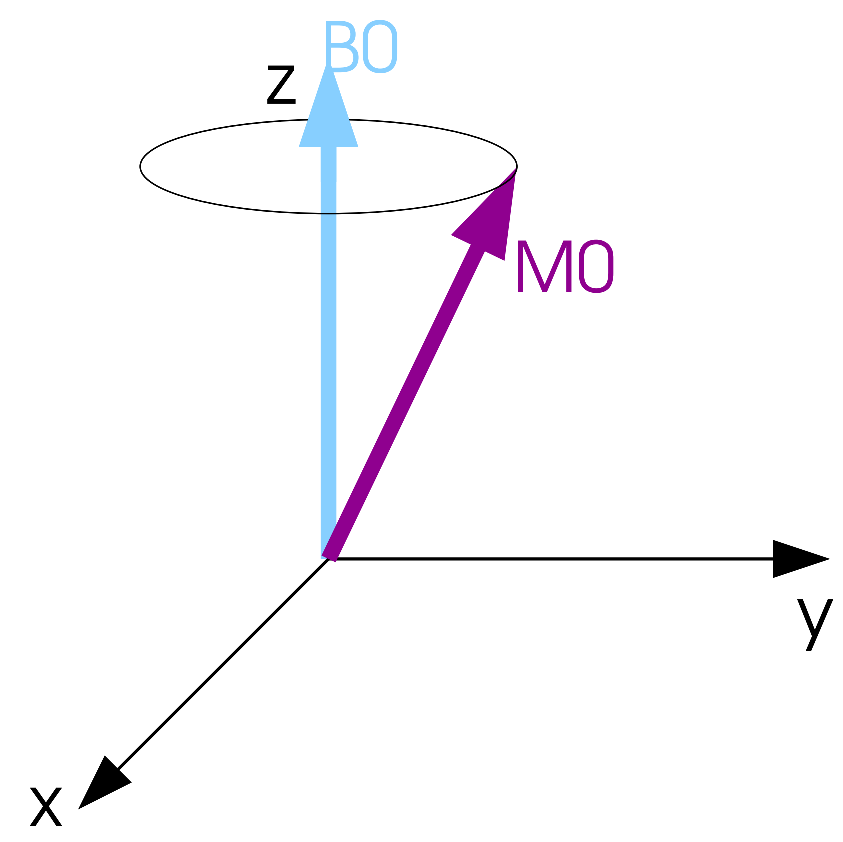
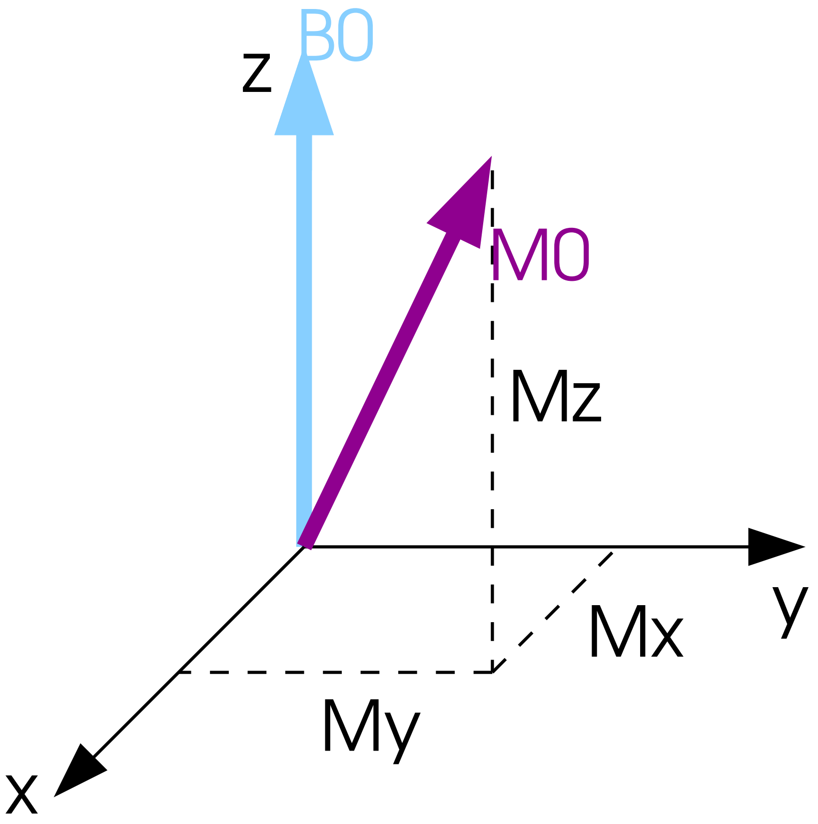
- Magnetization rotates towards (rotating-frame) y-axis
- Turn off B1 field when magnetization reaches the appropriate flip angle with respect to the z-axis
- Magnetization precesses and relaxes back to equilibrium



NMR: Classical view

MR signal

- **M** = $(M_x, M_y, M_z)^T$
- M_z is the longitudinal component
- M_x,M_y are transverse components



NMR: Classical view

MR signal

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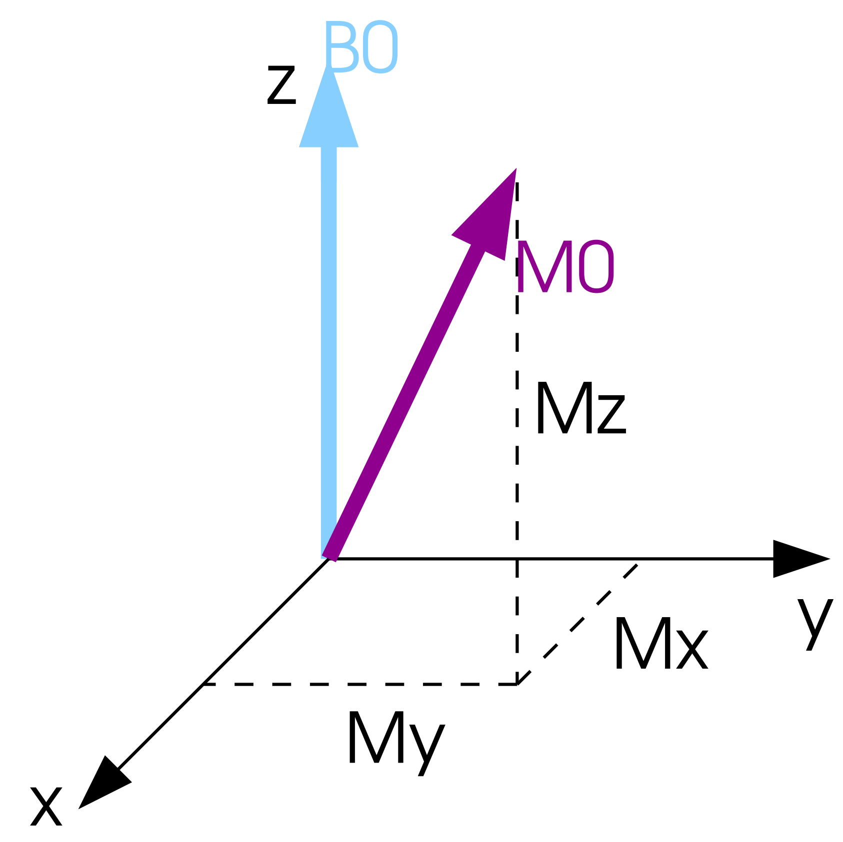
- M_z is the longitudinal component
- M_x, M_y are transverse components

- NMR signal is proportional to M_{xy} where:

$$M_{xy} = M_x + iM_y$$

- M_{xy} is considered to be a complex-valued signal induced in the receiver coil

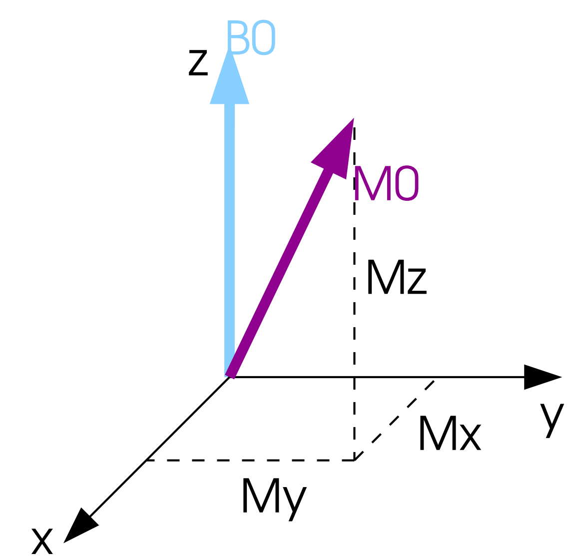


NMR: Classical view

MR relaxation

- M_z is the longitudinal component of **M**
- After excitation M_z relaxes back to M_0 by T_1 relaxation

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

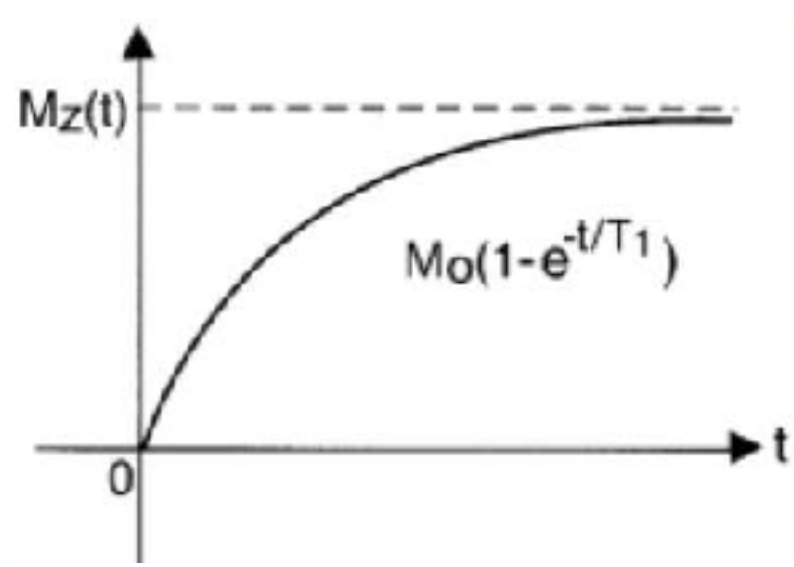


- So that:

$$M_z(t) = M_0 + (M_z(0) - M_0)e^{-t/T1}$$

- or, equivalently

$$M_z(t) = M_z(0)e^{-t/T1} + M_0(1 - e^{-t/T1})$$



NMR: Classical view

MR relaxation

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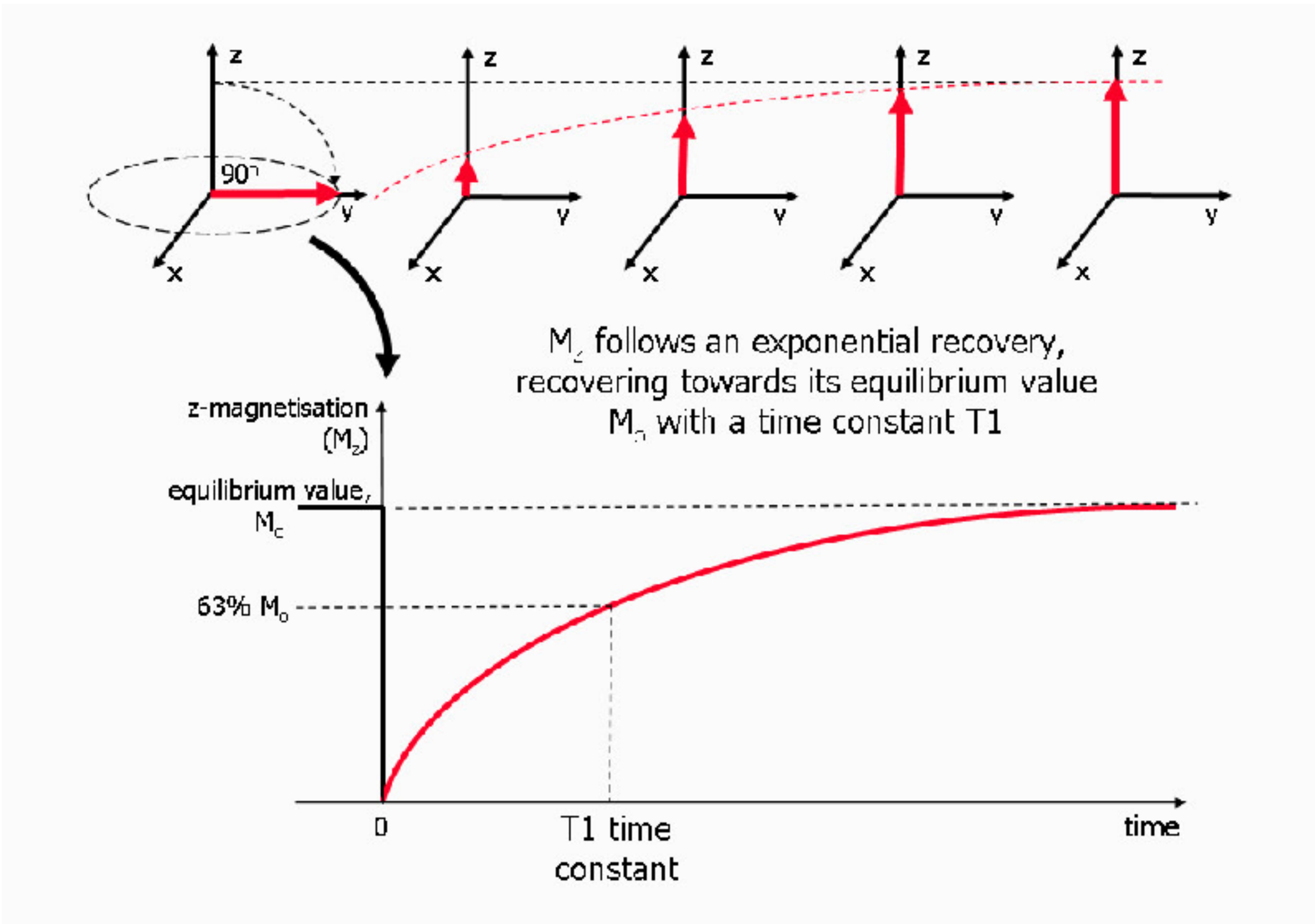


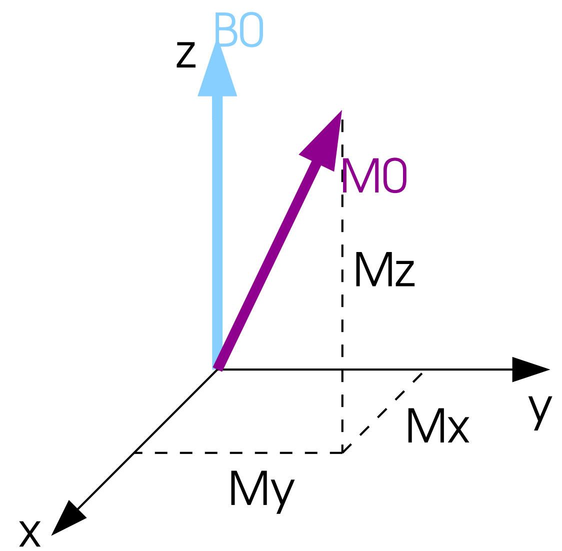
from J. Ridgeway, JCMR, **12**:71, 2010

NMR: Classical view

MR relaxation

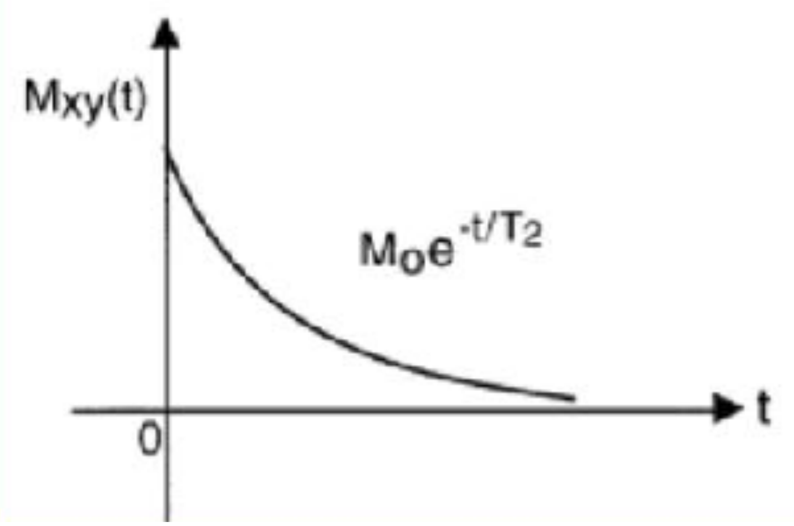
- M_{xy} is the transverse component of **M**
- After excitation M_{xy} relaxes back to zero by T_2 relaxation

$$\frac{dM_{xy}}{dt} = -\frac{M_{xy}}{T_2}$$



- So that:

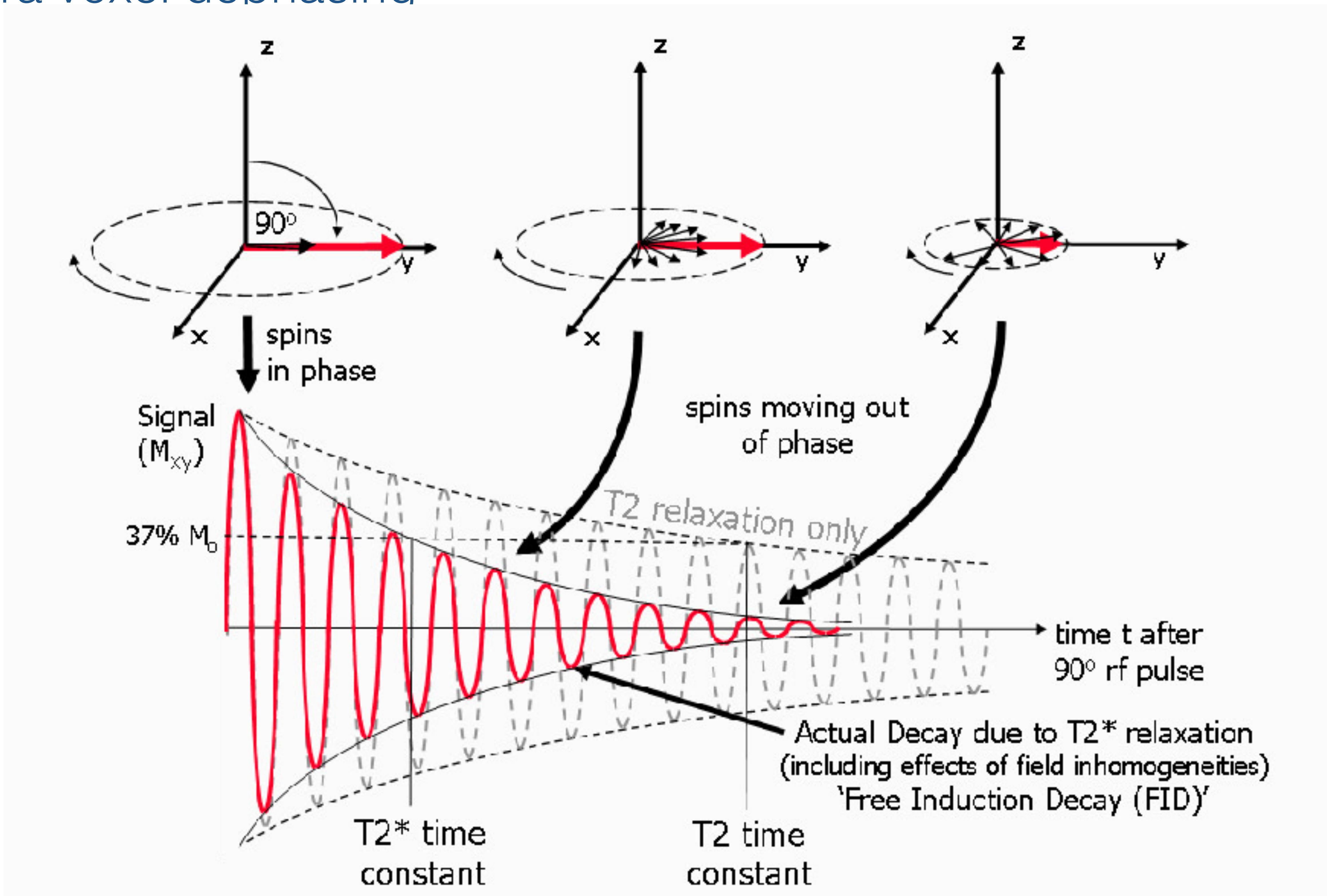
$$M_{xy}(t) = M_{xy}(0)e^{-t/T_2}$$



- Note that $T_2 <= T_1$ so that the MR signal generally dies faster than Mz regrows.

NMR: Classical view

Intra voxel dephasing



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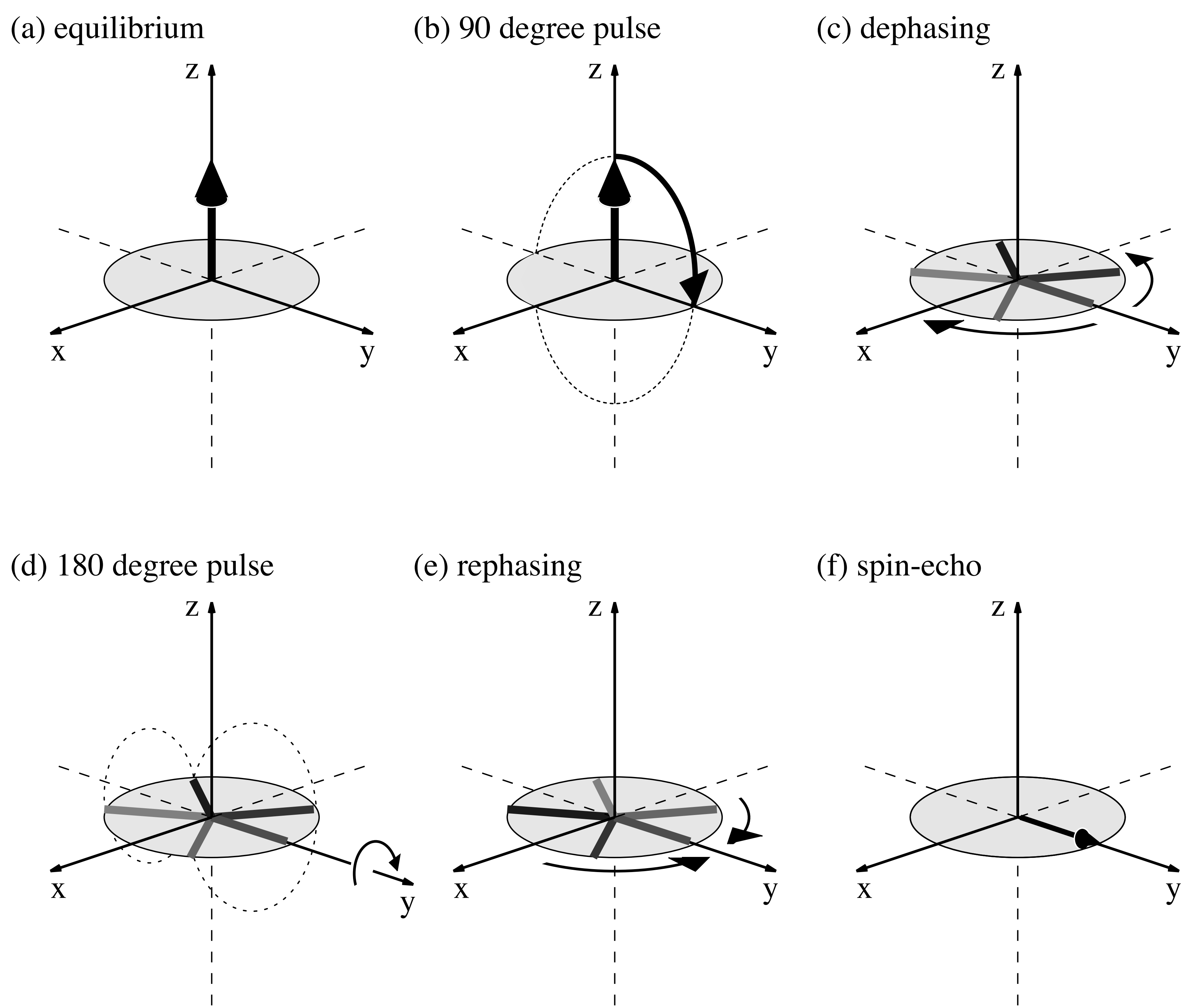
NMR: Classical view

Spin-echo



Outline

- NMR: Review of physics basics
- MR Imaging: tools and techniques
  - Gradients
  - Selective excitation
  - Gradient echo
- K-space trajectories
- Controlling the image contrast
- Other stuff

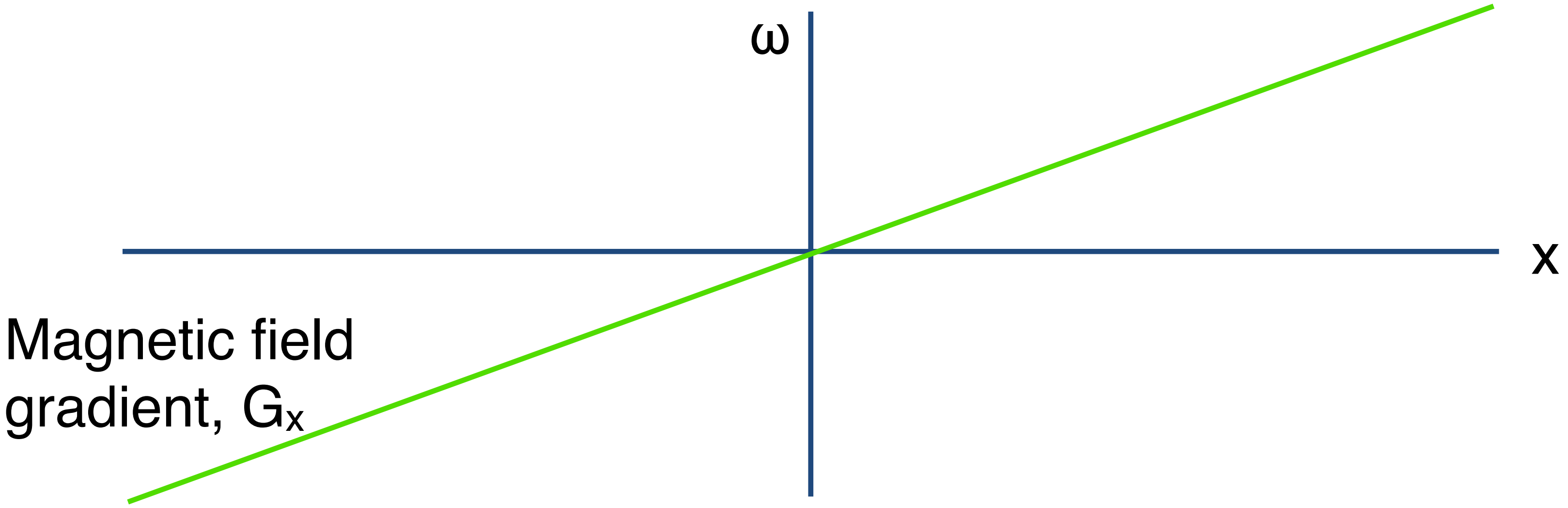
Gradients

- MR image formation is based on the equation: $\omega = \gamma B$
- In the main magnetic field, B_0, we have: $\omega_0 = \gamma B_0$
- Superimpose a spatial magnetic field gradient,

$$\mathbf{G} = (G_x\ ,\ G_y\ ,\ G_z)^T$$

then:

$$\omega = \gamma(G_x x + G_y y + G_z z) = \mathbf{G.r}$$



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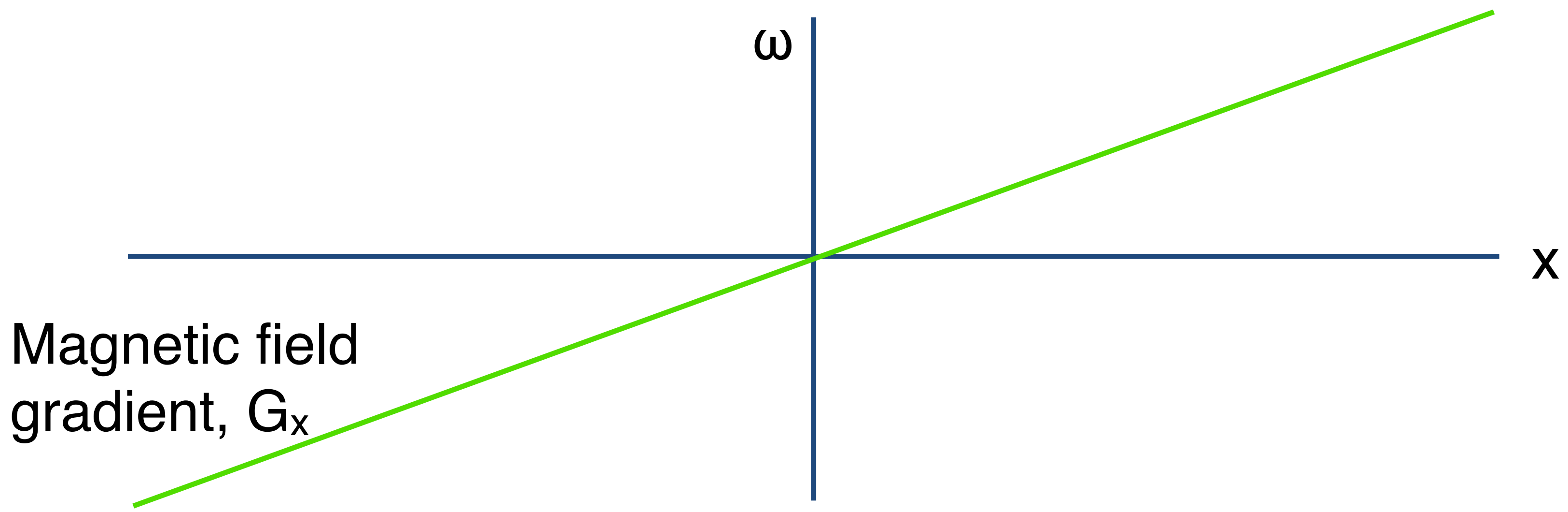
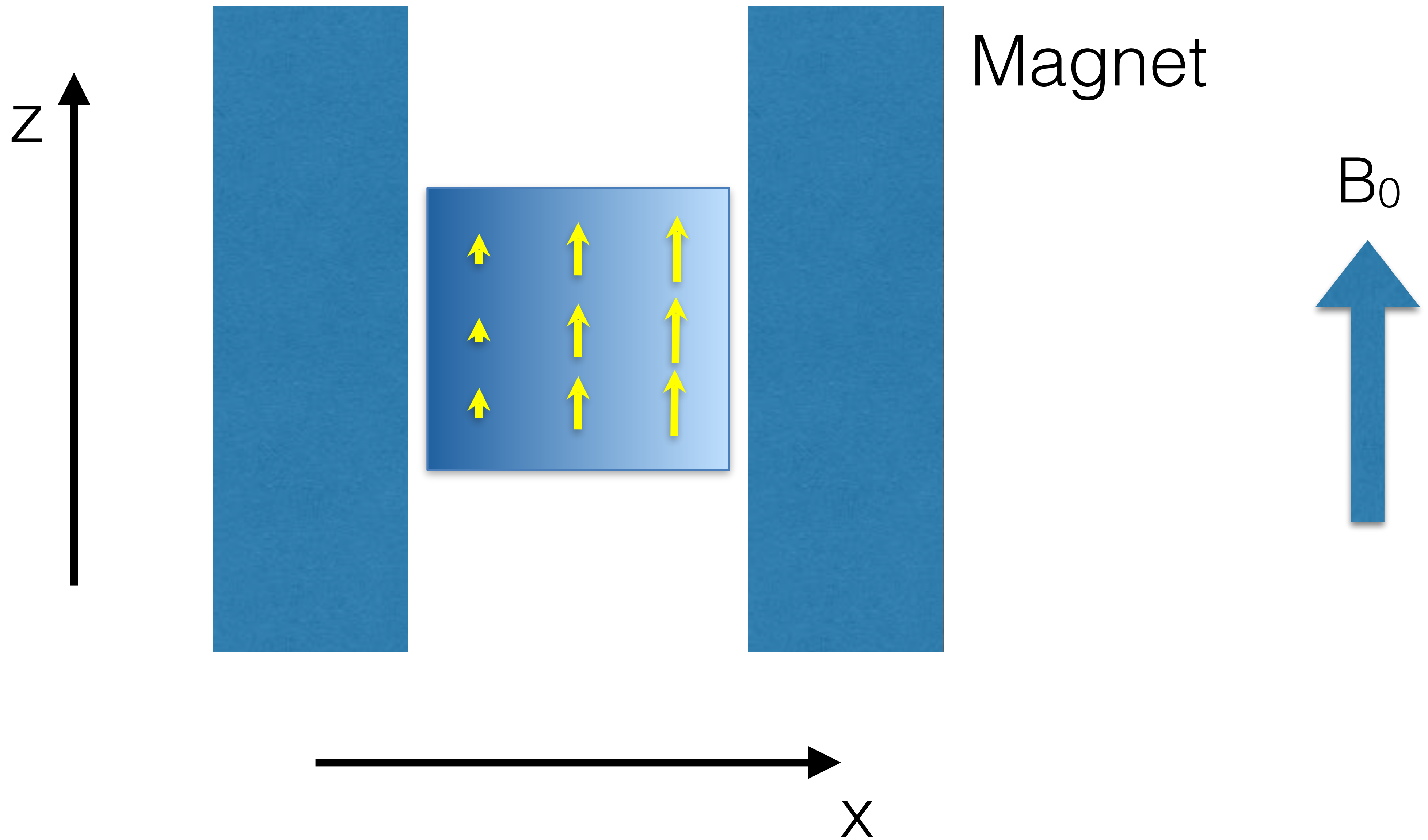


- Typical gradient fields are G = 30mT/m.
  i.e. +/-3mT at 10cm from isocenter.
- 1000 times smaller than B_0

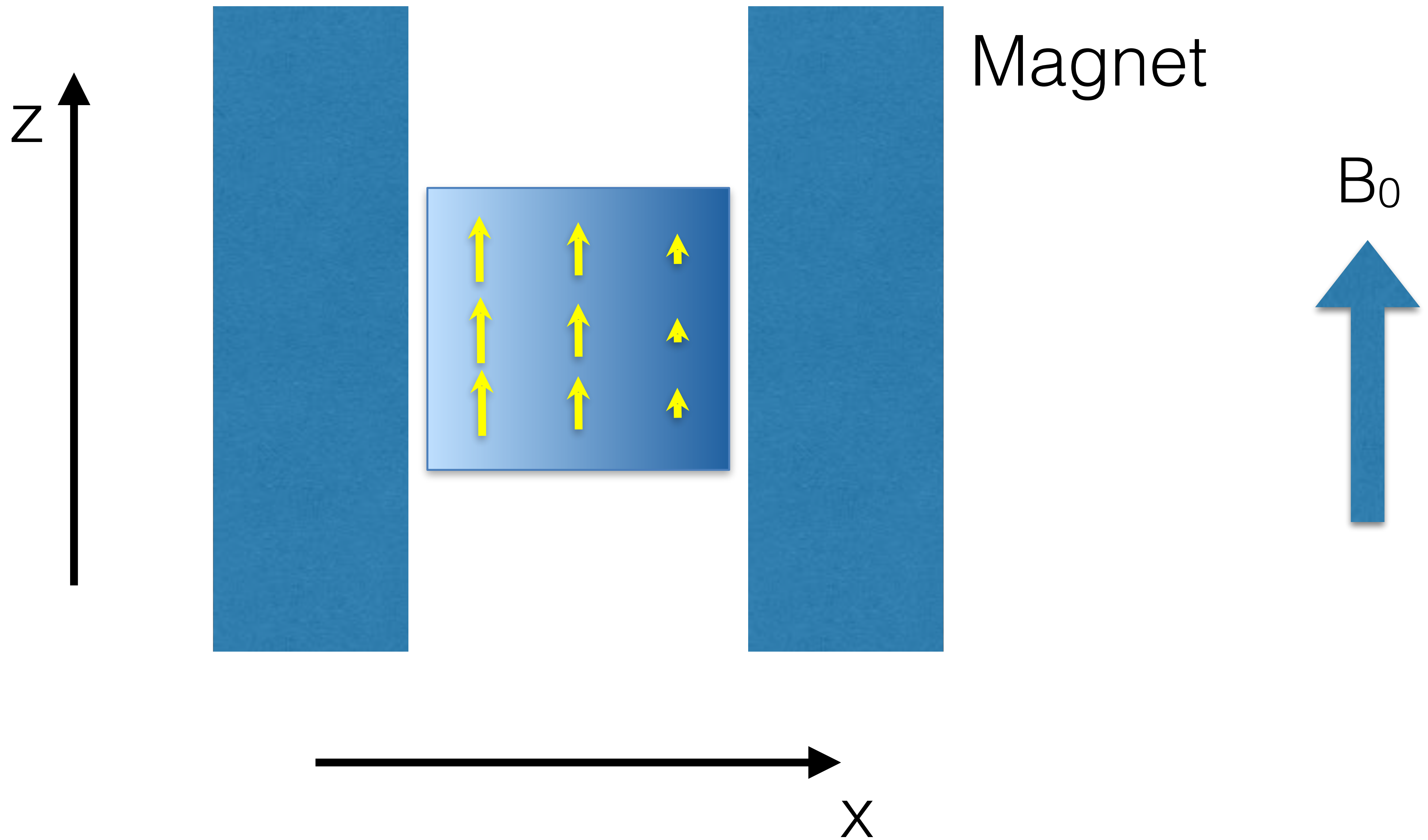
Gradients

- Gradient in +X



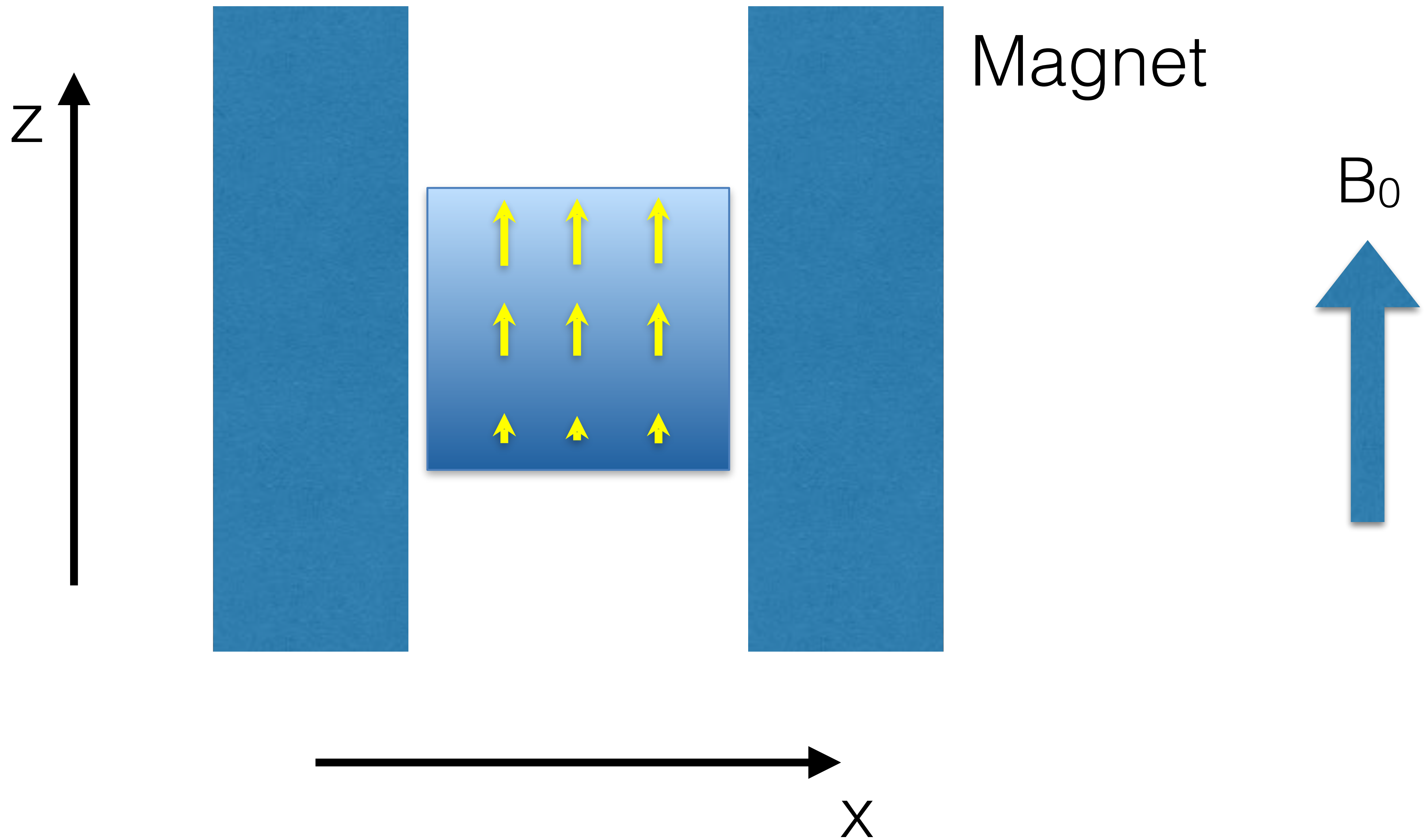
Gradients

- Gradient in -X



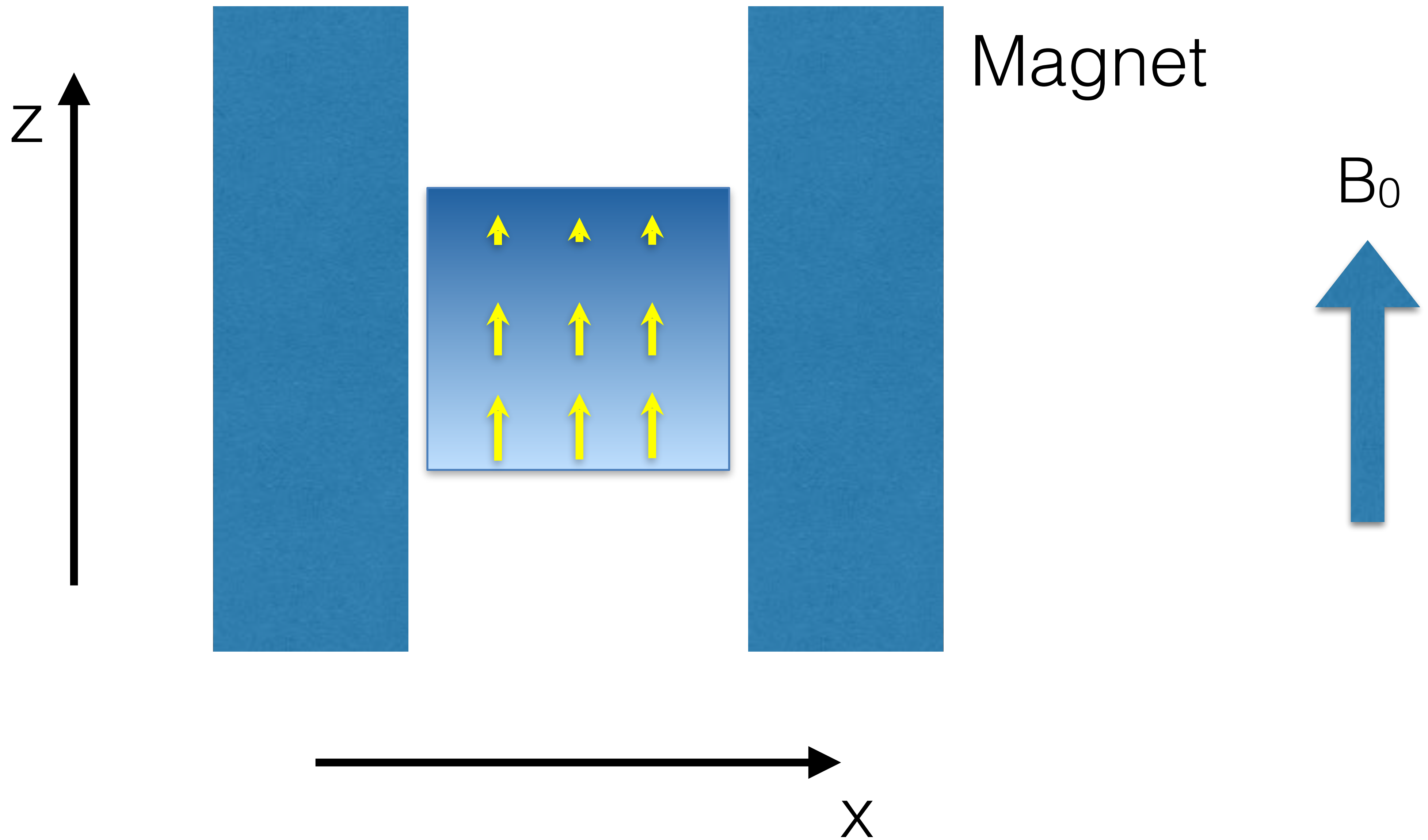
Gradients

- Gradient in +Z



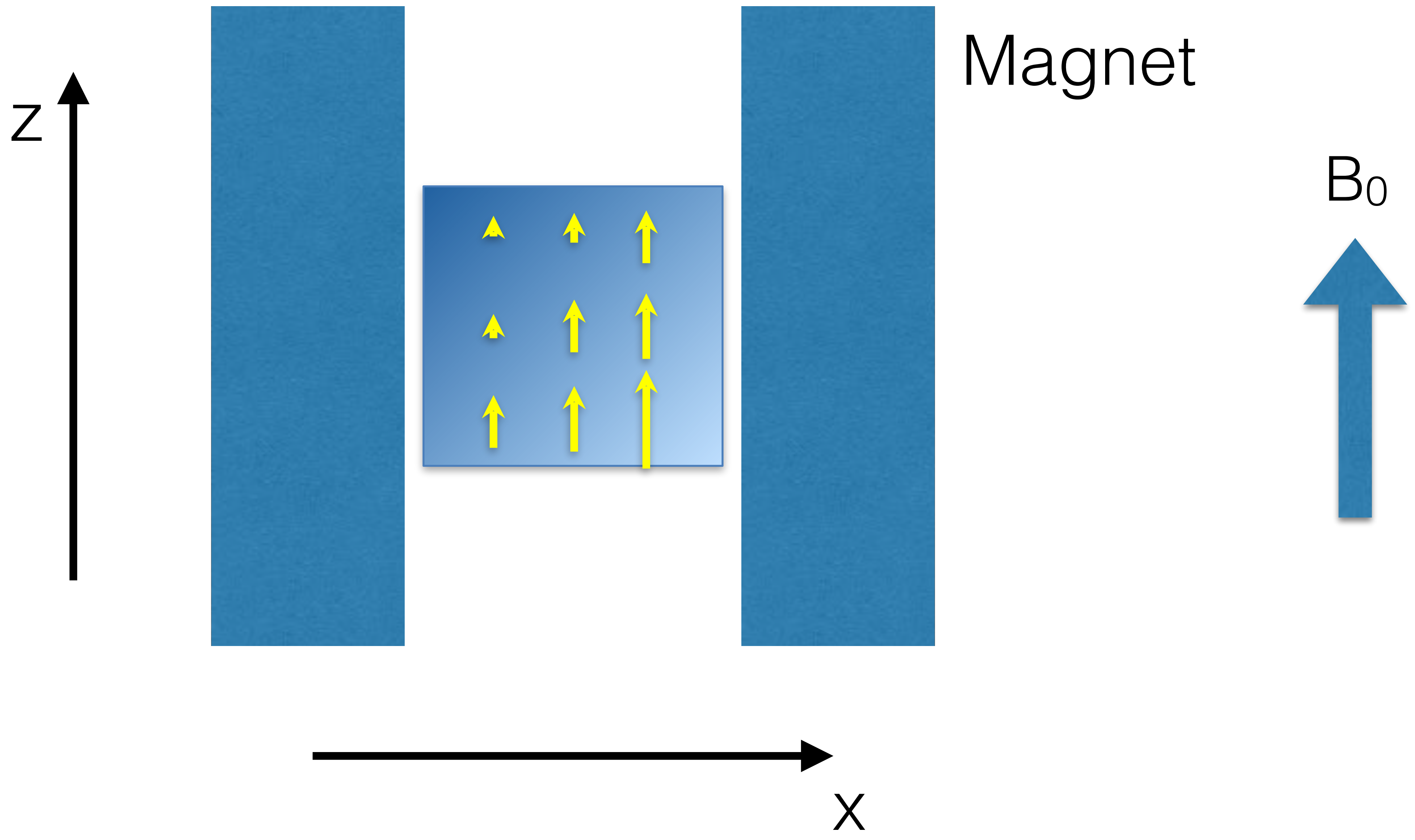
Gradients

- Gradient in -Z



Gradients

- Gradients in both X and -Z

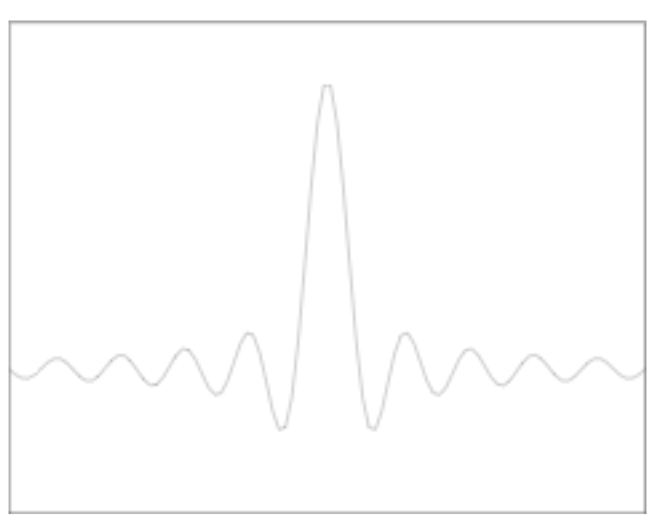


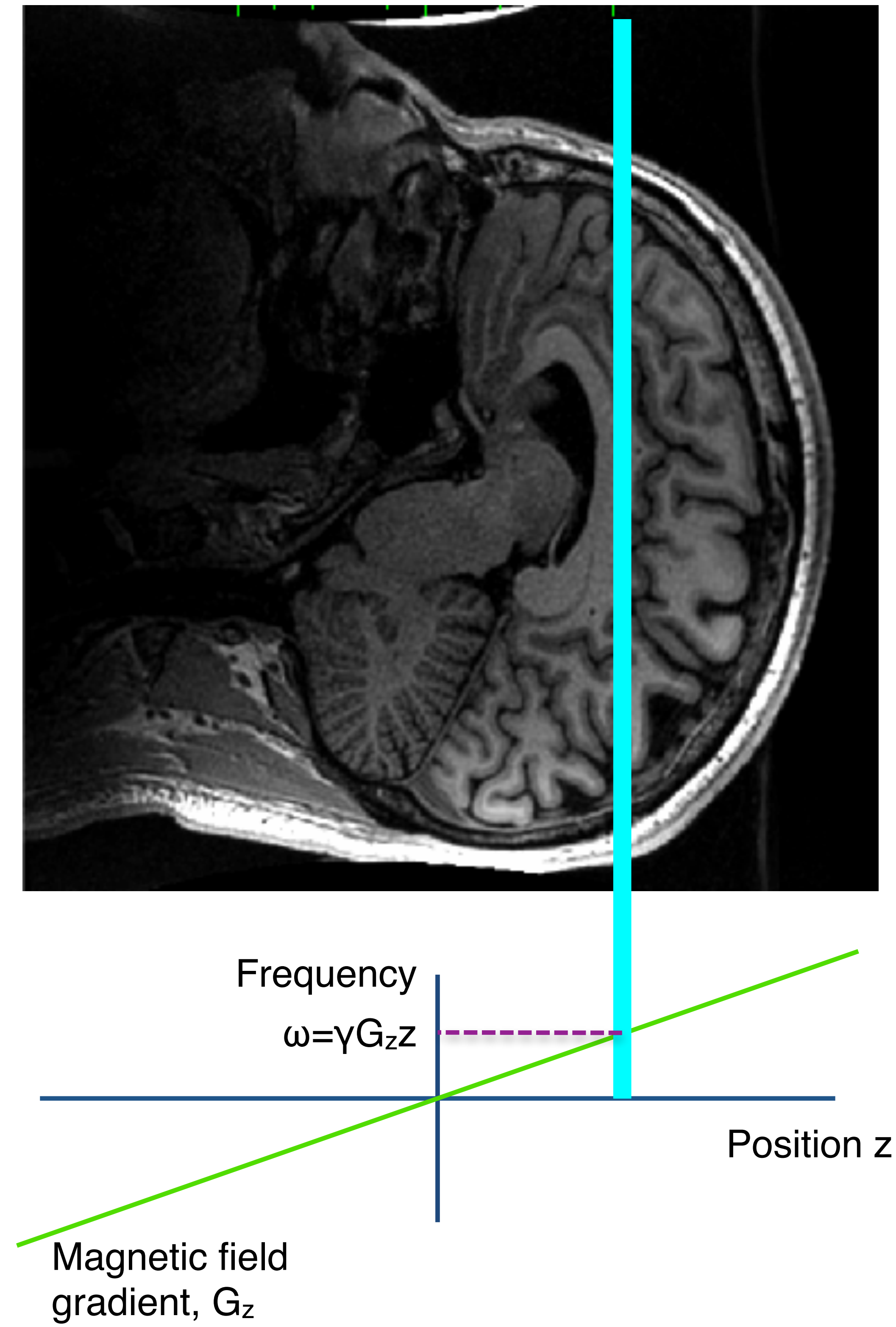
Slice selection

- Consider the slice of tissue at position z
- In the presence of gradient G_z, the local slice frequency is given by:

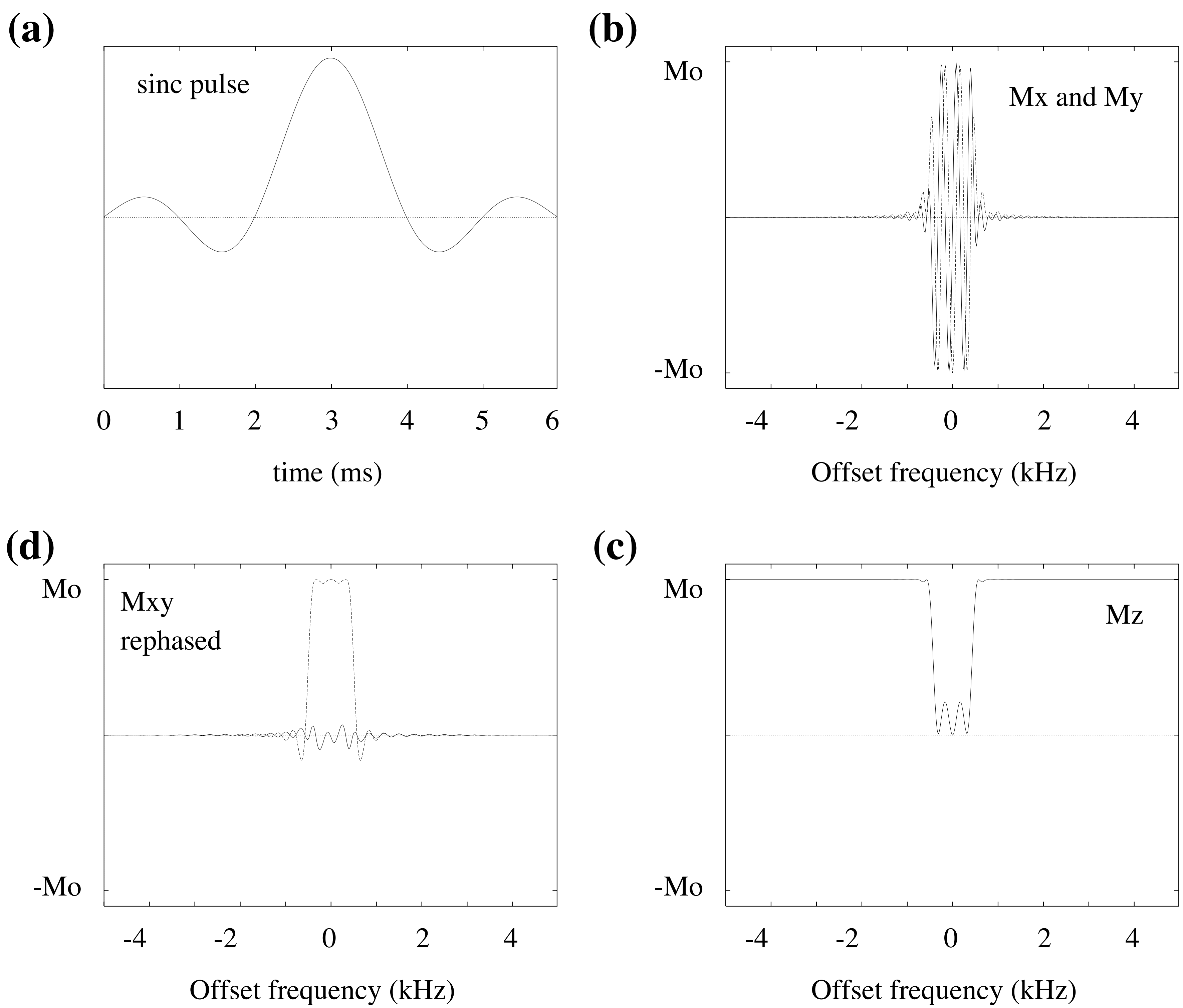
$$\delta\omega = \gamma G_z z$$

- Excite with frequency $\omega_0 + \delta\omega$ to move slice from isocenter to position of interest.
- Excite with an RF pulse that is the composition of a band of frequencies to define a particular slice width.
- Sinc pulse:

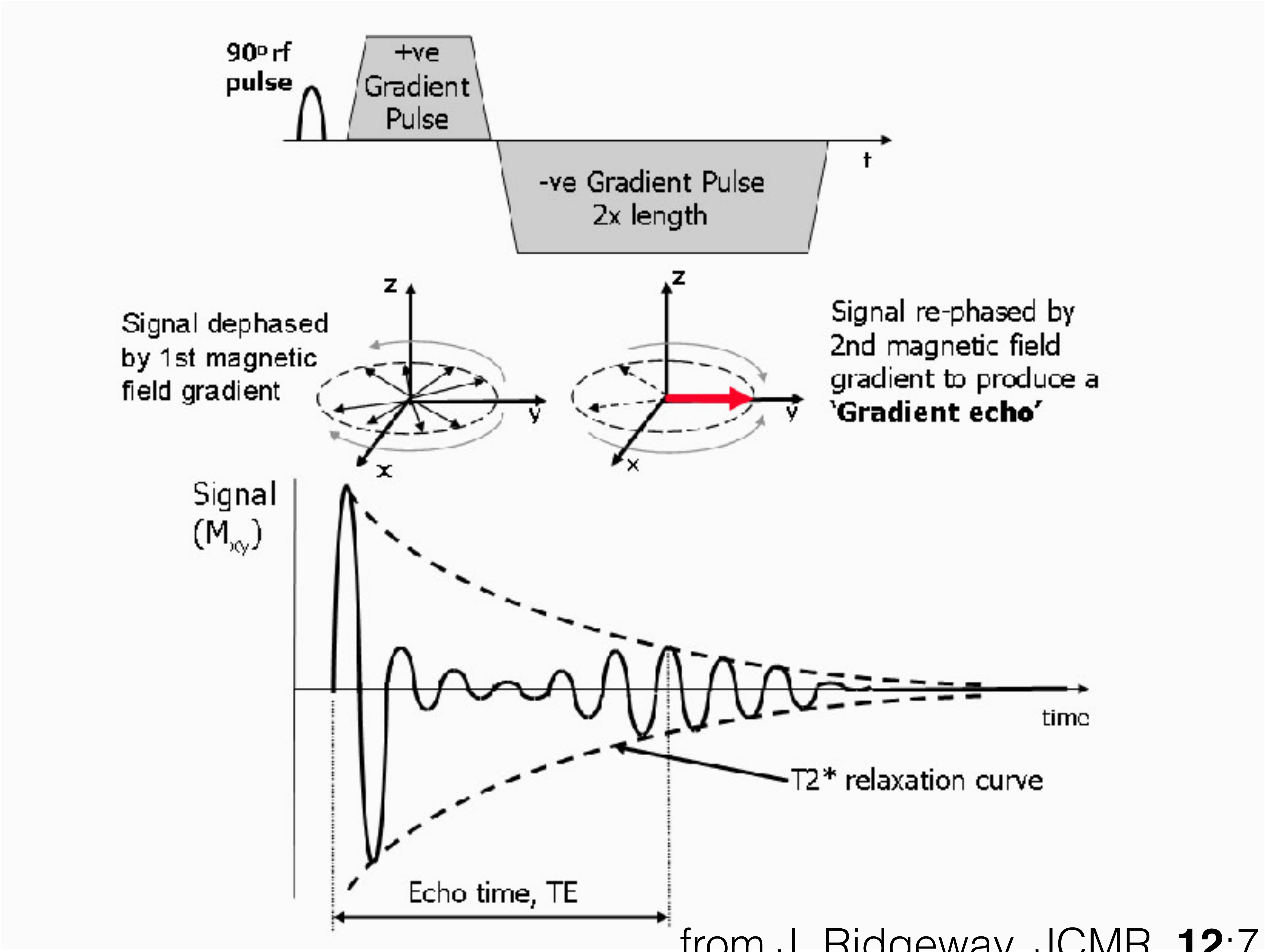




Selective excitation



Gradient echo



from J. Ridgeway, JCMR, **12**:71, 2010

Outline

- NMR: Review of physics basics
- MR Imaging: tools and techniques
- K-space trajectories
  - Theory: MR signal and reconstruction equations
  - Fourier Imaging: readout and phase encoding
  - Echo planar imaging
  - Spiral Imaging
- Image acceleration
- Controlling the image contrast

Effect of imaging gradient

- MR image formation is based on the equation: $\omega = \gamma B$
- In the main magnetic field, B_0, we have: $\omega_0 = \gamma B_0$
- Superimpose a spatial magnetic field gradient,

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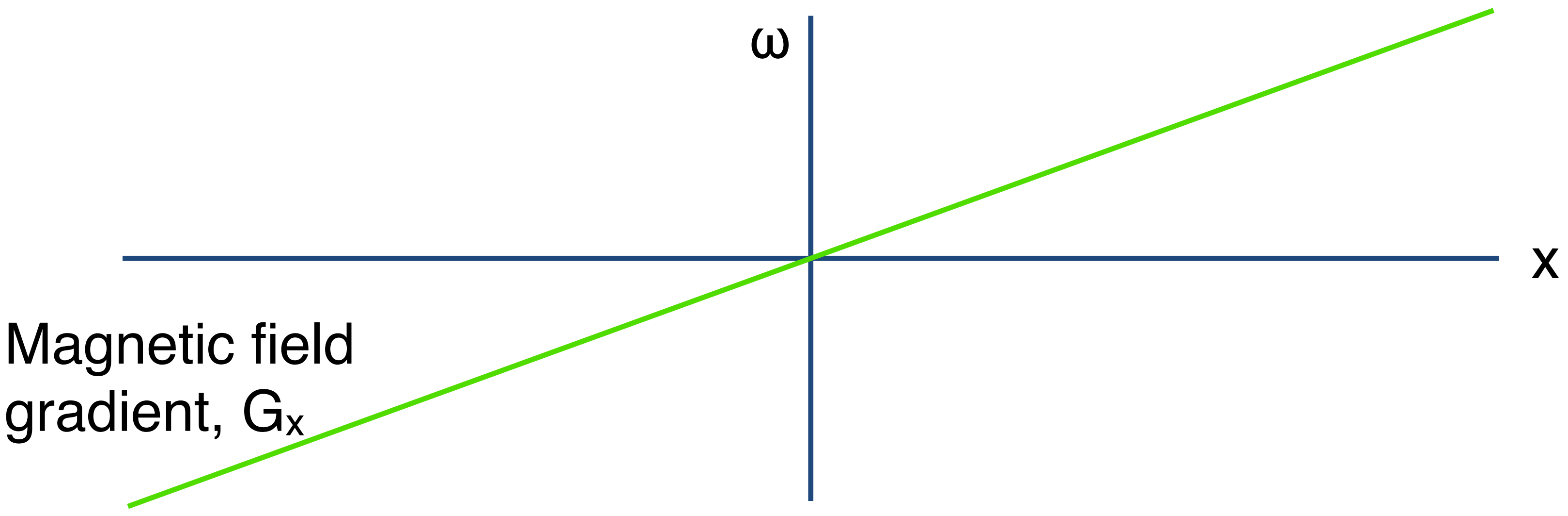
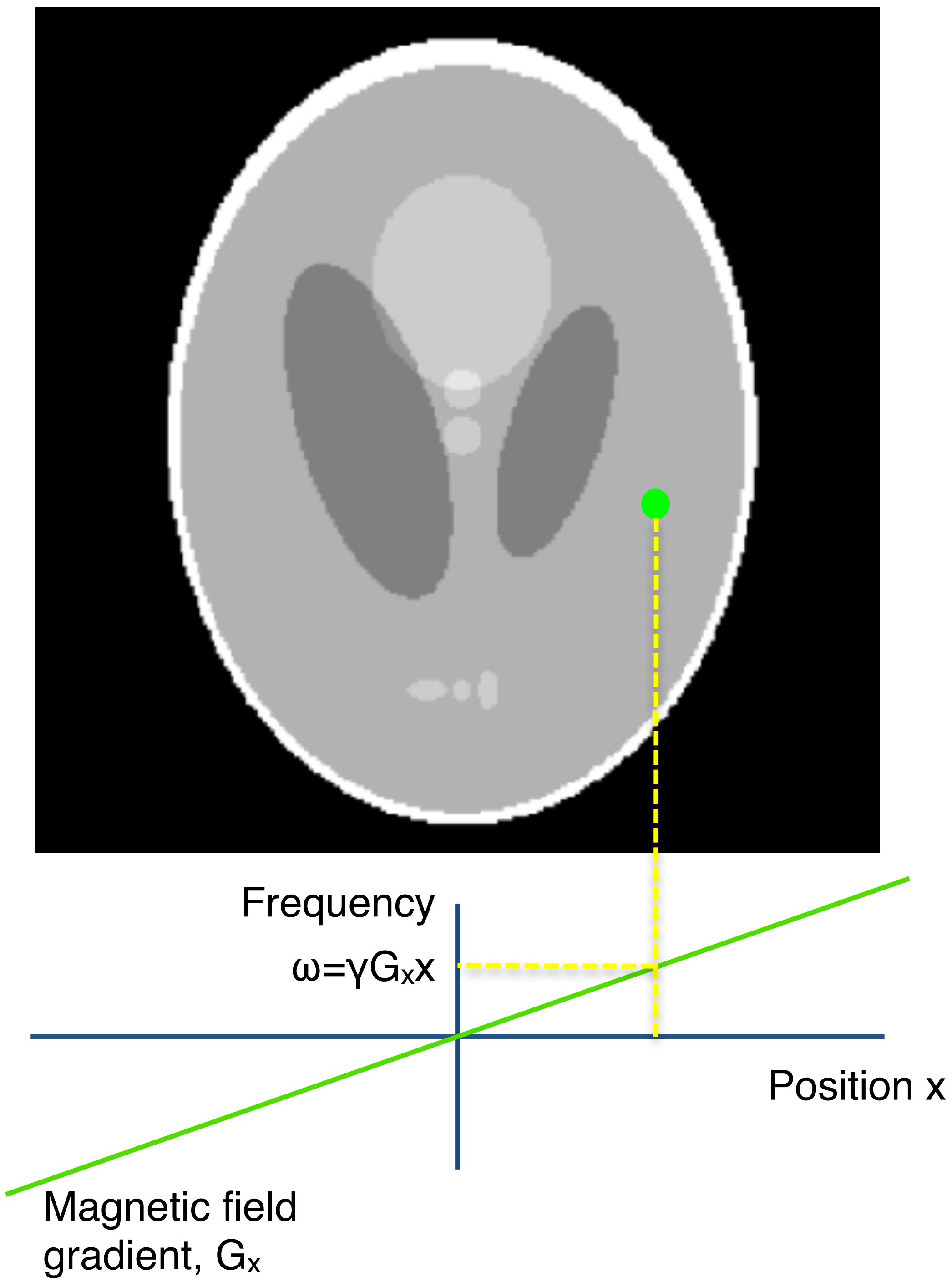


MR Imaging Theory

- Consider the green blob of tissue...
- The frequency is given by:

$$\delta\omega = \gamma \mathbf{G}(t) \cdot \mathbf{r}(t)$$



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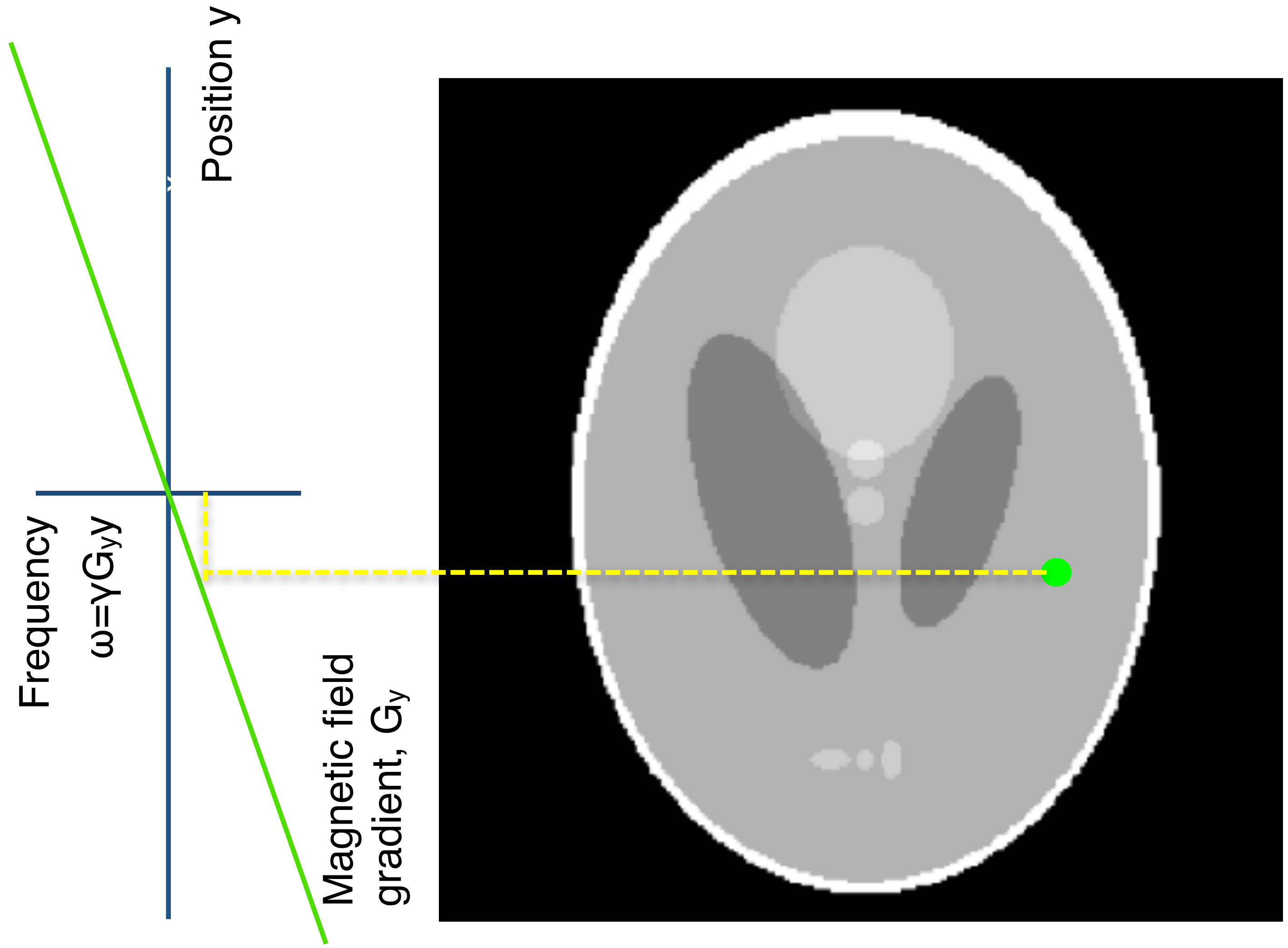

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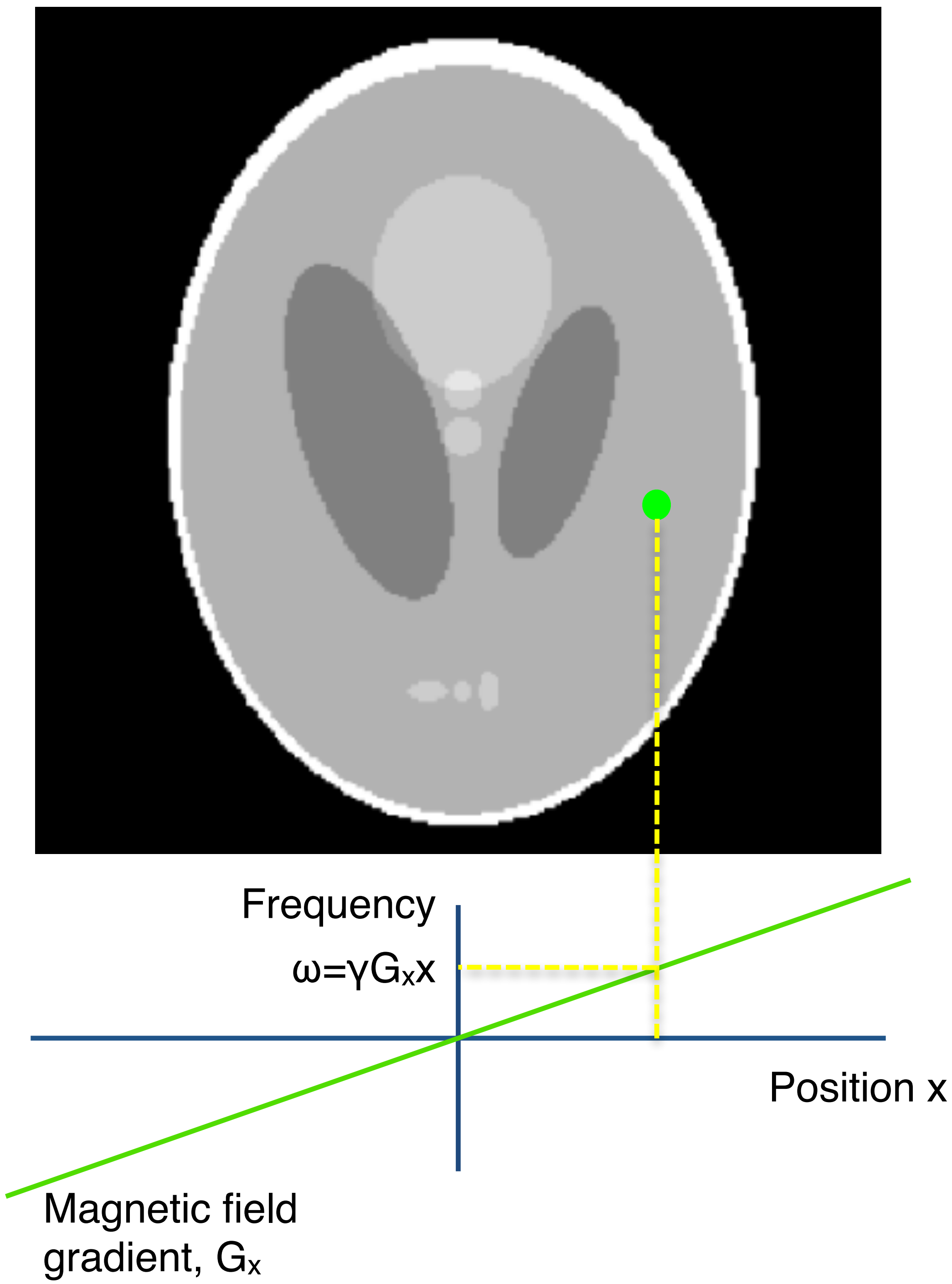
- Over time, phase accumulates as:

$$\delta\theta = \gamma \int_0^t \mathbf{G}(t')dt' \cdot \mathbf{r}(t)$$



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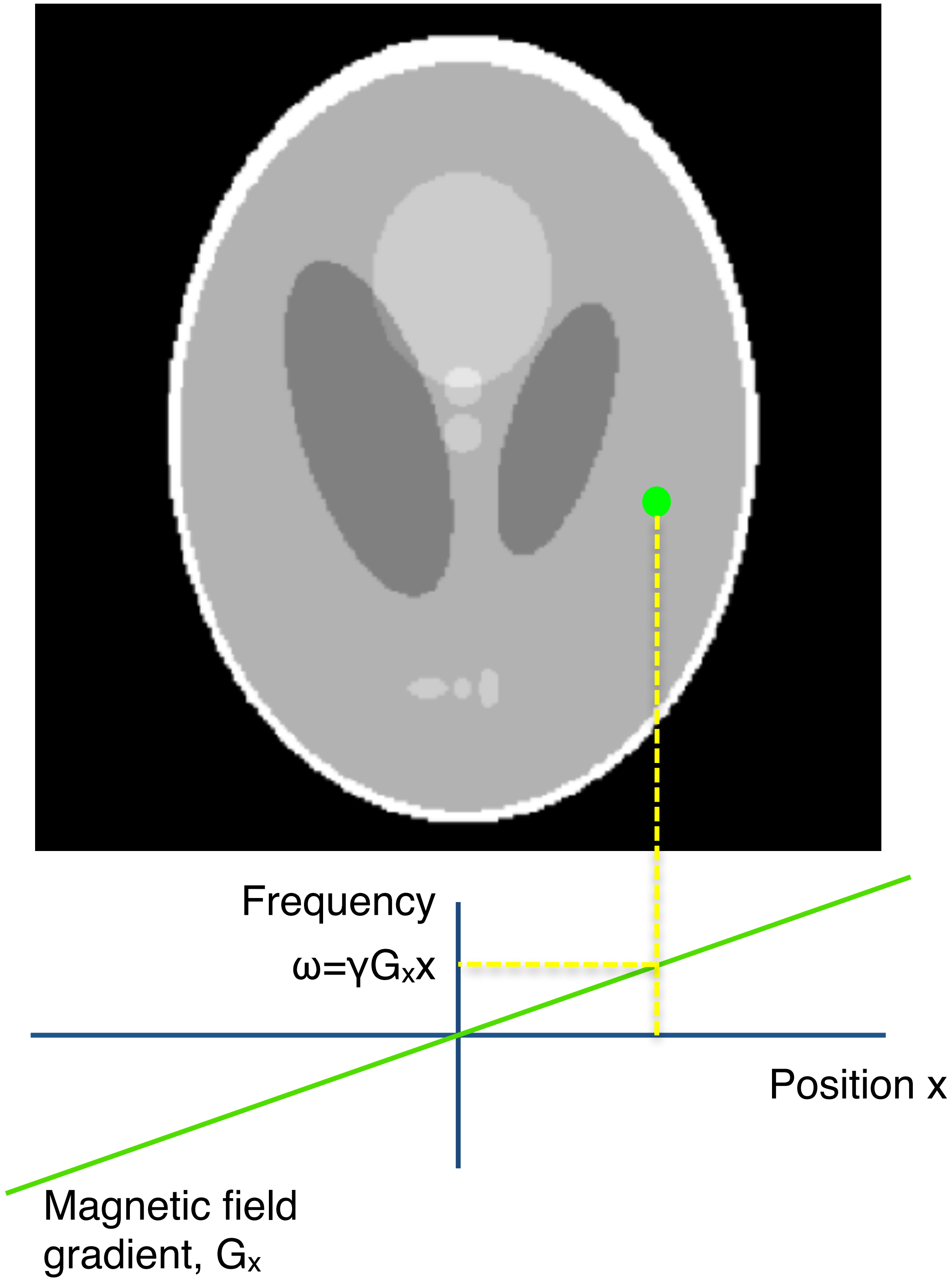
- Signal from the whole slice is given by:

$$S(G,t) = A \int_V \rho(\mathbf{r}) \exp\left[i\gamma \int_0^t \mathbf{G}(t')dt' \cdot \mathbf{r} \right] d^3\mathbf{r}$$



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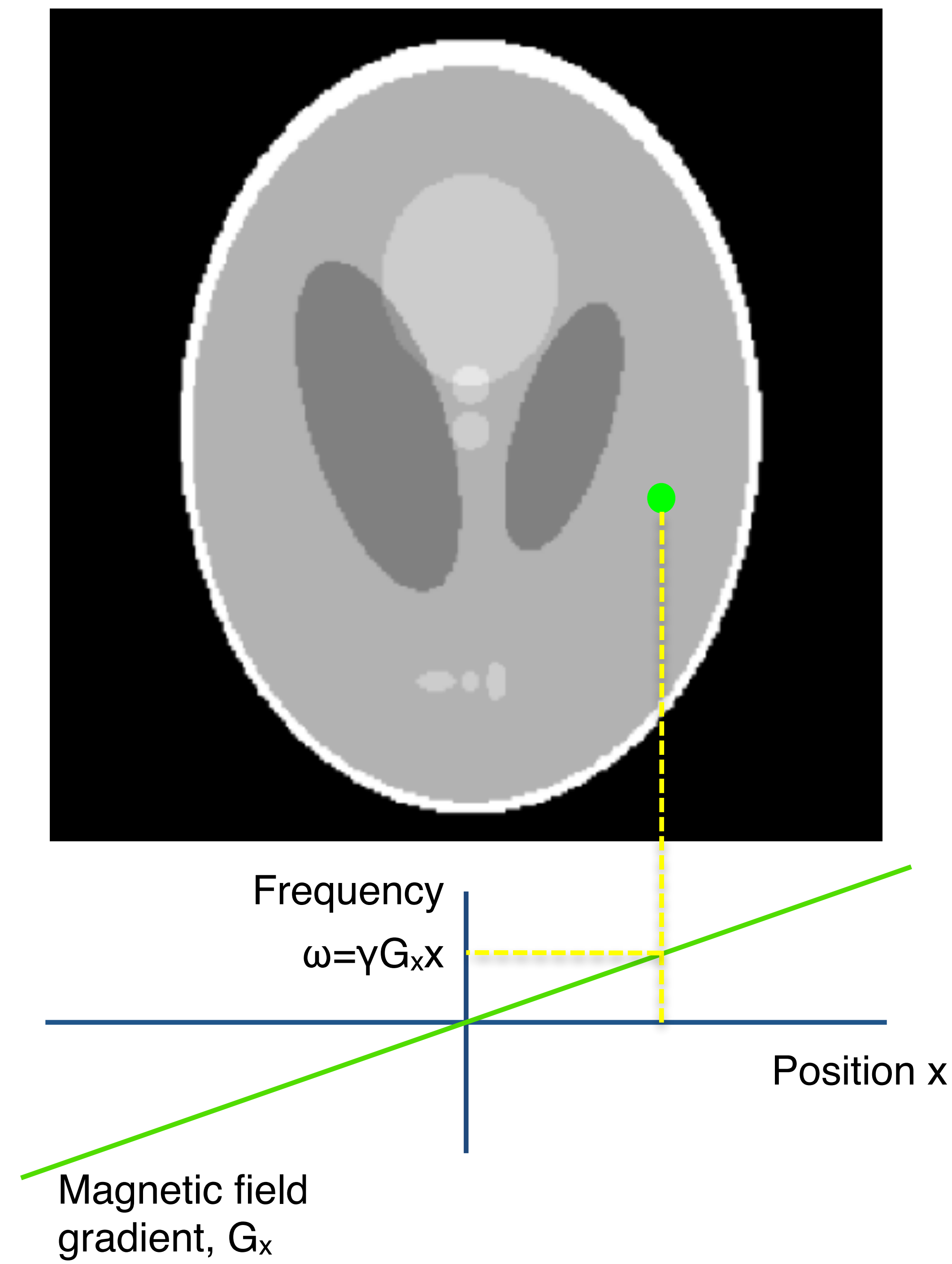
- Write:

$$\mathbf{k}(t) = \frac{\gamma}{2\pi} \int_0^t \mathbf{G}(t')dt'$$



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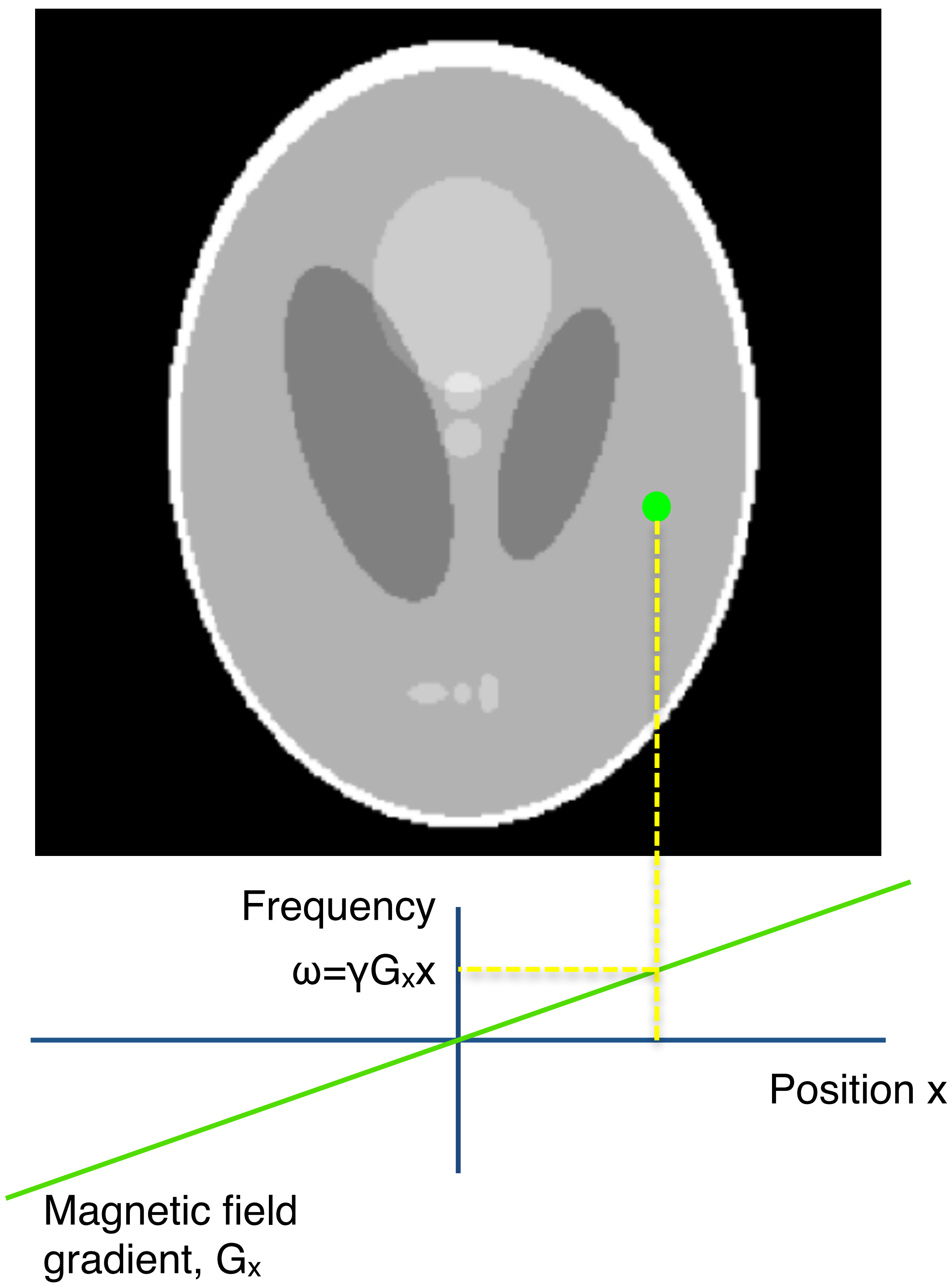
$$S(\mathbf{k}) = \int_V \rho(\mathbf{r}) \exp(i2\pi \mathbf{k} \cdot \mathbf{r})\, d^3\mathbf{r}$$



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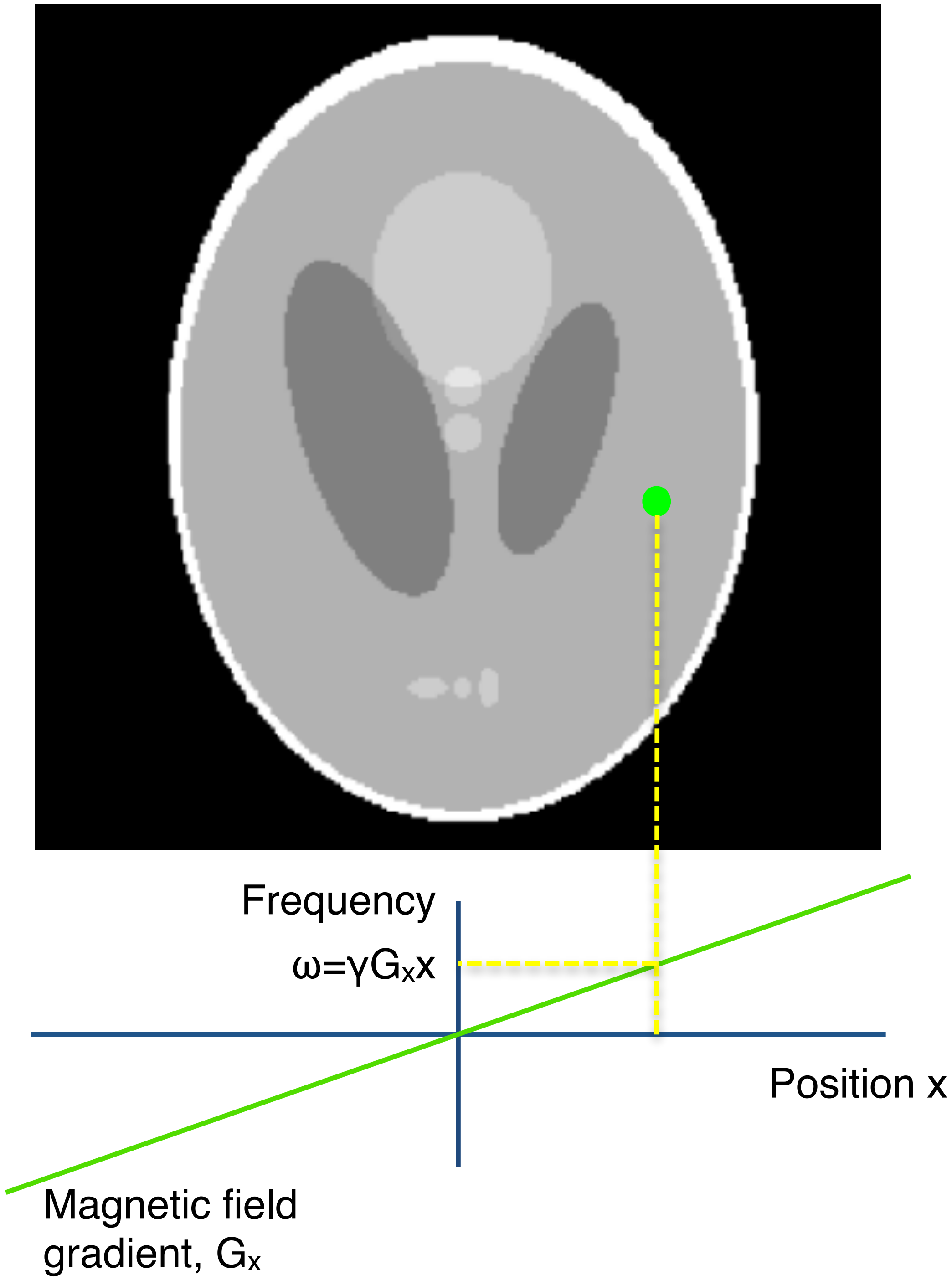
$$\rho(\mathbf{r}) = \int_{\mathbb{R}^3} S(\mathbf{k}) \exp\left(-i2\pi\mathbf{k}\cdot\mathbf{r}\right) d^3\mathbf{k}$$



Fourier transform to k-space

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r}) \exp(i2\pi\mathbf{k}\cdot\mathbf{r})\, d^3\mathbf{r}$$

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

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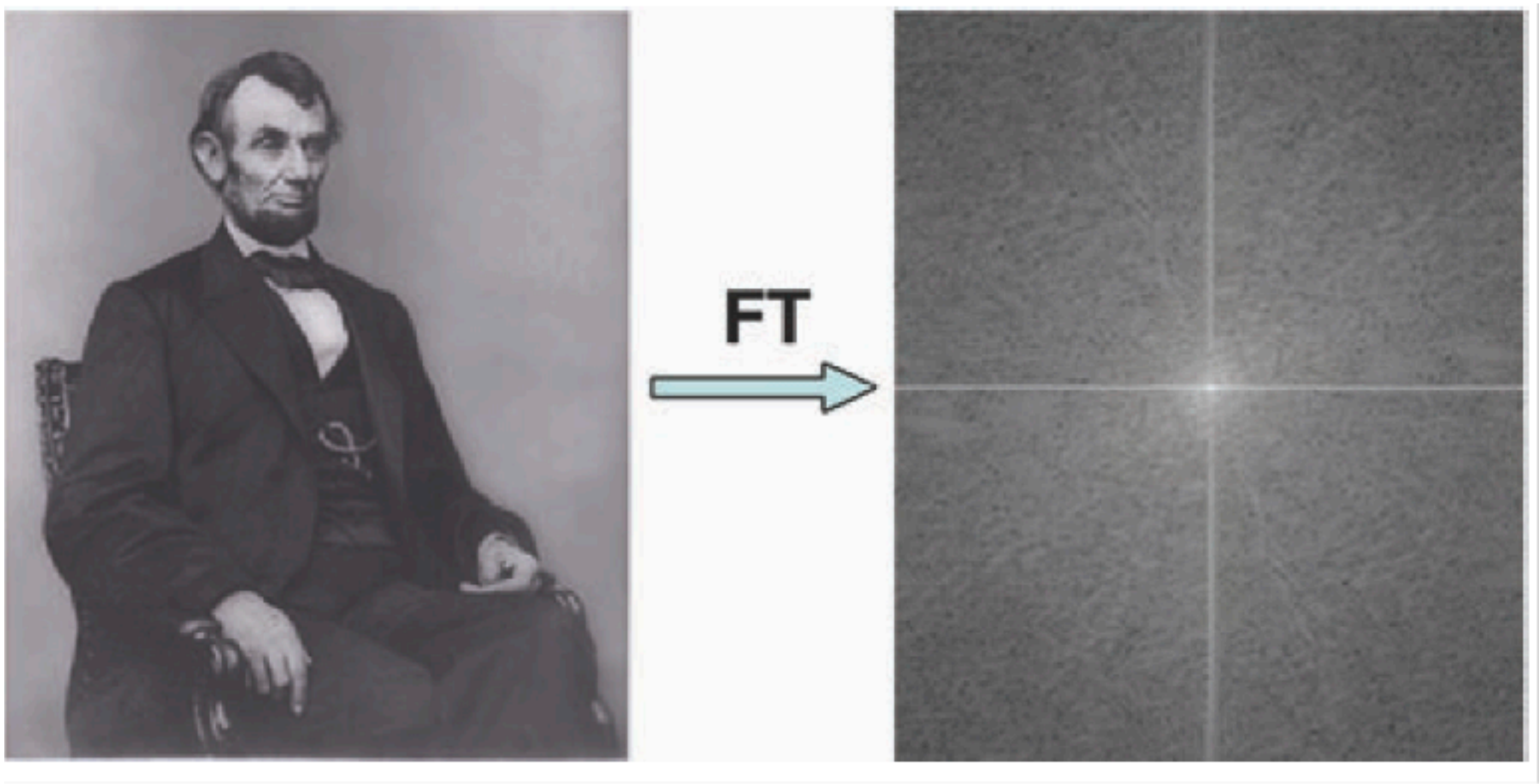
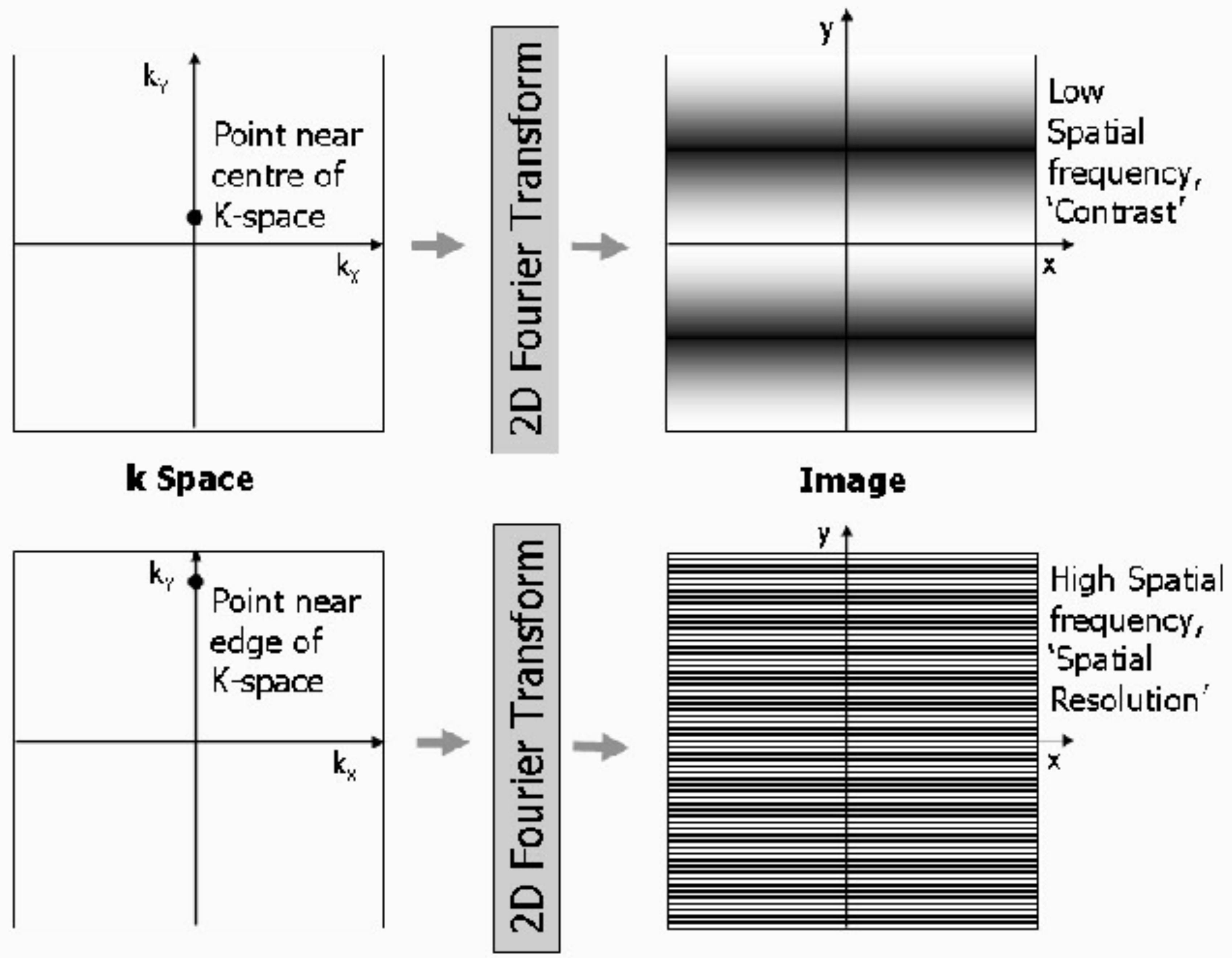


K-space

Spatial frequencies

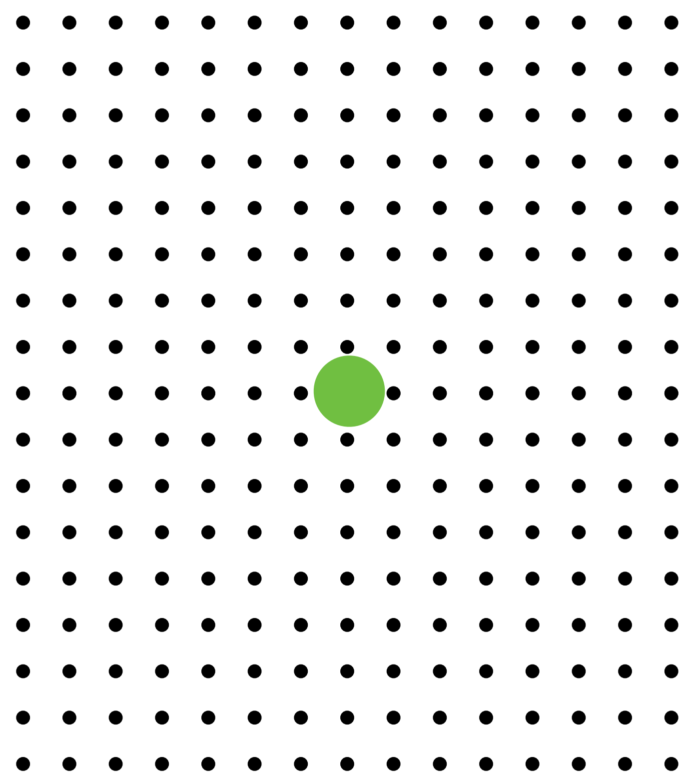
$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r})\exp(i2\pi\mathbf{k}\cdot\mathbf{r})\,d^3\mathbf{r}$$

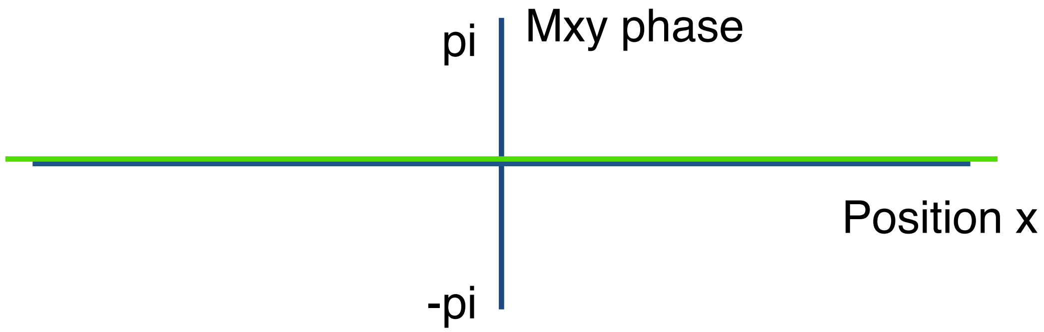


from J. Ridgeway, JCMR, **12**:71, 2010

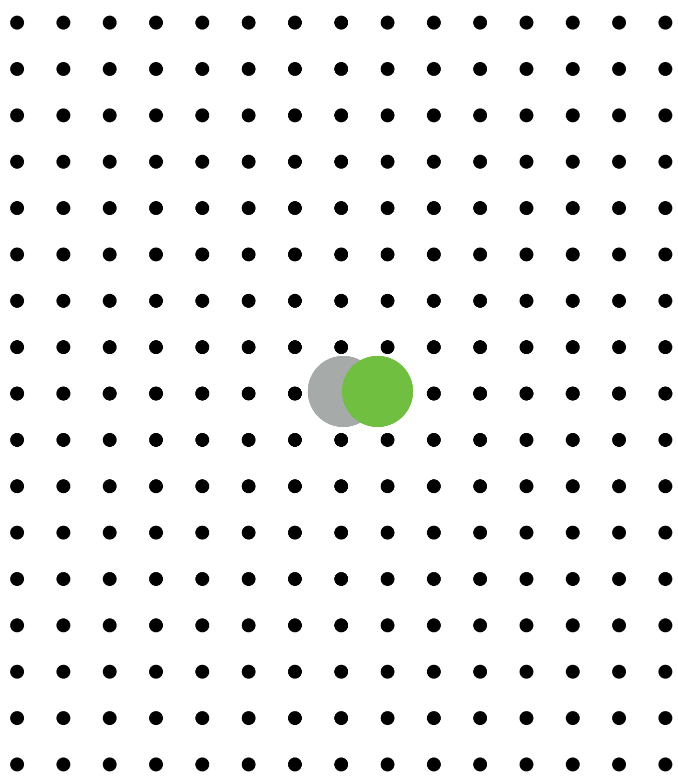
What does it mean?



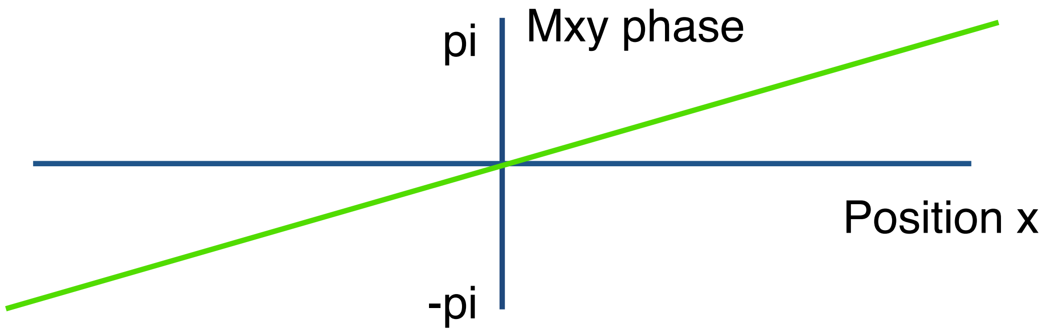
- At center of k-space there is no Mxy phase across the image



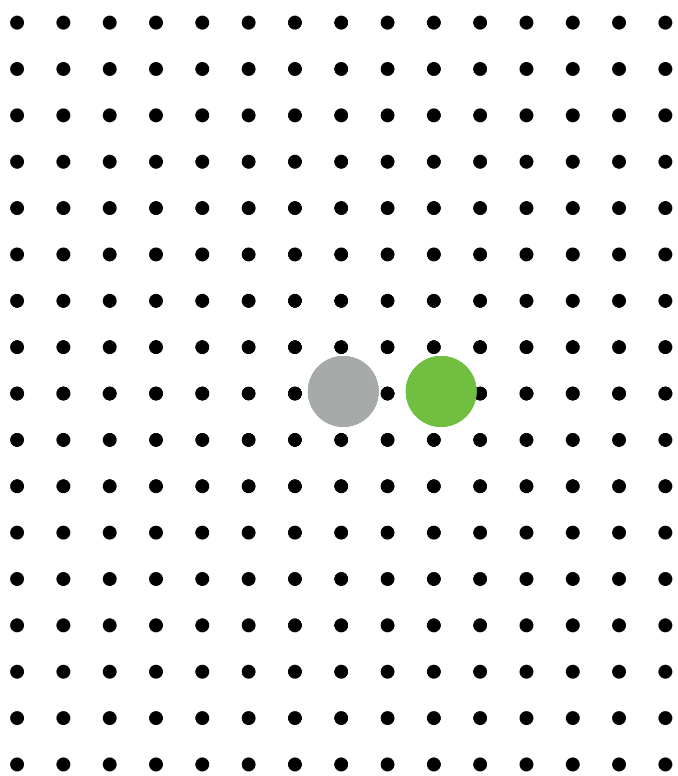
What does it mean?



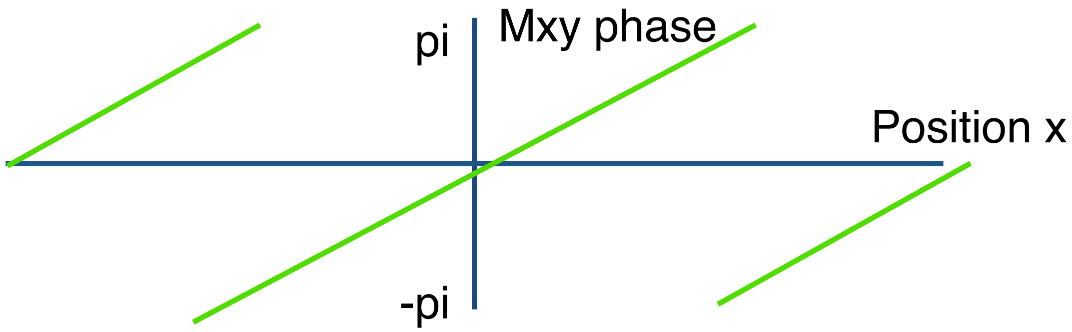
- Moving one k-space sample puts 2pi (-pi to pi) across the whole image FOV



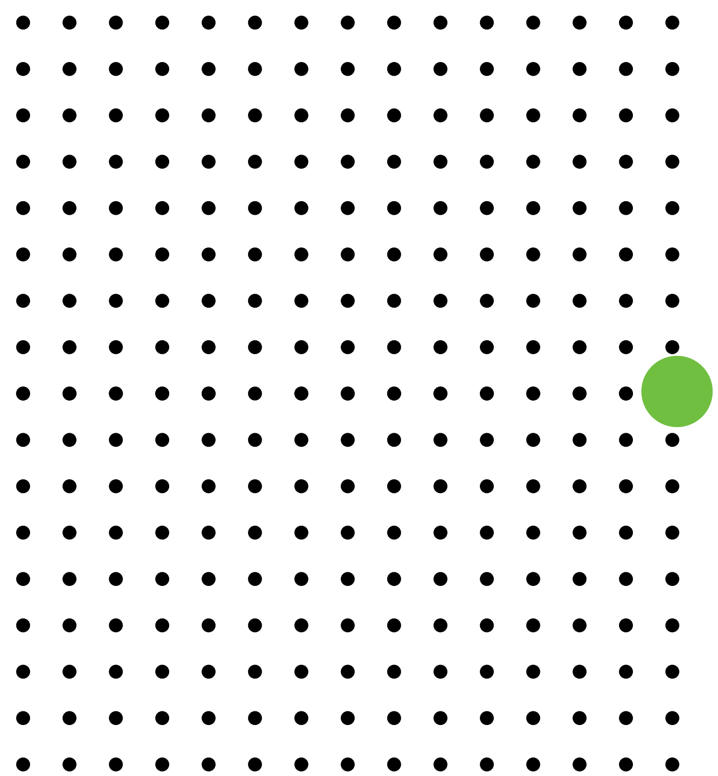
What does it mean?



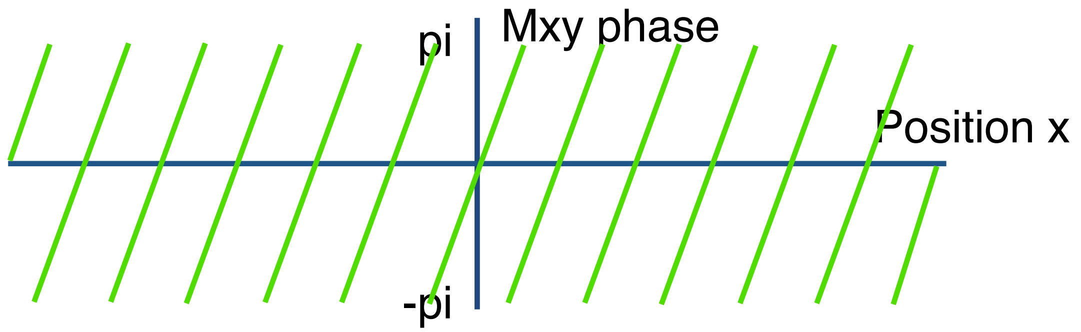
- Moving two k-space sample puts 4pi across the whole image FOV



What does it mean?



- At the edge of the k-space sampling region we have pi phase across each pixel/voxel.

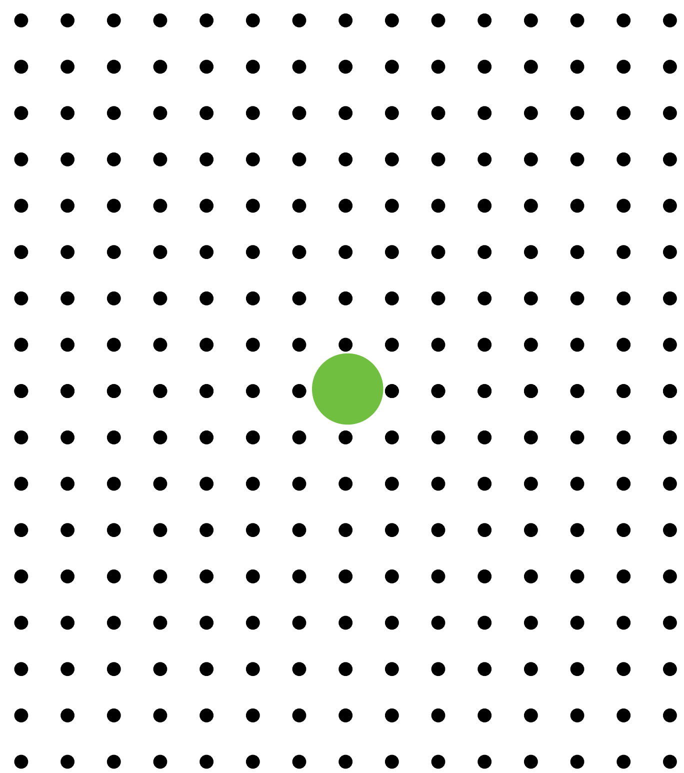


Encoding strategies

k-space trajectories

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r})\exp(i2\pi\mathbf{k}\cdot\mathbf{r})\,d^3\mathbf{r}$$



- Sample all points in k-space to acquire sufficient data for image reconstruction.
- Initial position: origin
- **k**(t) is the sampling position
- **G**(t) is the velocity through k-space
- Sample spacing: δ**k** = 1/FOV
- Sampling extent: Δ**k** = 1/pixelsize

Encoding strategies

k-space trajectories

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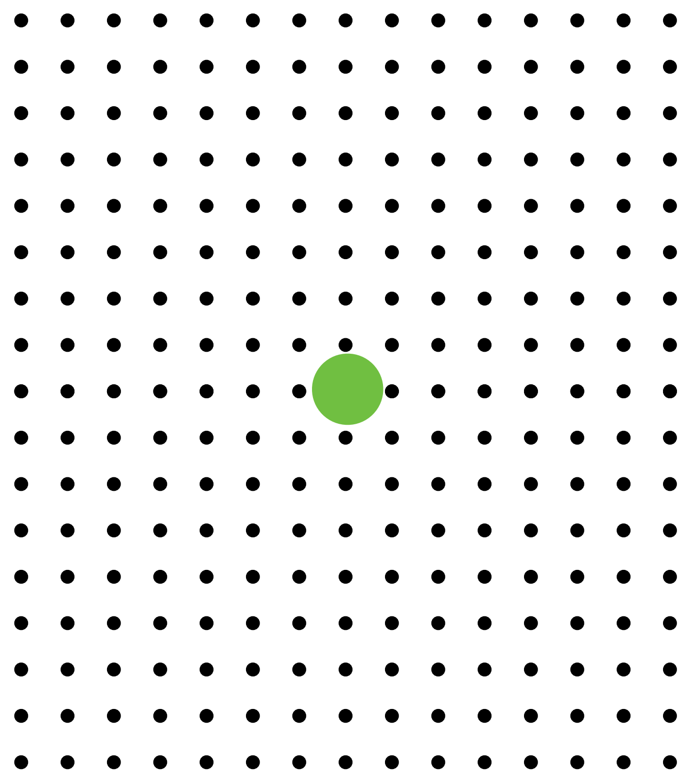


- Sample all points in k-space to acquire sufficient data for image reconstruction.
- Initial position: origin
- **k**(t) is the sampling position
- **G**(t) is the velocity through k-space
- Sample spacing: δ**k** = 1/FOV
- Sampling extent: Δ**k** = 1/pixelsize

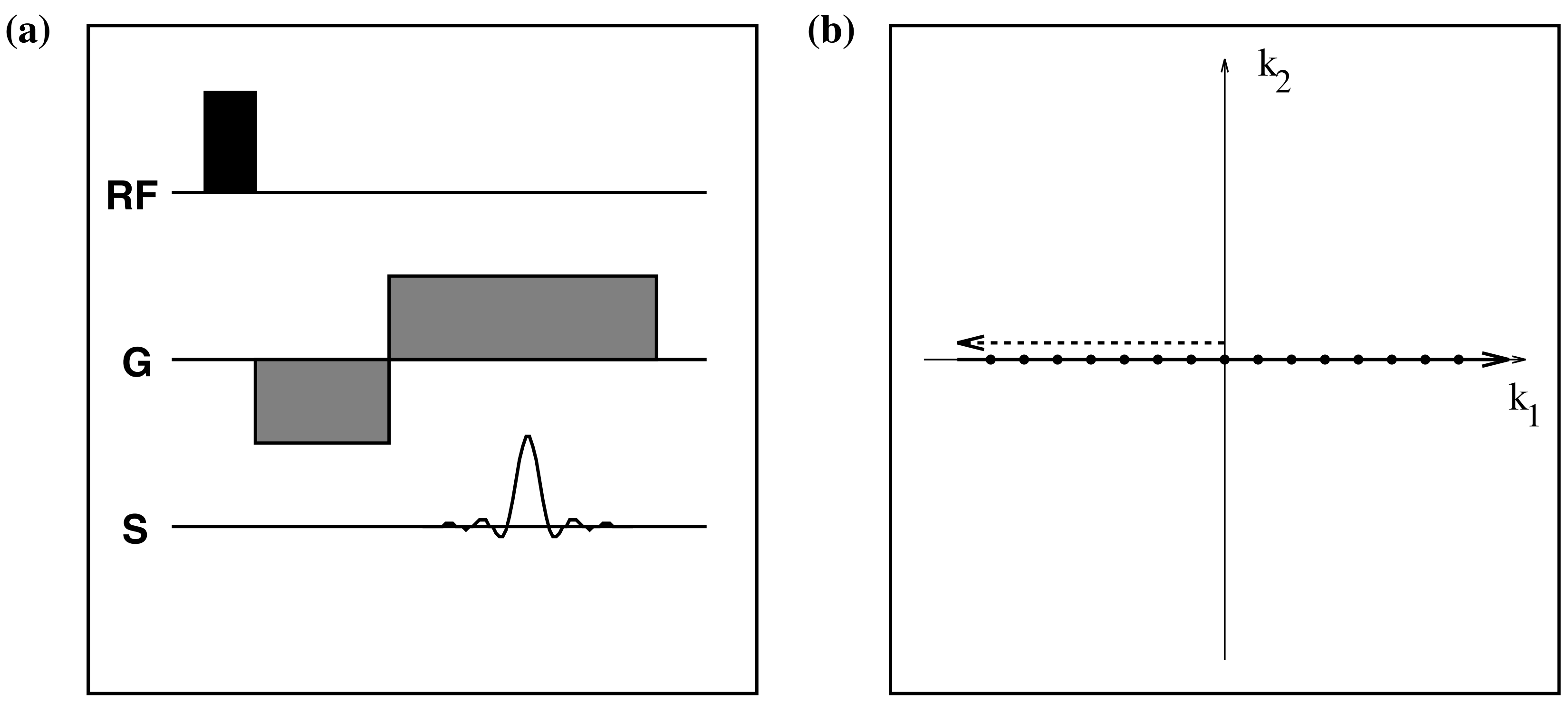
Encoding strategies

Gradient echo

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

- Forms echo signal with spatial encoding in the gradient direction

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r}) \exp\left(i2\pi\mathbf{k}\cdot\mathbf{r}\right) d^3\mathbf{r}$$



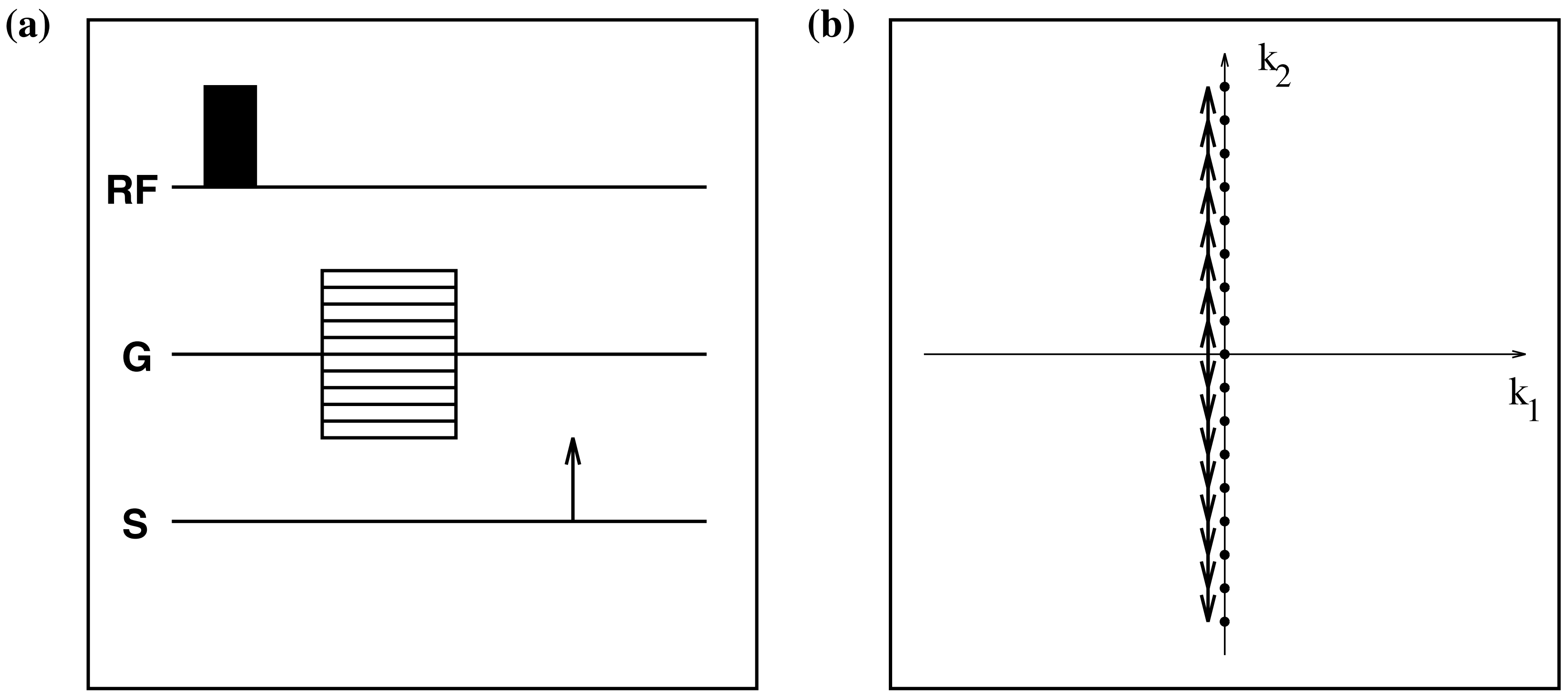
Encoding strategies

Phase encoding

- The phase encoding gradient offsets each acquisition in an orthogonal direction to the readout and slice

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r})\exp\left(i2\pi\mathbf{k}\cdot\mathbf{r}\right)d^3\mathbf{r}$$



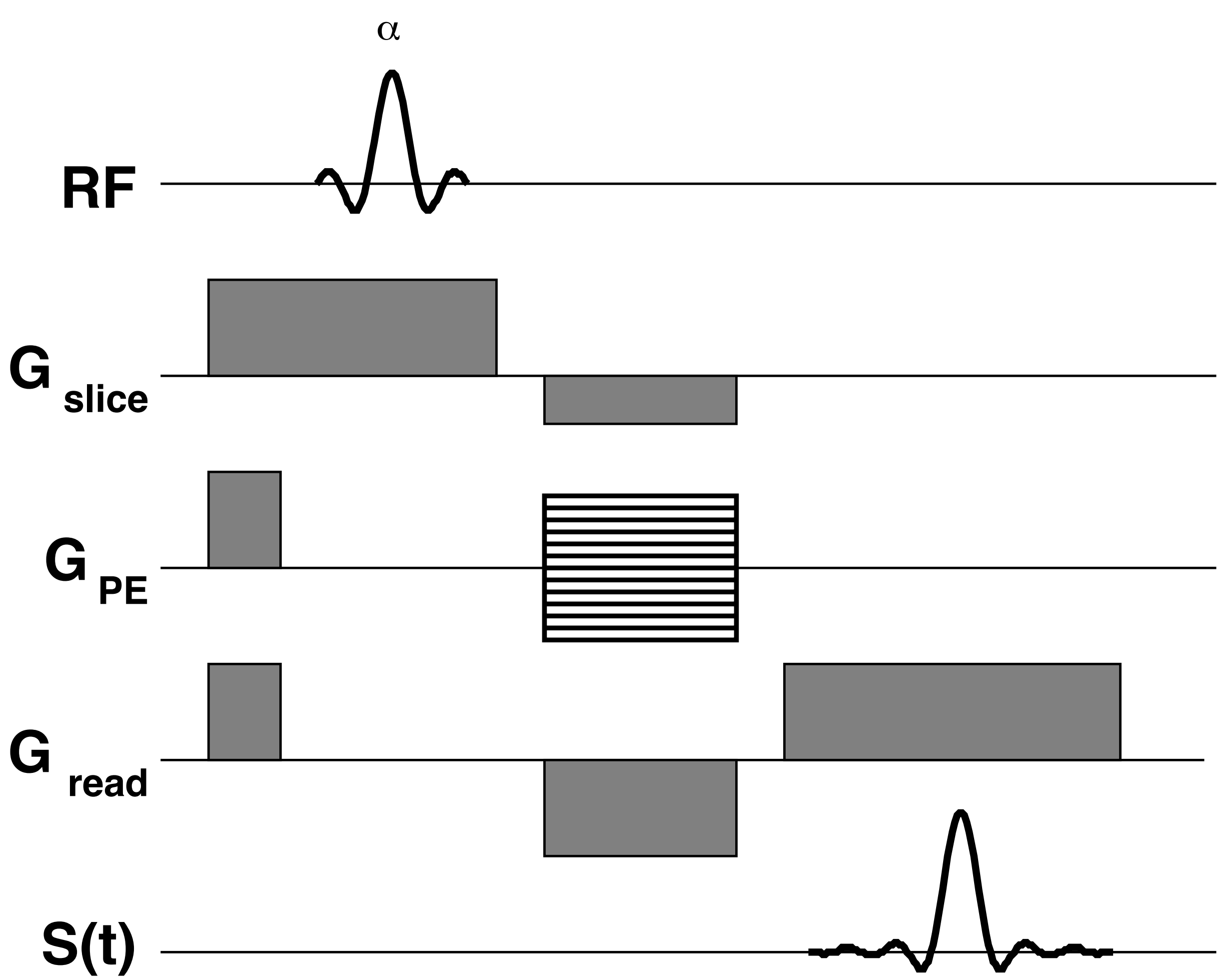
Encoding strategies

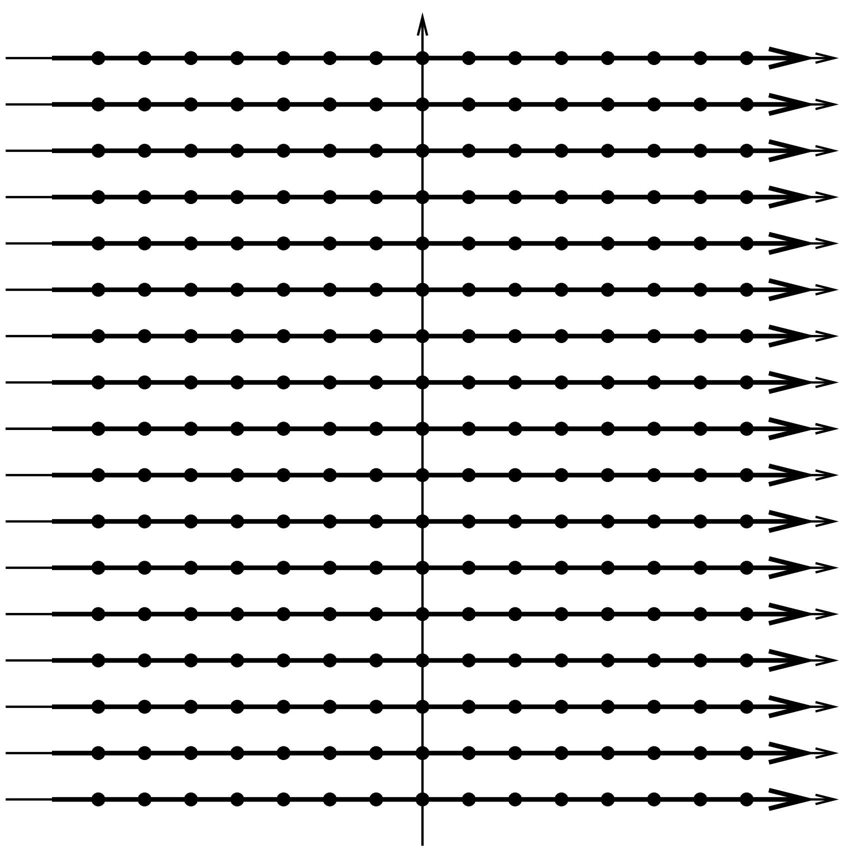
2D Gradient echo imaging

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r}) \exp\left(i2\pi\mathbf{k}\cdot\mathbf{r}\right) d^3\mathbf{r}$$

- Slice selective excitation combined with gradient echo in one direction and phase encoding in the other





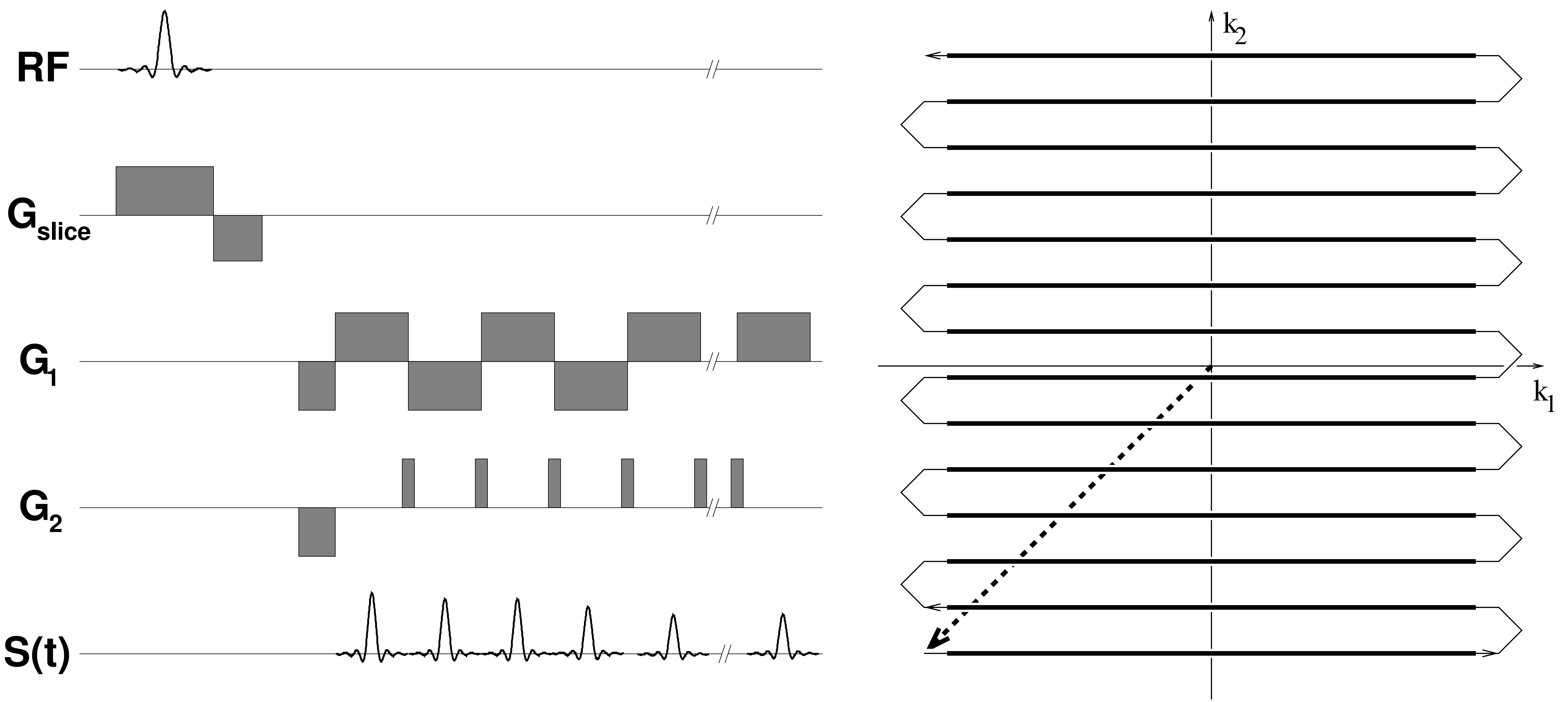
Encoding strategies

Echo planar Imaging

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r}) \exp(i2\pi\mathbf{k}\cdot\mathbf{r})\, d^3\mathbf{r}$$

- Acquire the whole 2D k-space after excitation
- Time varying gradients during the acquisition

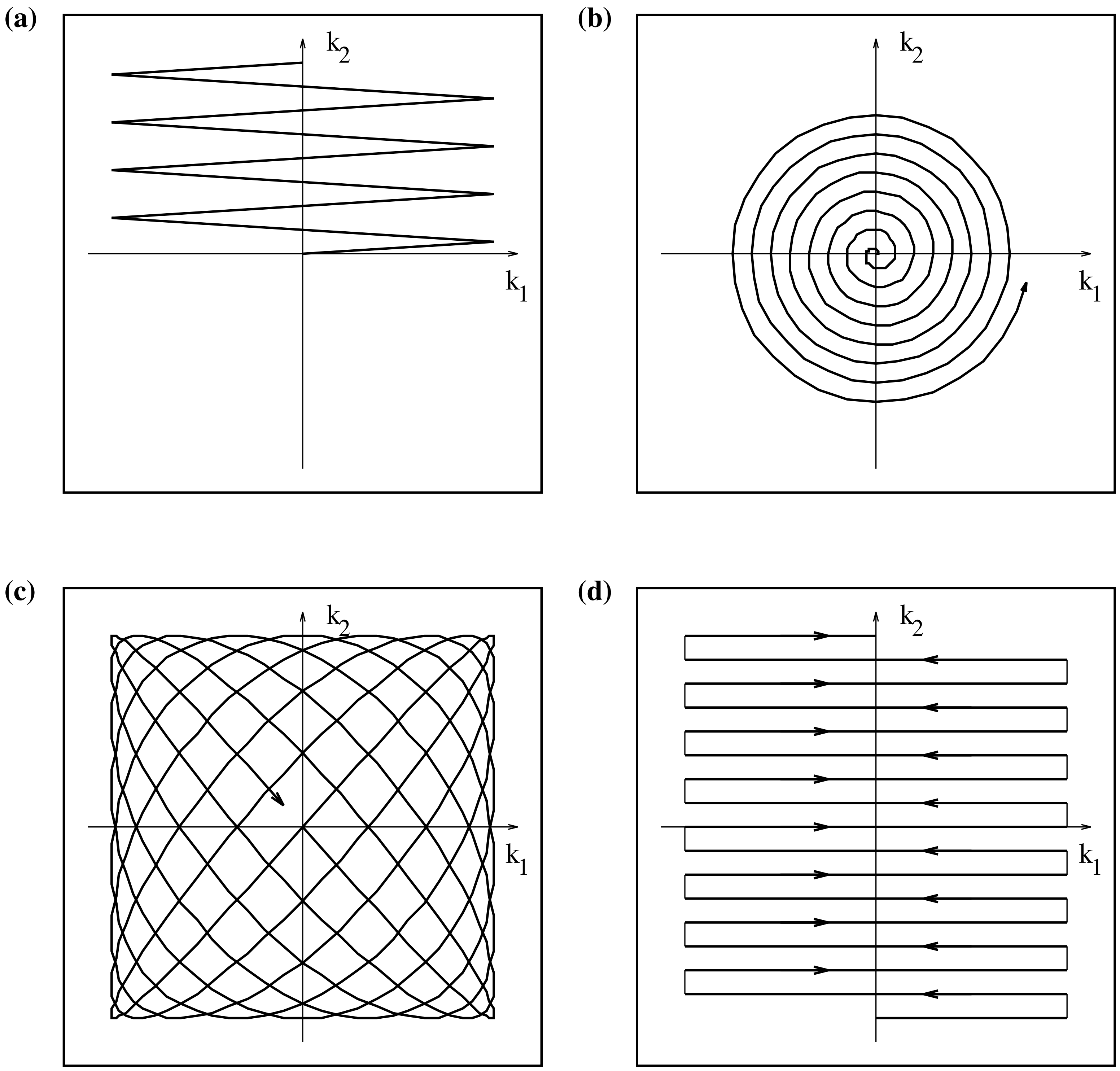


Encoding strategies

Spirals etc...

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r})\exp(i2\pi\mathbf{k}\cdot\mathbf{r})\,d^3\mathbf{r}$$



Accelerated Acquisitions

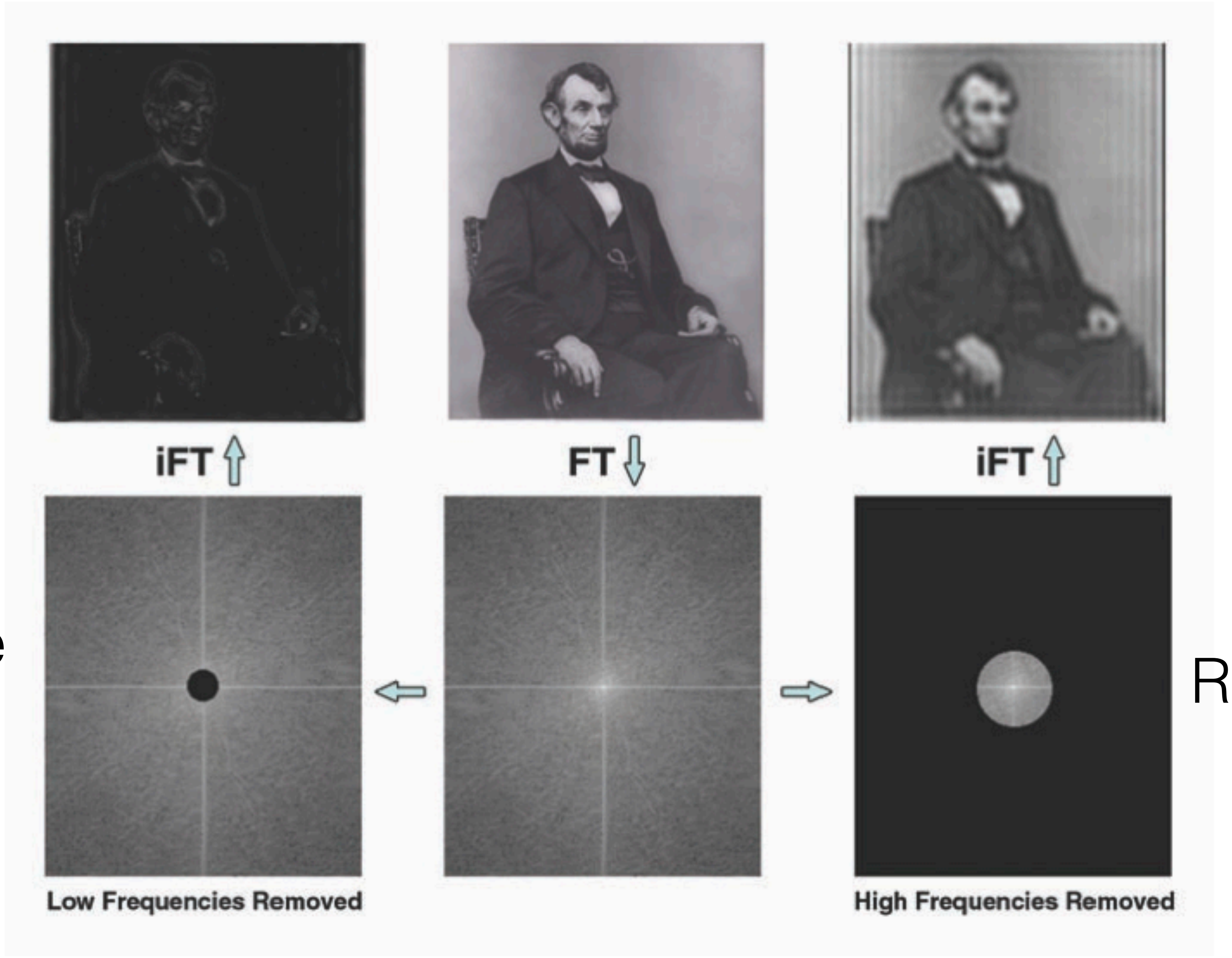
Improving the temporal resolution

- Long image acquisition times
  - low temporal resolution
  - reduce coverage
  - increase sensitivity to subject motion
  - EPI: increase distortion
  - fMRI: reduce sensitivity to rapid events
- Partial k-space acquisition
  - partial echo
  - partial k-space
- Parallel imaging
  - SENSE
  - GRAPPA
- Multiband / Simultaneous multi-slice imaging

Acceleration techniques

keyhole imaging

Acquire once



Repeat

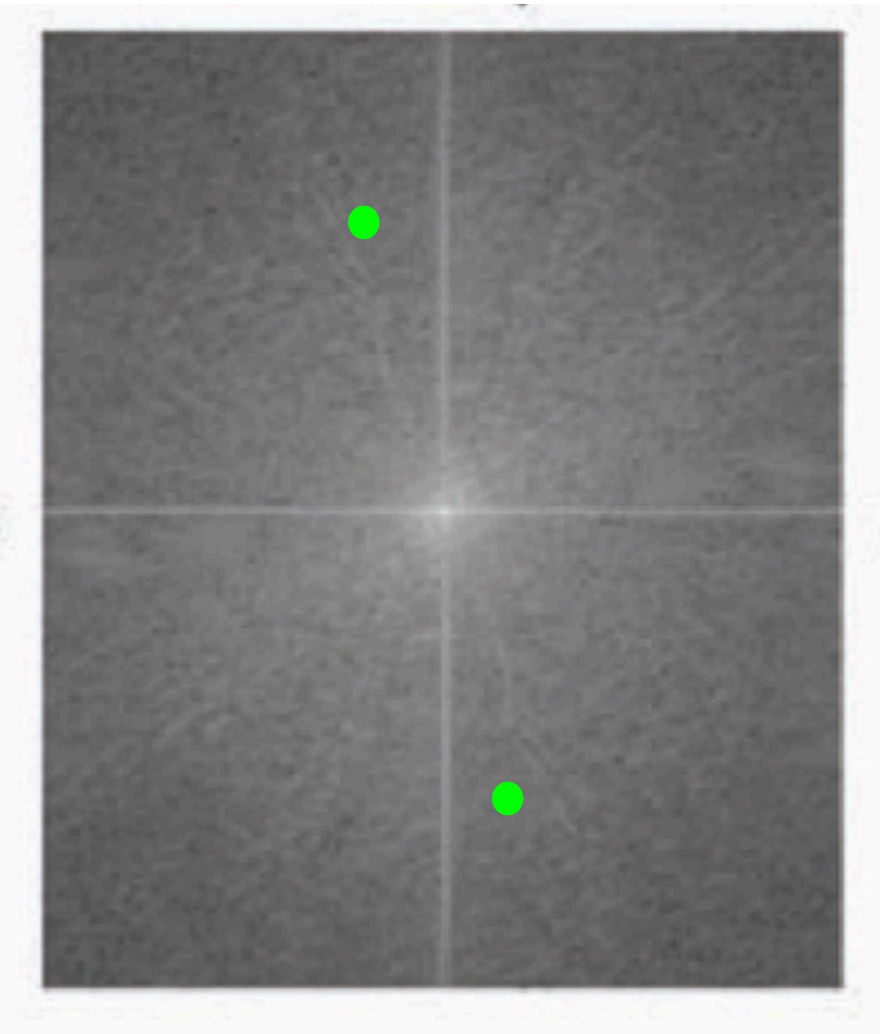
AJNR, Gallagher, DOI:10.2214/AJR.07.2874

Acceleration techniques

Partial k-space

- Assume underlying image is real-valued
- Then k-space should be conjugate symmetric

$$S(kx, ky) = S^*(-kx, -ky)$$

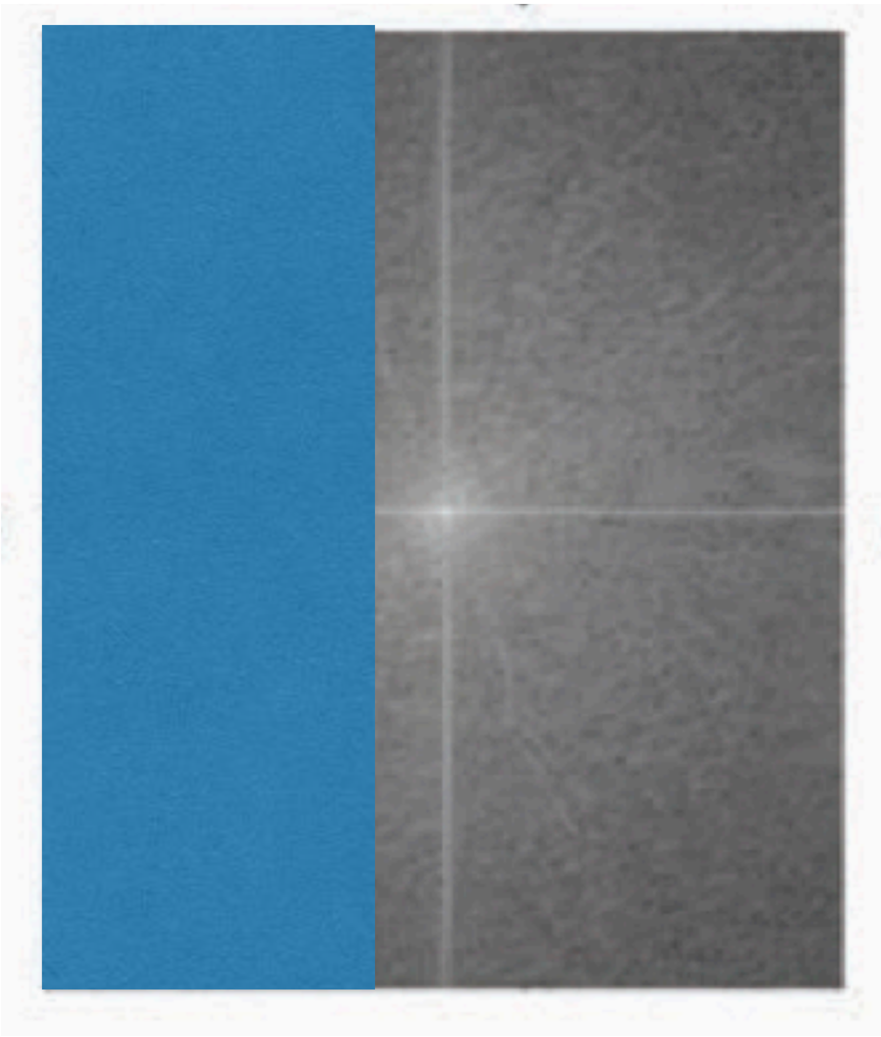


- Synthesize missing conjugate data during image reconstruction
- Need some additional lines near center of k-space to account for mild background phase roll across FOV (e.g. due to receiver coils)
-

Acceleration techniques

Partial k-space

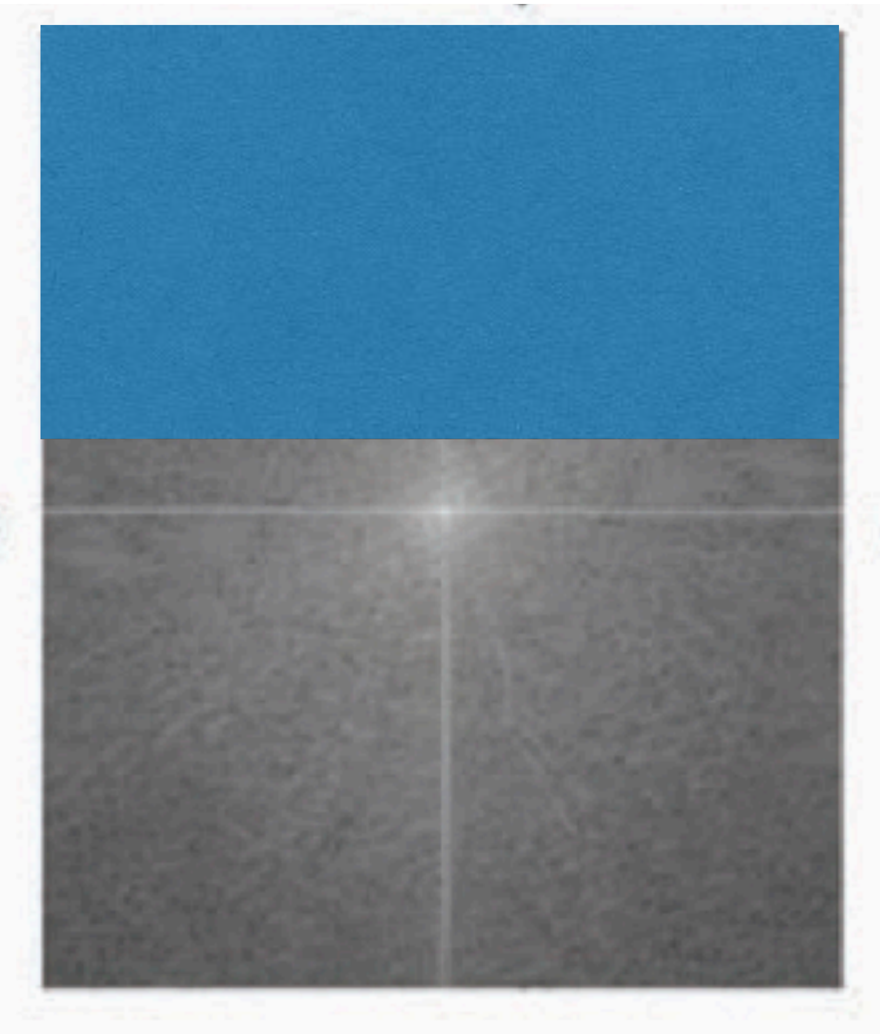
- Partial echo / Asymmetric echo
  - Short TE



Acceleration techniques

Partial k-space

- Partial NEX / Fractional k-space
  - Less TRs needed to form image



Parallel Imaging

Multiple RF receiver coils

- Modern MRI systems acquire signals from multiple receiver coils
- Typically 8-64 channels

Parallel Imaging

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- Each coil has a distinct spatial sensitivity profile in space
- The MR signal from each receiver channel is weighted by the corresponding coil's sensitivity profile
- This provides an additional mechanism for positional encoding the MR signal

Image from coil 1

Parallel Imaging

Multiple RF receiver coils

- Modern MRI systems acquire signals from multiple receiver coils
- Typically 8-64 channels
- Each coil has a distinct spatial sensitivity profile in space
- The MR signal from each receiver channel is weighted by the corresponding coil's sensitivity profile
- This provides an additional mechanism for positional encoding the MR signal

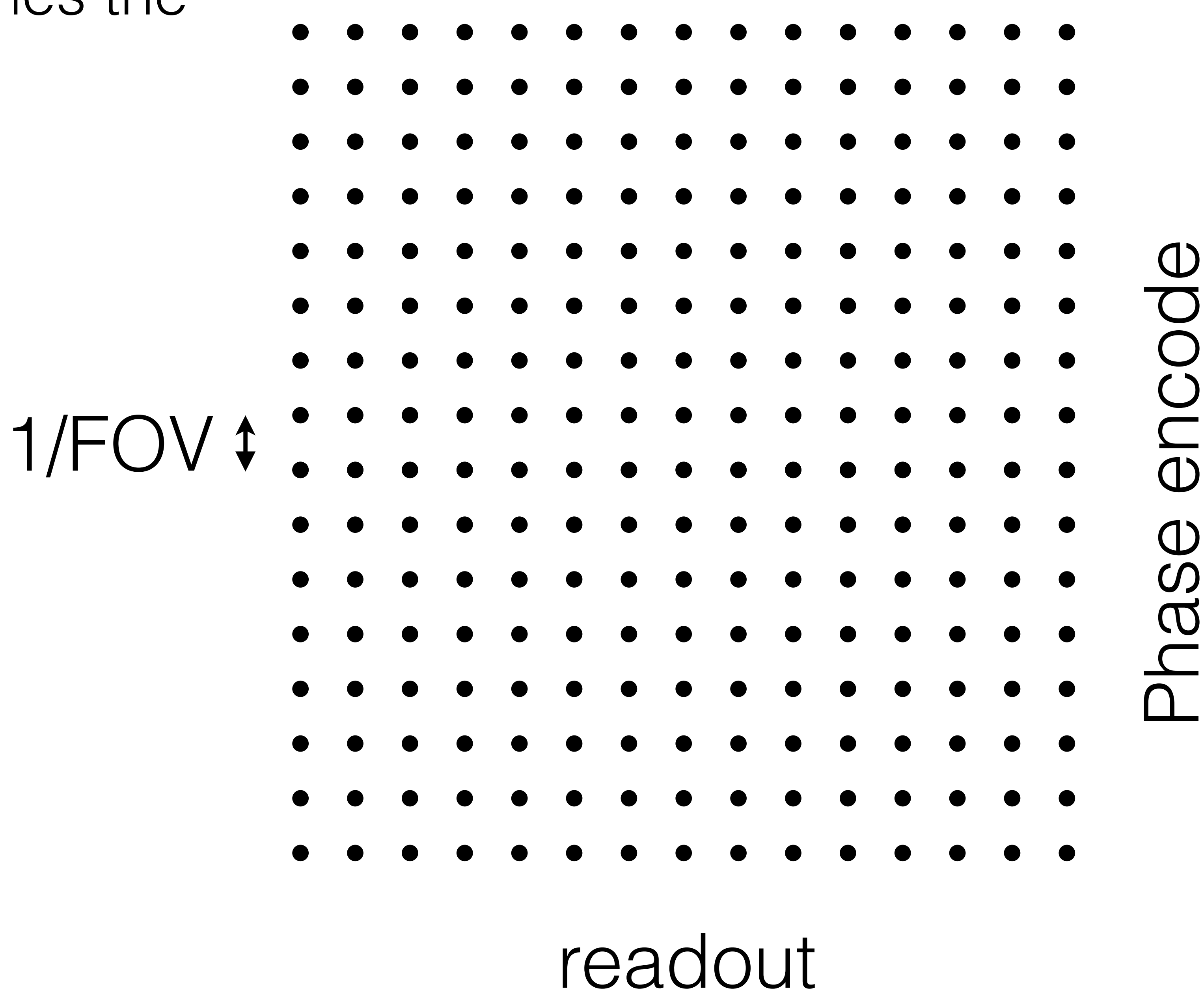
Image from coil 2

Parallel Imaging

Undersampling k-space

- Regular acquisition
- Inter-sample spacing defines the FOV of the final image
- Inverse relationship

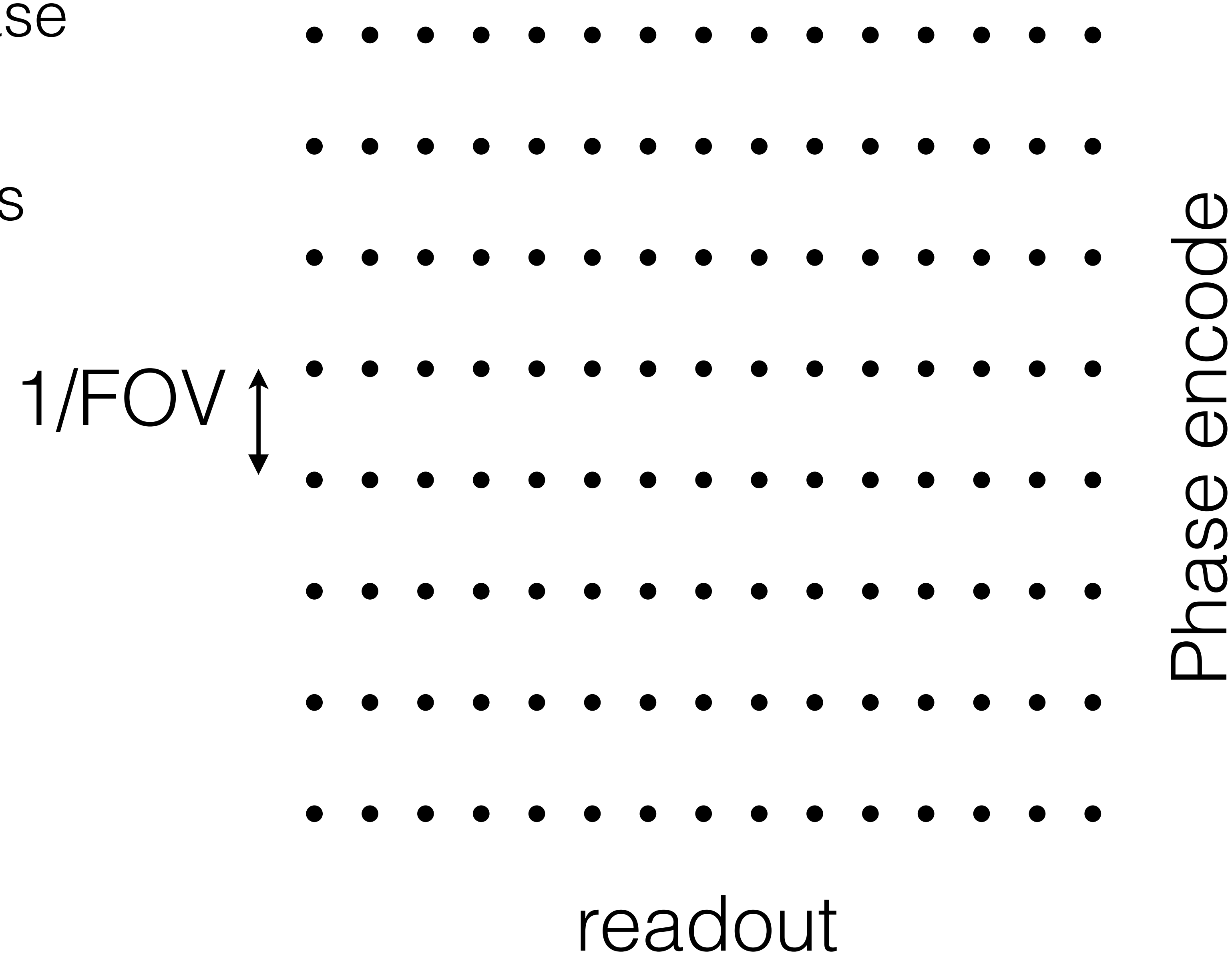
  dk = 1/FOV



Parallel Imaging

Undersampling k-space

- Acquire only even-numbered k-space lines
- Effectively 1/2 FOV in phase encoding direction
- FOV in readout direction is unaffected



Parallel Imaging

SENSE example (R=2)

- Acquire only even-numbered k-space lines
  - effectively 1/2 FOV in phase encoding direction
  - aliased images
- Solve linear system to unwrap the aliasing at I(x, y) - green dot.

  $S_1(x,y) = w_{11}\ I(x,y) + w_{12}\ I(x, y+N/2)$

  $S_2(x,y) = w_{21}\ I(x,y) + w_{22}\ I(x, y+N/2)$

  etc.

  Matrix form: $S = E\ I$

  Solution: $I = \mathrm{pinv}(E).S$

- E is the coil encoding matrix
- Serious noise amplification (g-factor) for large R

Image from coil 1

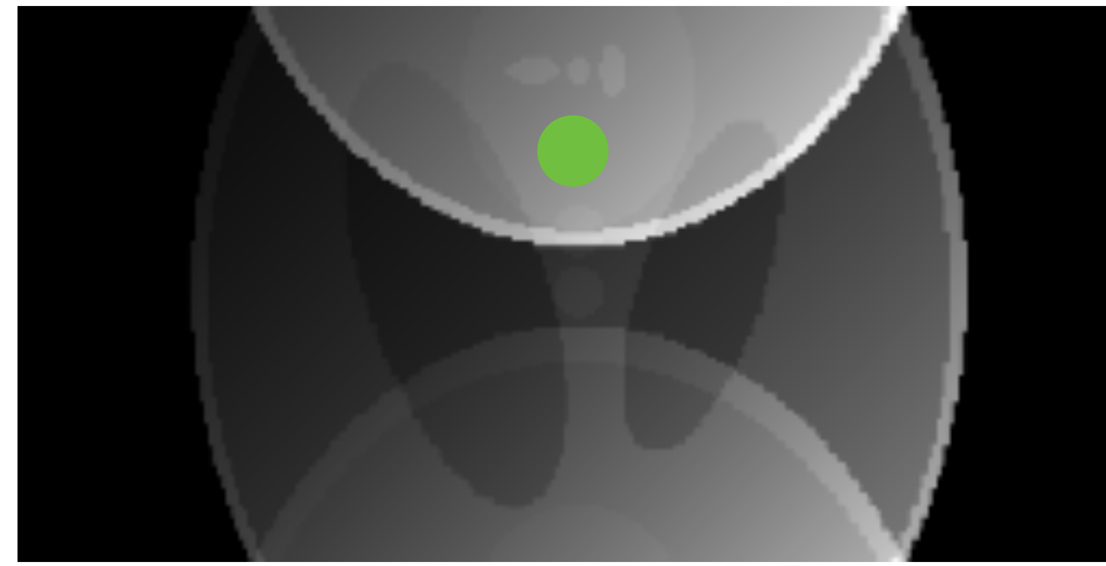
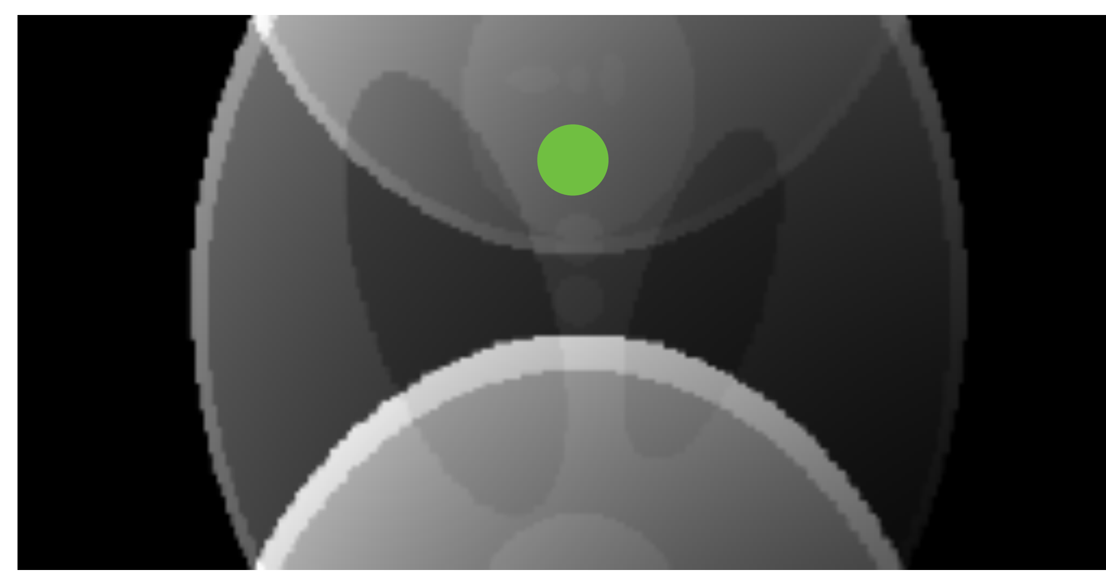


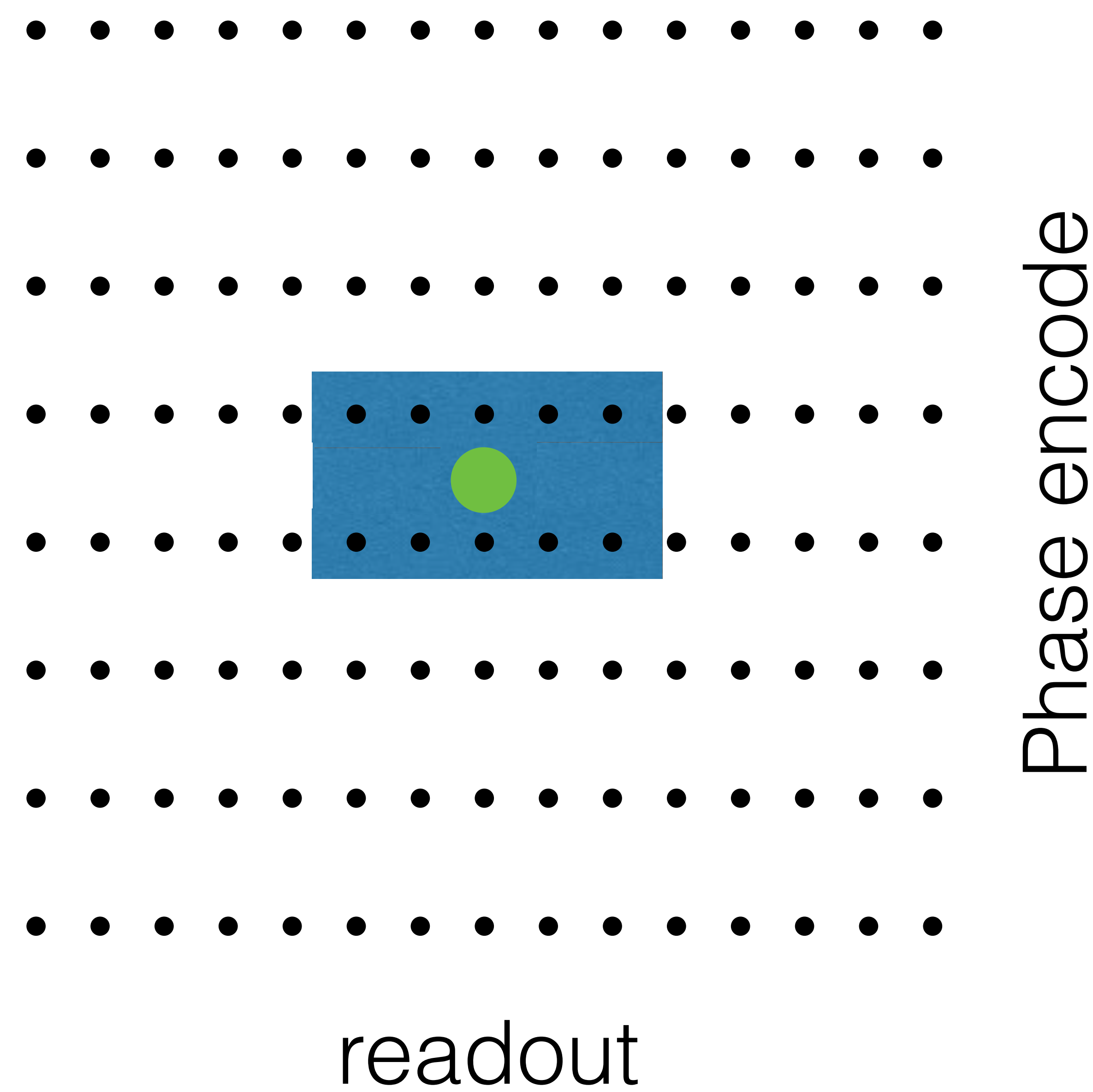
Image from coil 2



Parallel Imaging

GRAPPA (R=2 example)

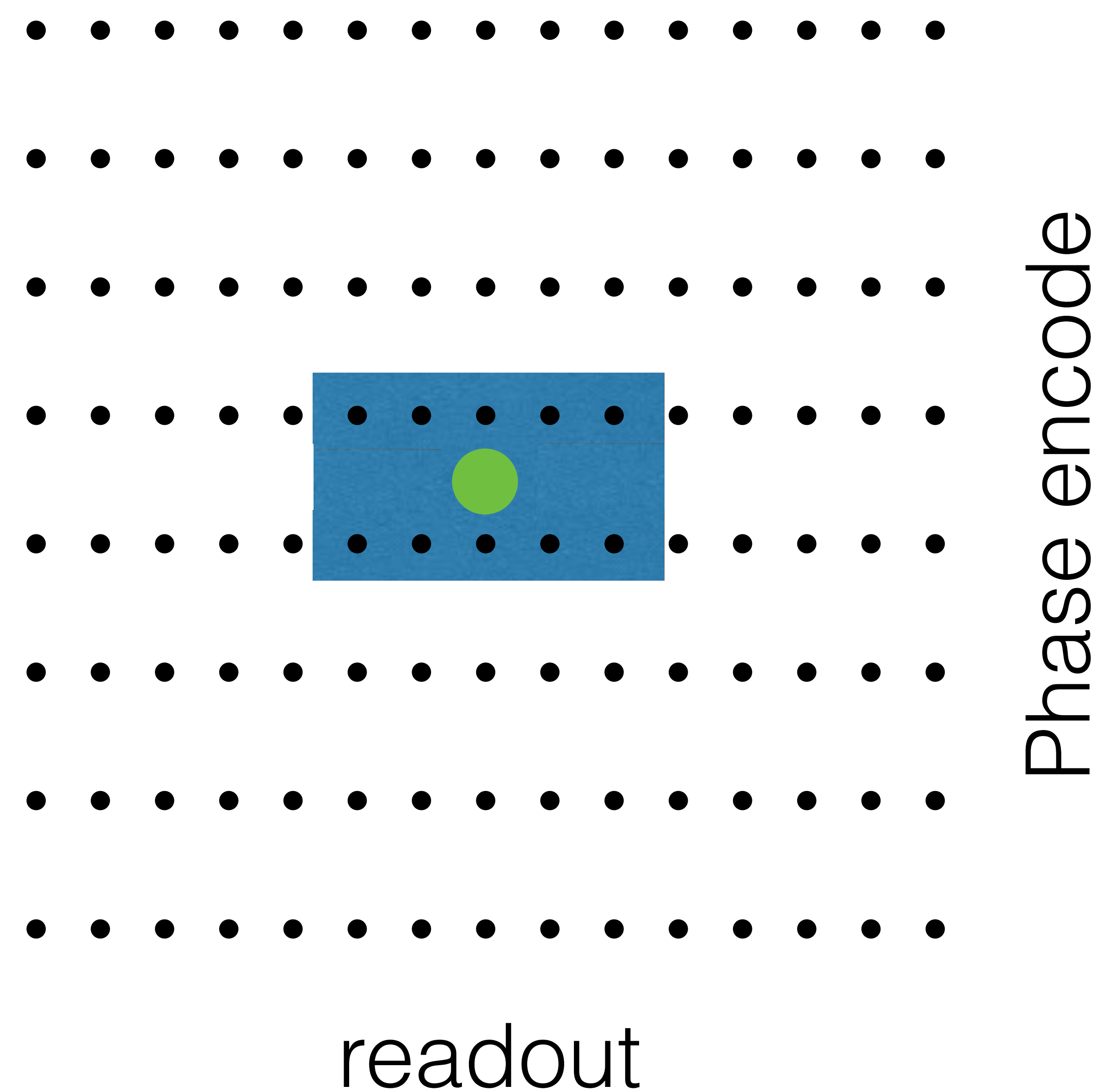
- Similar k-space undersampling as SENSE
- Data is interpolated in k-space from neighboring samples



Parallel Imaging

GRAPPA (R=2 example)

- Similar k-space undersampling as SENSE
- Data is interpolated in k-space from neighboring samples
- Both SENSE and GRAPPA require additional *calibration* data to perform their reconstruction



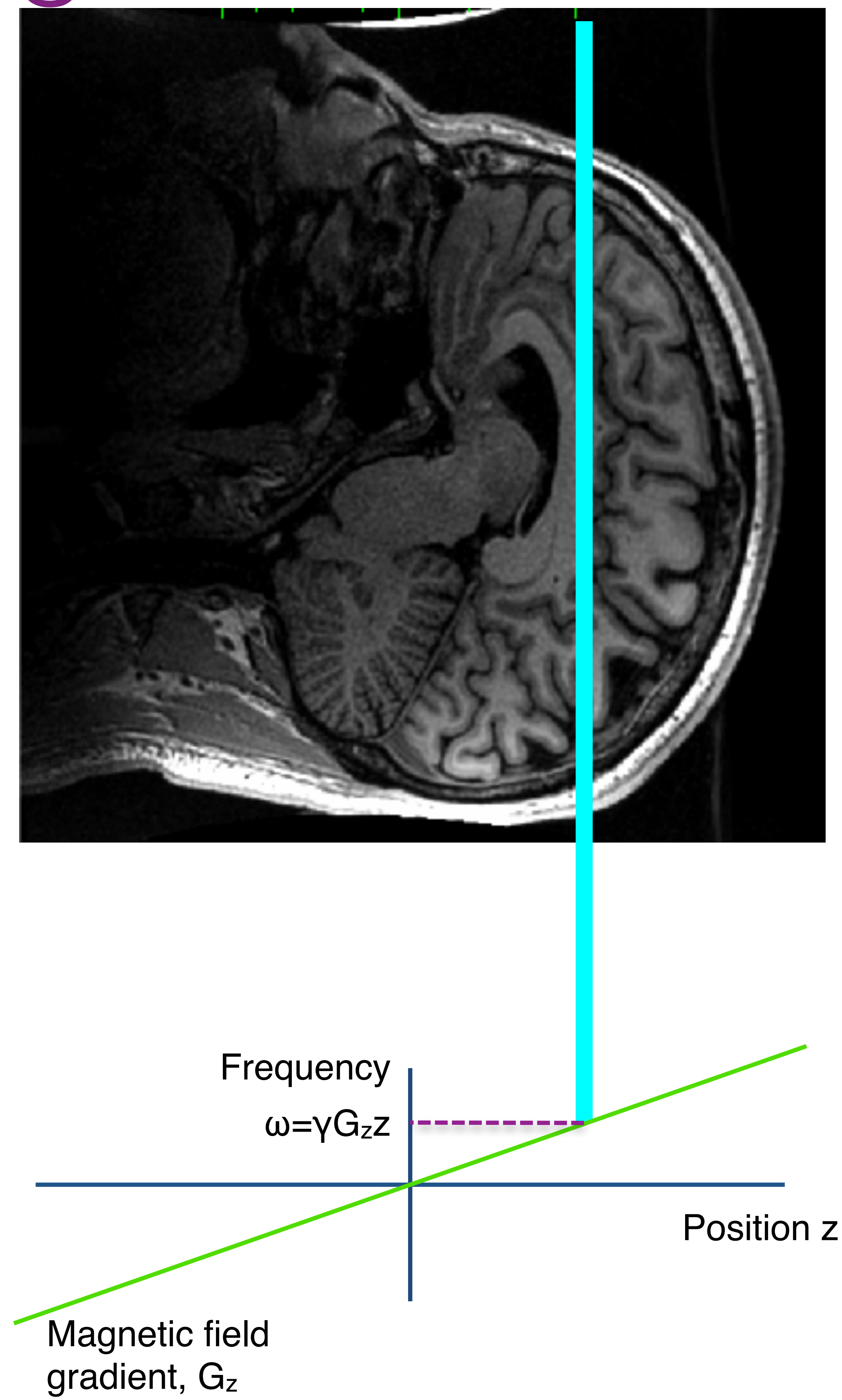
Multiband/SMS imaging

Simultaneous multi-slice imaging

- Consider the slice of tissue at position z
- In the presence of gradient G_z, the local slice frequency is given by:

$$\delta\omega = \gamma G_z z$$

- Excite with frequency $\omega_0 + \delta\omega$ to move slice from isocenter to position of interest.
- Excite with an RF pulse that is the composition of a band of frequencies to define a particular slice width.



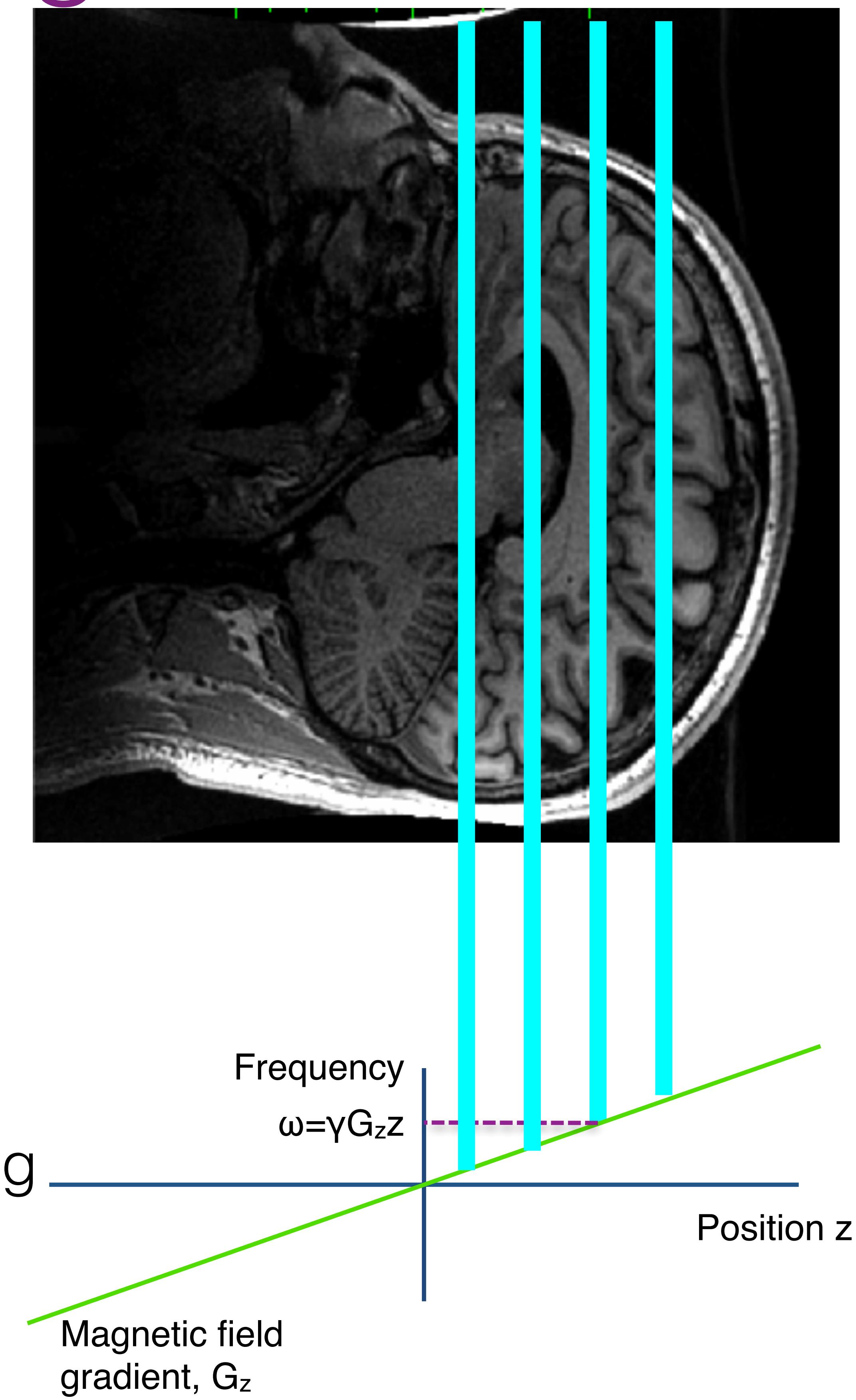
Multiband/SMS imaging

Simultaneous multi-slice imaging

- Create RF pulse composed from the superposition of several (typically 2-12) bands of frequencies.
- Basically add a bunch of RF pulses (with specific frequency modulations) together

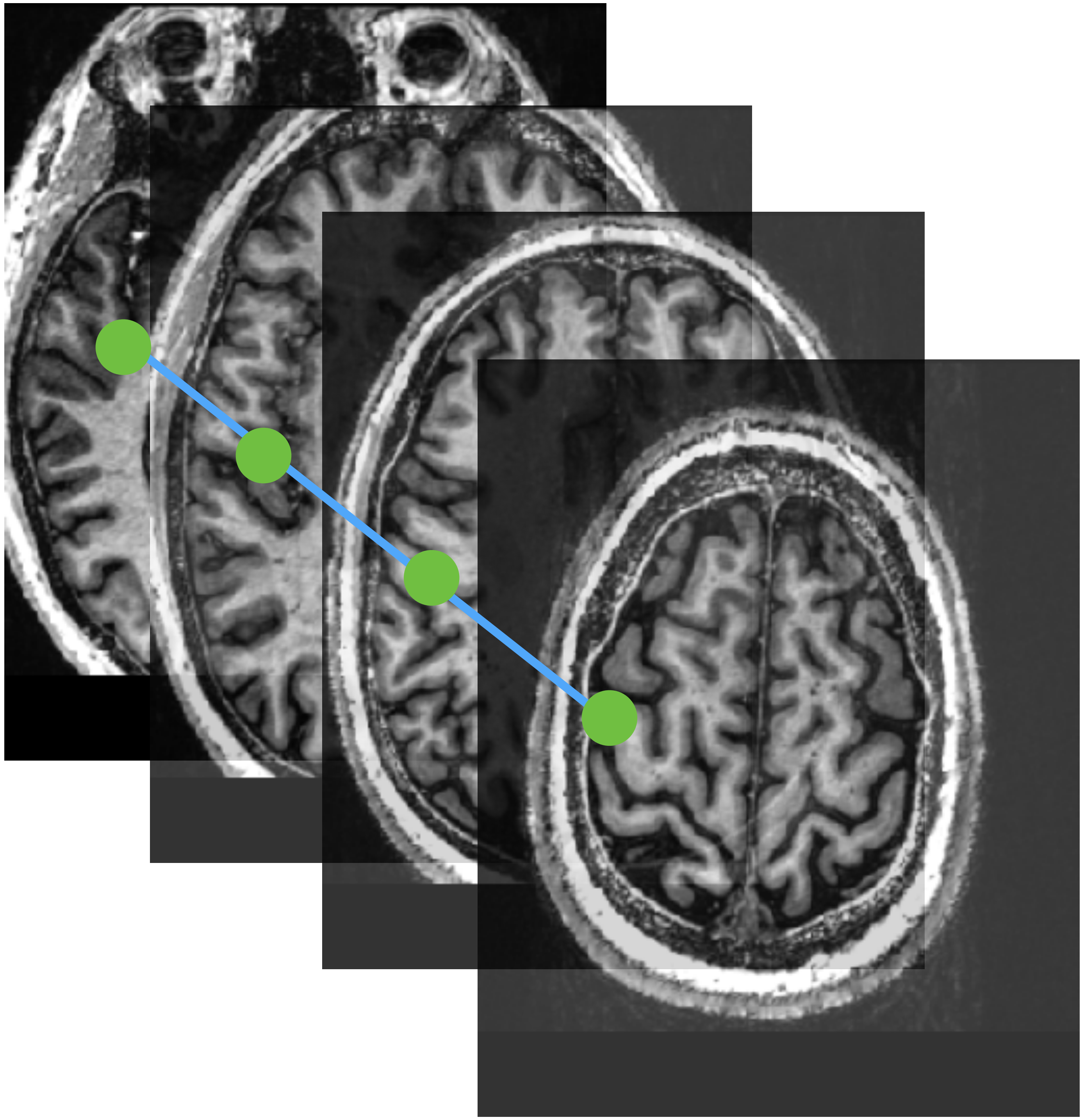
- Multiple slices excited with each RF pulse
- Works reasonably well when slices separated by >3 slice widths.

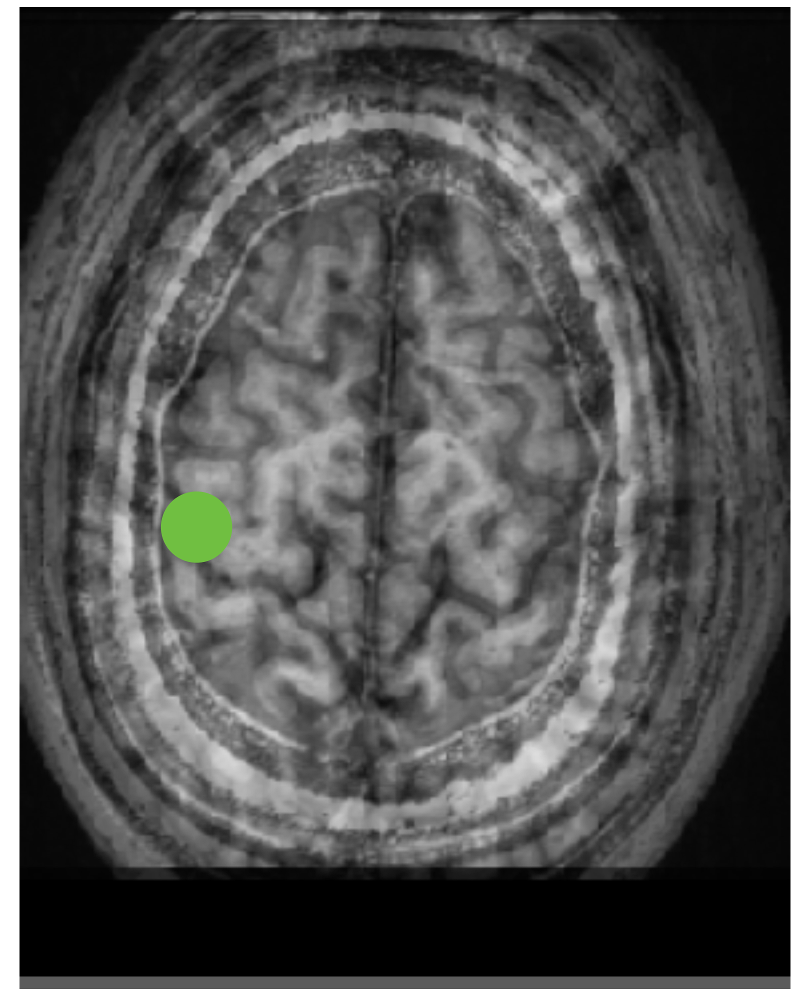
- Side effect: SMS RF pulses have significantly higher SAR than corresponding single slice excitations



Multiband/SMS imaging

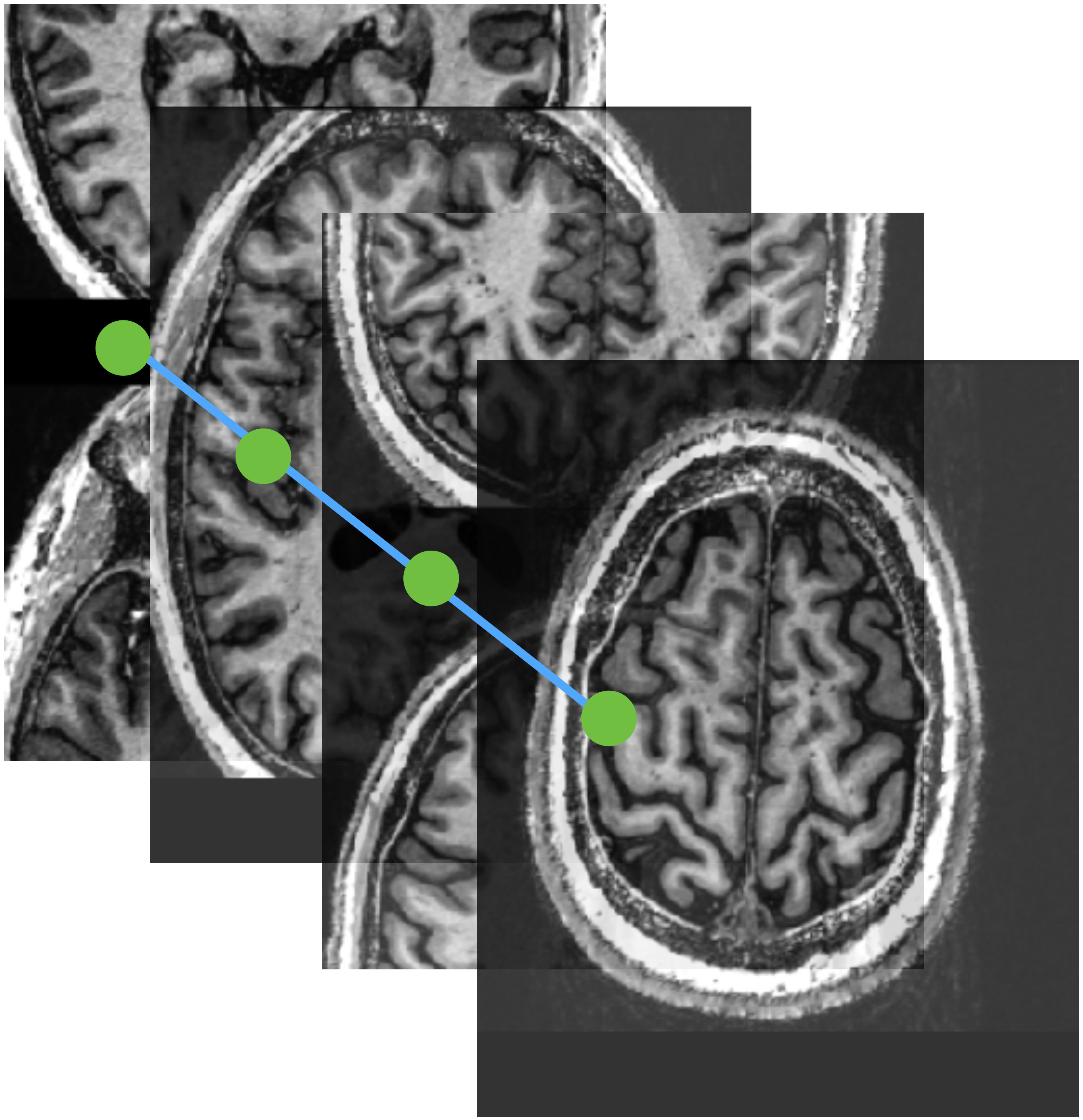
Simultaneous multi-slice imaging



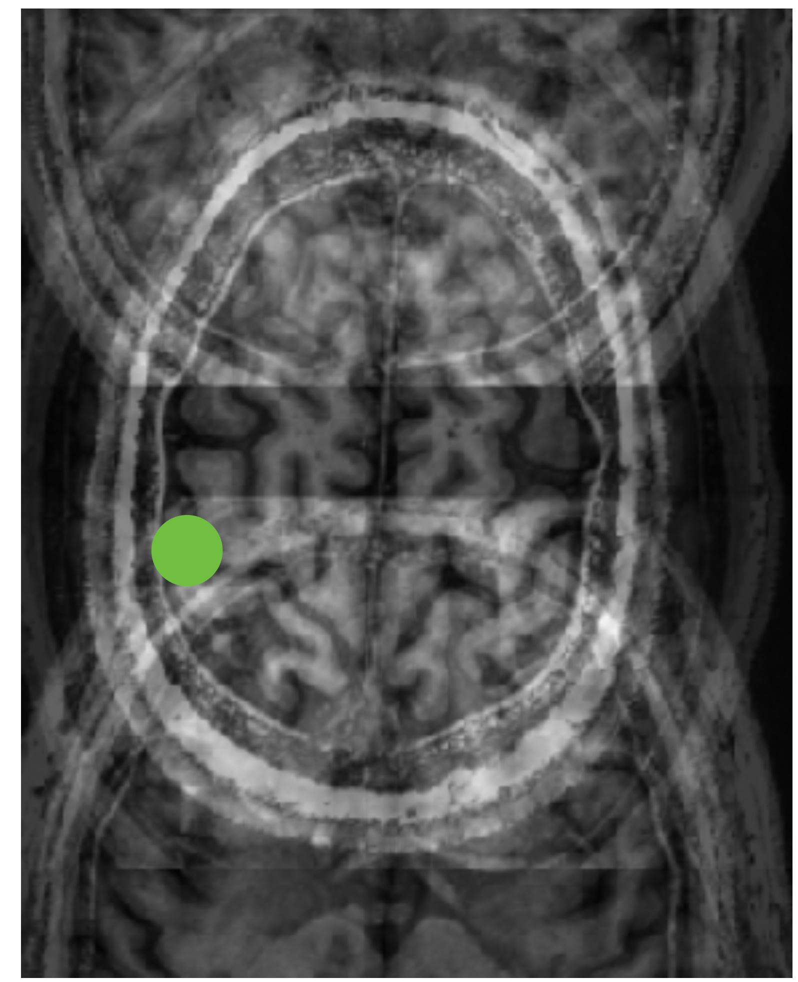


Multiband/SMS imaging

Simultaneous multi-slice imaging



CAIPI

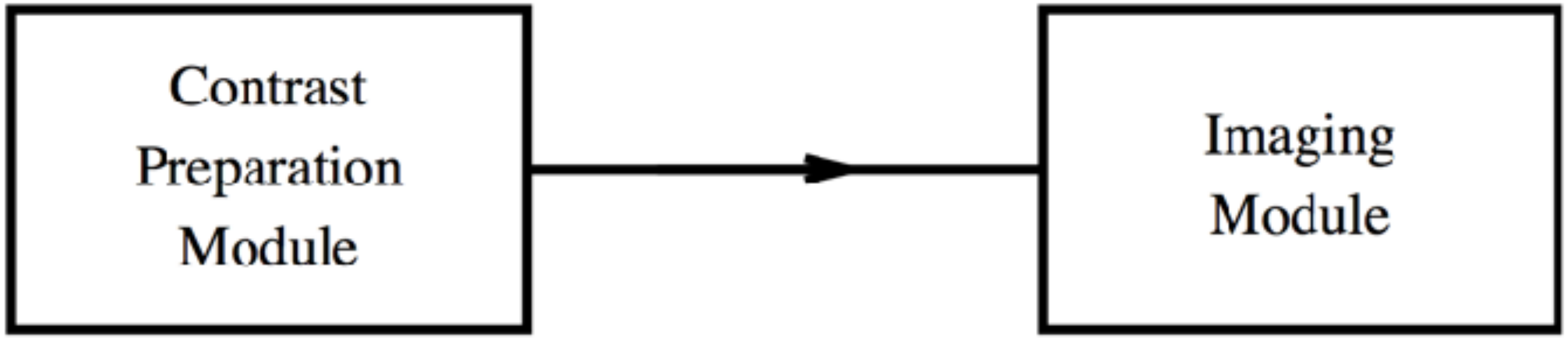


Improves g-factor
(i.e. less noise)

Controlling image contrast

- Intrinsic contrast of the pulse sequence
  - Gradient-echo and spin-echo sequences
  - Effects of TE and TR

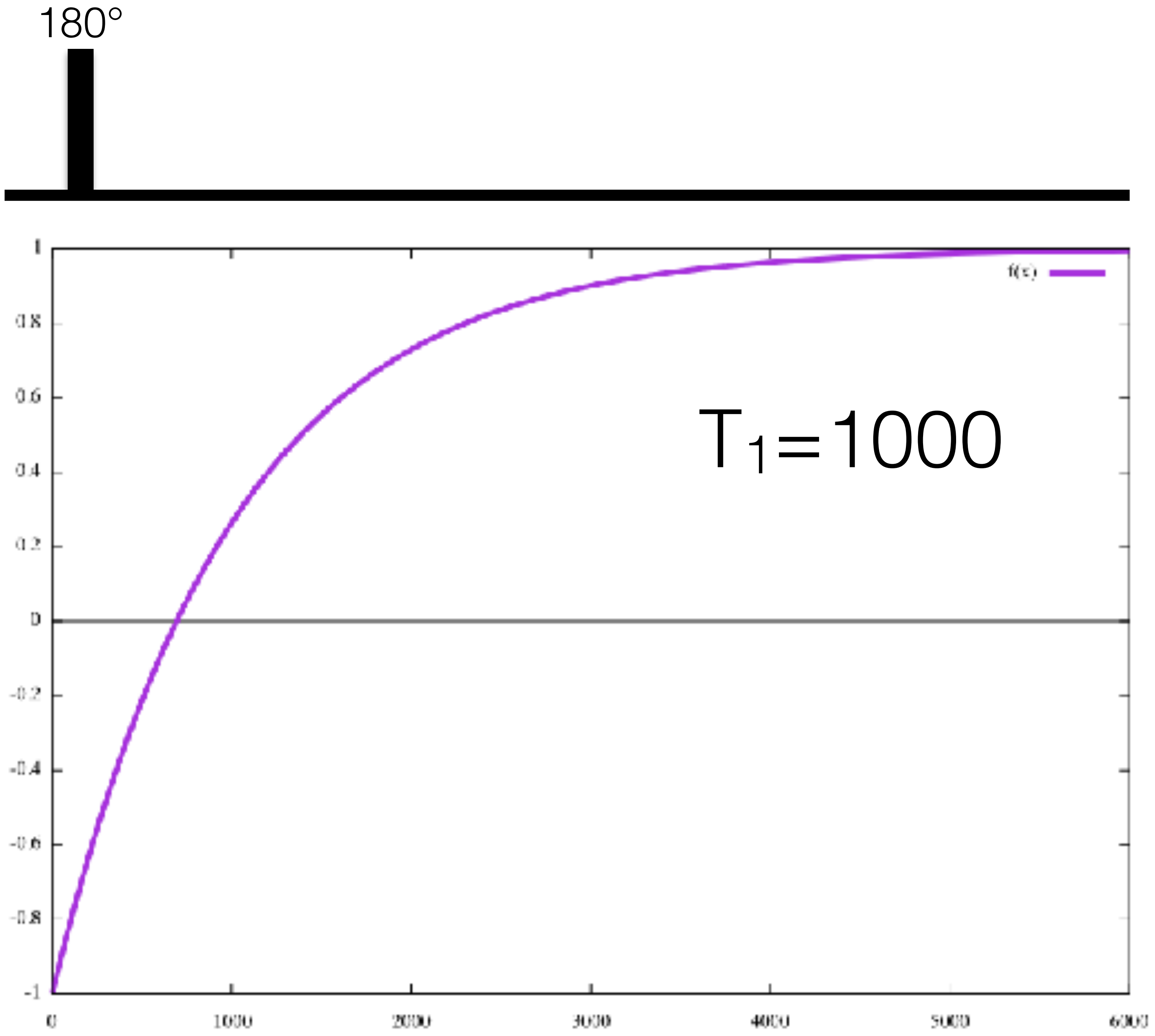
- Magnetization Preparation methods (examples)
  - Fat saturation
  - Flow preparation
  - Diffusion preparation



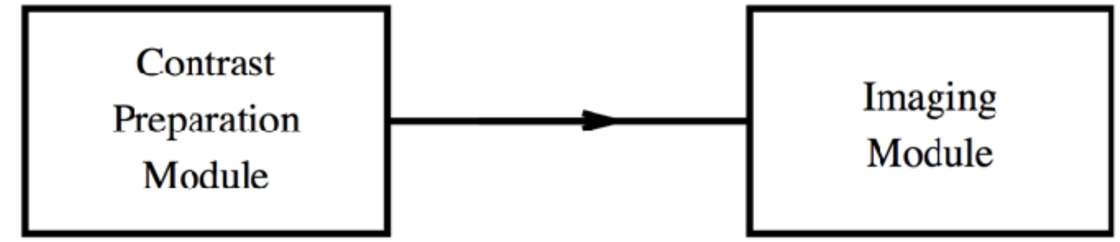
Inversion recovery

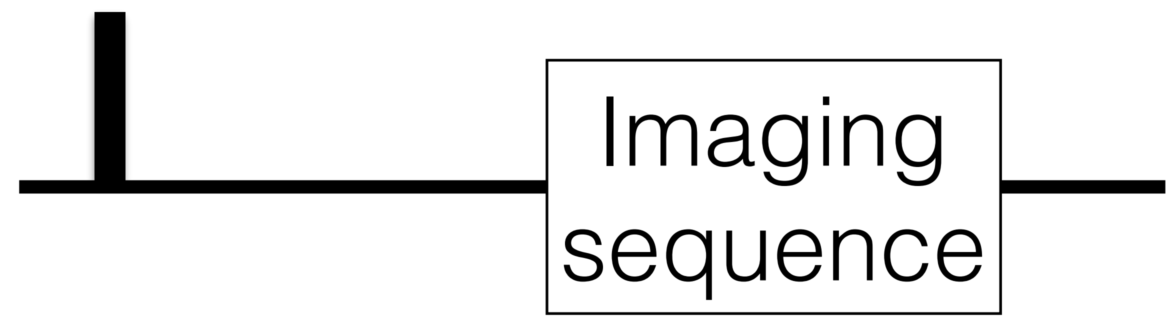
$$M_z(t) = M_0 + (M_z(0) - M_0)e^{-t/T1}$$

- Invert magnetization with 180° inversion pulse
- Magnetization slowly recovers by T_1

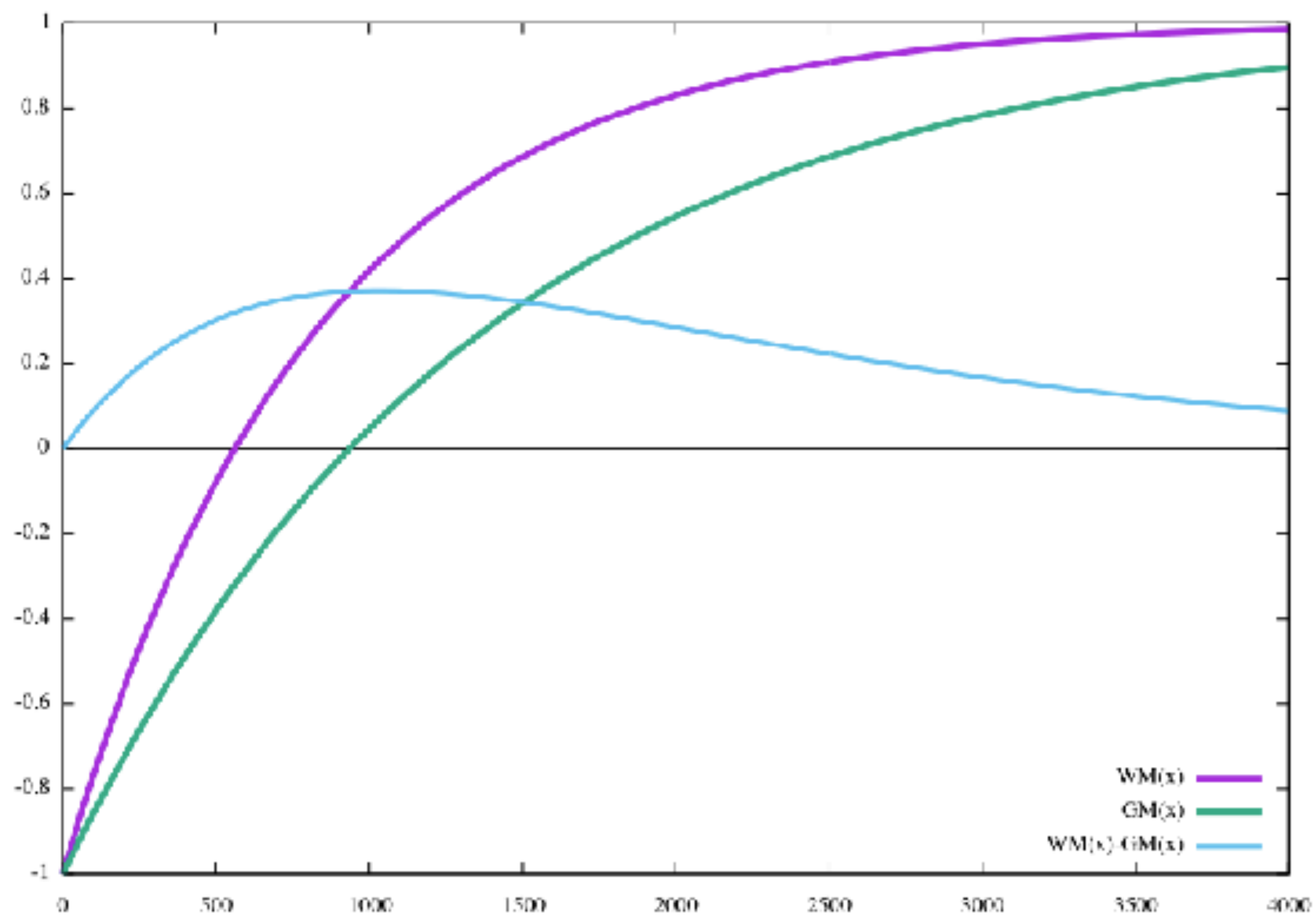


Inversion recovery



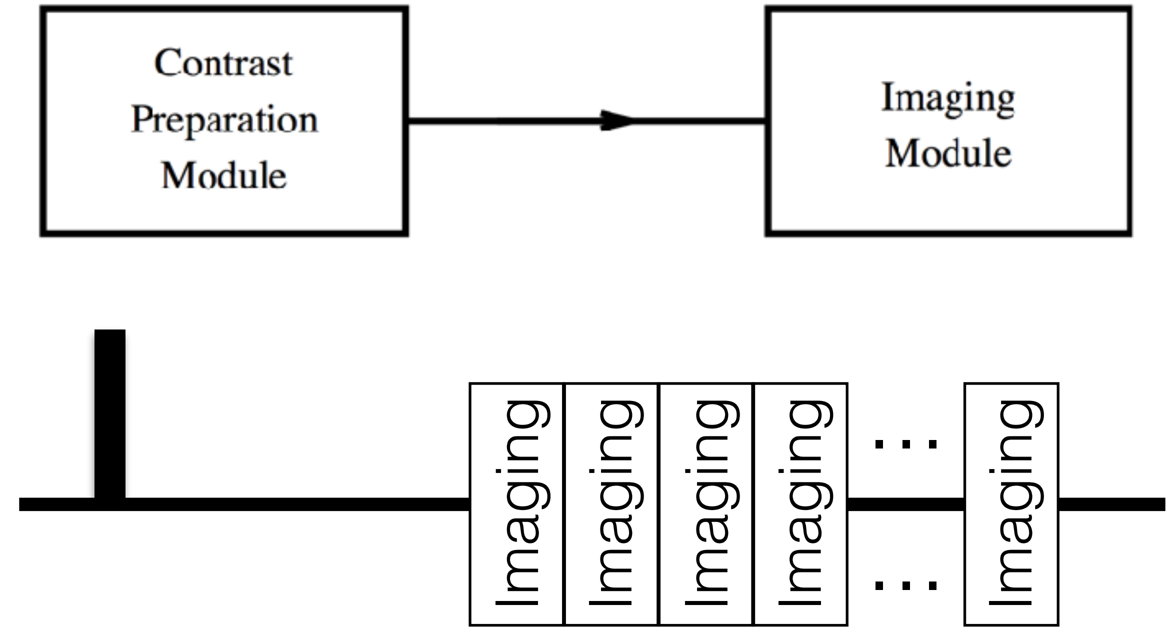


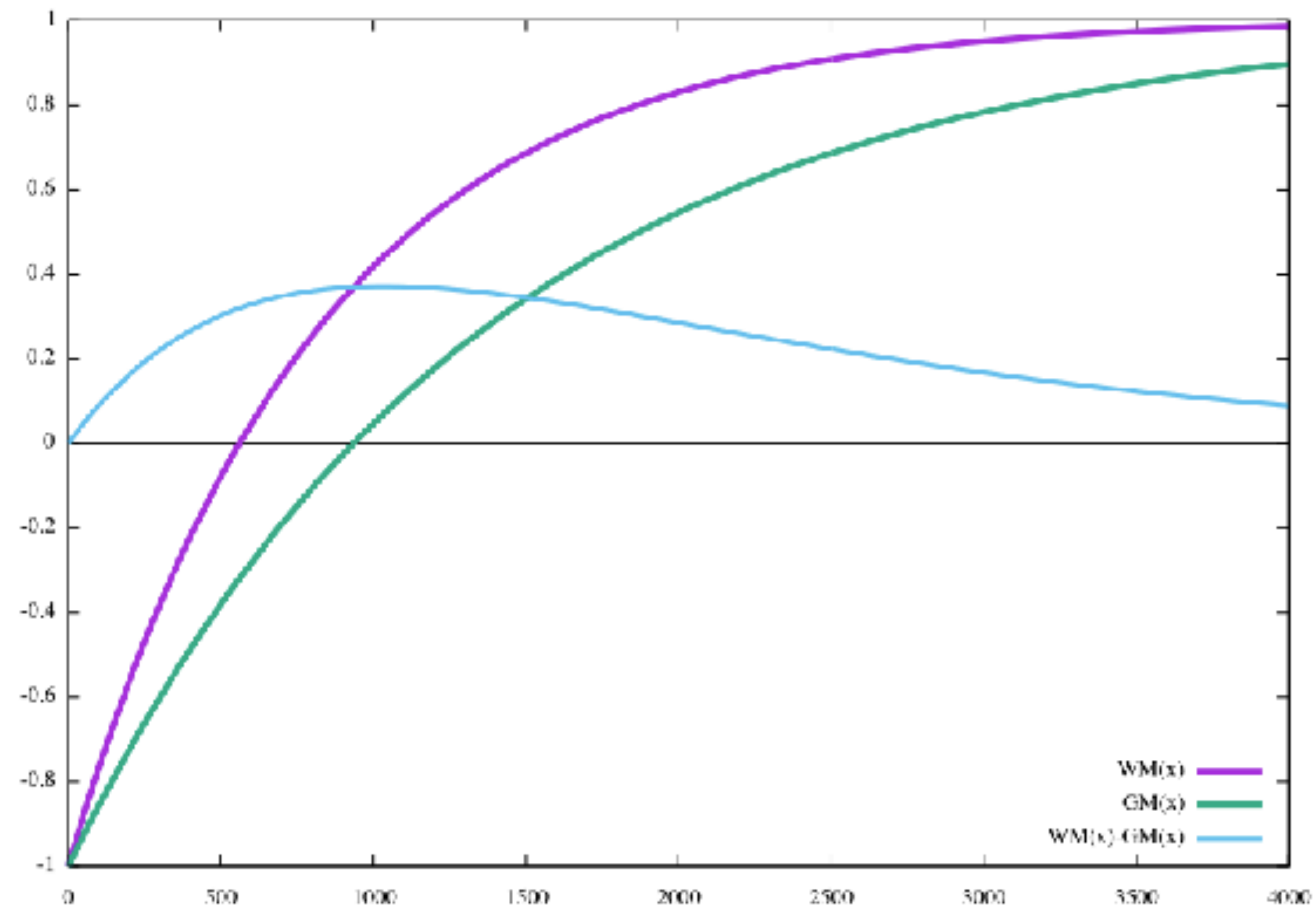
- Prepare magnetization with 180° inversion pulse
- Wait ...
  TI=inversion time.
- Perform imaging sequence.
- Obtain images with IR contrast



MPRAGE

- Prepare magnetization with 180° inversion pulse
- Wait TI for constrast to evolve
- Perform multiple imaging sequence acquisitions (typically do all kz, slice encodes)
- Repeat for all ky (phase encodes).



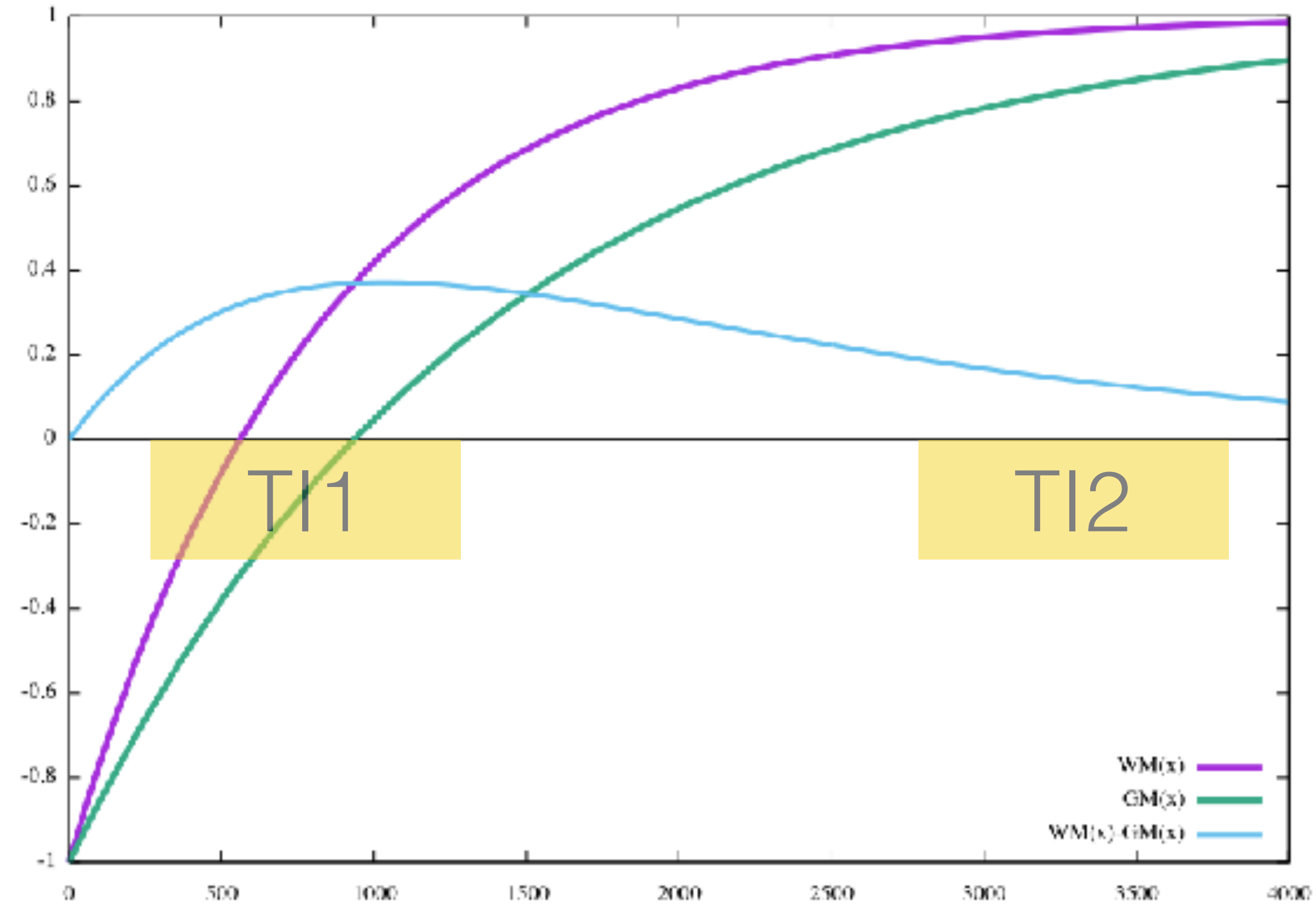


MP-RAGE

Magnetization Prepared Rapid Acquisition with Gradient Echoes

- 3D anatomical scan with white/grey matter contrast
- Typically:
  - 0.8-1.25mm isotropic resolution
  - 6 -12 minutes scan time
- **Multi-echo MPRAGE**
  - acquires images at multiple TEs during each imaging module
  - Different T2* weightings
- **MP2RAGE**
  - acquires images at two TIs (short, long) after each inversion pulse
  - combined images can be used to compensate for variations in image intensity (e.g. due to coil profiles)
  - Can generate estimated spatial maps of T_1 and T_2.

MP2-RAGE

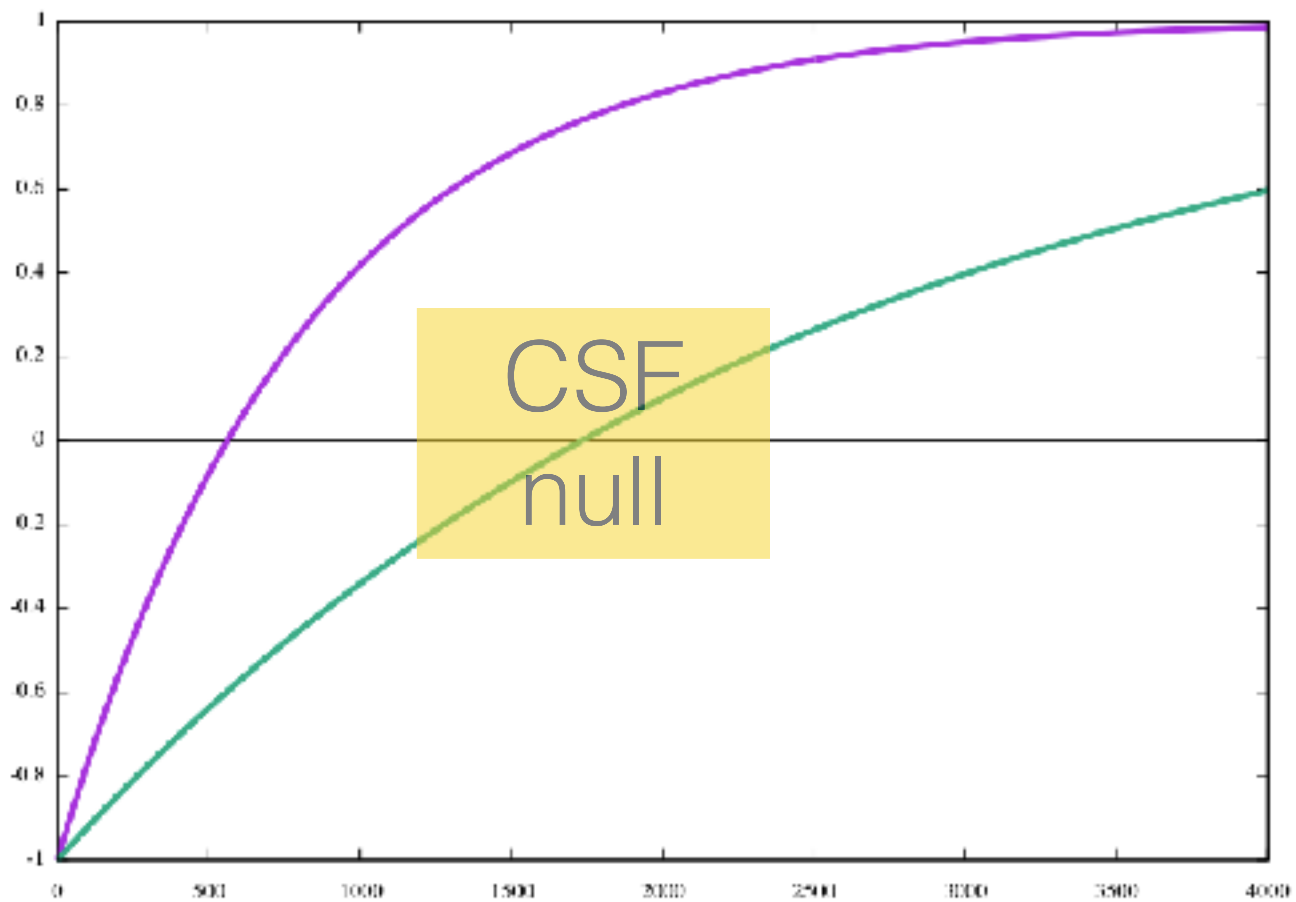


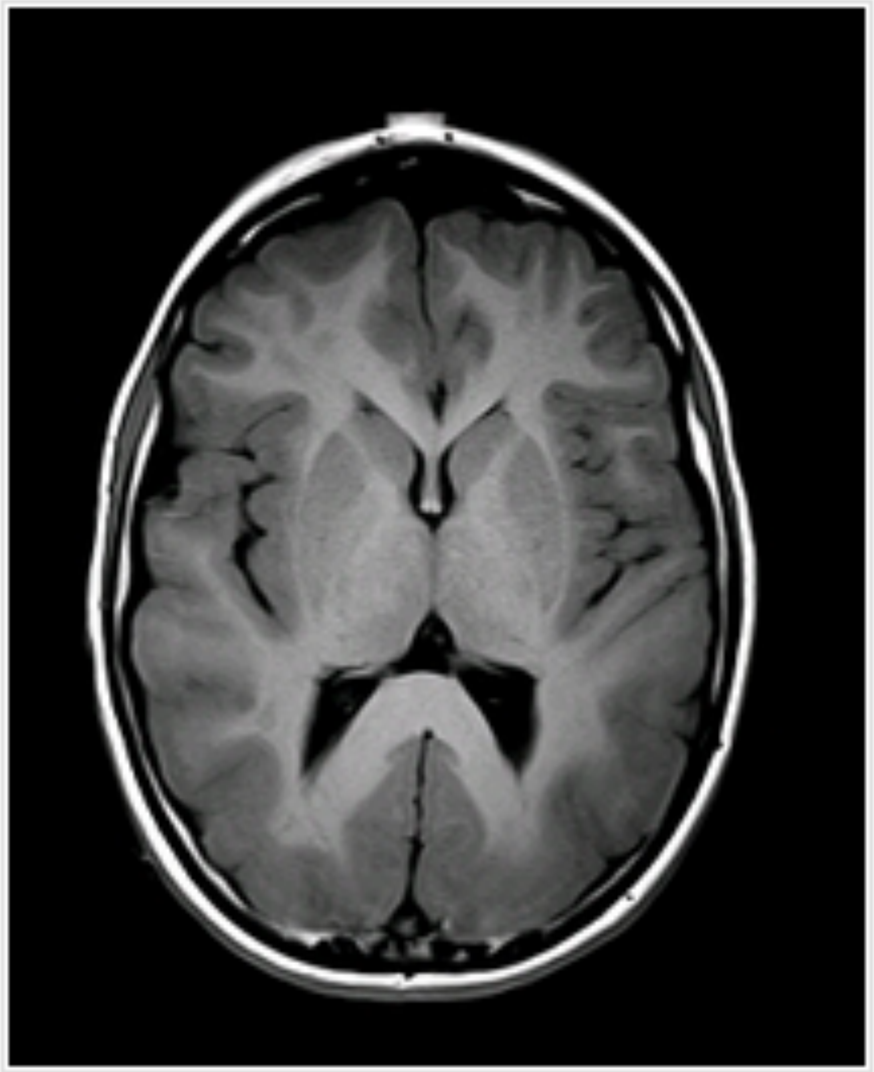
- **MP2RAGE**
  - acquires images at two TIs (short, long) after each inversion pulse
  - combined images can be used to compensate for variations in image intensity (e.g. due to coil profiles)
  - Can generate estimated spatial maps of T_1 and T_2.

FLAIR

Fluid attenuated inversion recovery

- Null (long T1) signals due to fluid (e.g. CSF)





T1-FLAIR brain image at 3T with TR=2100, TE=9, TI=880.

- Often uses FSE/TSE for the imaging sequence
  - short TR for T1w-FLAIR
  - long TR for T2w-FLAIR

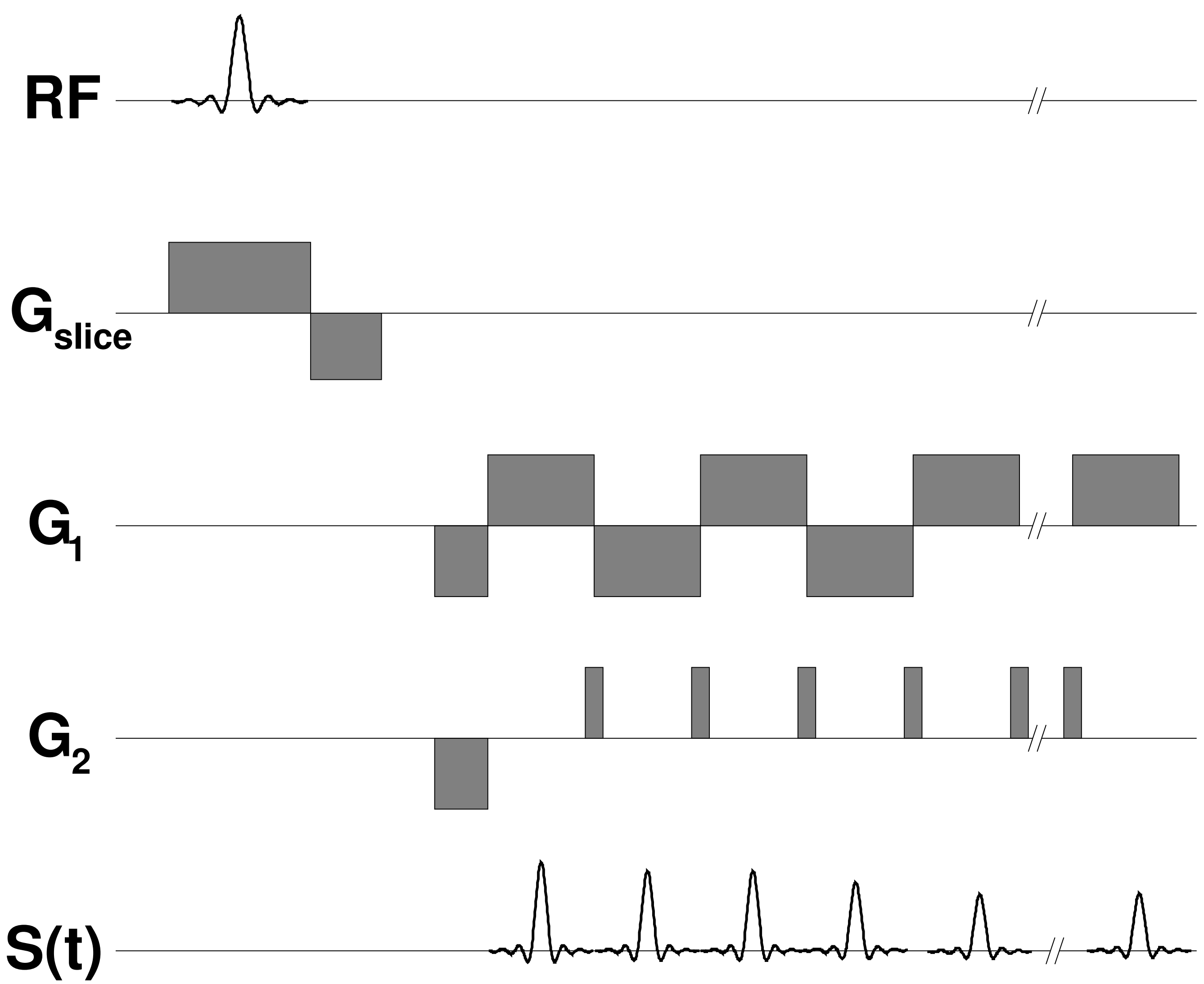
EPI

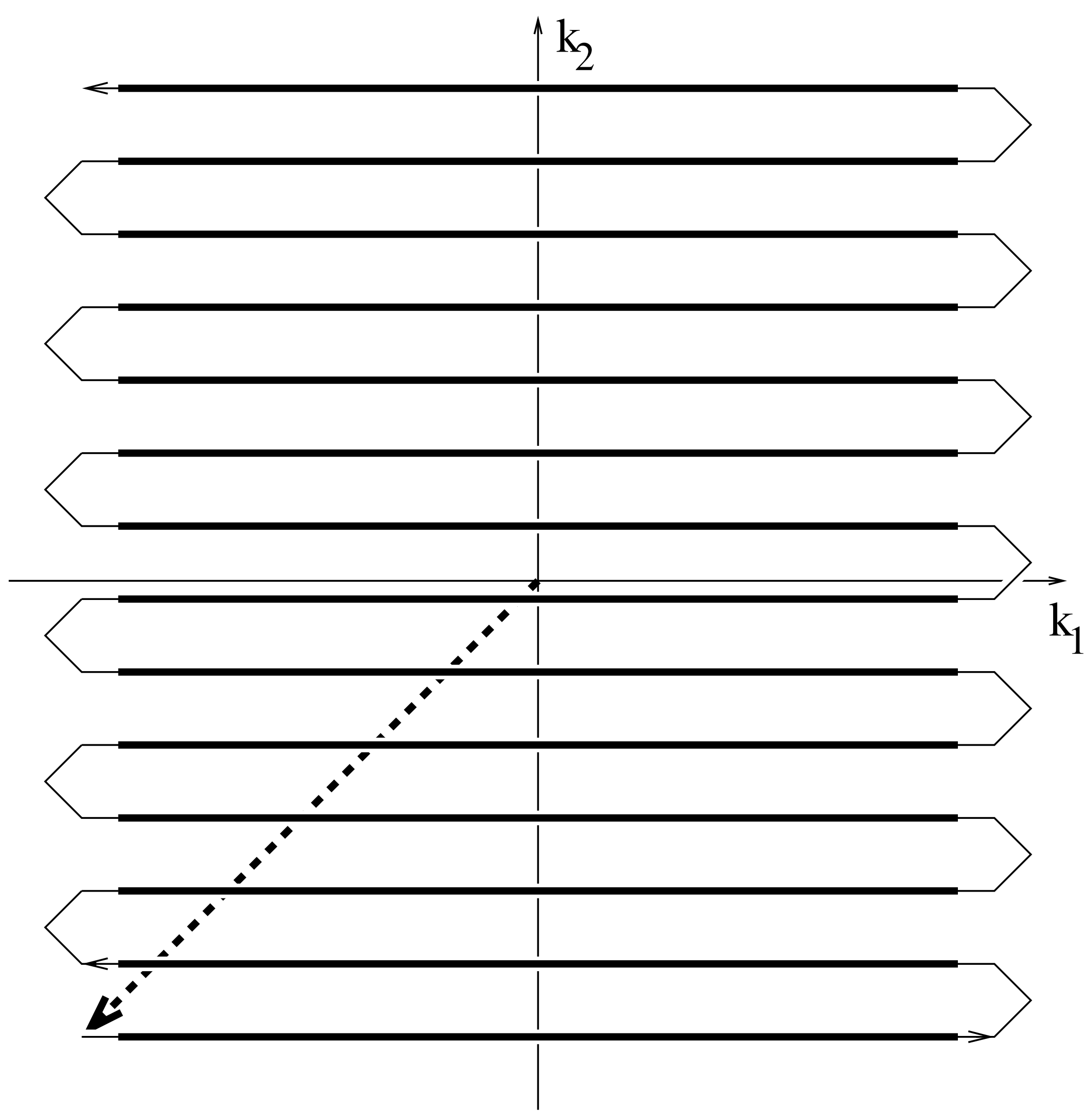
Echo planar Imaging

$$\mathbf{k}(t) = \frac{\gamma}{2\pi}\int_0^t \mathbf{G}(t')dt'$$

$$S(\mathbf{k}) = \int_V \rho(\mathbf{r})\exp(i2\pi\mathbf{k}\cdot\mathbf{r})\,d^3\mathbf{r}$$

- Acquire the whole 2D k-space after excitation
- Strong T2* weighting -- BOLD contrast





Safety

SAR and Gradient stimulation

- **Specific absorption rate** (**SAR**) is a measure of the rate at which energy is absorbed by the subject due to the RF (B1) field

- **Peripheral nerve stimulation** is caused rapidly changing magnetic fields (typically gradient) inducing electric fields in tissue causing stimulation of peripheral nerves.

Other stuff

Not covered in this talk

- Artifacts
- Motion monitoring/suppression
- Diffusion imaging
- Anything involving deeper NMR phenomena
- System engineering

Thanks for your attention

National Institute
of Mental Health
Functional MRI Facility