

Advanced MRI and fMRI Acquisition Methods

J. Andrew Derbyshire

*Functional MRI Facility
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

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

$$\omega = \gamma B$$

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

NMR: Classical view

NMR: Nuclear Magnetic Resonance

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

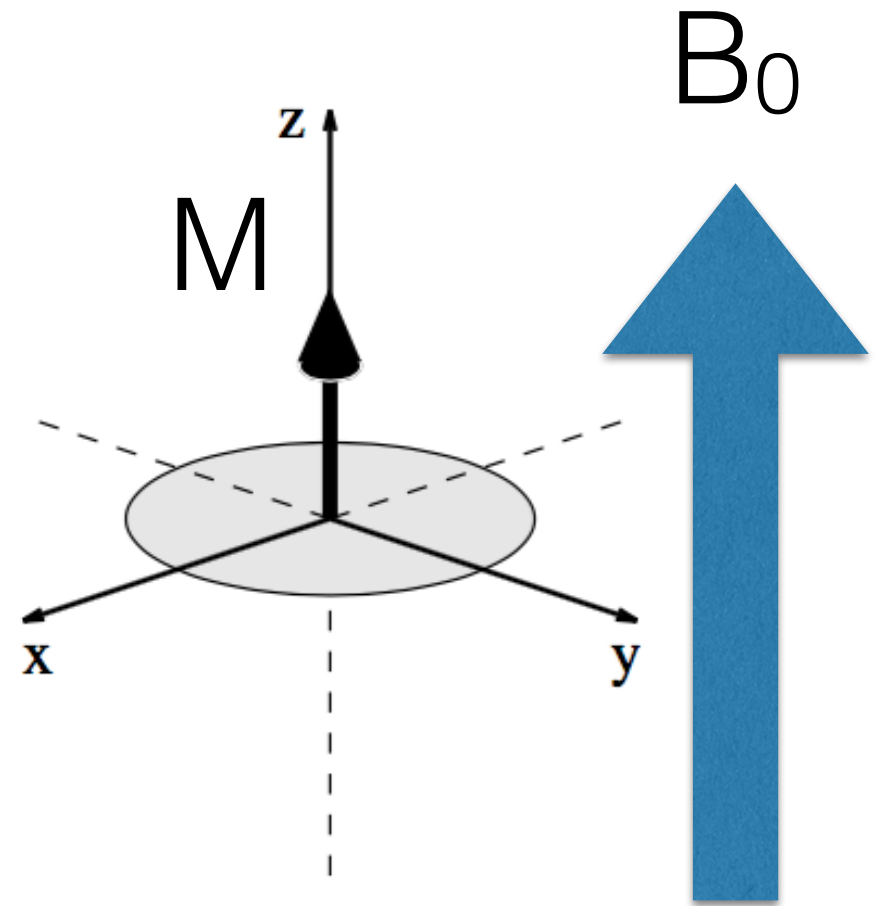
$$\mathbf{M} = (M_x, M_y, M_z)^T$$

- 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: 63MHz
3.0T: 127MHz
7.0T: 298MHz

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

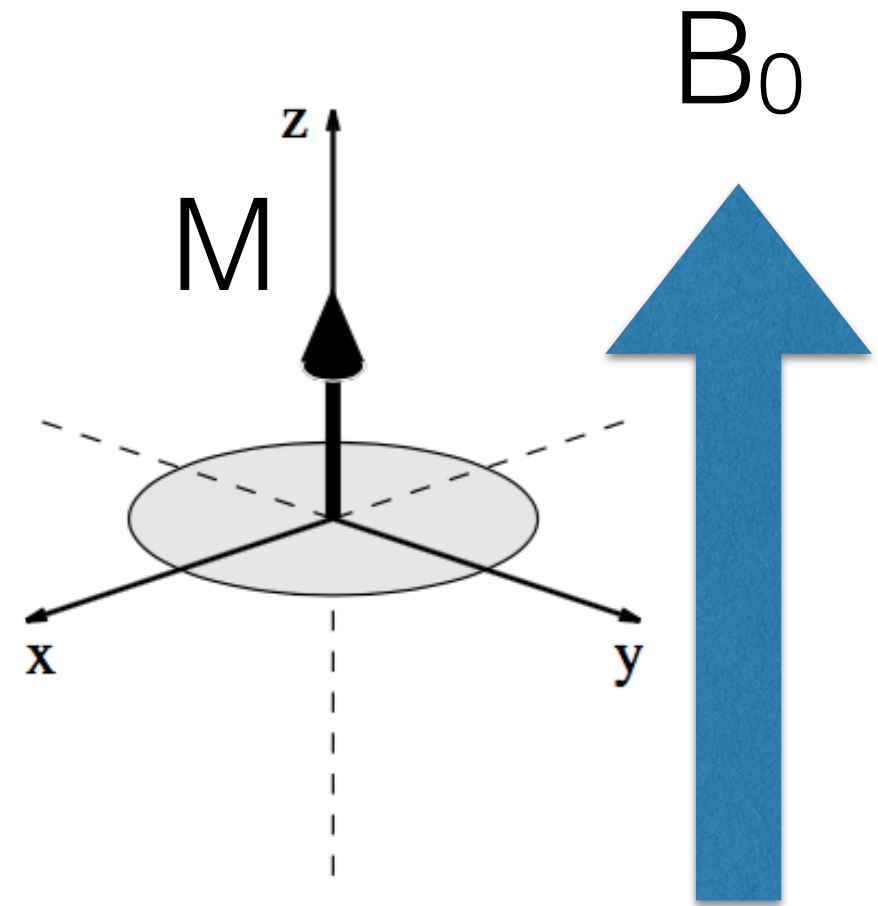
$$\mathbf{M} = (M_x, M_y, M_z)^T$$

- At equilibrium:

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

where

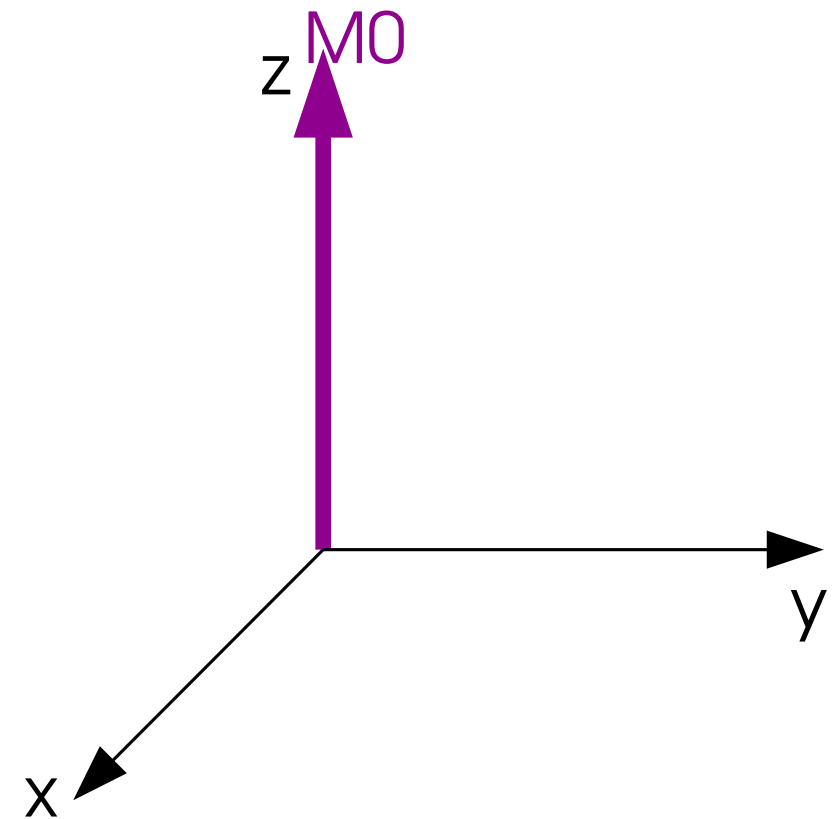
$$\frac{M_0 = N\gamma\hbar^2 I_z(I_z + 1)B_0}{3kT}$$



NMR: Classical view

Excitation, Precession and the Rotating Frame

- In rotating frame at equilibrium

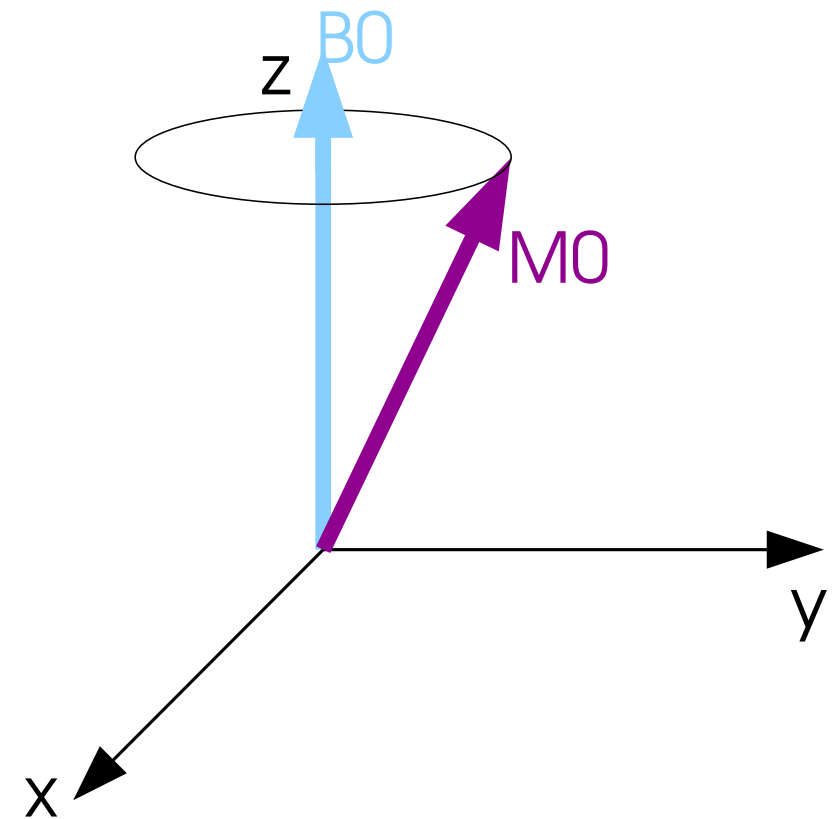


NMR: Classical view

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

$$\omega = \gamma B$$



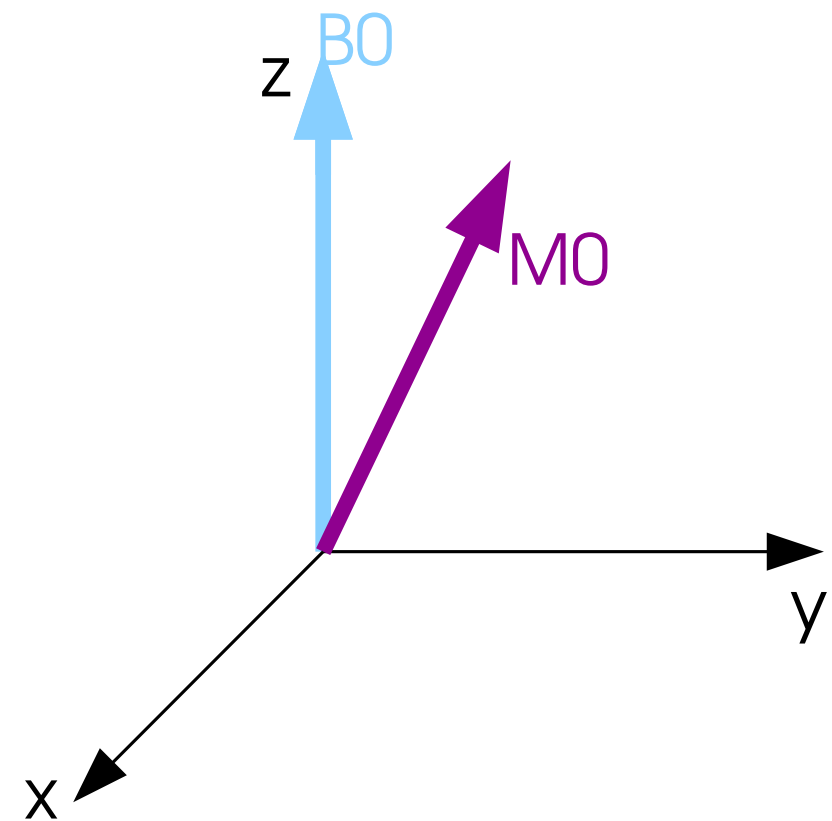
NMR: Classical view

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

$$\omega = \gamma B$$

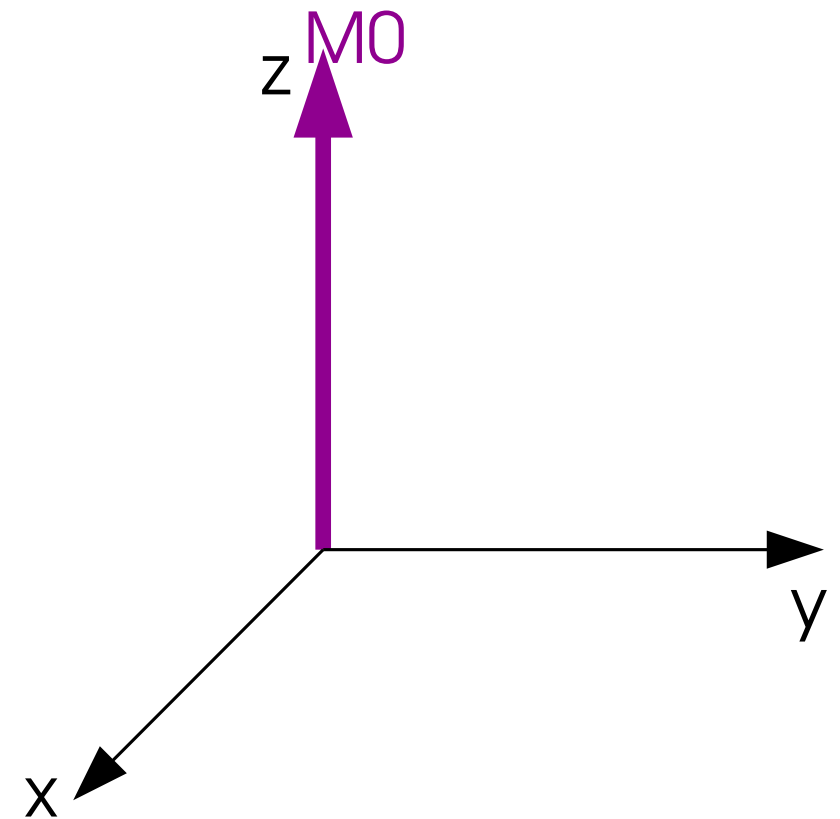
- It is convenient to work in a frame of reference rotating at $\omega = \gamma B$



NMR: Classical view

Excitation, Precession and the Rotating Frame

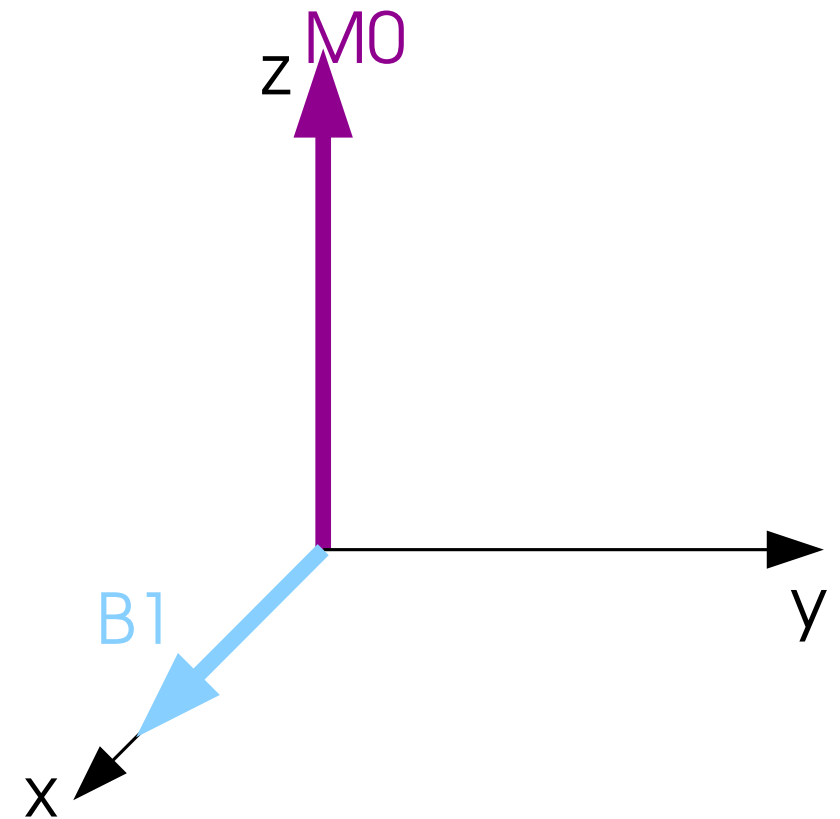
- In rotating frame at equilibrium



NMR: Classical view

Excitation, Precession and the Rotating Frame

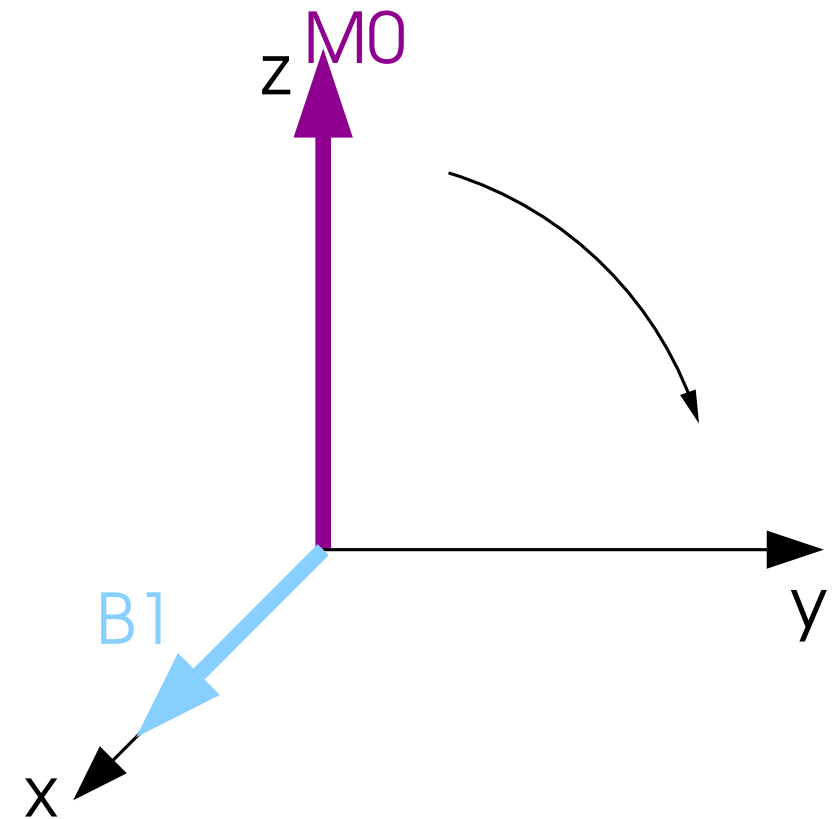
- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating frame) x-axis



NMR: Classical view

Excitation, Precession and the Rotating Frame

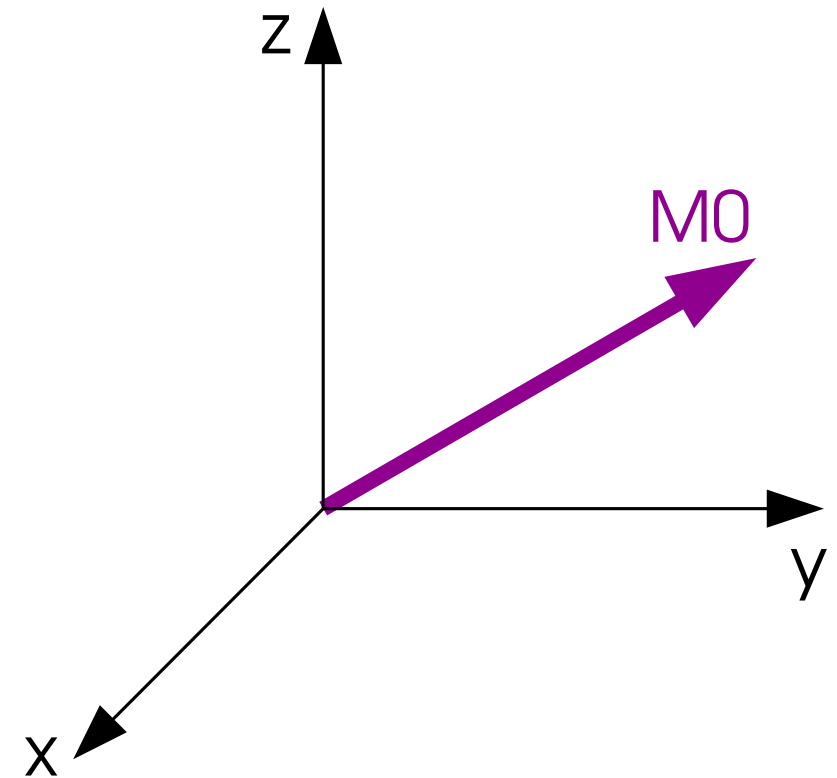
- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating-frame) x-axis
- $\omega_1 = \gamma B_1$
- Magnetization rotates towards (rotating-frame) y-axis



NMR: Classical view

Excitation, Precession and the Rotating Frame

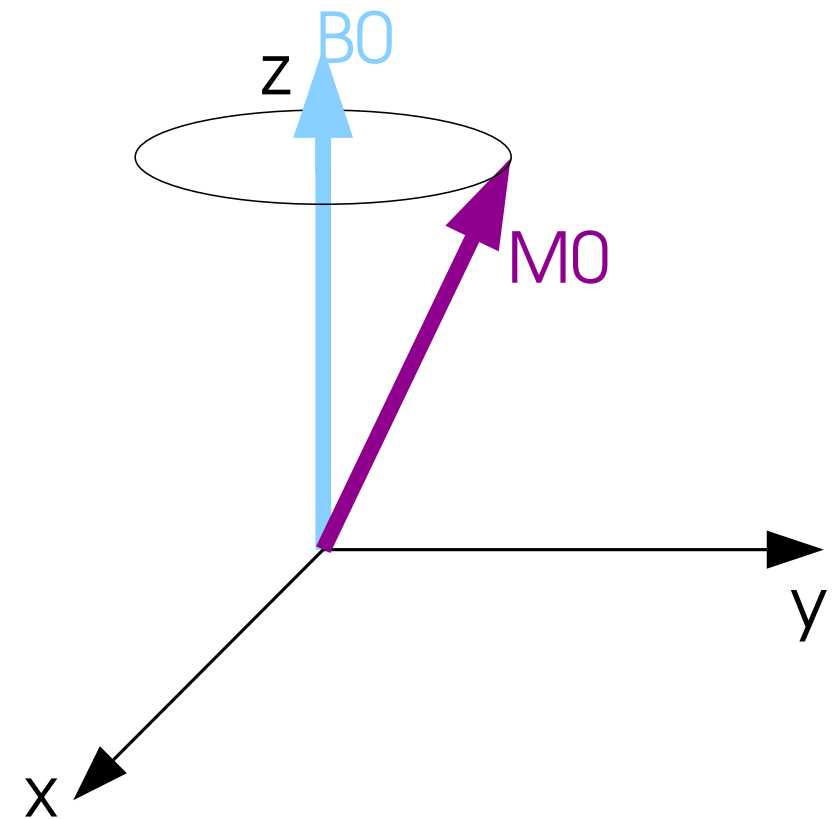
- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating-frame) x-axis
- $\omega_1 = \gamma B_1$
- Magnetization rotates towards (rotating-frame) y-axis
- Turn off B_1 field when magnetization reaches the appropriate **flip angle** with respect to the z-axis



NMR: Classical view

Excitation, Precession and the Rotating Frame

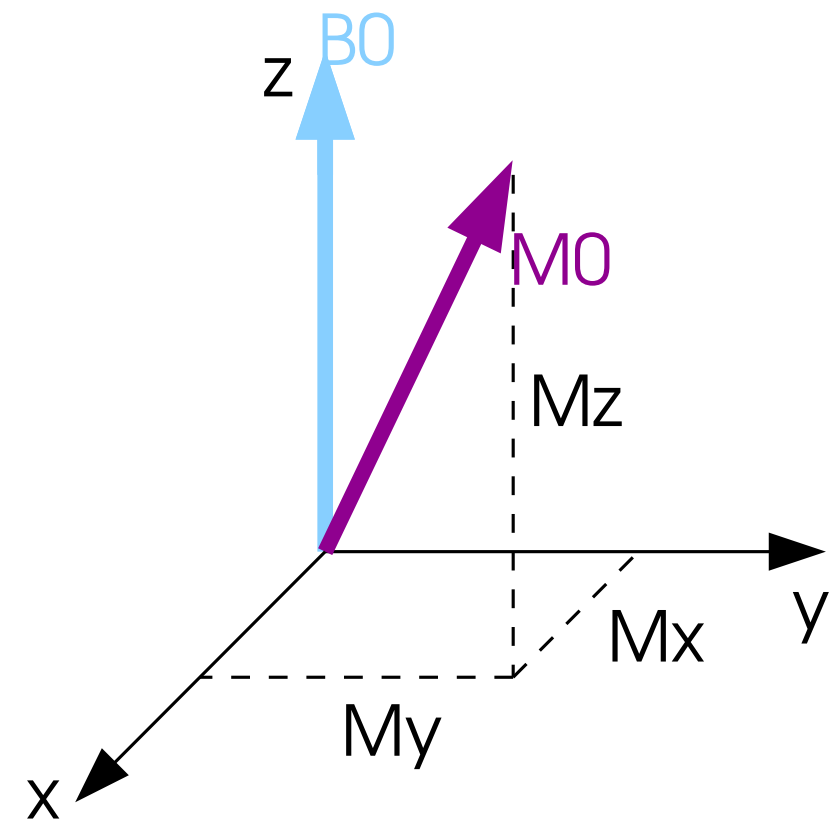
- In rotating frame at equilibrium
- Apply B_1 magnetic field along (rotating-frame) x-axis
- $\omega_1 = \gamma B_1$
- Magnetization rotates towards (rotating-frame) y-axis
- Turn off B_1 field when magnetization reaches the appropriate **flip angle** with respect to the z-axis
- Magnetization precesses and relaxes back to equilibrium



NMR: Classical view

MR signal

- $\mathbf{M} = (M_x, M_y, M_z)^T$
- M_z is the longitudinal component
- M_x, M_y are transverse components



NMR: Classical view

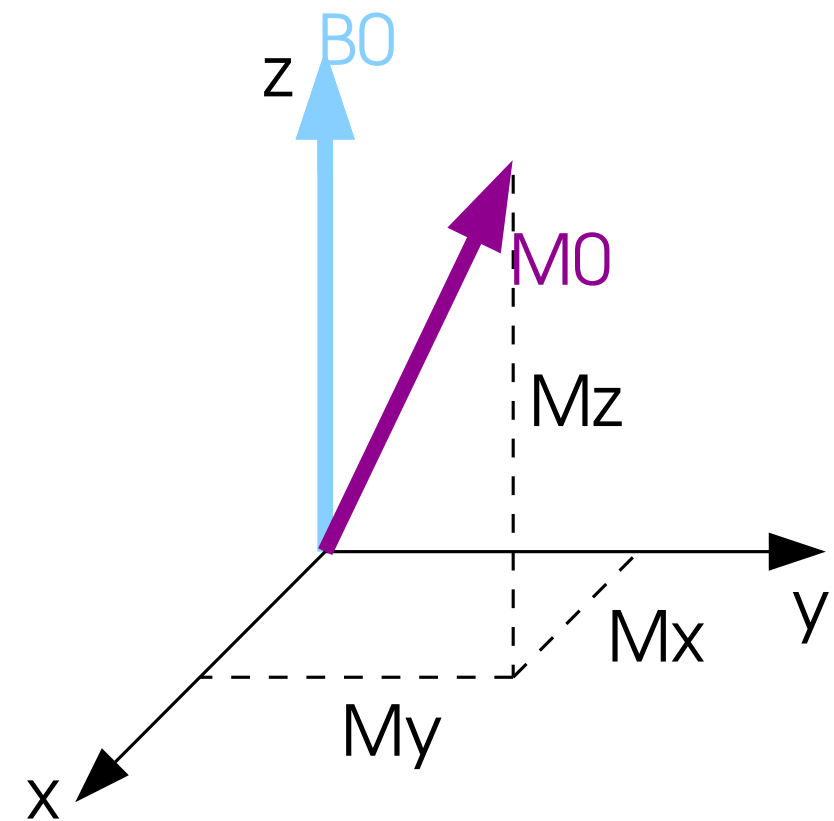
MR signal

- $\mathbf{M} = (M_x, M_y, M_z)^T$
- 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 **$\mathbf{M}$**
- After excitation M_z relaxes back to M_0 by T_1 relaxation

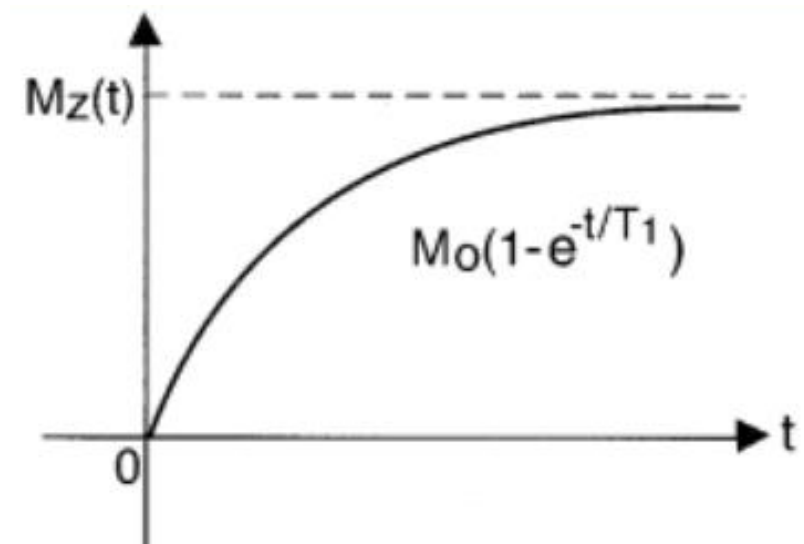
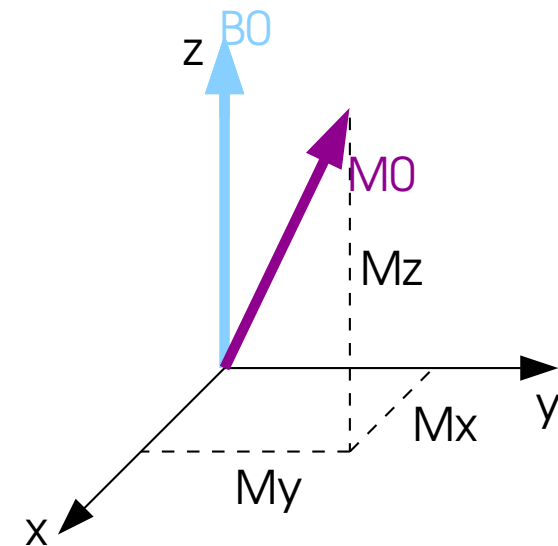
$$\frac{dM_z}{dt} = \frac{(M_0 - M_z)}{T_1}$$

- So that:

$$M_z(t) = M_0 + (M_z(0) - M_0)e^{-t/T_1}$$

- or, equivalently

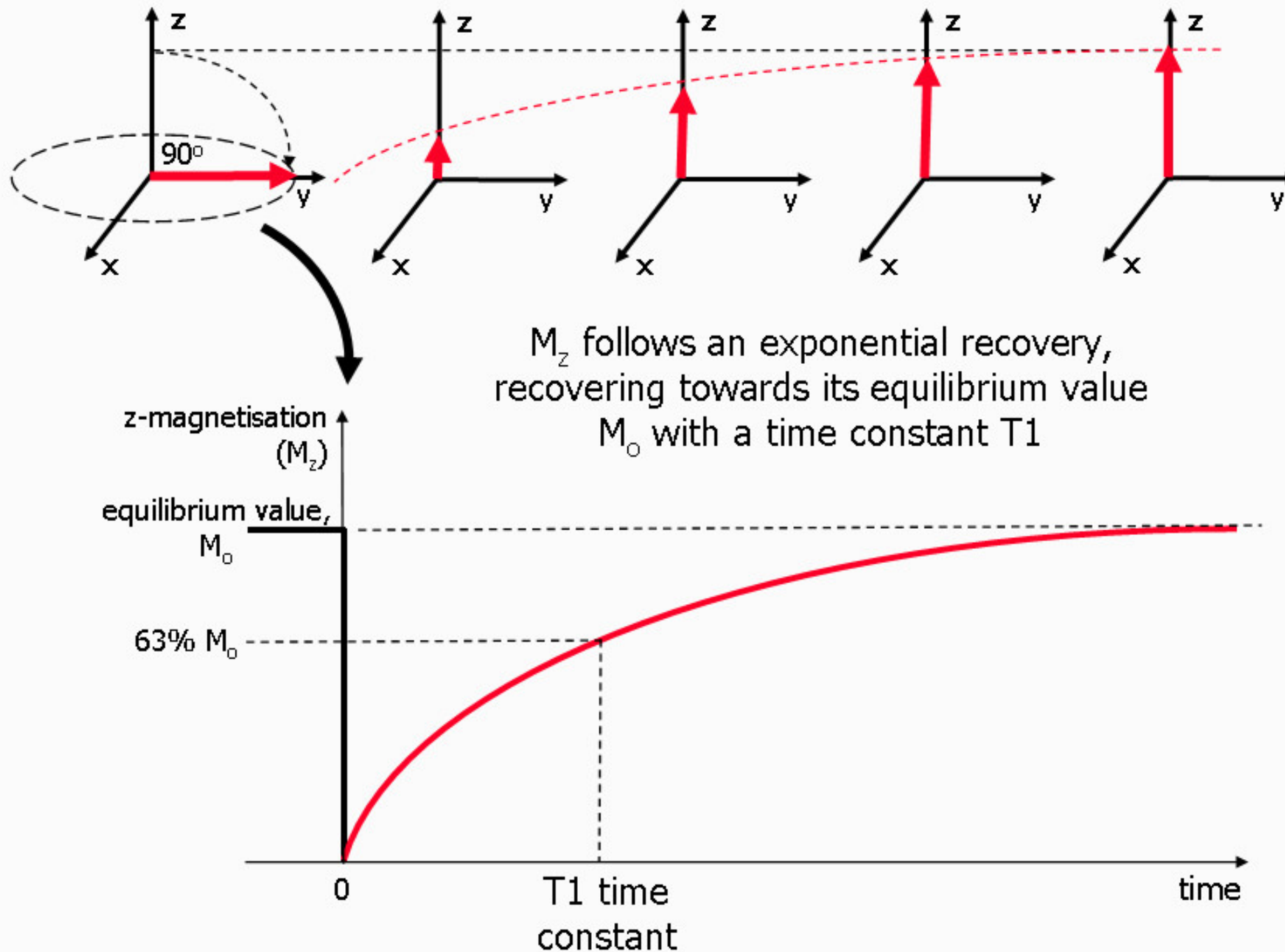
$$M_z(t) = M_z(0)e^{-t/T_1} + M_0(1 - e^{-t/T_1})$$



NMR: Classical view

MR relaxation

$$M_z(t) = M_0 + (M_z(0) - M_0)e^{-t/T_1}$$



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

NMR: Classical view

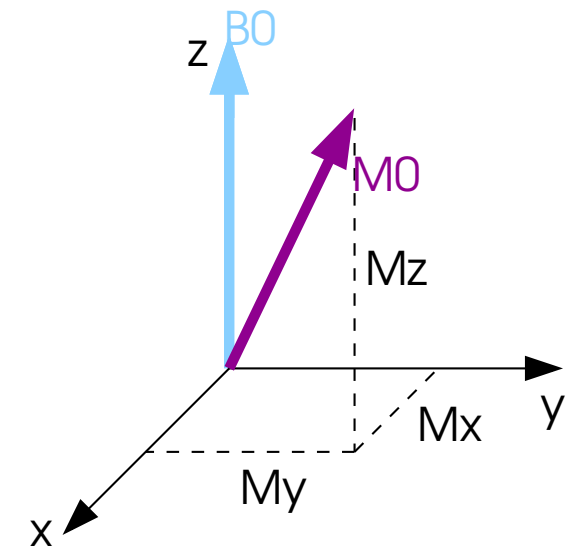
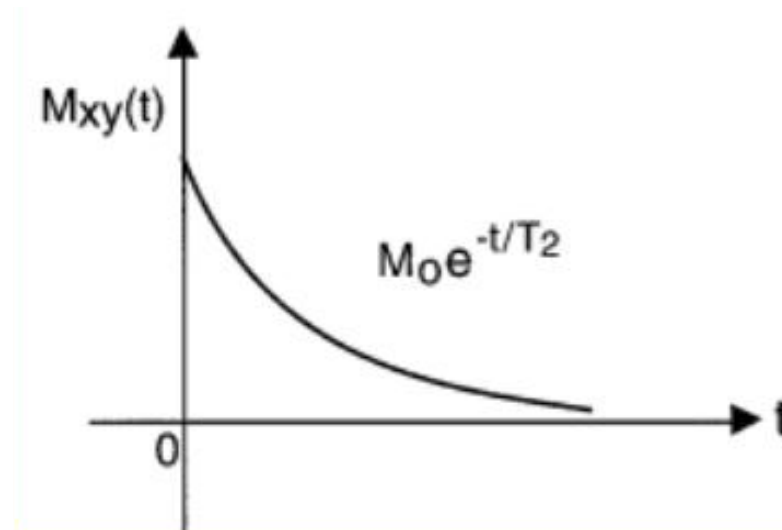
MR relaxation

- 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:

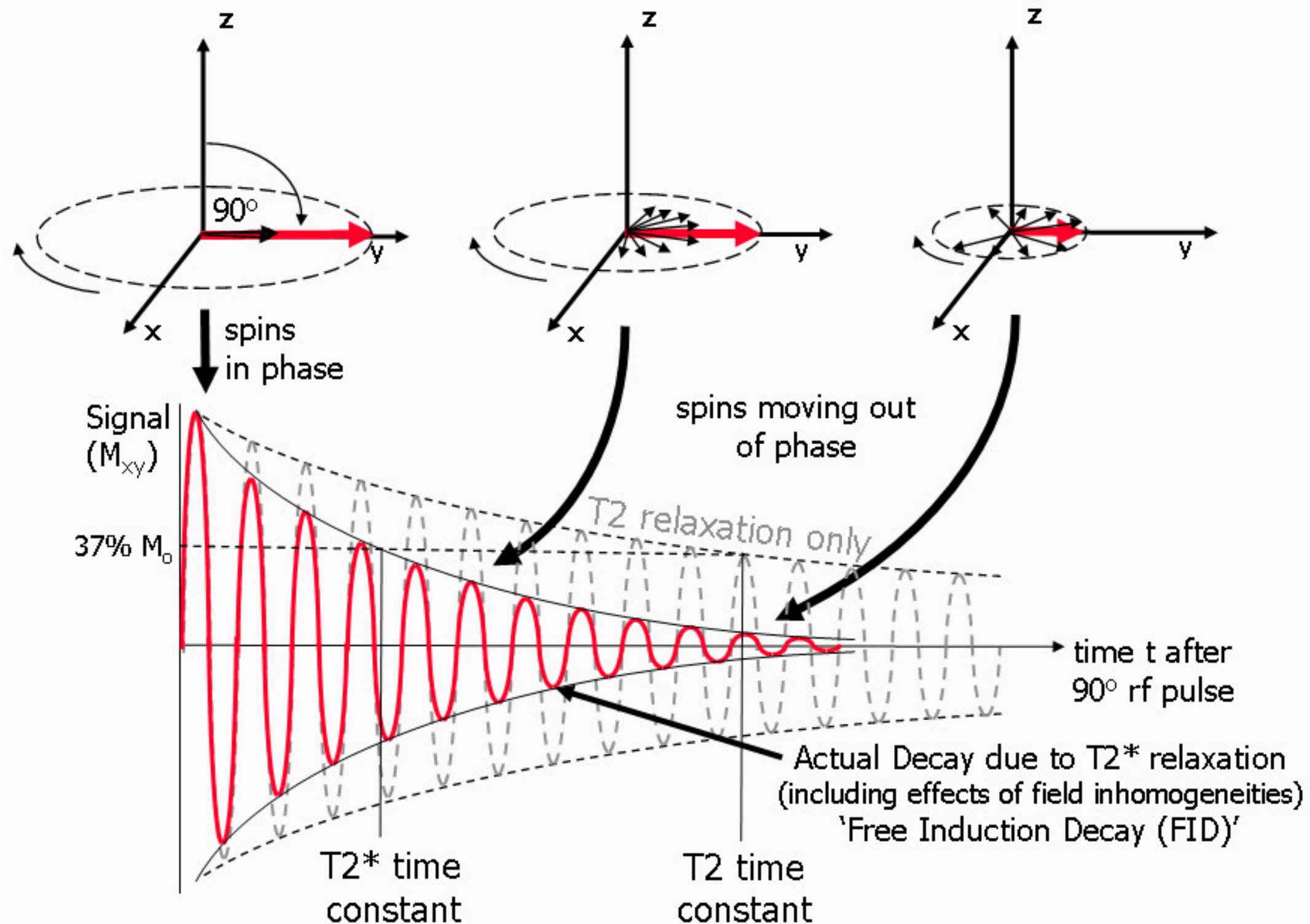
$$M_{xy}(t) = M_{xy}(0)e^{-t/T_2}$$



- Note that $T_2 \leq T_1$ so that the MR signal generally dies faster than M_z regrows.

NMR: Classical view

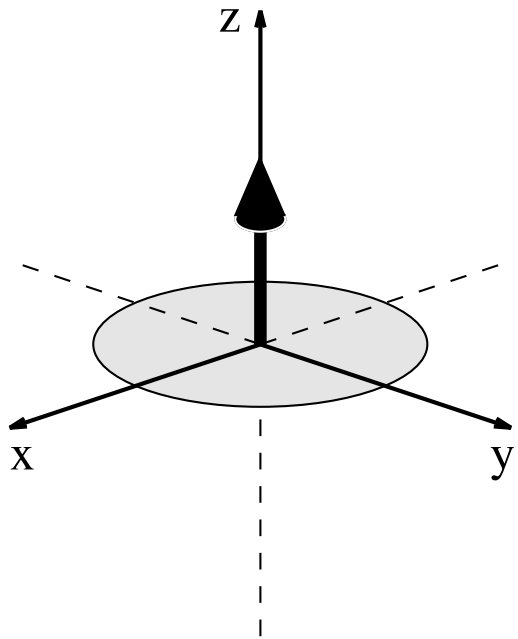
Intra voxel dephasing



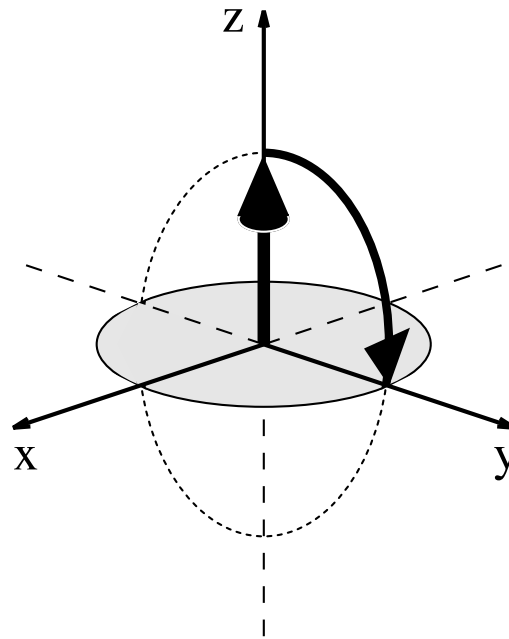
NMR: Classical view

Spin-echo

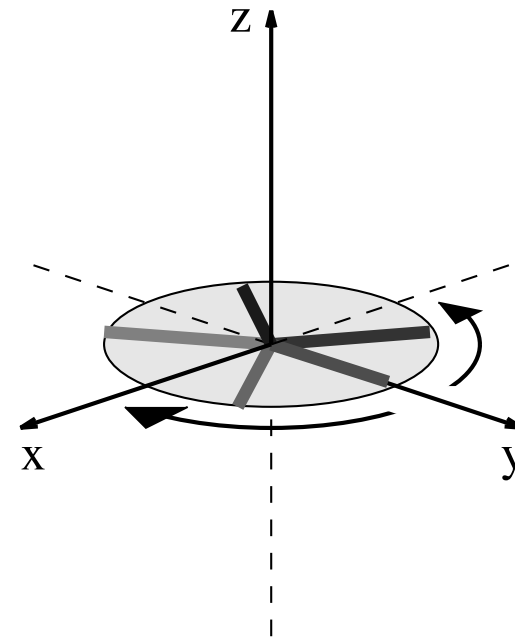
(a) equilibrium



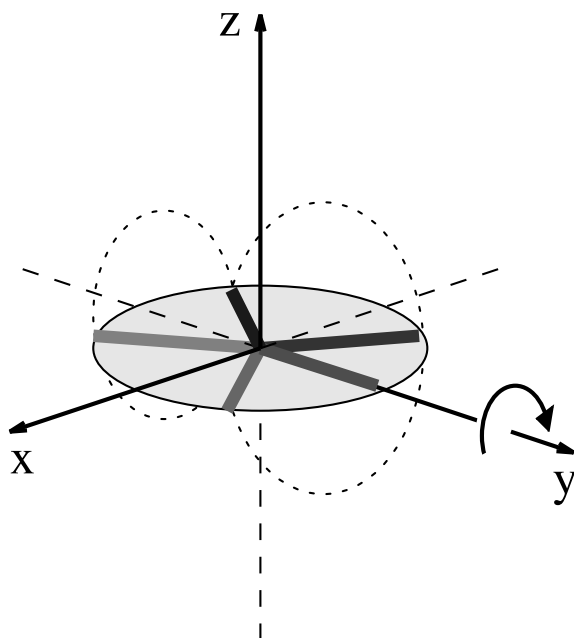
(b) 90 degree pulse



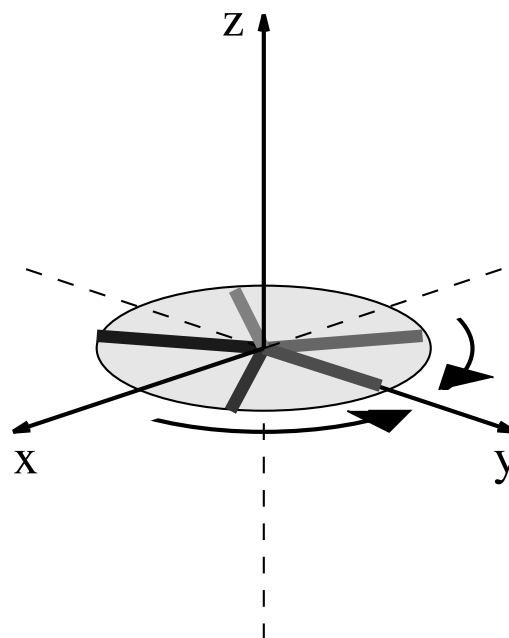
(c) dephasing



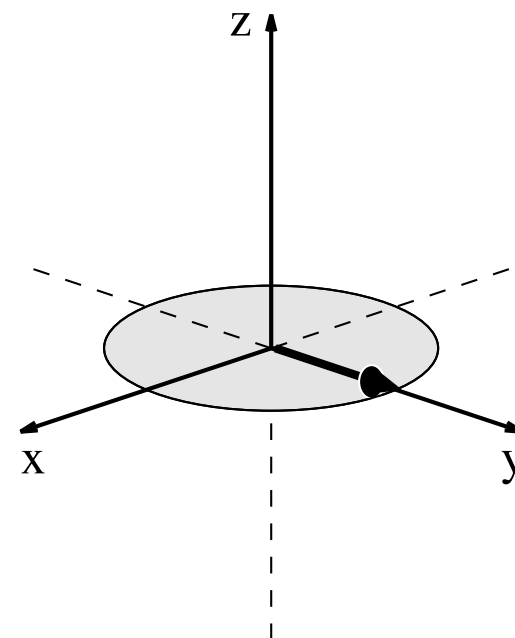
(d) 180 degree pulse



(e) rephasing



(f) 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

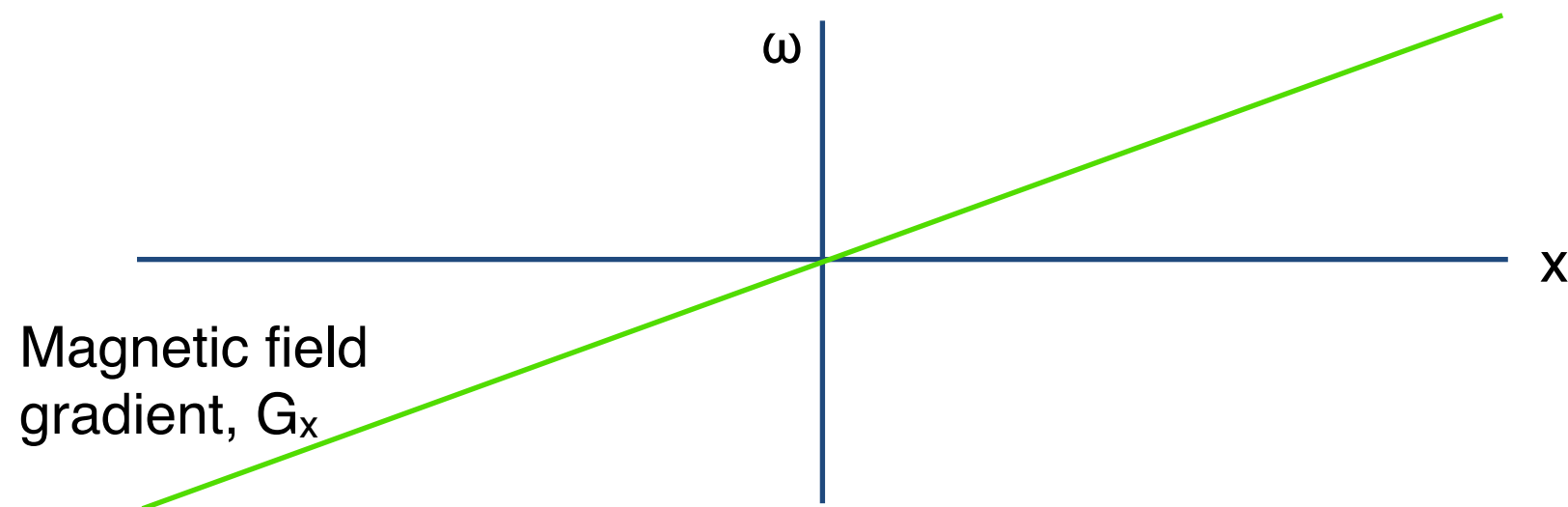
Gradients

- 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:

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



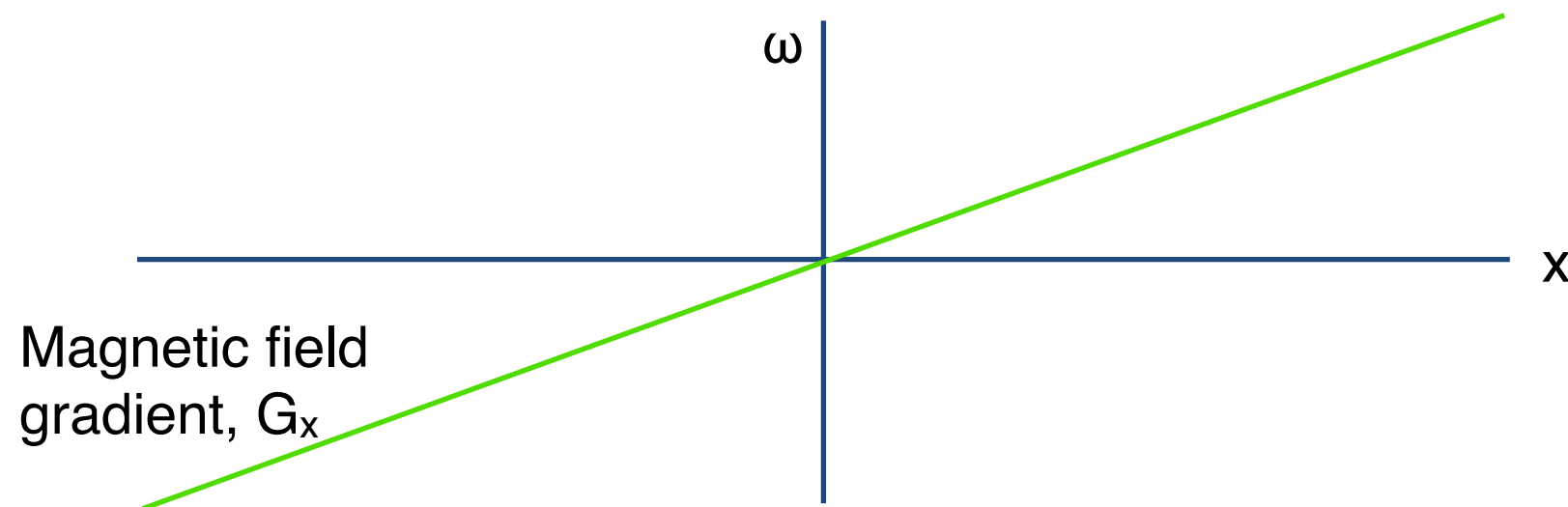
Gradients

- 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:

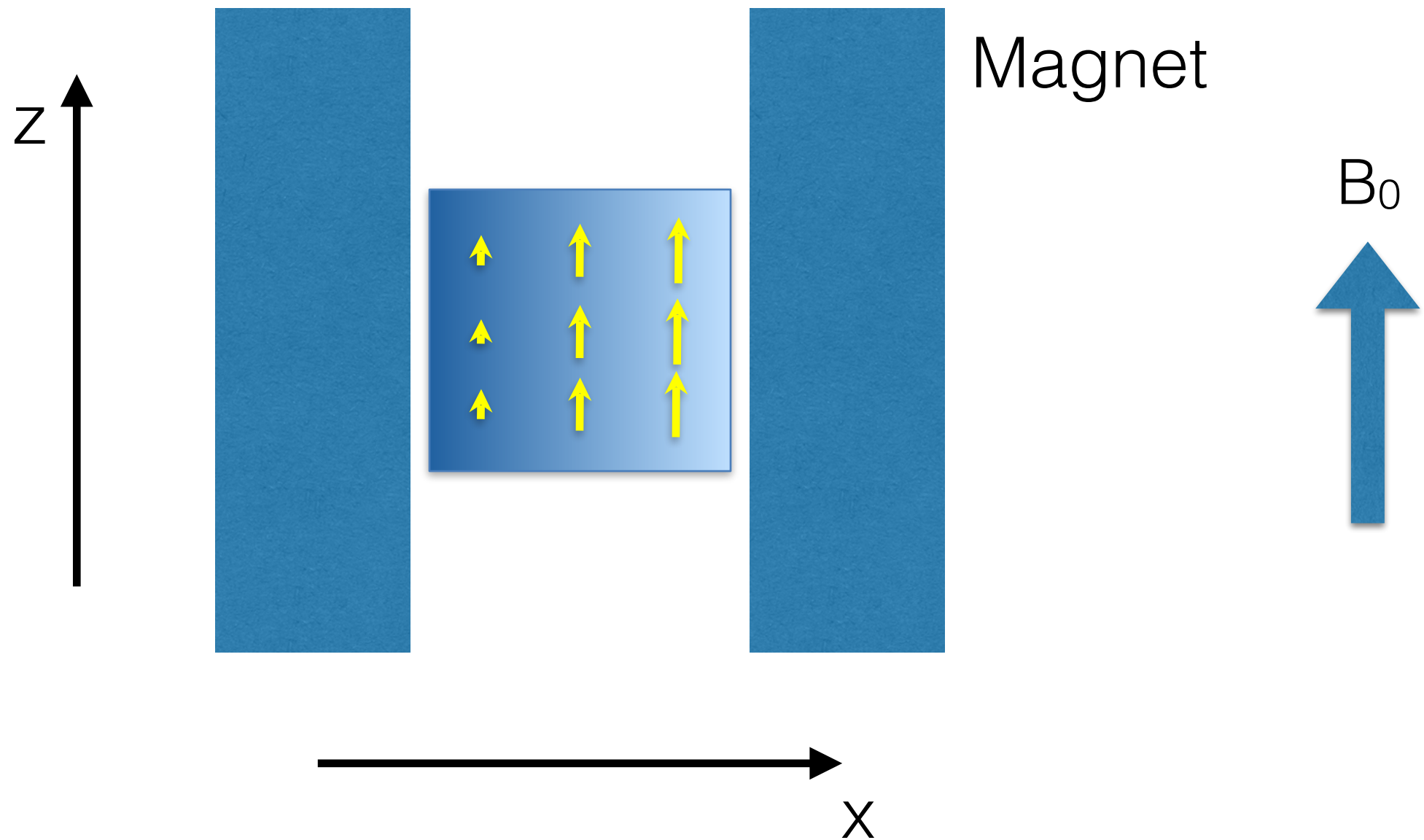
$$\omega = \gamma(G_x x + G_y y + G_z z) = \mathbf{G} \cdot \mathbf{r}$$



- Typical gradient fields are $G = 30\text{mT/m}$.
i.e. $\pm 3\text{mT}$ at 10cm from isocenter.
- 1000 times smaller than B_0

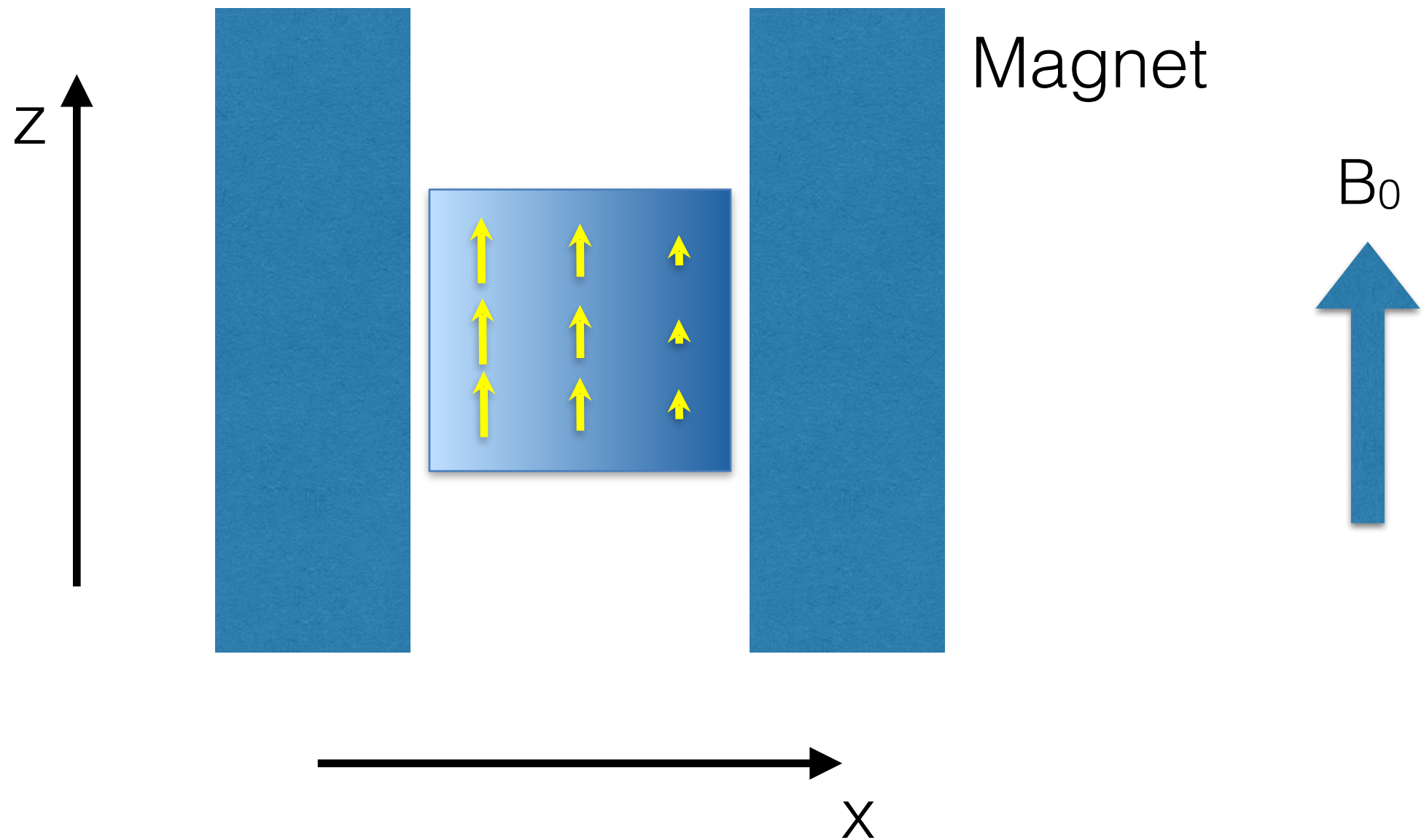
Gradients

- Gradient in +X



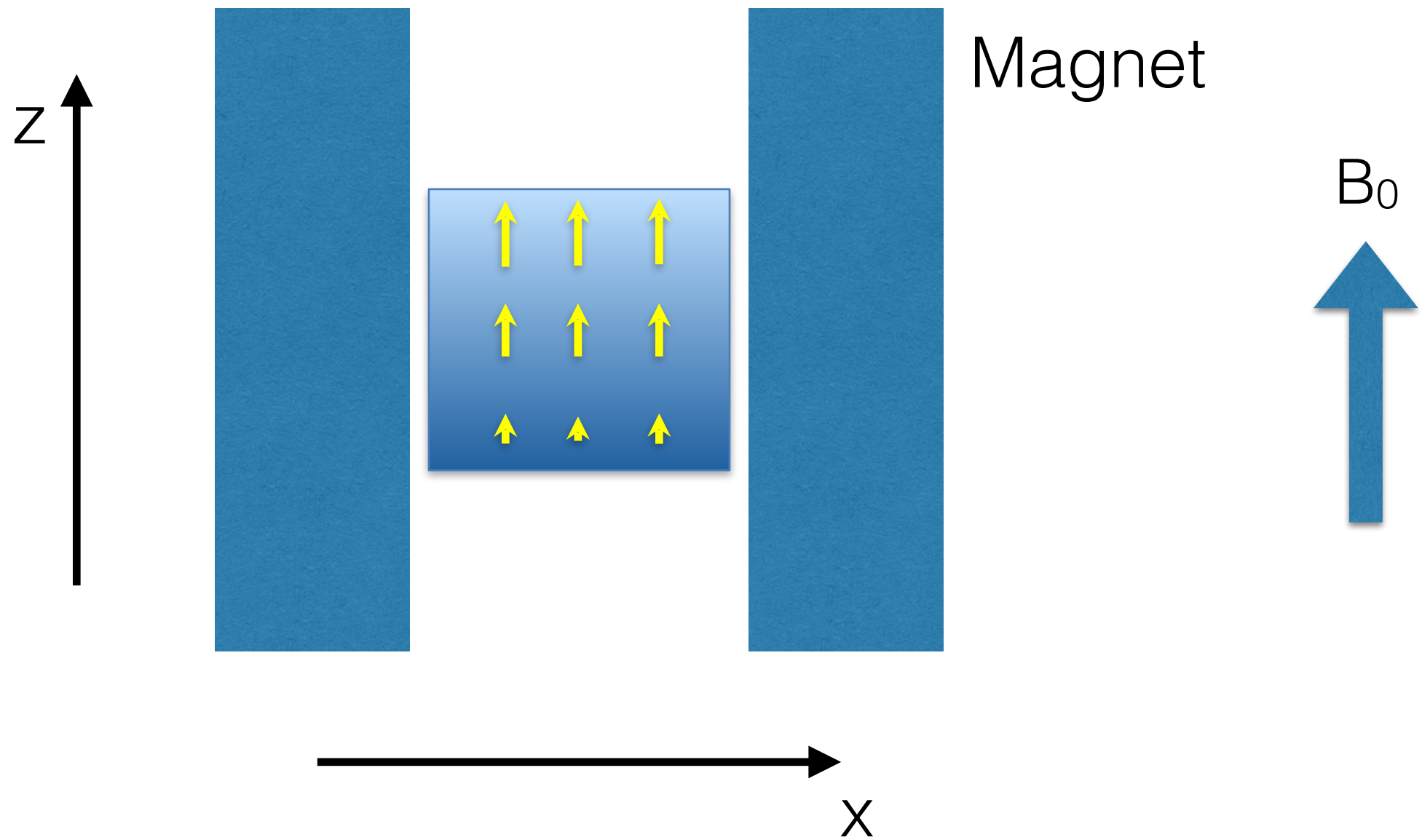
Gradients

- Gradient in $-X$



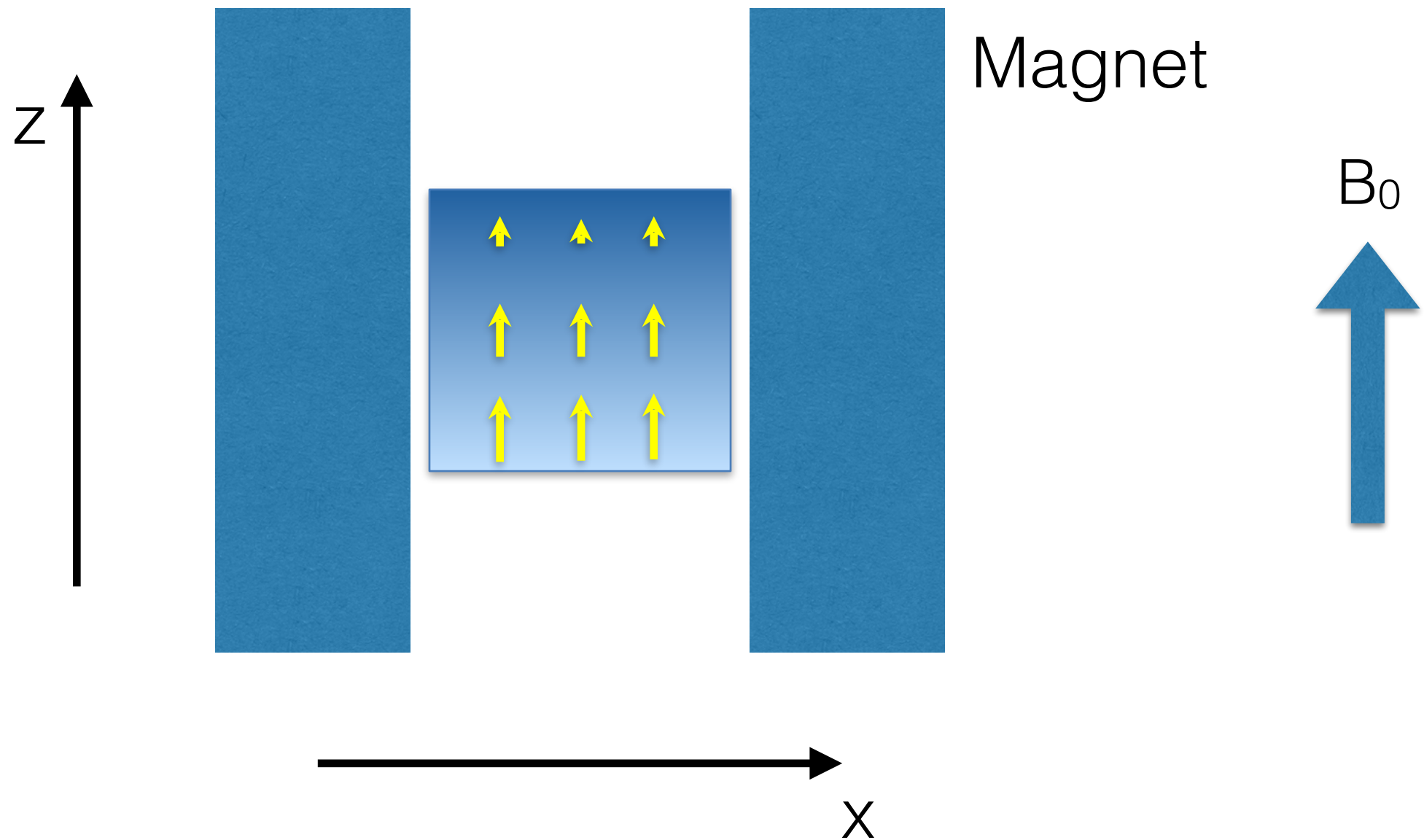
Gradients

- Gradient in +Z



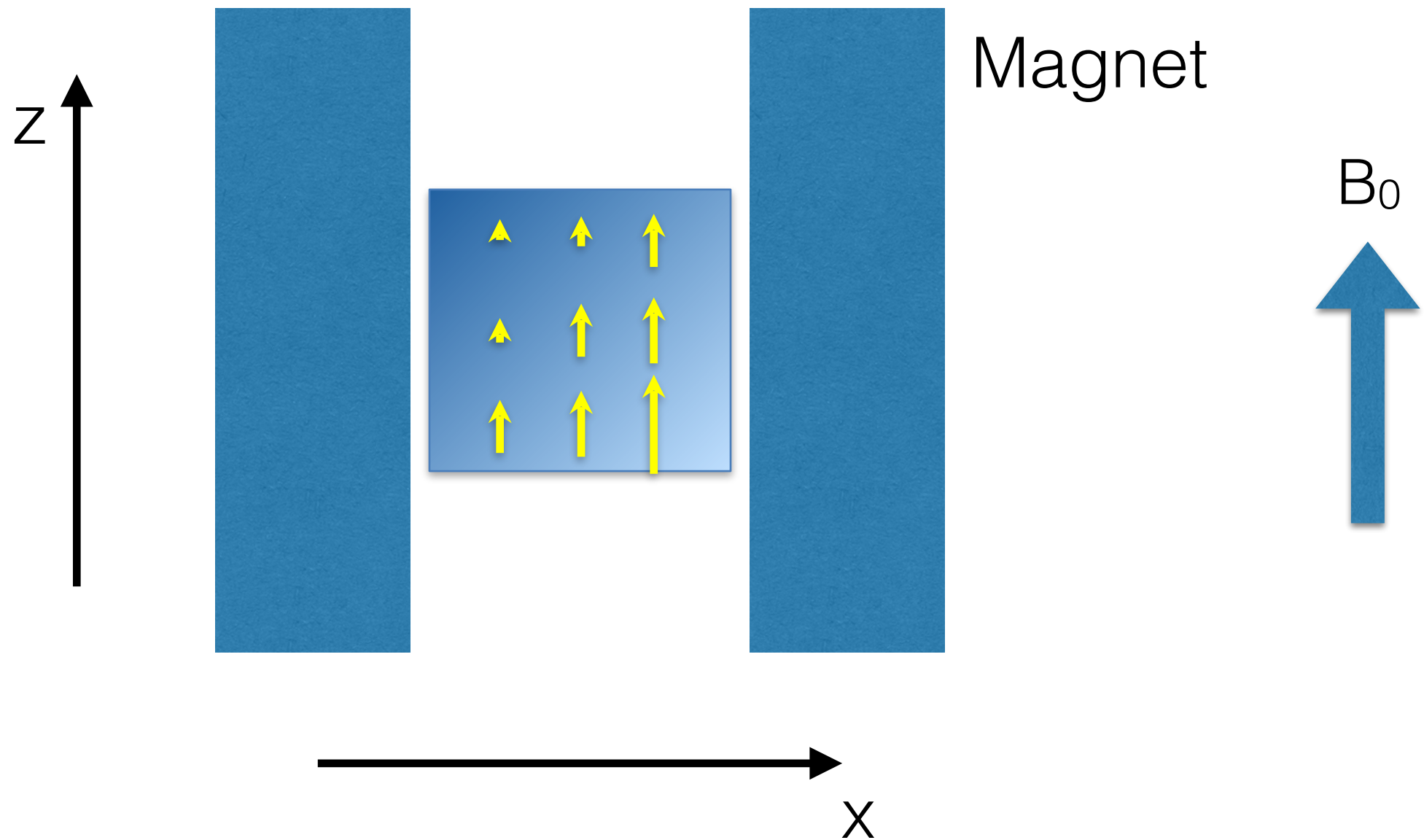
Gradients

- Gradient in -Z



Gradients

- Gradients in both X and -Z

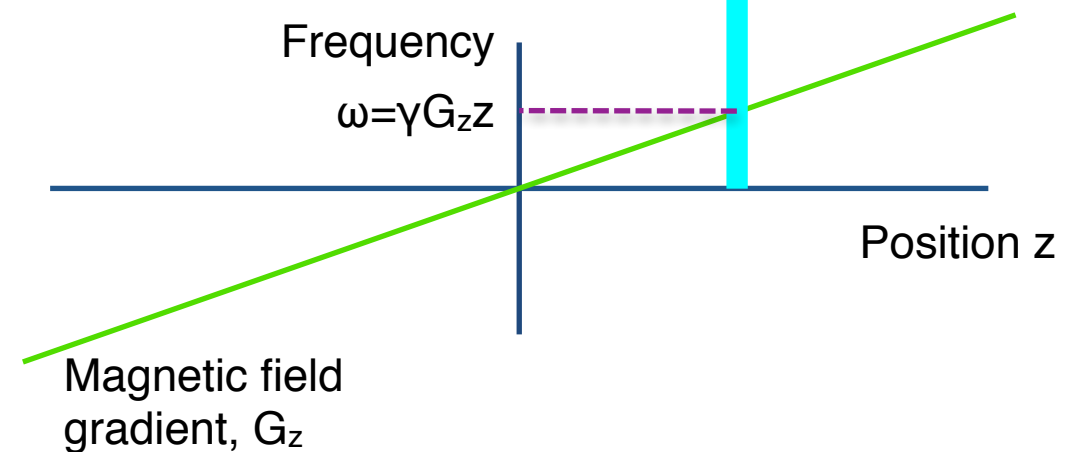
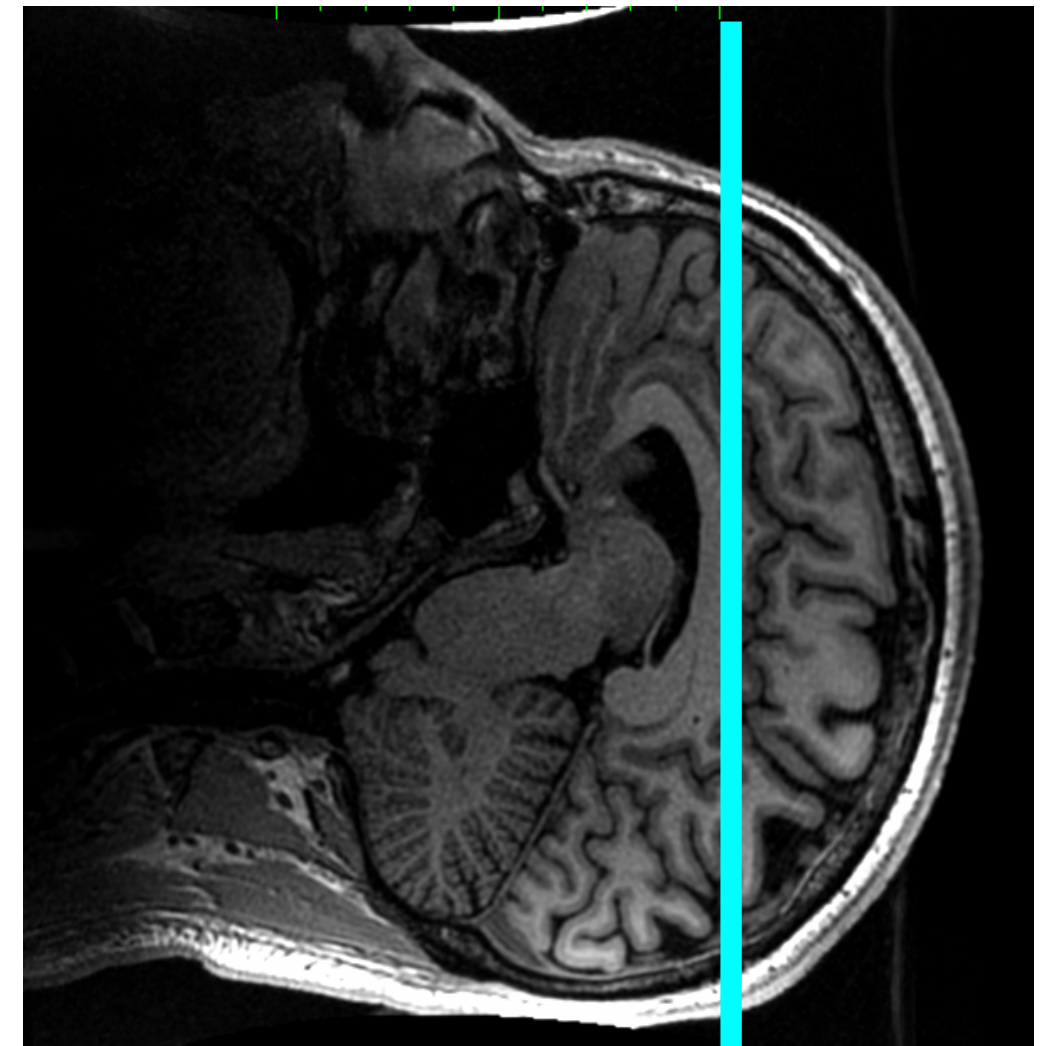


Slice selection

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

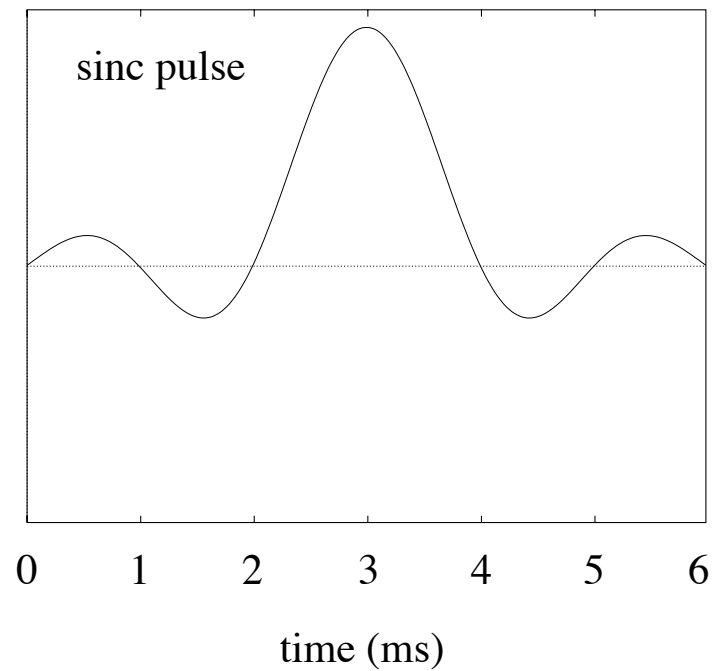
$$\delta\omega = \gamma G_z z$$

- 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:

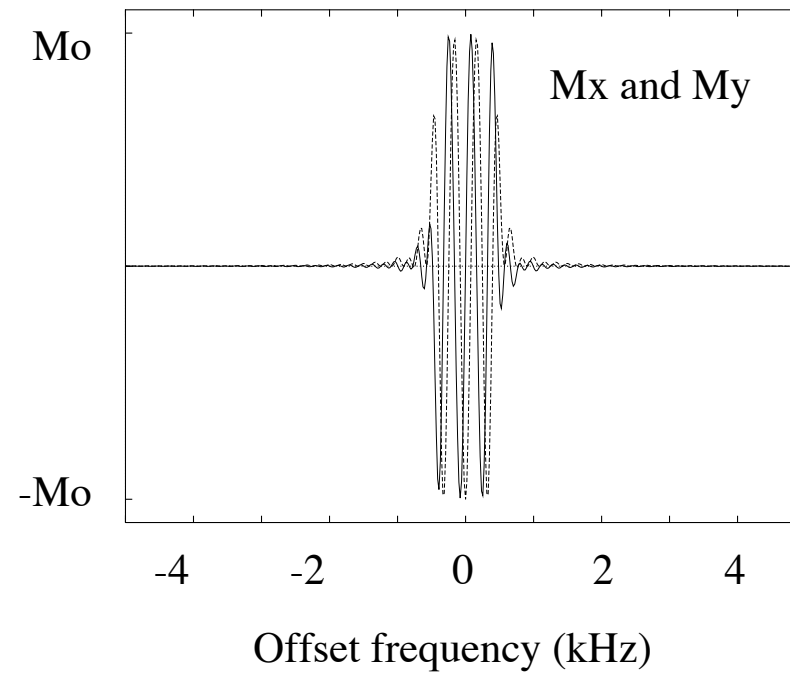


Selective excitation

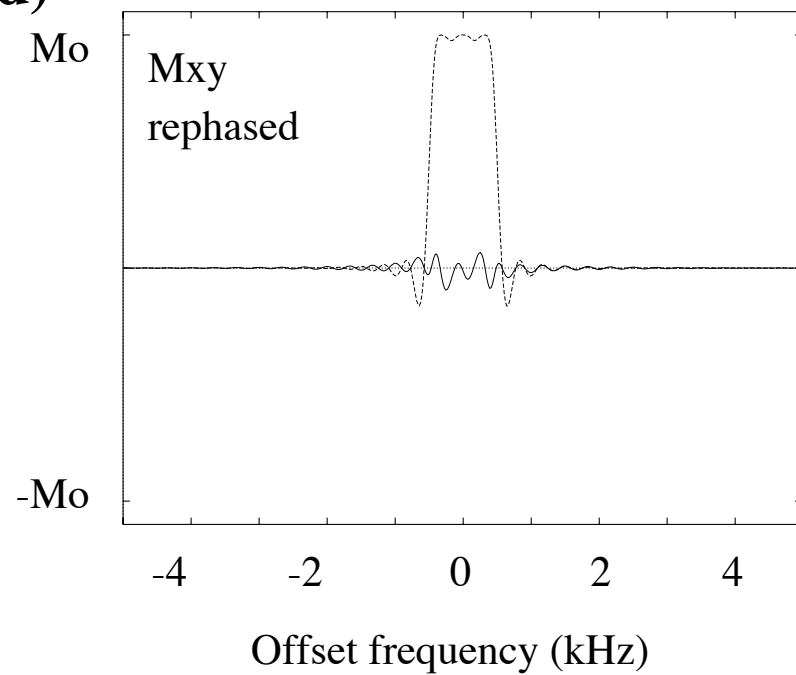
(a)



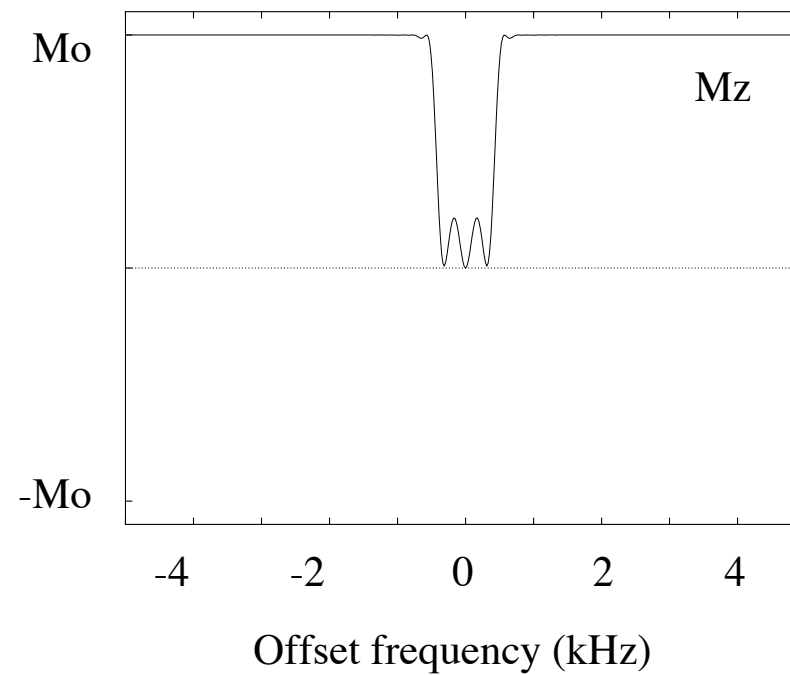
(b)



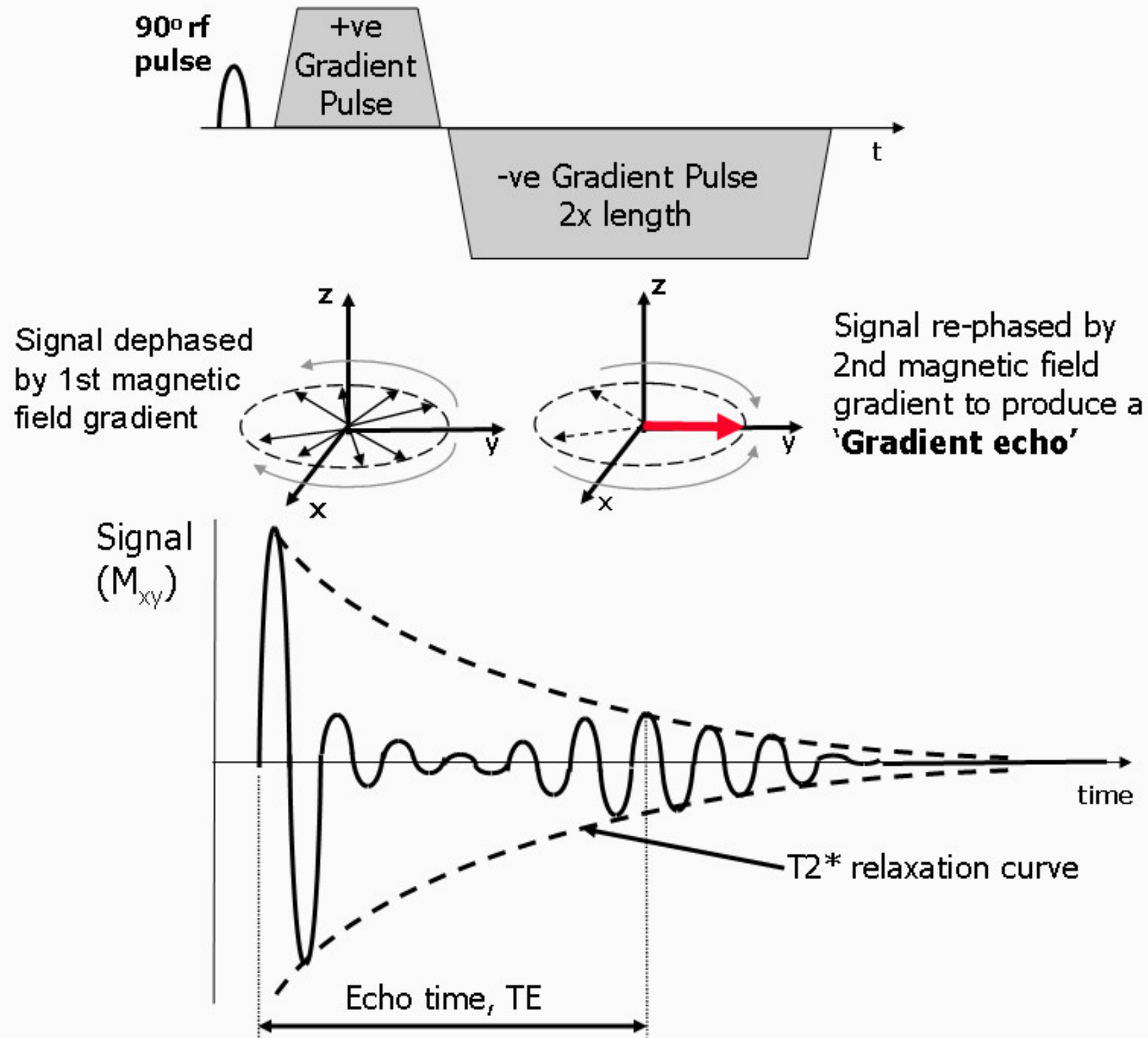
(d)



(c)



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

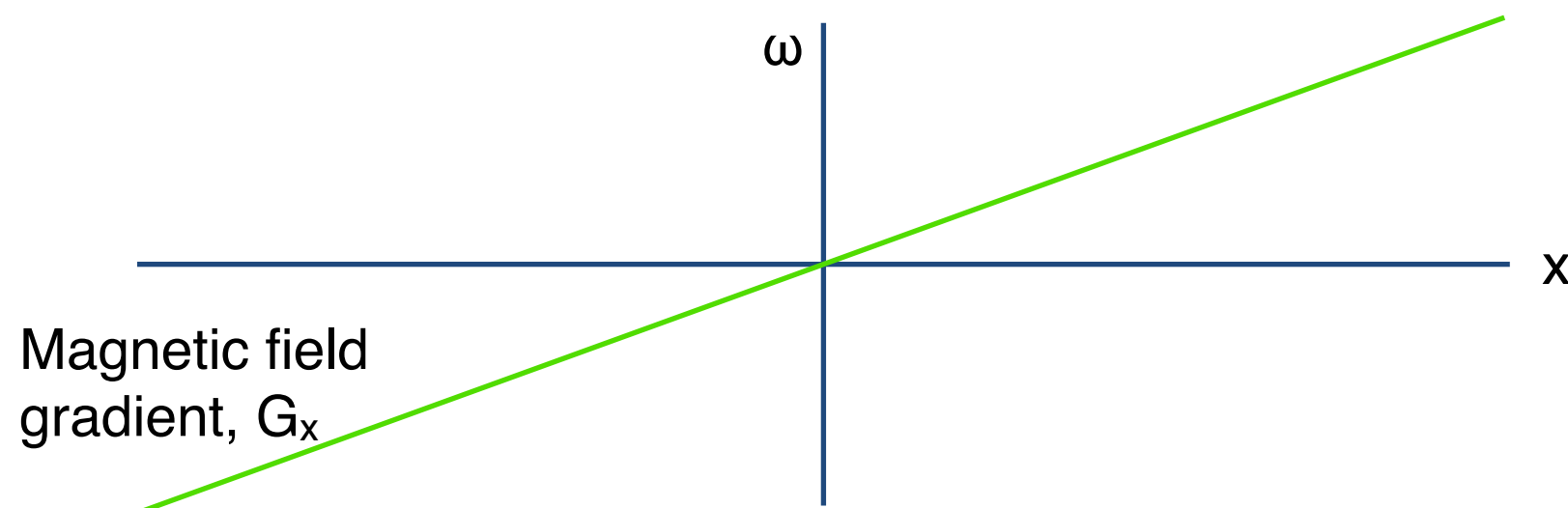
Effect of imaging gradient

- 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,

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

then:

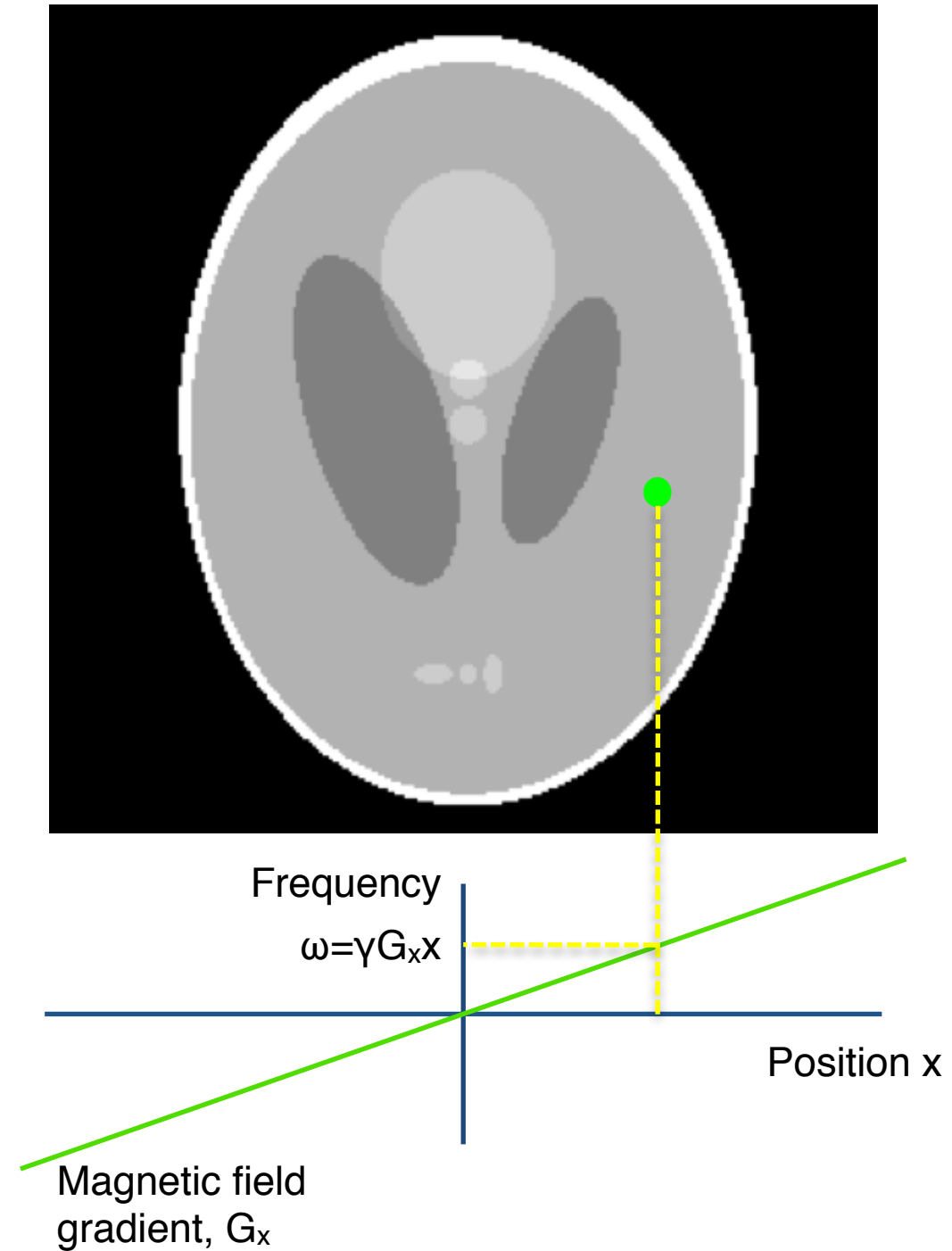
$$\omega = \gamma(G_x x + G_y y + G_z z) = \mathbf{G} \cdot \mathbf{r}$$



MR Imaging Theory

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

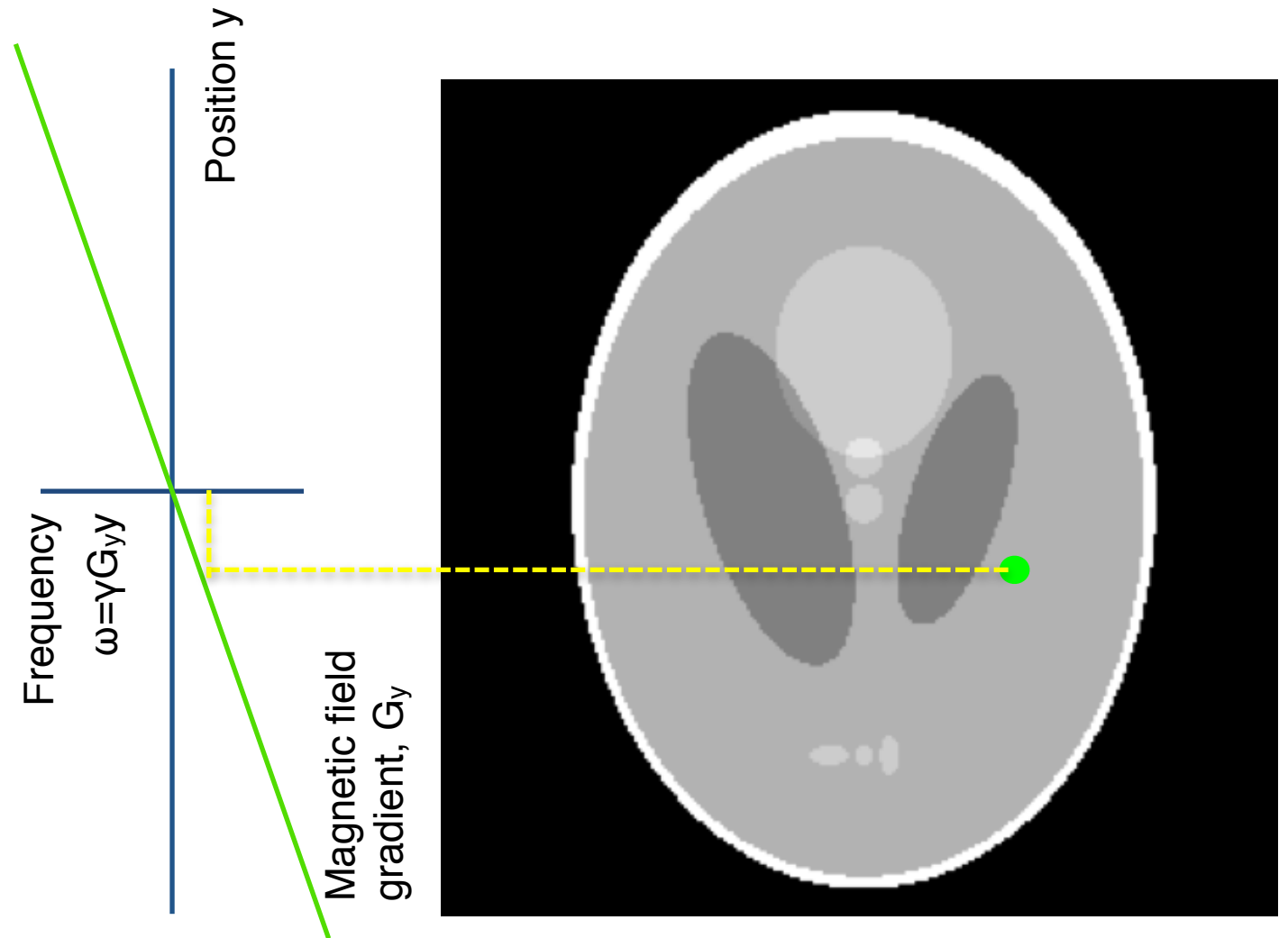
$$\delta\omega = \gamma \mathbf{G}(t) \cdot \mathbf{r}(t)$$



MR Imaging Theory

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

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



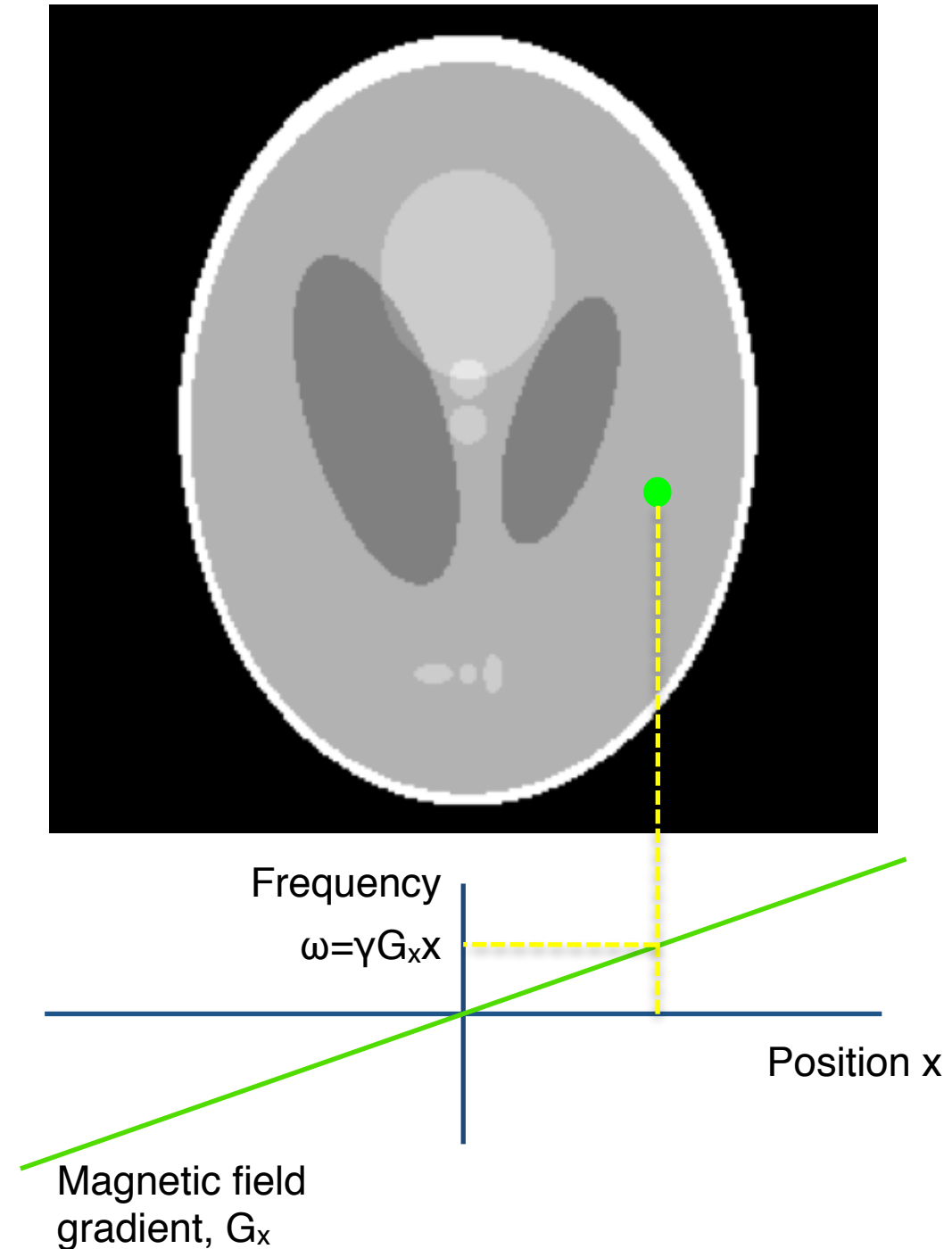
MR Imaging Theory

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

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

- Over time, phase accumulates as:

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



MR Imaging Theory

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

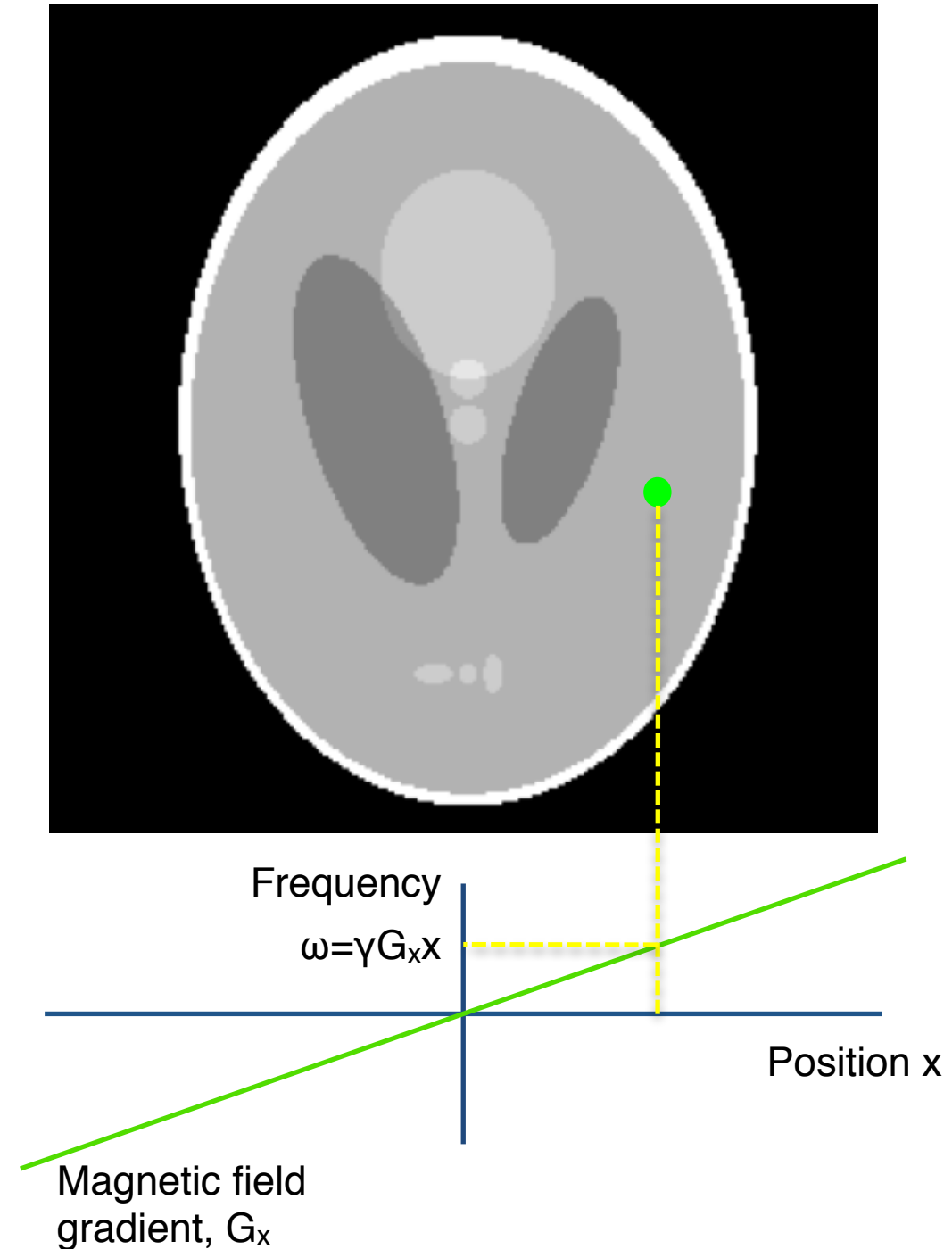
$$\delta\omega = \gamma \mathbf{G}(t) \cdot \mathbf{r}(t)$$

- Over time, phase accumulates as:

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

- Signal from the whole slice is given by:

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



MR Imaging Theory

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

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

- Over time, phase accumulates as:

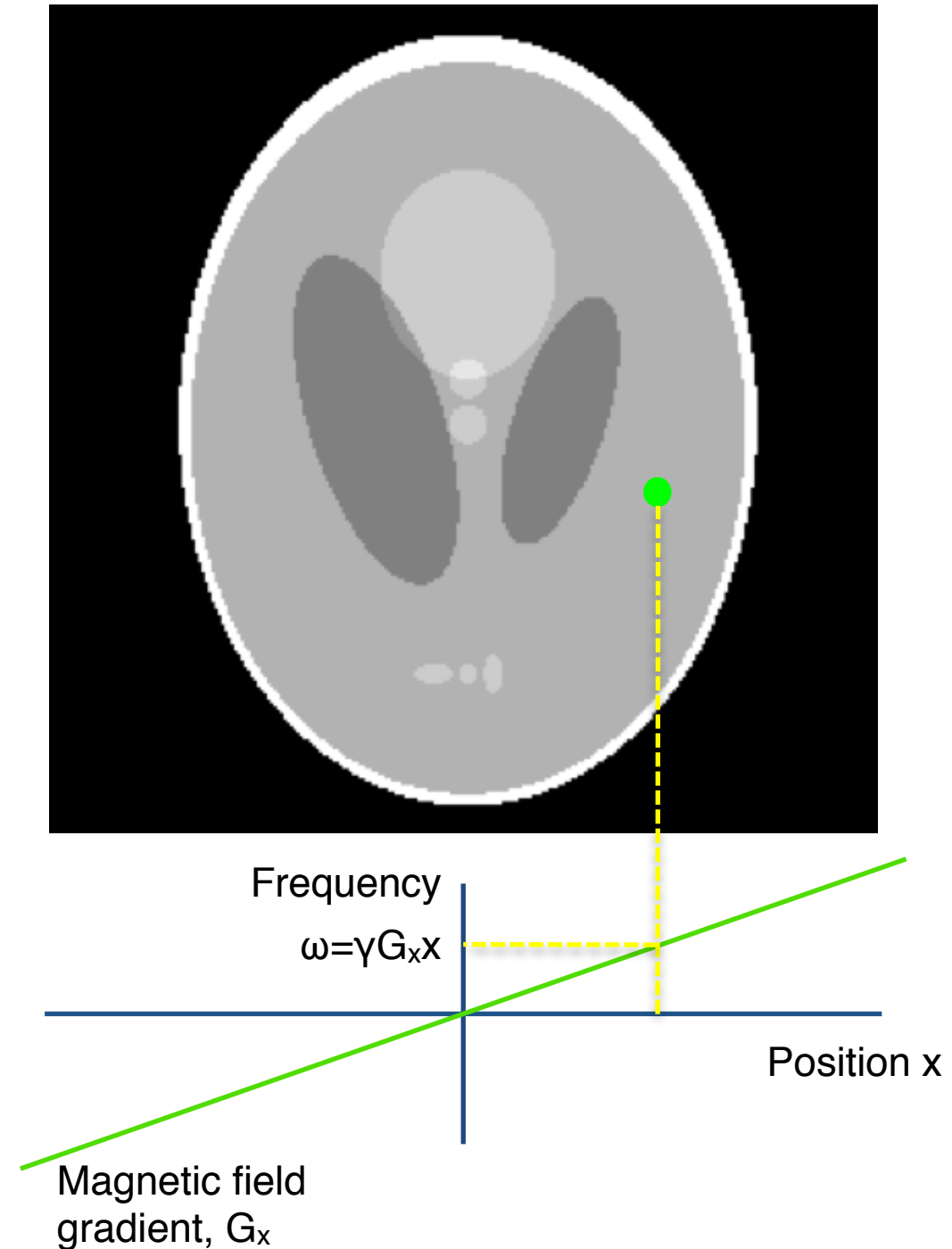
$$\delta\theta = \gamma \int_0^t \mathbf{G}(t') dt' \cdot \mathbf{r}(t)$$

- Signal from the whole slice is given by

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

- Write:

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



MR Imaging Theory

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

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

- Over time, phase accumulates as:

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

- Signal from the whole slice is given by

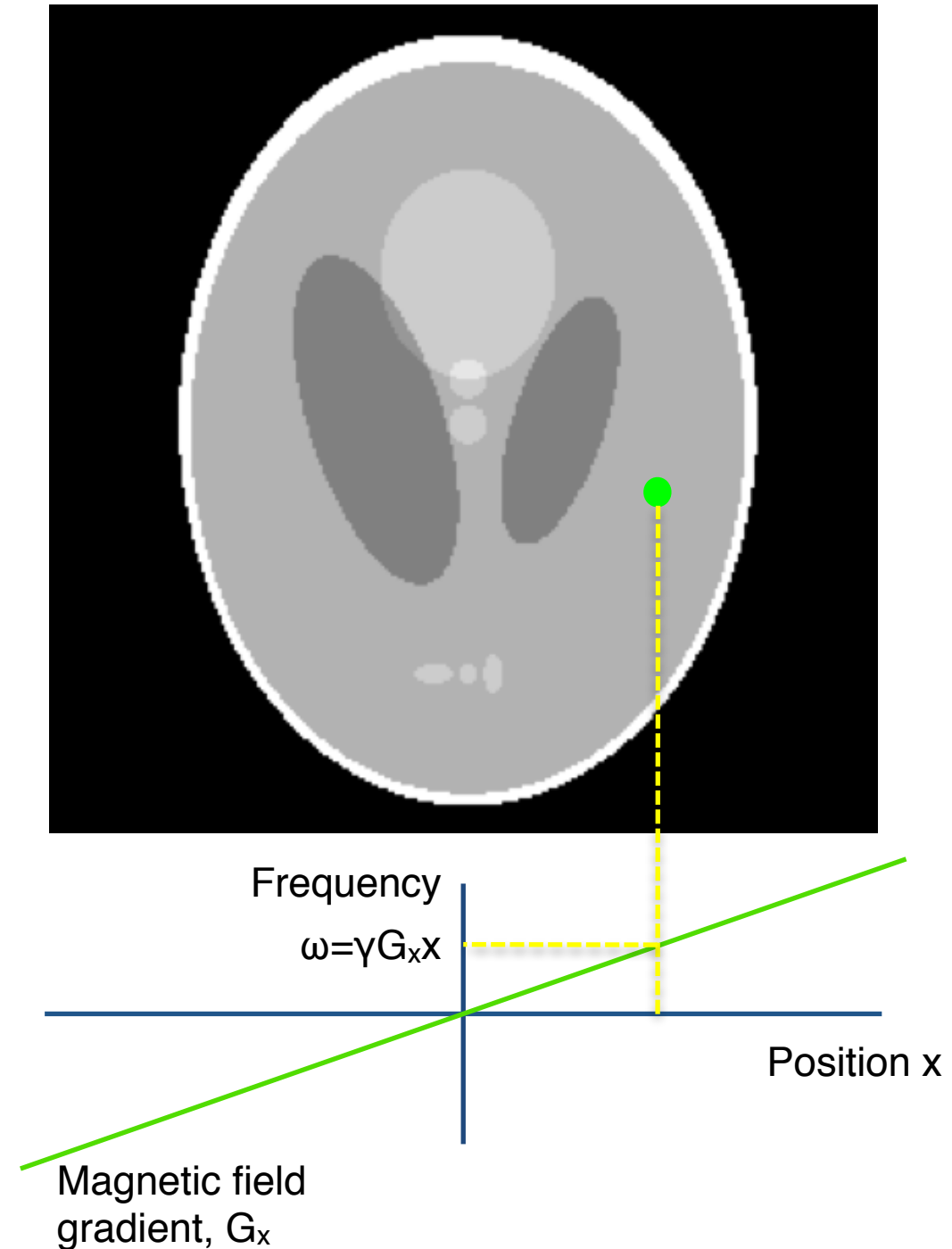
$$S(G, t) = A \int_V \rho(\mathbf{r}) \exp \left[i\gamma \int_0^t \mathbf{G}(t') dt' \cdot \mathbf{r} \right] d^3\mathbf{r}$$

- Write:

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

- Then:

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



MR Imaging Theory

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

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

- Over time, phase accumulates as:

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

- Signal from the whole slice is given by

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

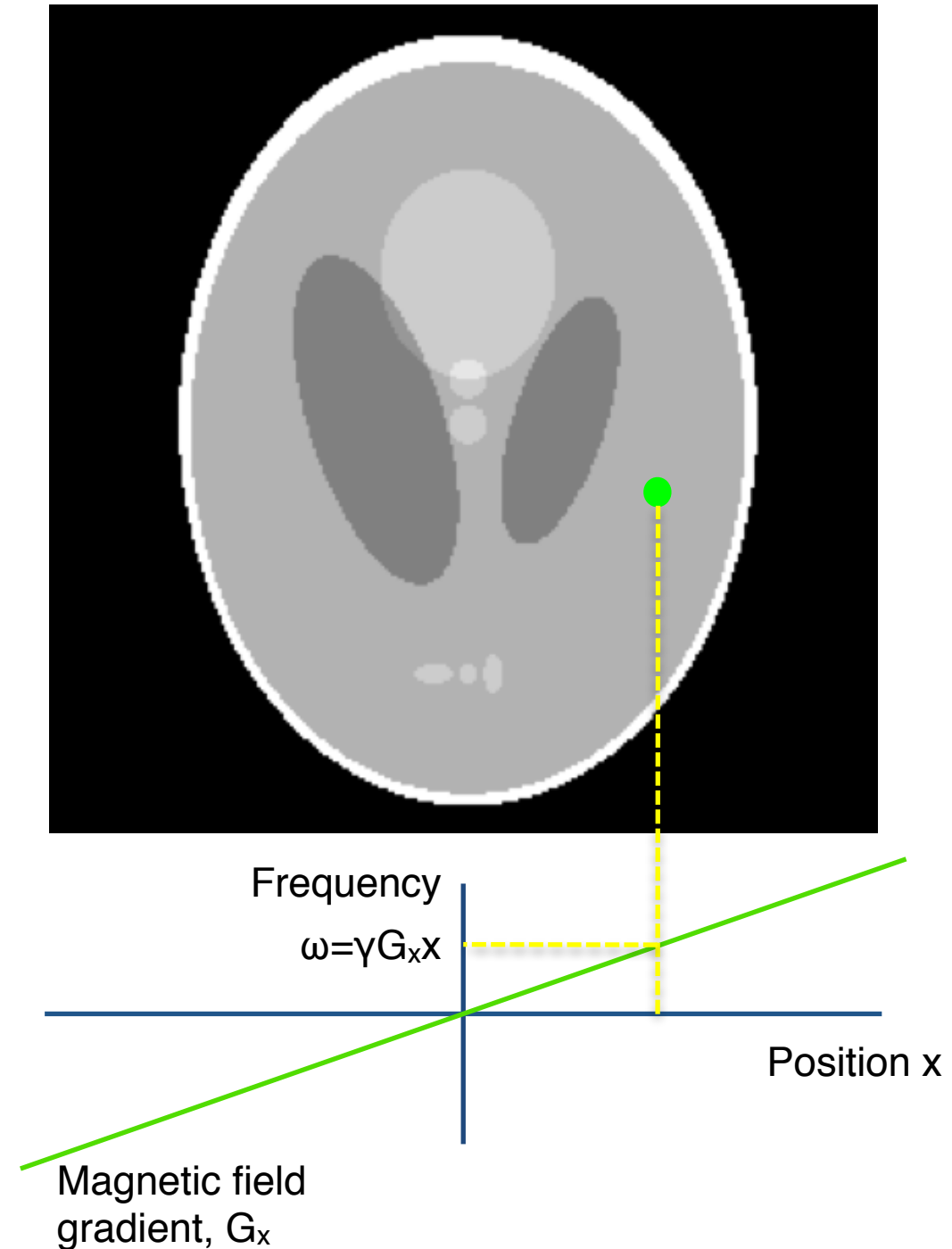
- Write:

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

- Then:

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

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

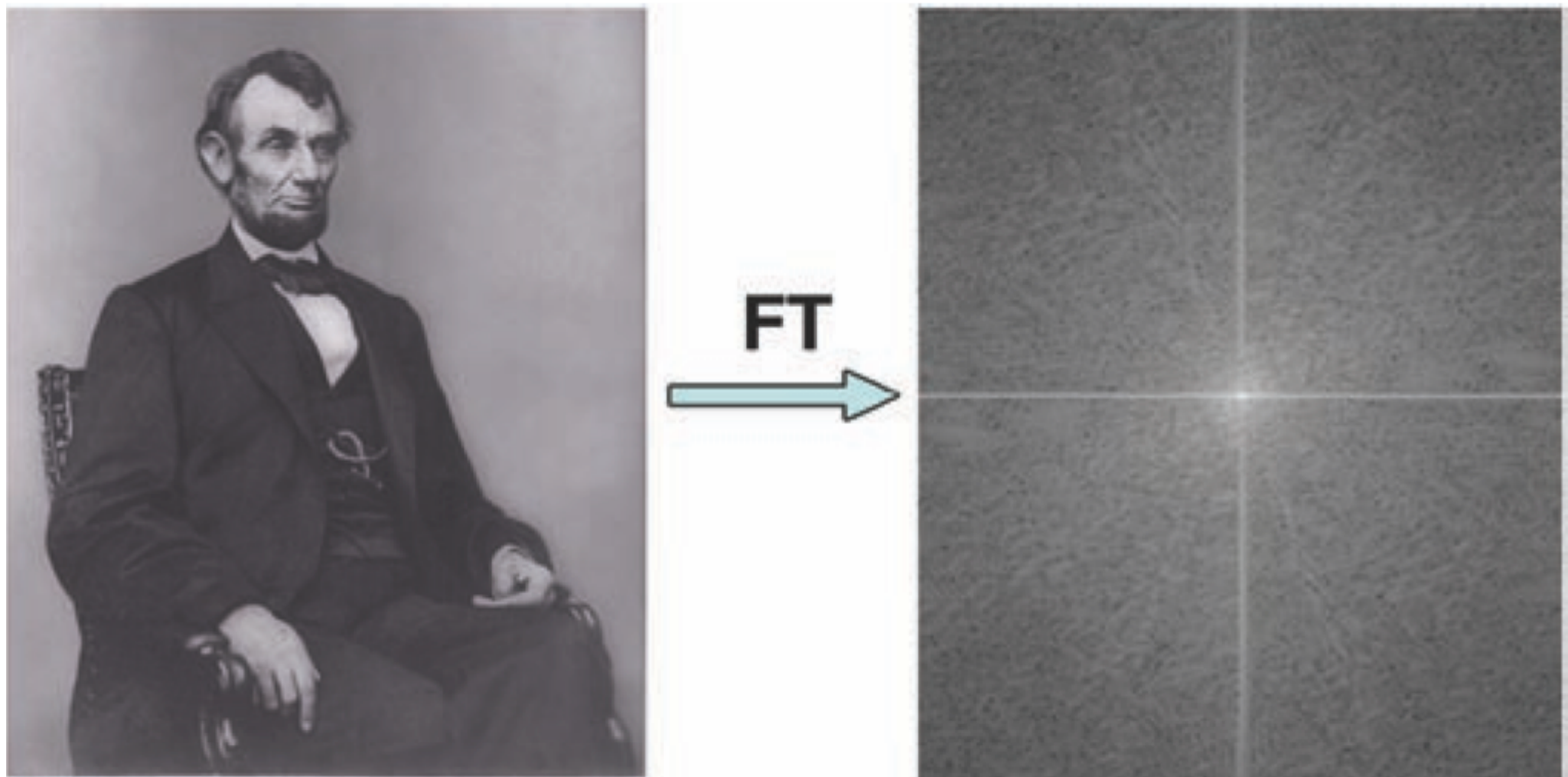


Fourier transform to k-space

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

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

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

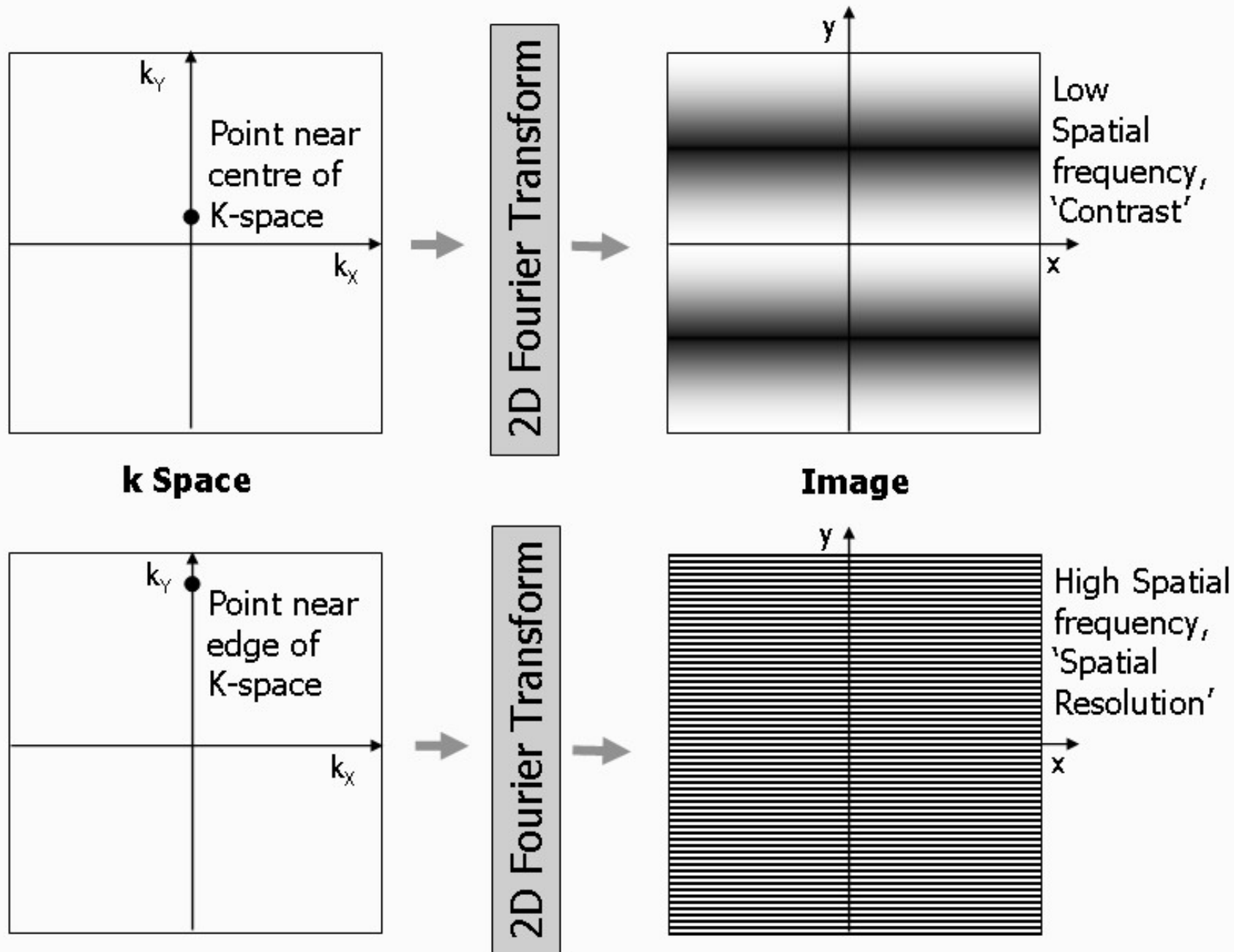


K-space

Spatial frequencies

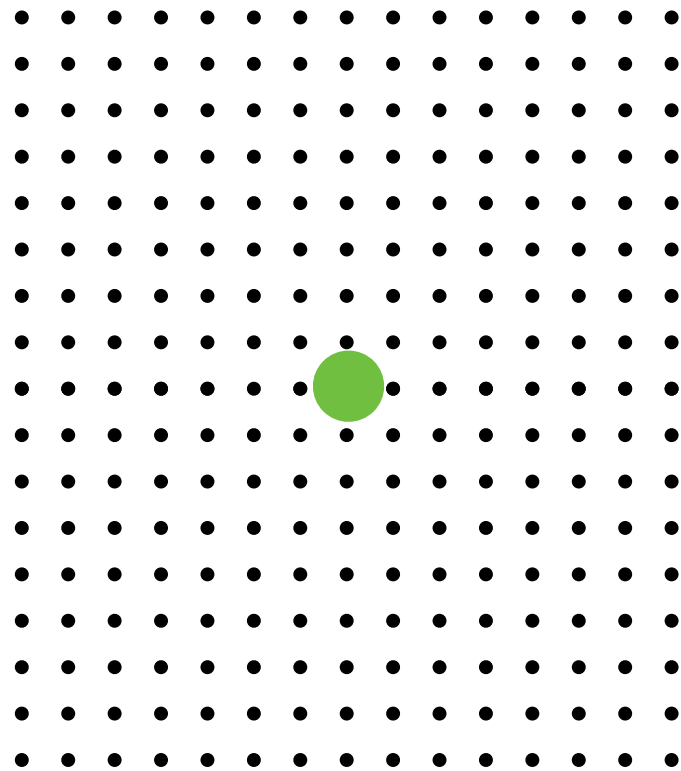
$$\mathbf{k}(t) = \frac{\gamma}{2\pi} \int_0^t \mathbf{G}(t') dt'$$

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

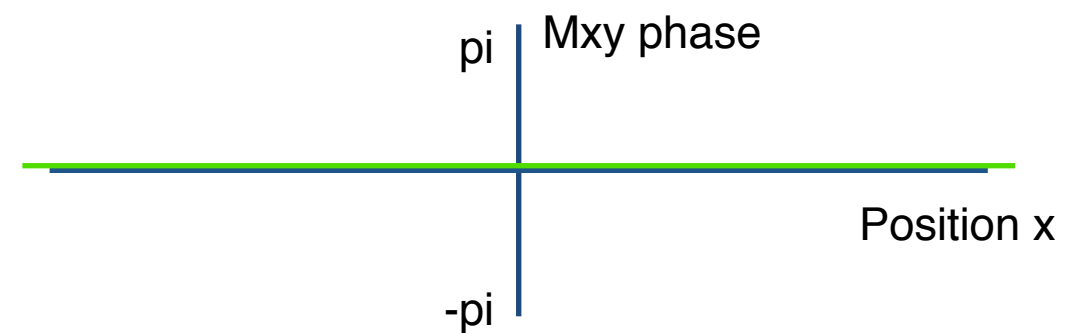


from J. Ridgeway, JCMR, **12**:71, 2010

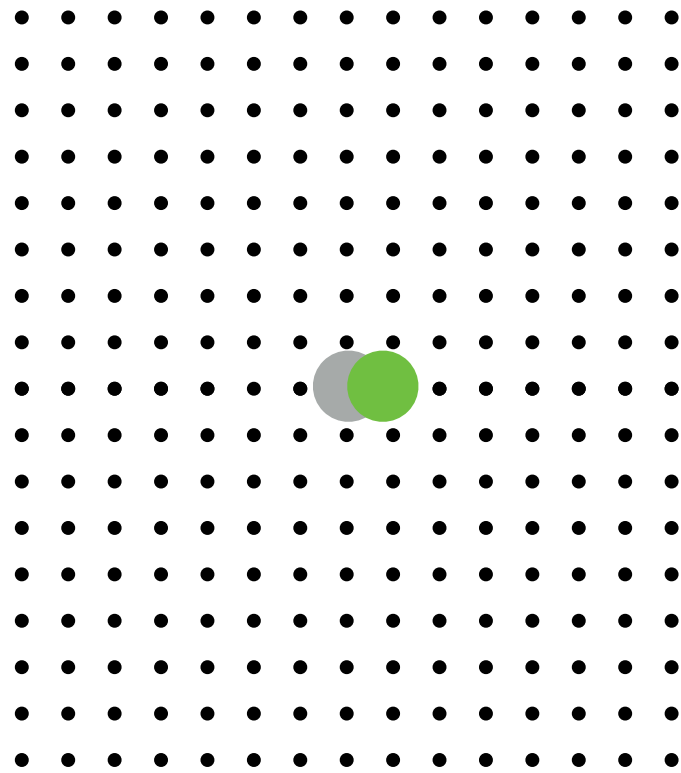
What does it mean?



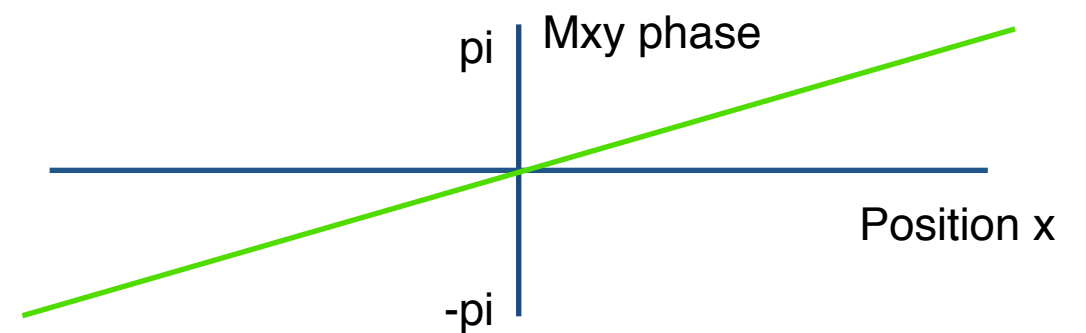
- At center of k-space there is no Mxy phase across the image



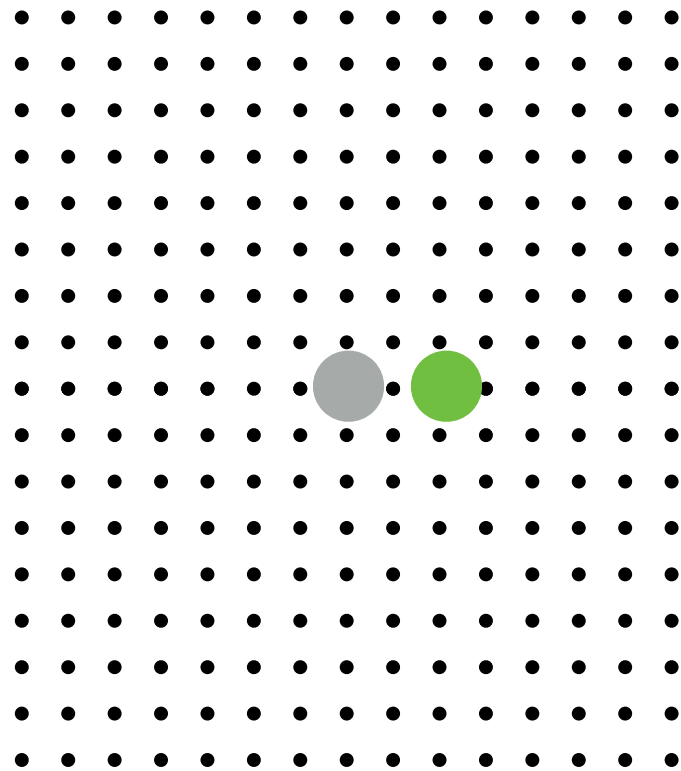
What does it mean?



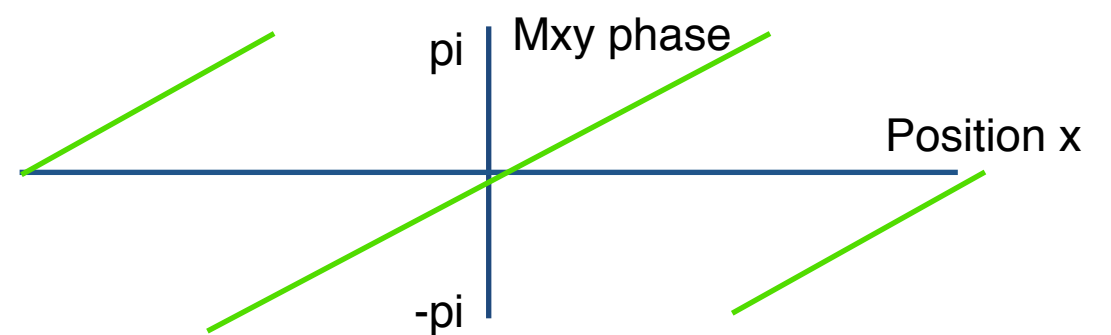
- Moving one k-space sample puts 2π ($-\pi$ to π) across the whole image FOV



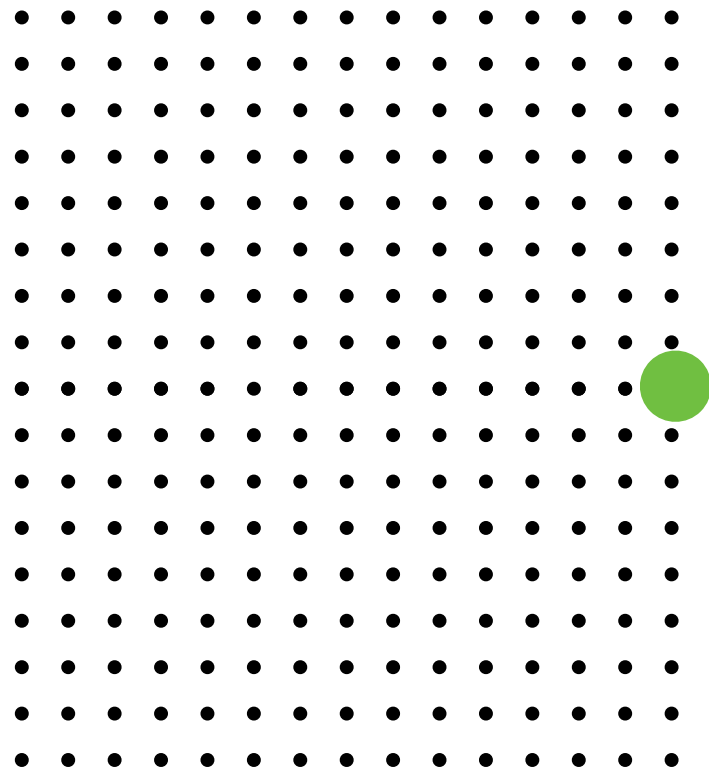
What does it mean?



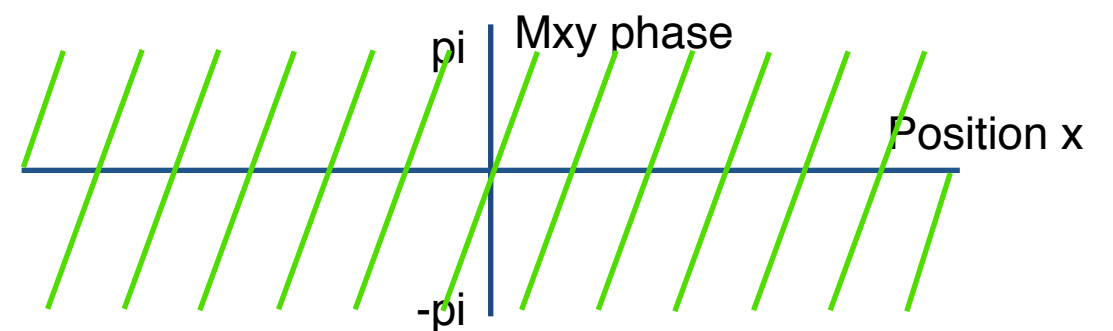
- Moving two k-space sample puts 4π across the whole image FOV



What does it mean?



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



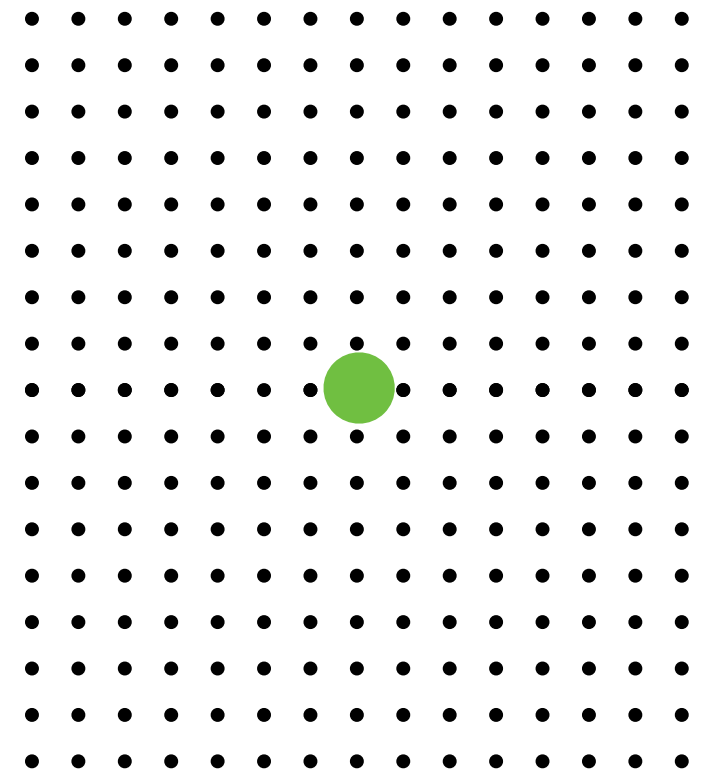
Encoding strategies

k-space trajectories

- Sample all points in k-space to acquire sufficient data for image reconstruction.
- Initial position: origin
- $\mathbf{k}(t)$ is the sampling position
- $\mathbf{G}(t)$ is the velocity through k-space
- Sample spacing: $\delta\mathbf{k} = 1/\text{FOV}$
- Sampling extent: $\Delta\mathbf{k} = 1/\text{pixelsize}$

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

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



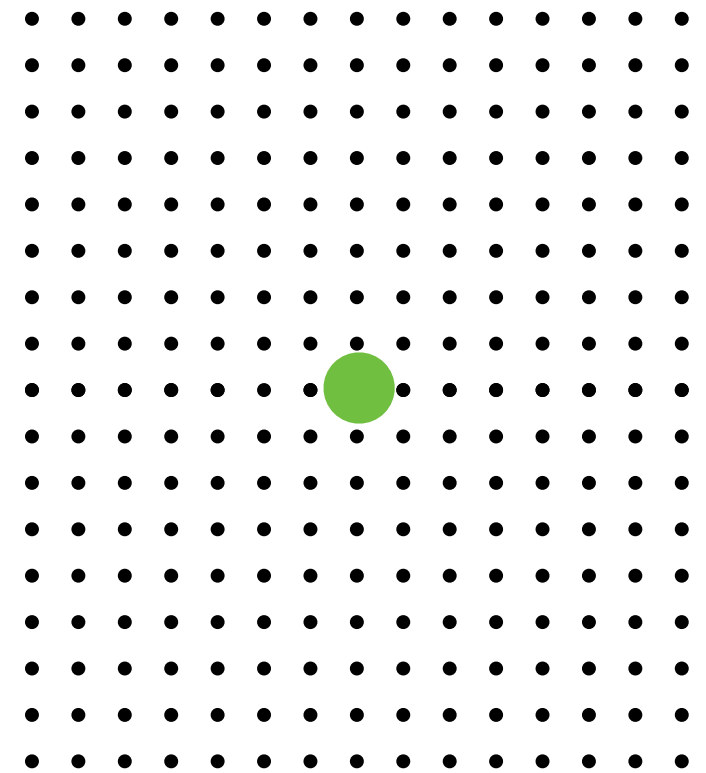
Encoding strategies

k-space trajectories

- Sample all points in k-space to acquire sufficient data for image reconstruction.
- Initial position: origin
- $\mathbf{k}(t)$ is the sampling position
- $\mathbf{G}(t)$ is the velocity through k-space
- Sample spacing: $\delta\mathbf{k} = 1/\text{FOV}$
- Sampling extent: $\Delta\mathbf{k} = 1/\text{pixelsize}$

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

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



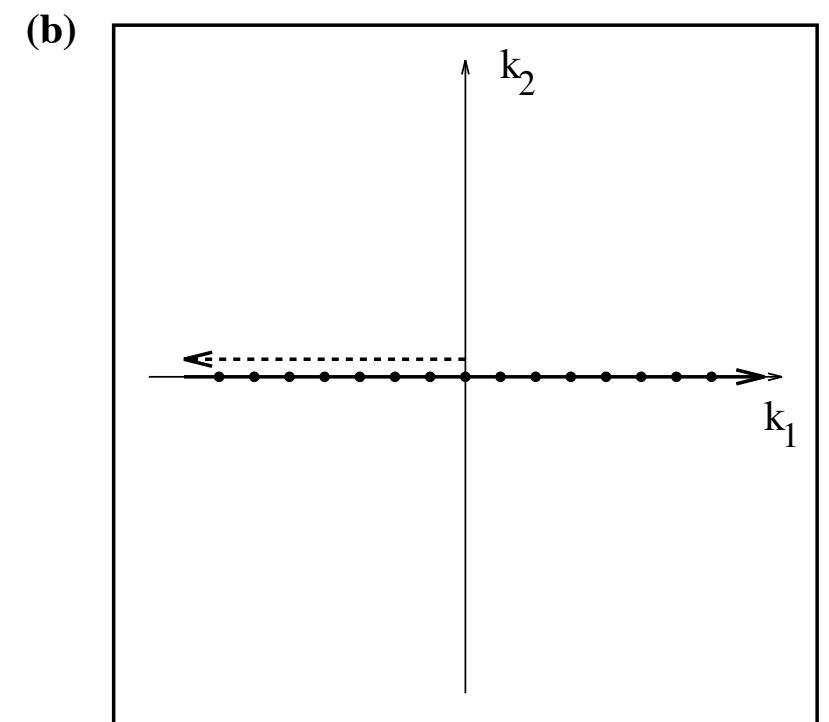
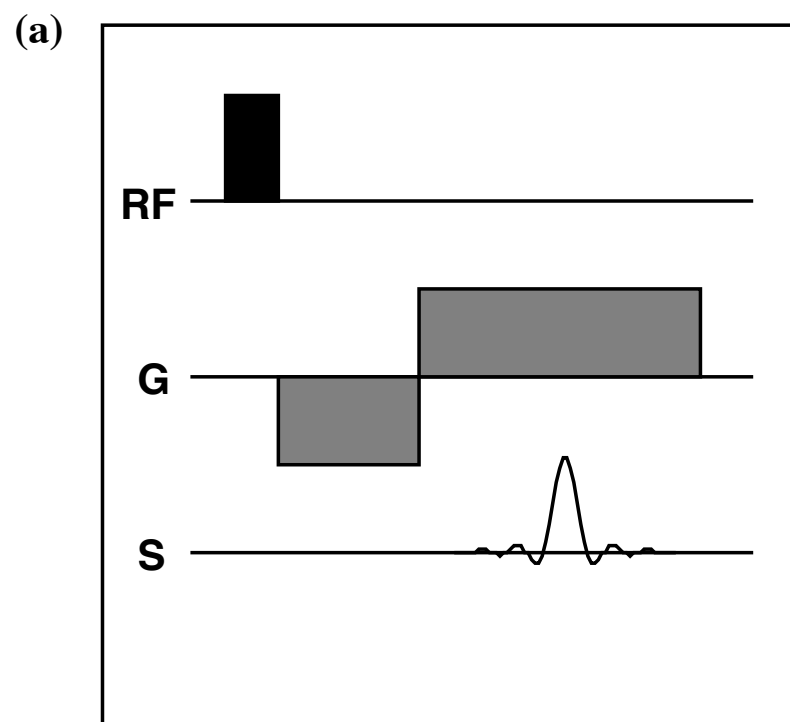
Encoding strategies

Gradient echo

- Forms echo signal with spatial encoding in the gradient direction

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

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



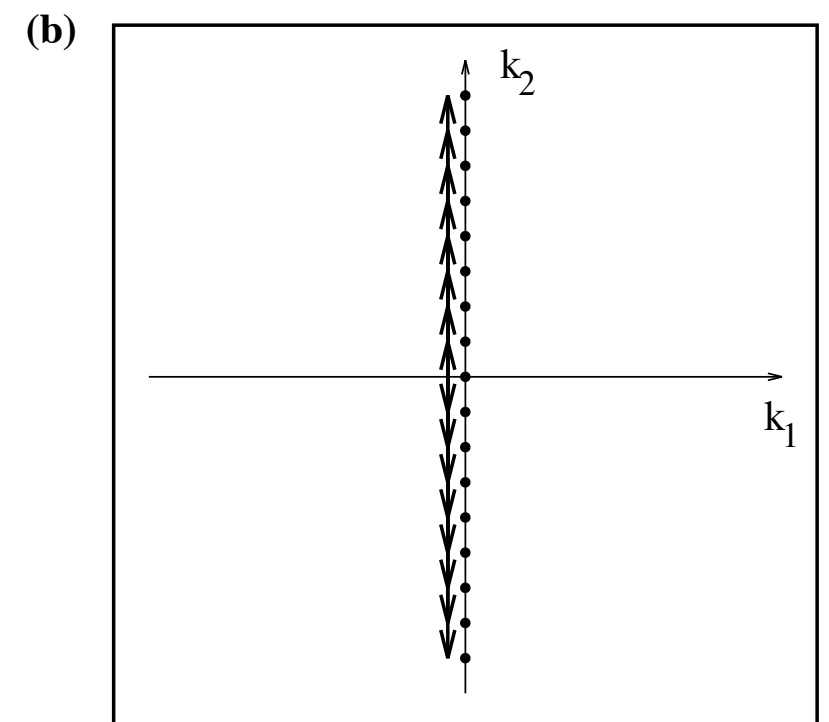
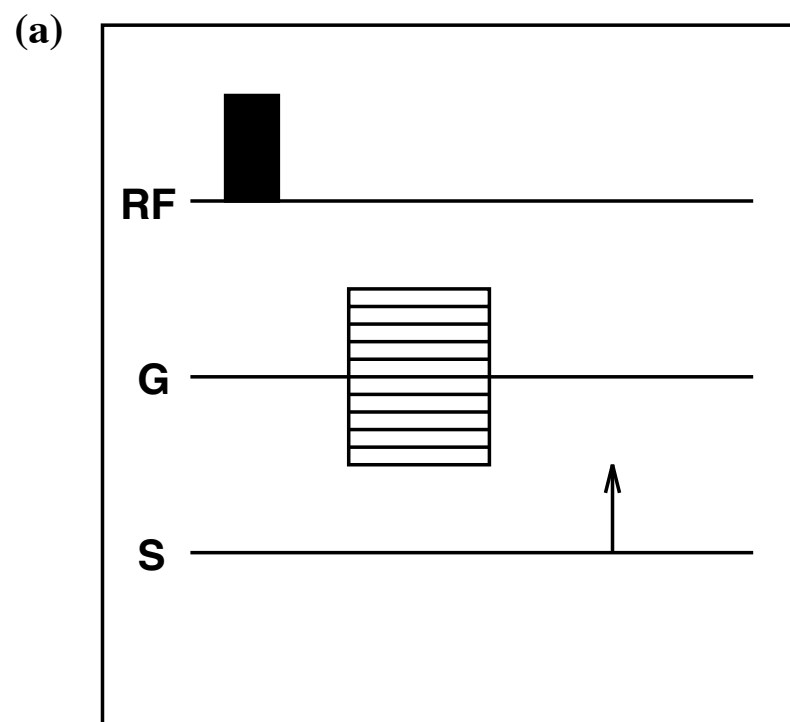
Encoding strategies

Phase encoding

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

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

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



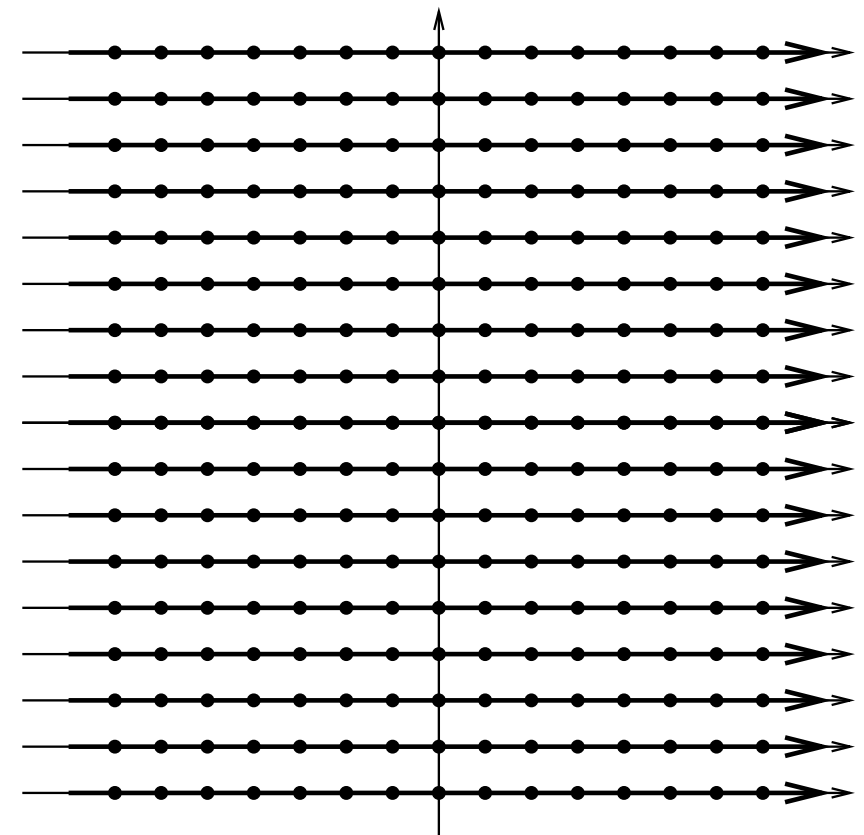
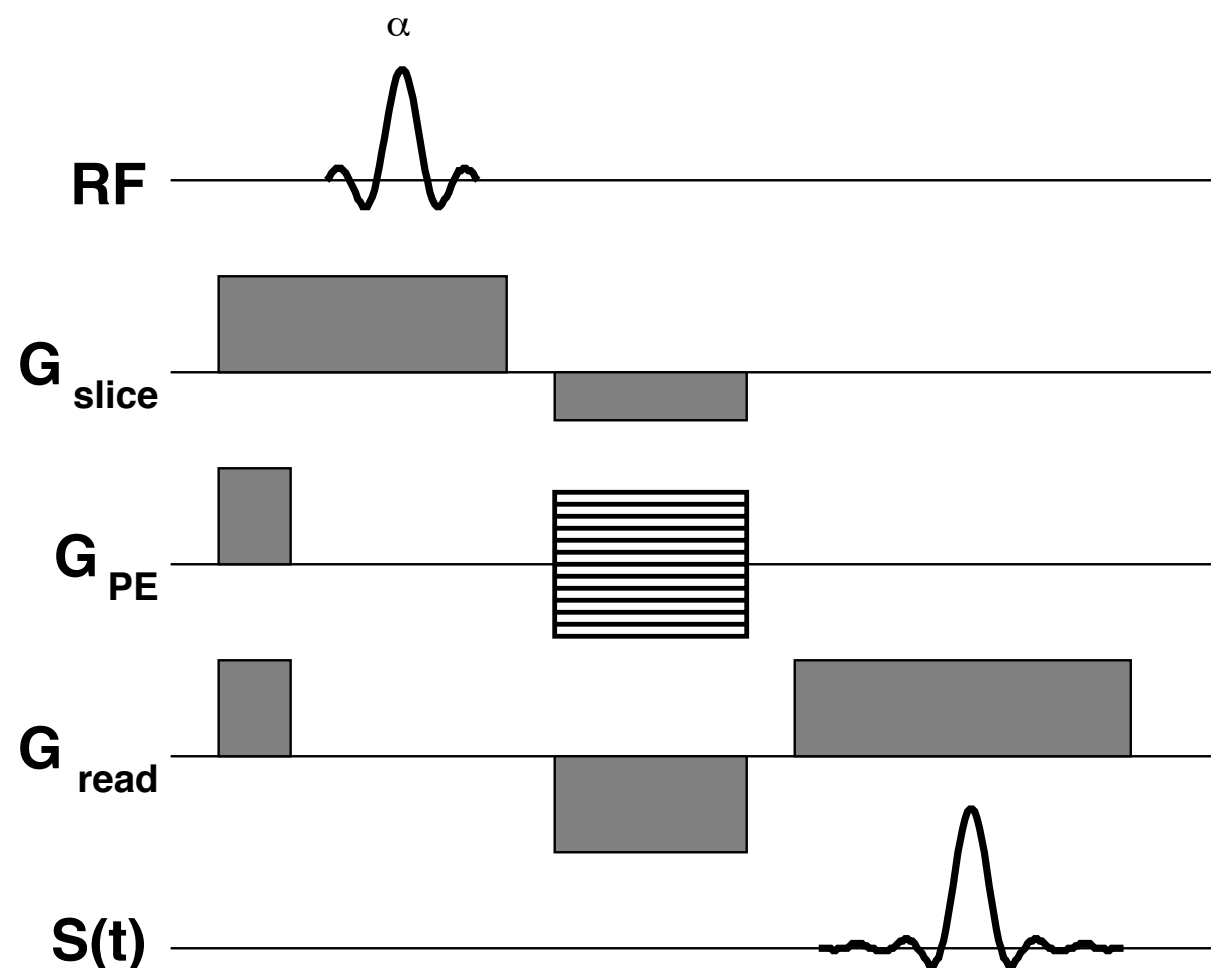
Encoding strategies

2D Gradient echo imaging

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

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

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



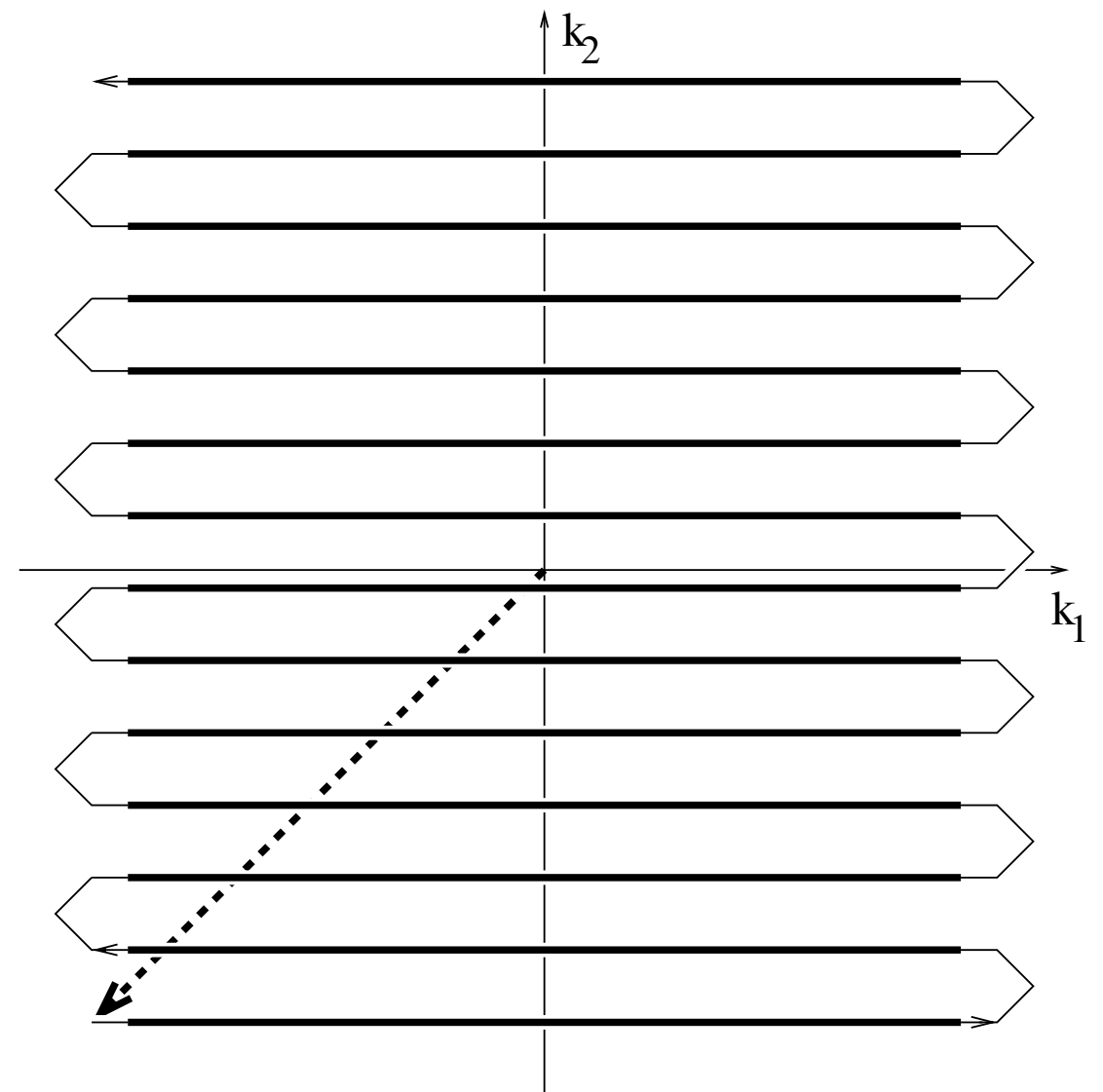
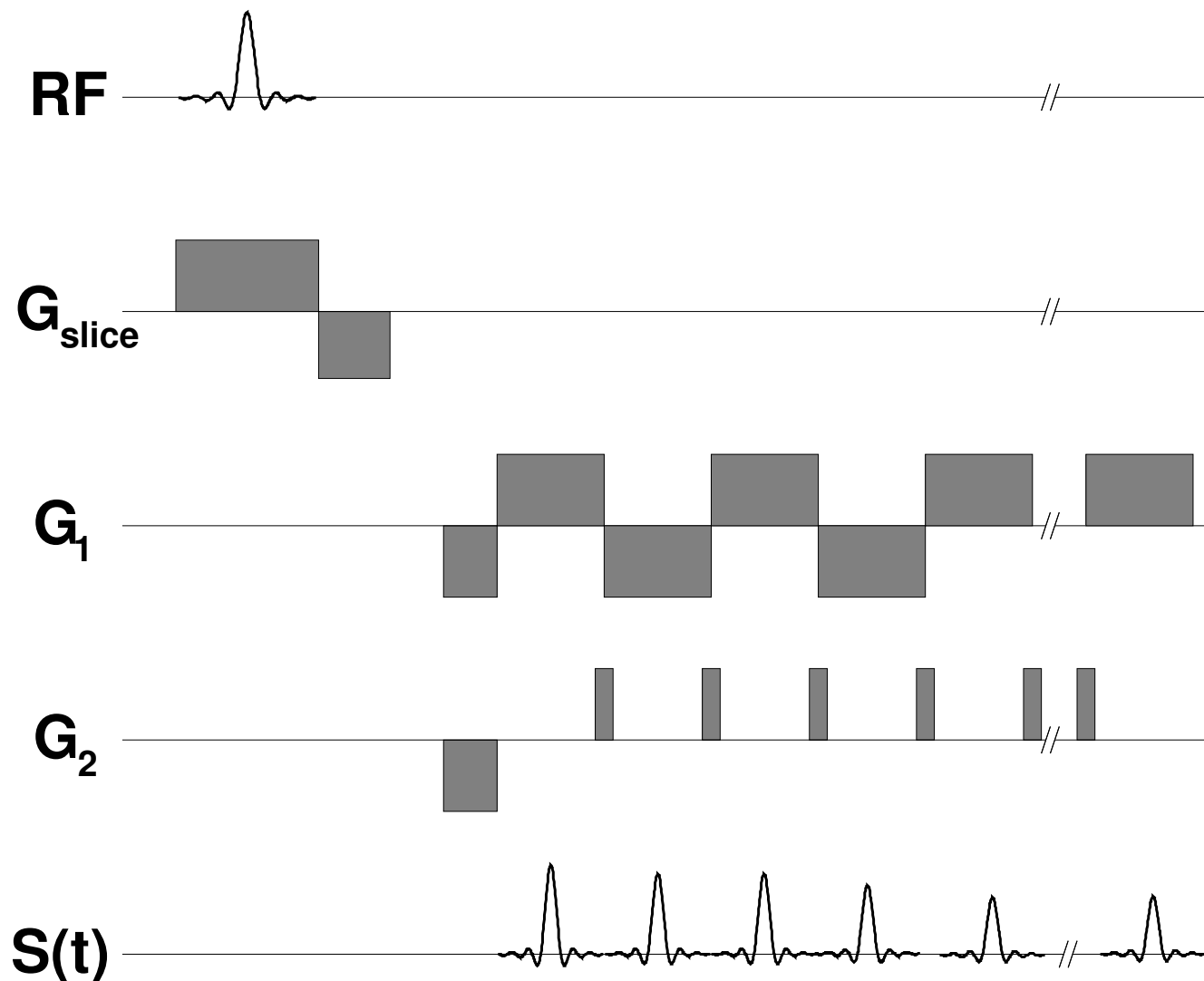
Encoding strategies

Echo planar Imaging

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

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

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

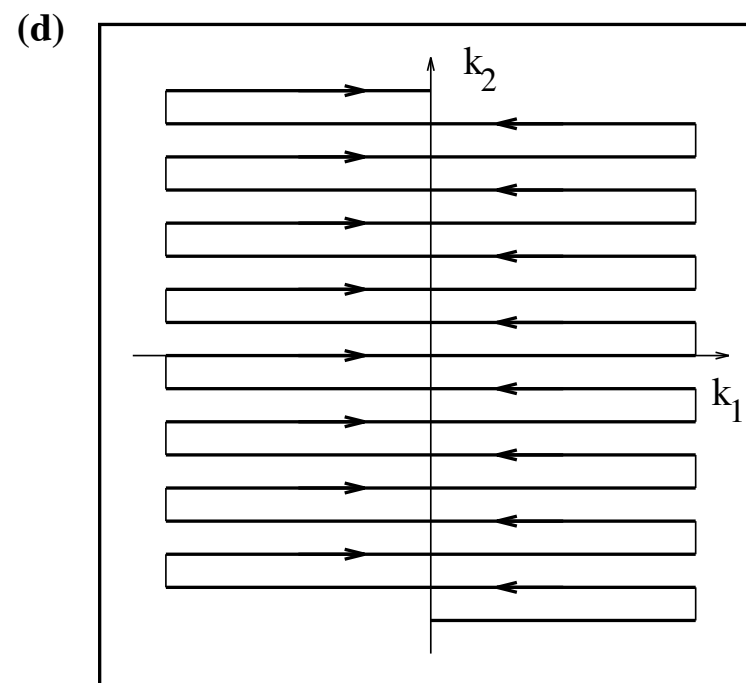
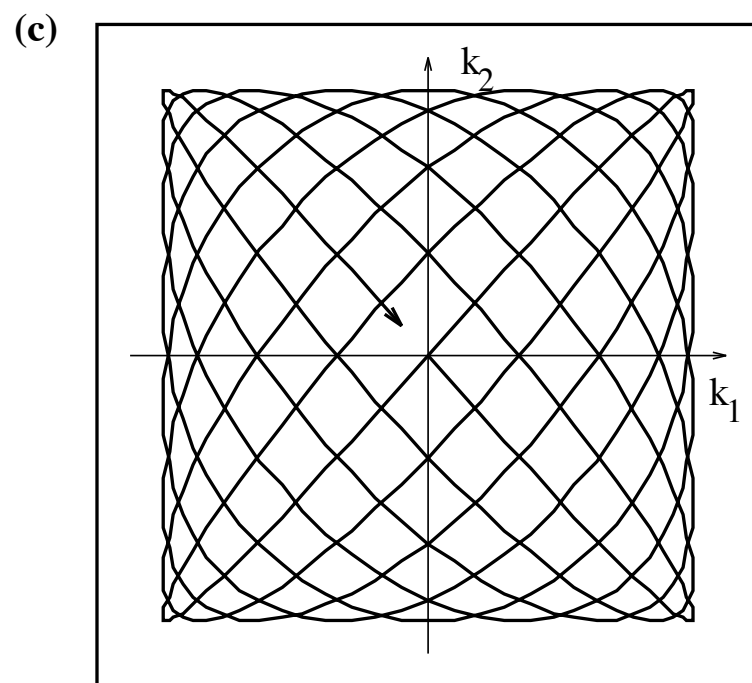
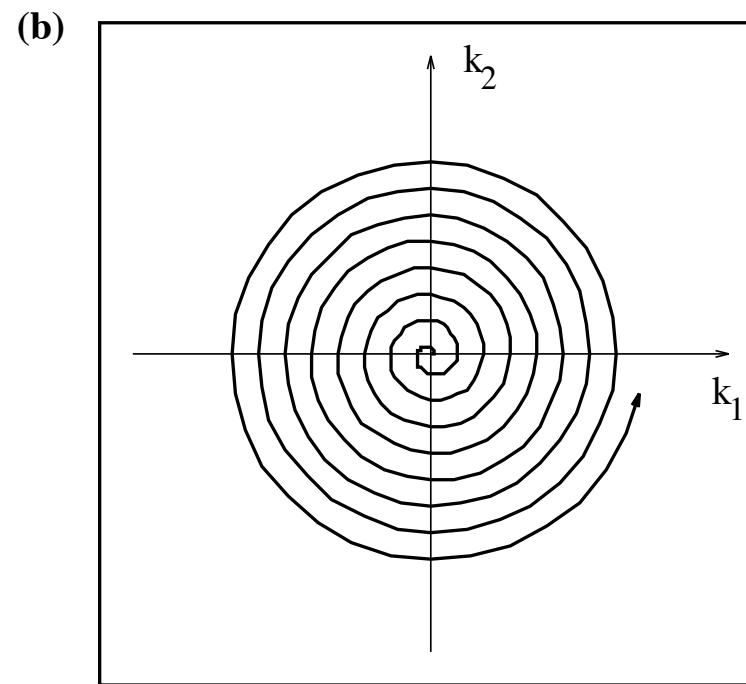
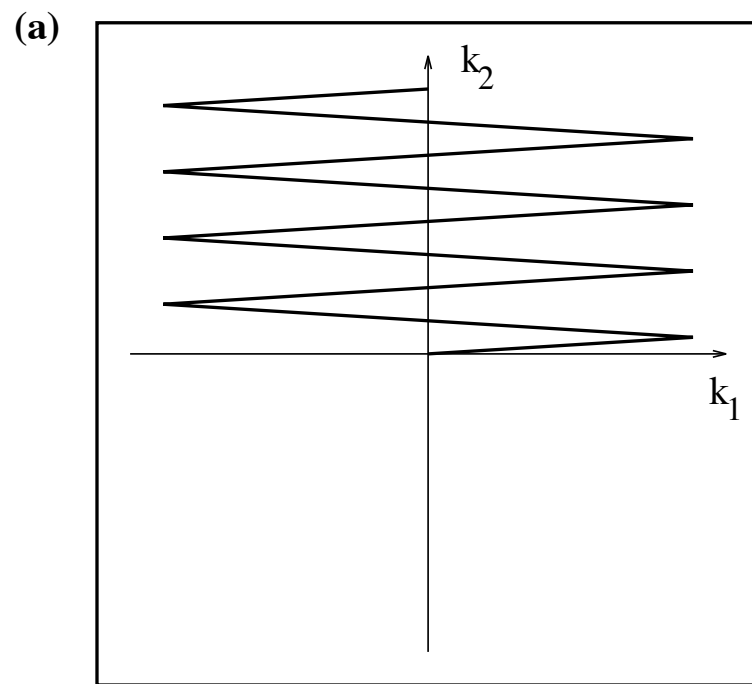


Encoding strategies

Spirals etc...

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

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



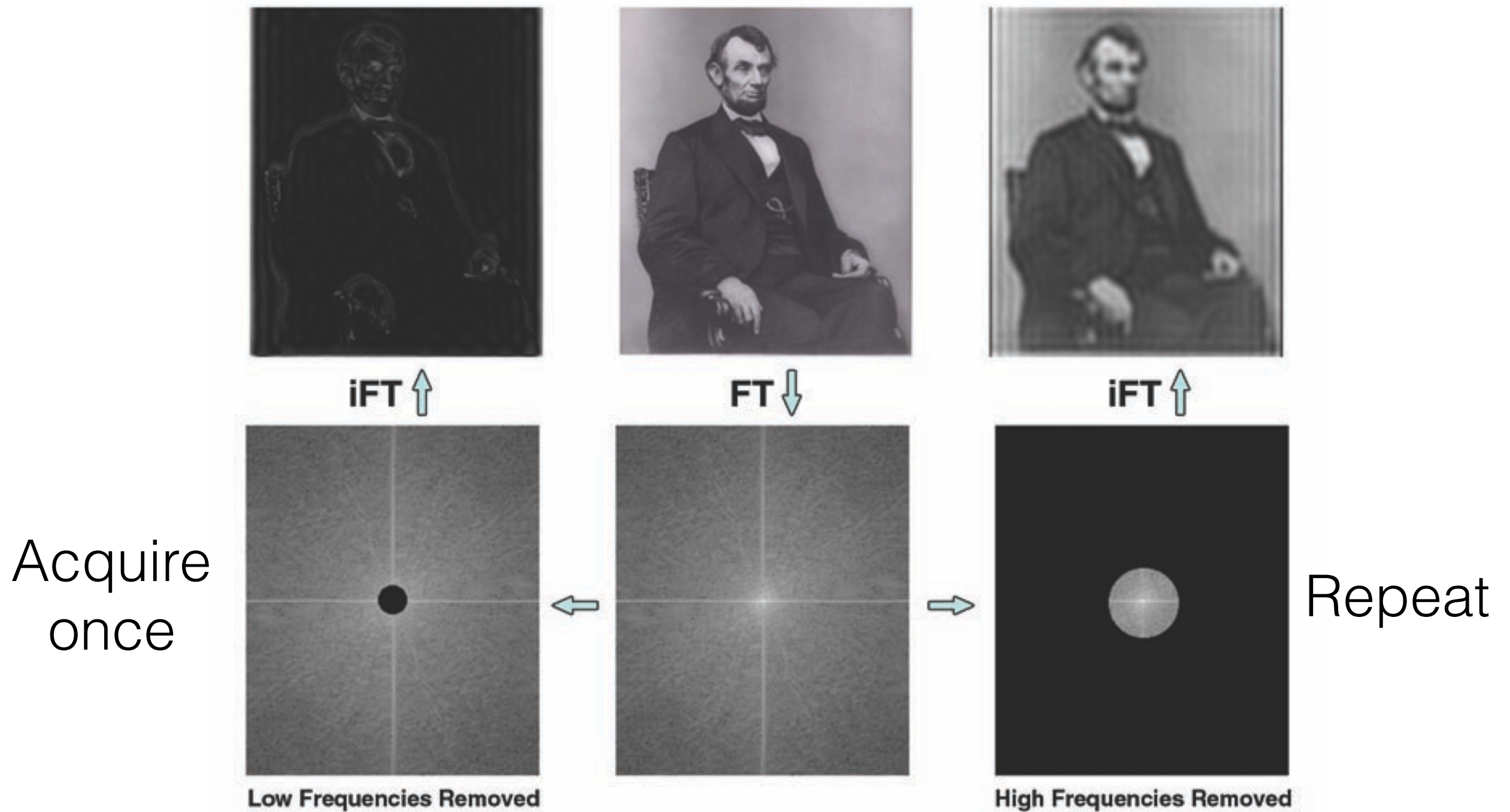
Accelerated Acquisitions

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



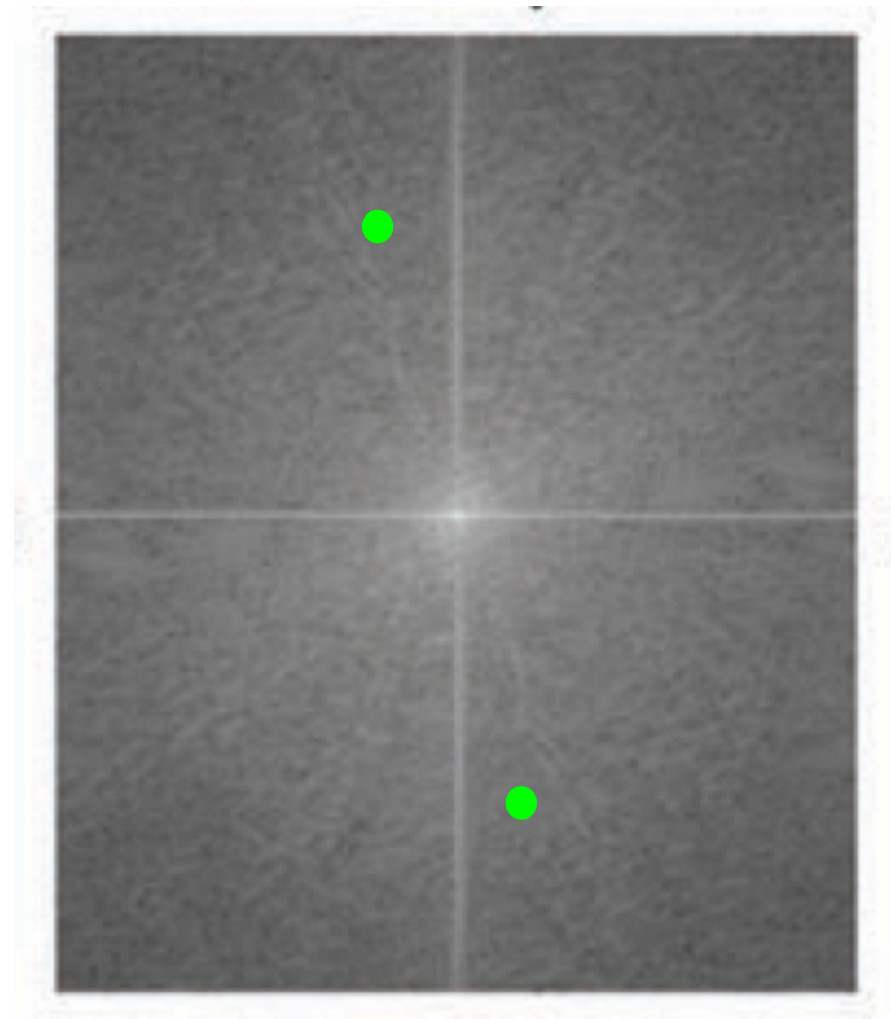
Acceleration techniques

Partial k-space

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

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

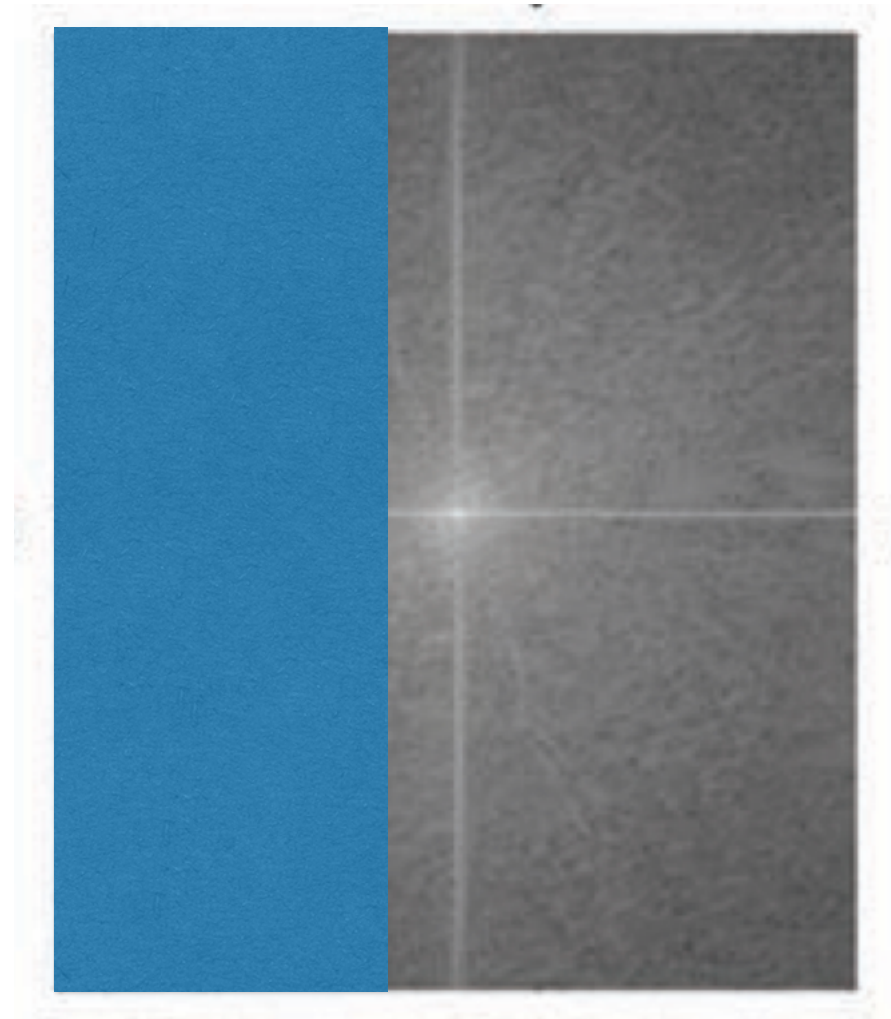
- Synthesize missing conjugate data during image reconstruction
- Need some additional lines near center of k-space to account for mild background phase roll across FOV (e.g. due to receiver coils)



Acceleration techniques

Partial k-space

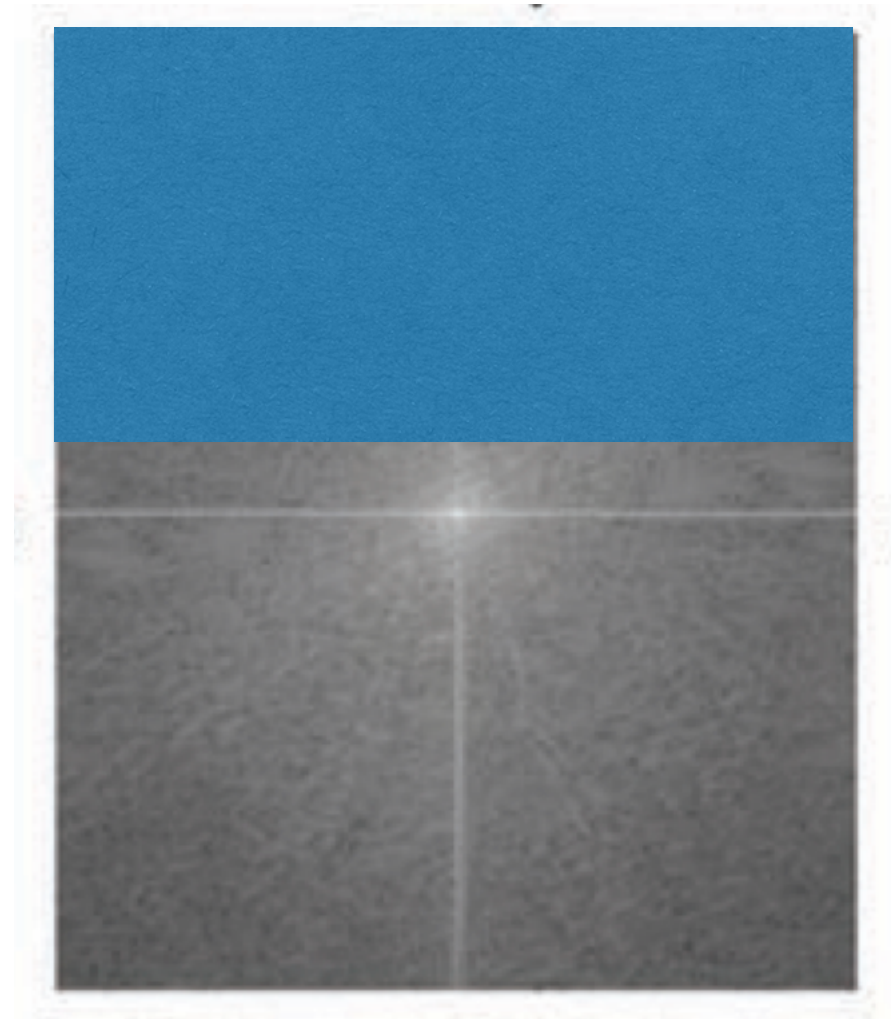
- Partial echo / Asymmetric echo
 - Short TE



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

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 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



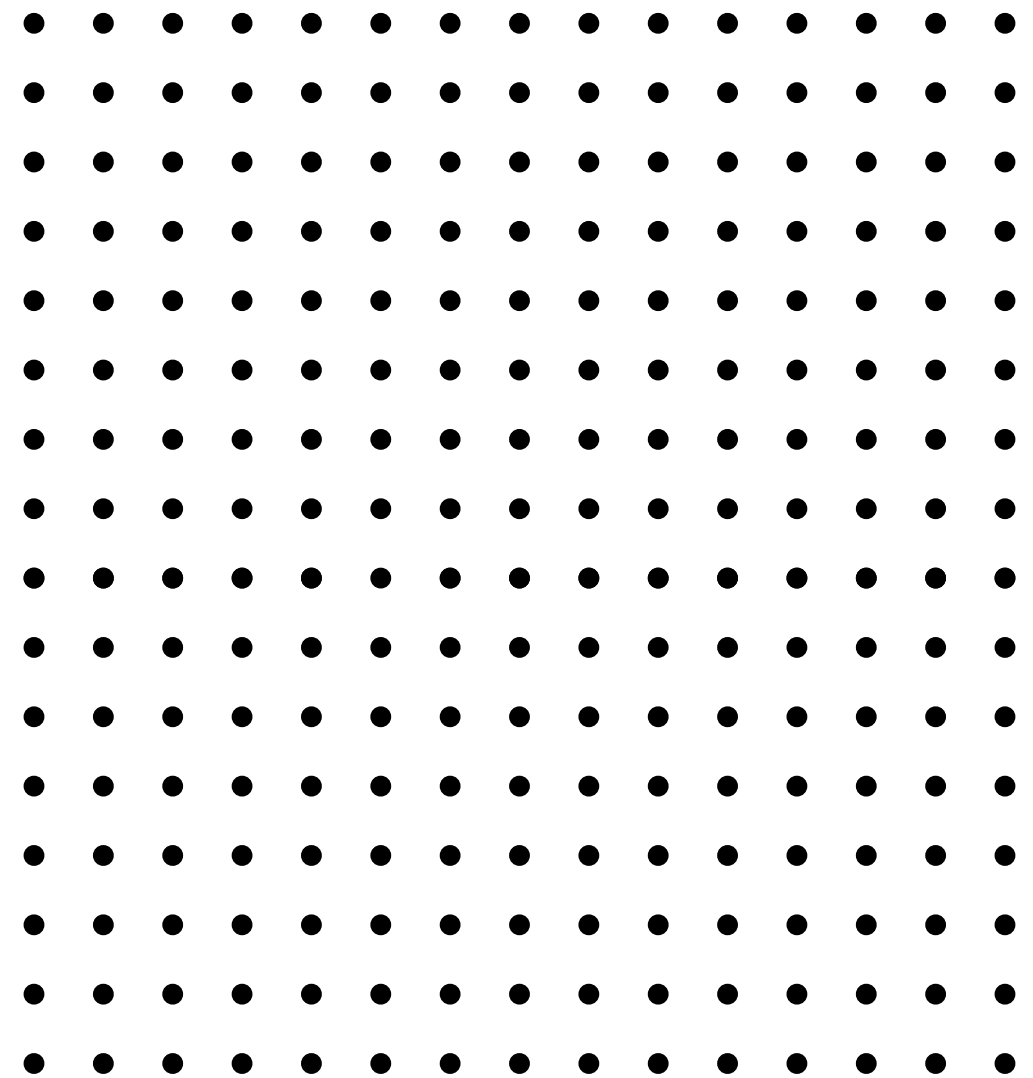
Parallel Imaging

Undersampling k-space

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

$$dk = 1/\text{FOV}$$

$1/\text{FOV} \updownarrow$



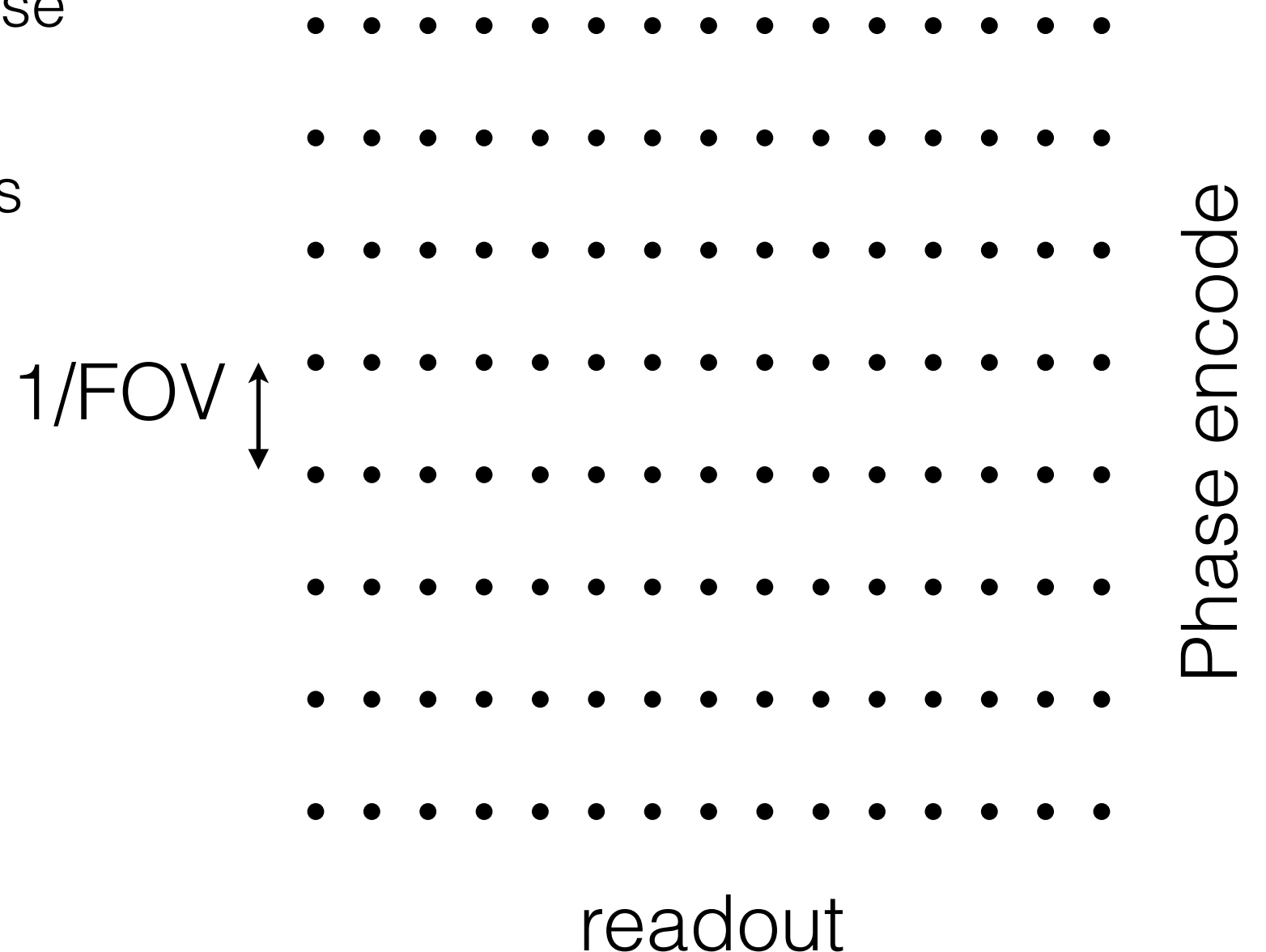
Phase encode

readout

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: $S = E I$

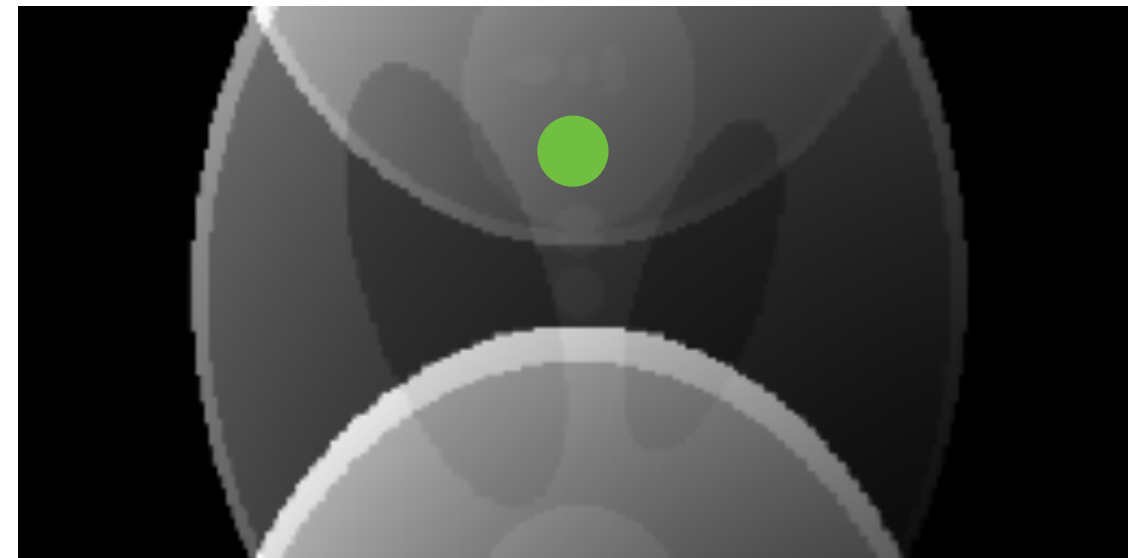
Solution: $I = \text{pinv}(E).S$

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

Image from coil 1



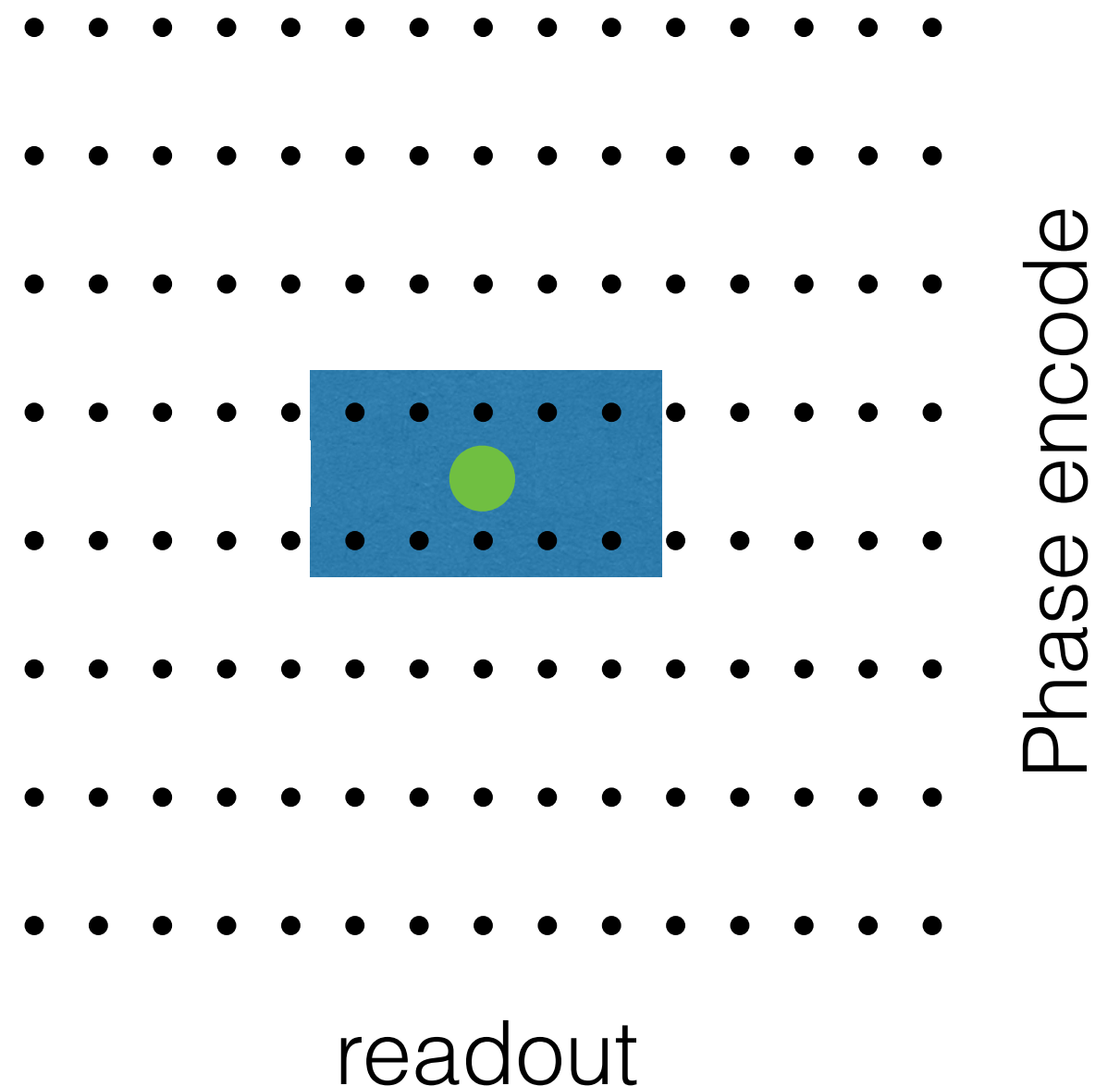
Image from coil 2



Parallel Imaging

GRAPPA (R=2 example)

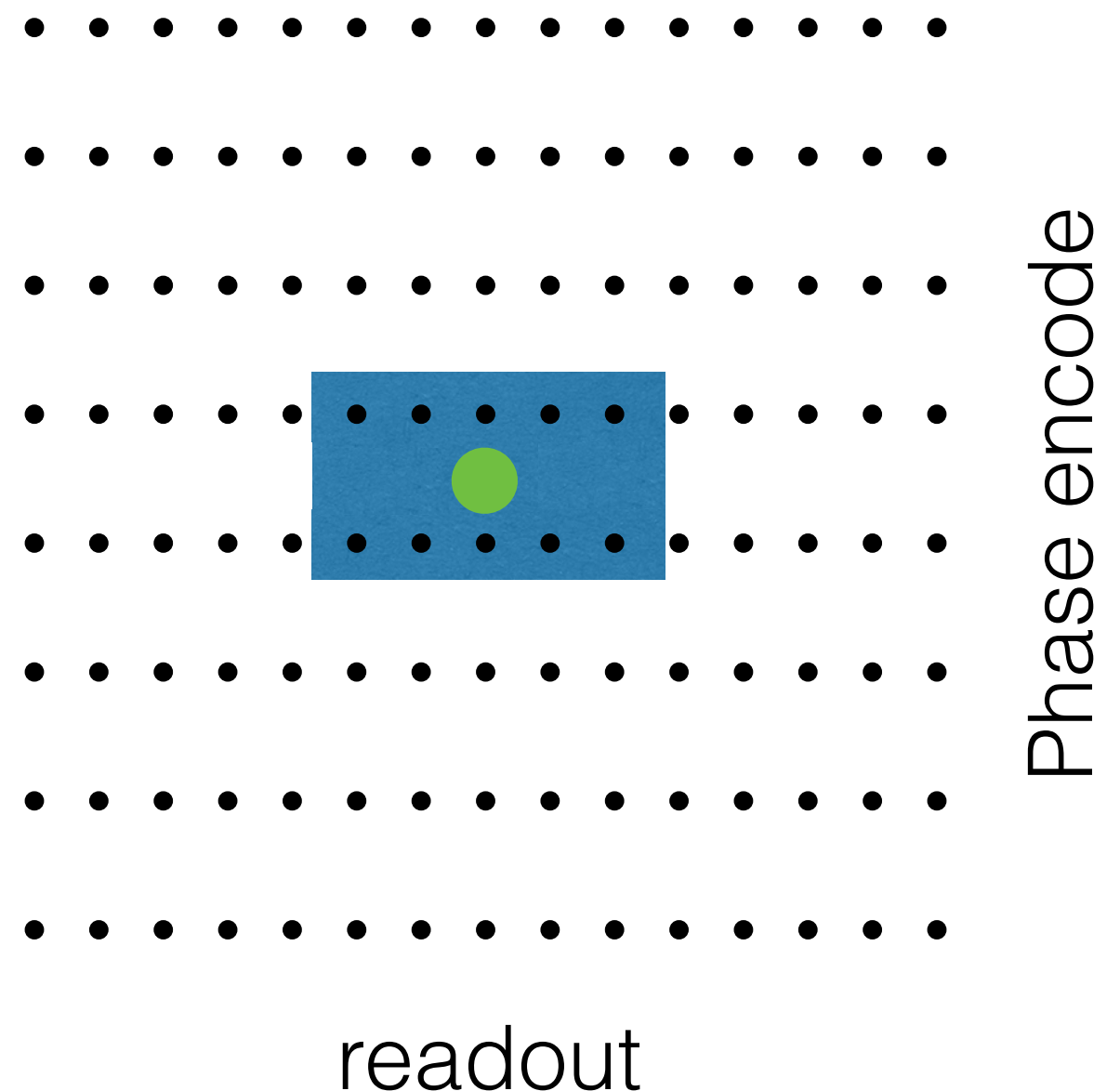
- Similar k-space undersampling as SENSE
- Data is interpolated in k-space from neighboring samples



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



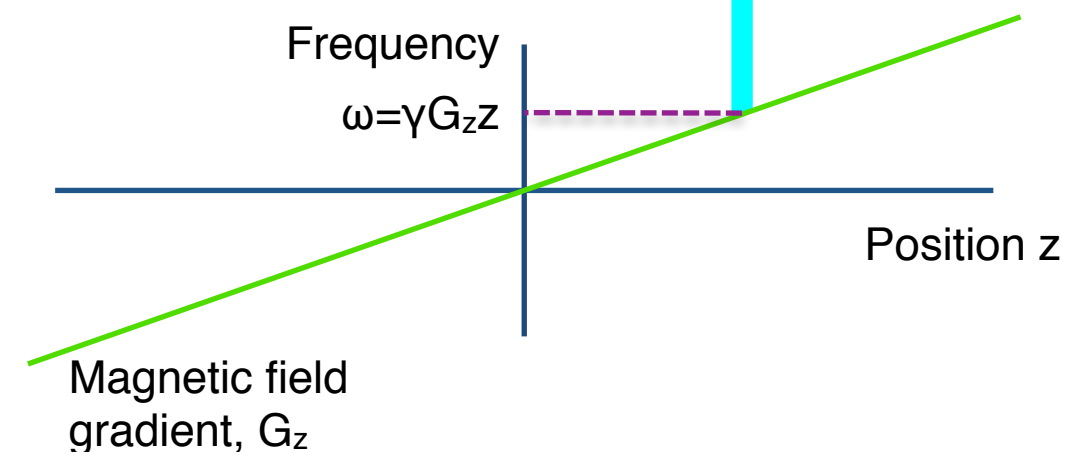
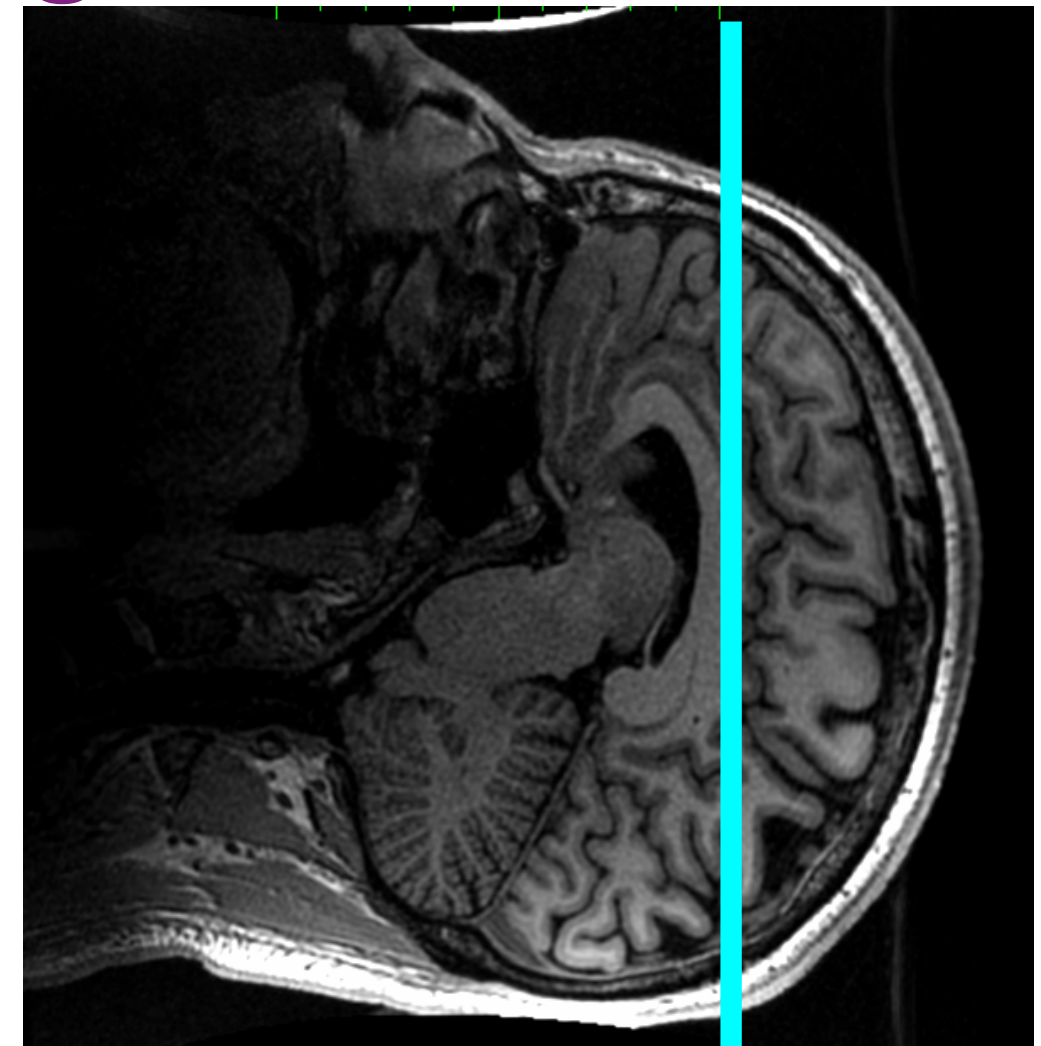
Multiband/SMS imaging

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:

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

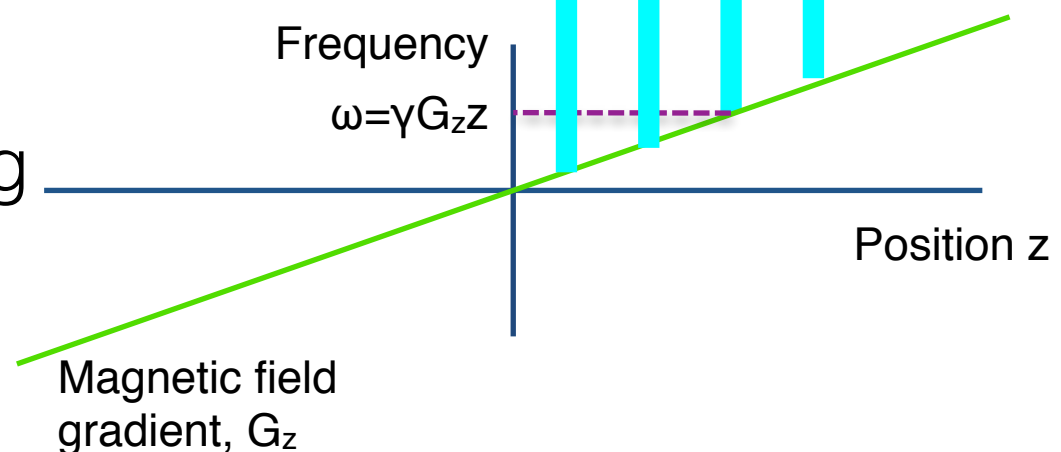
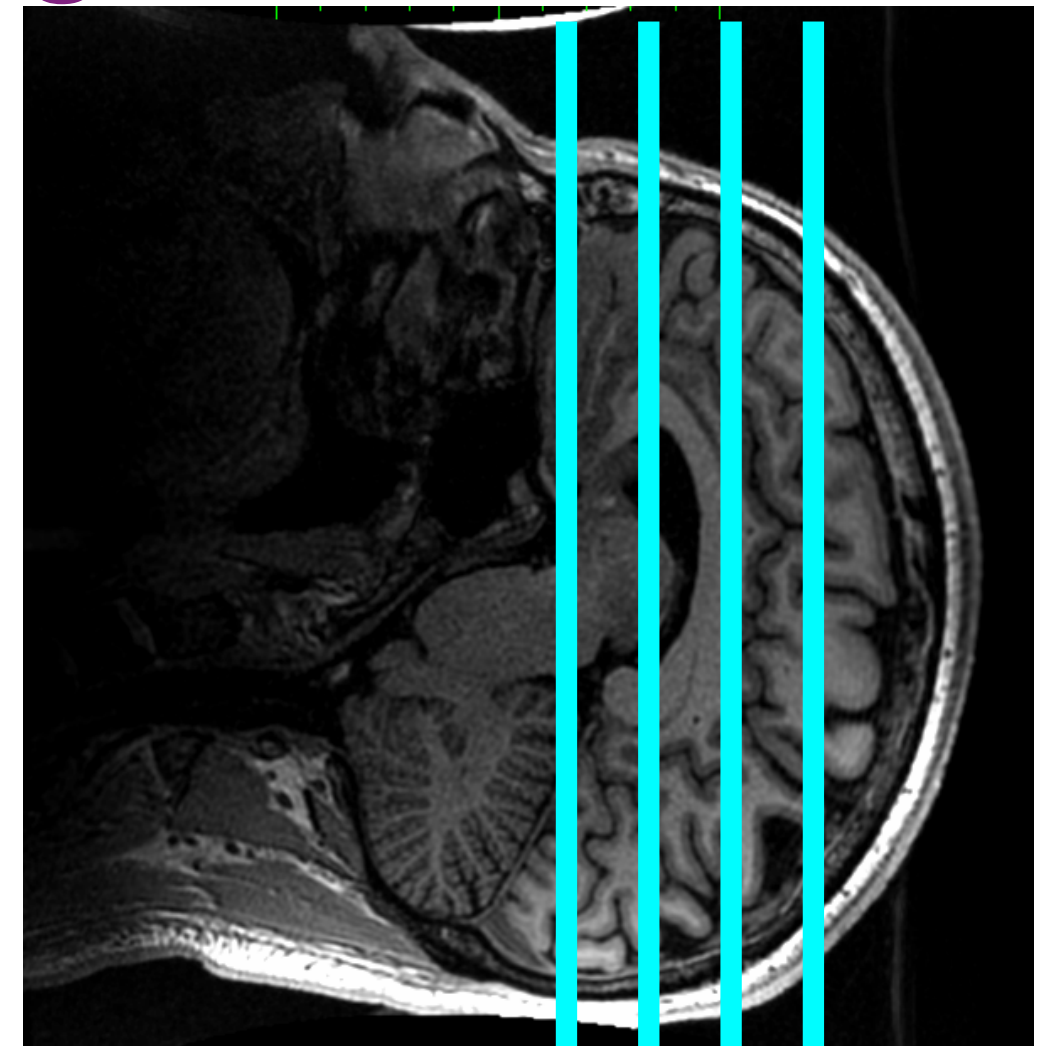
- Excite with frequency $\omega_0 + \delta\omega$ to move slice from isocenter to position of interest.
- Excite with an RF pulse that is the composition of a band of frequencies to define a particular slice width.



Multiband/SMS imaging

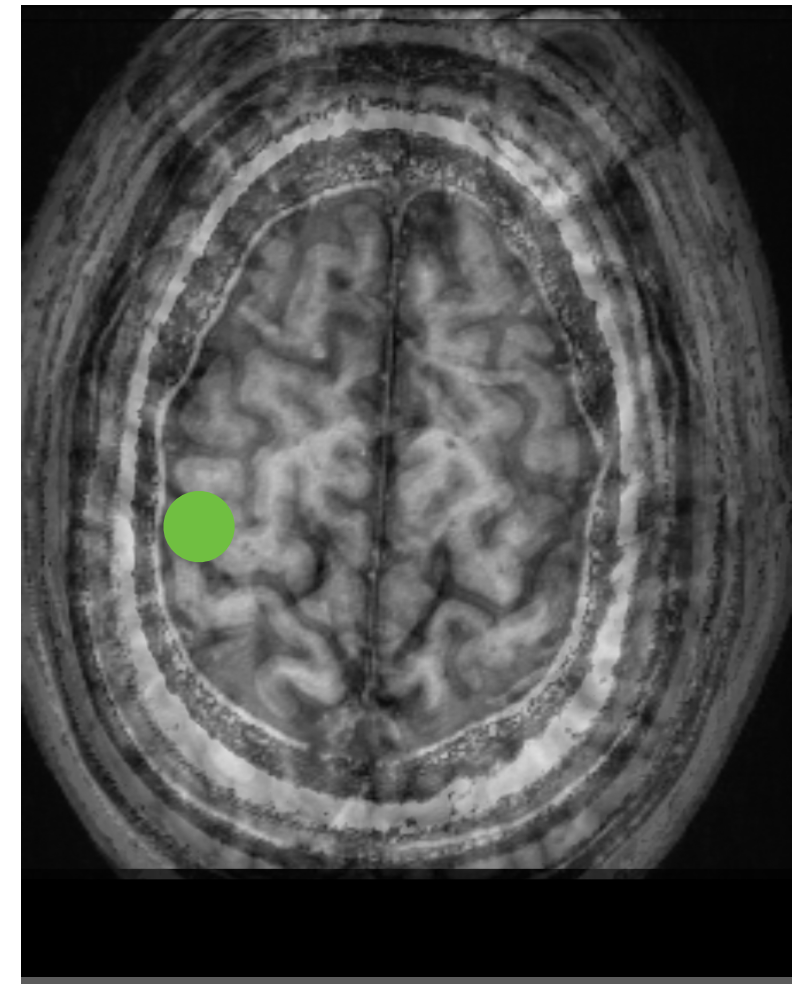
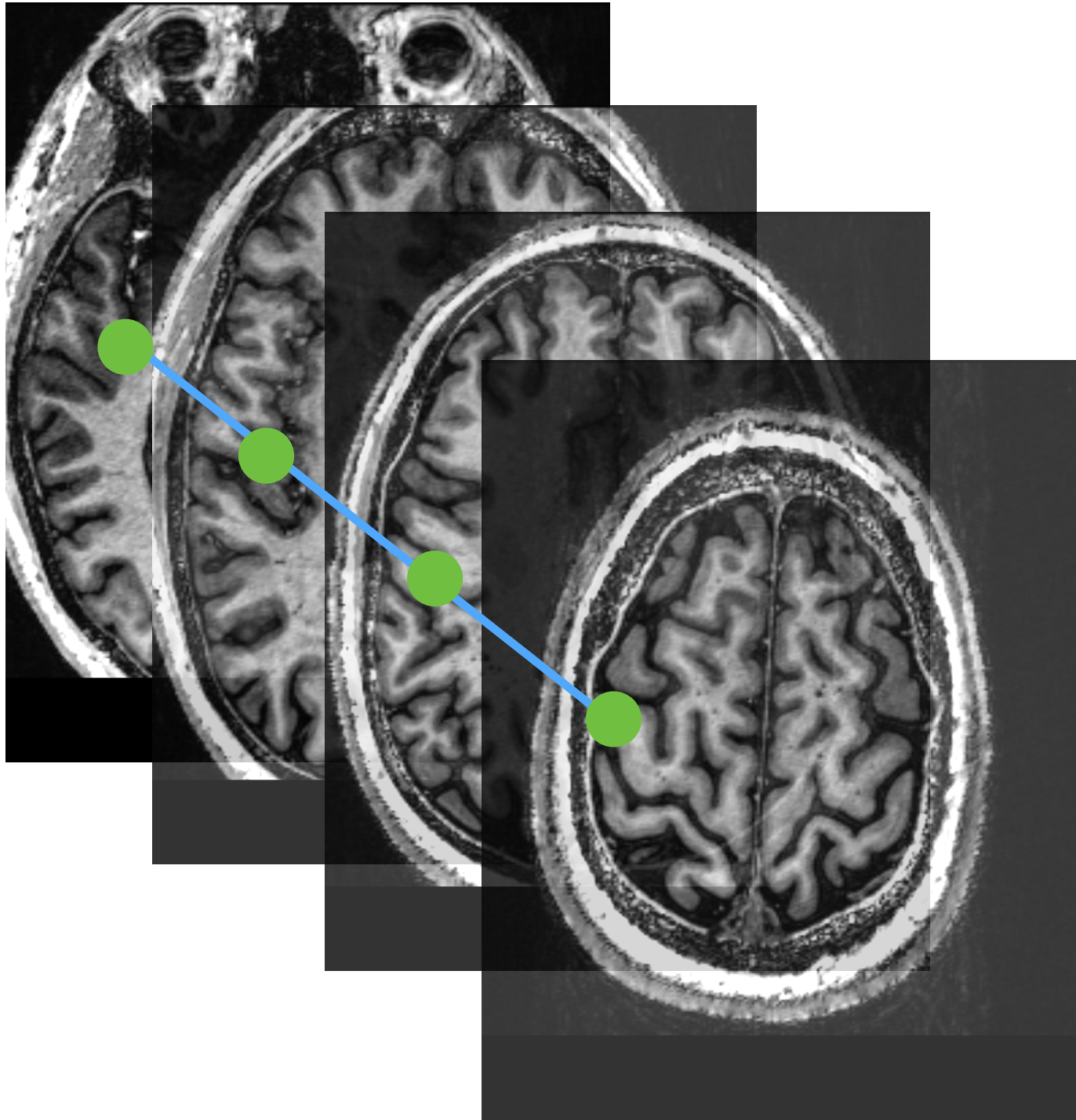
Simultaneous multi-slice imaging

- 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



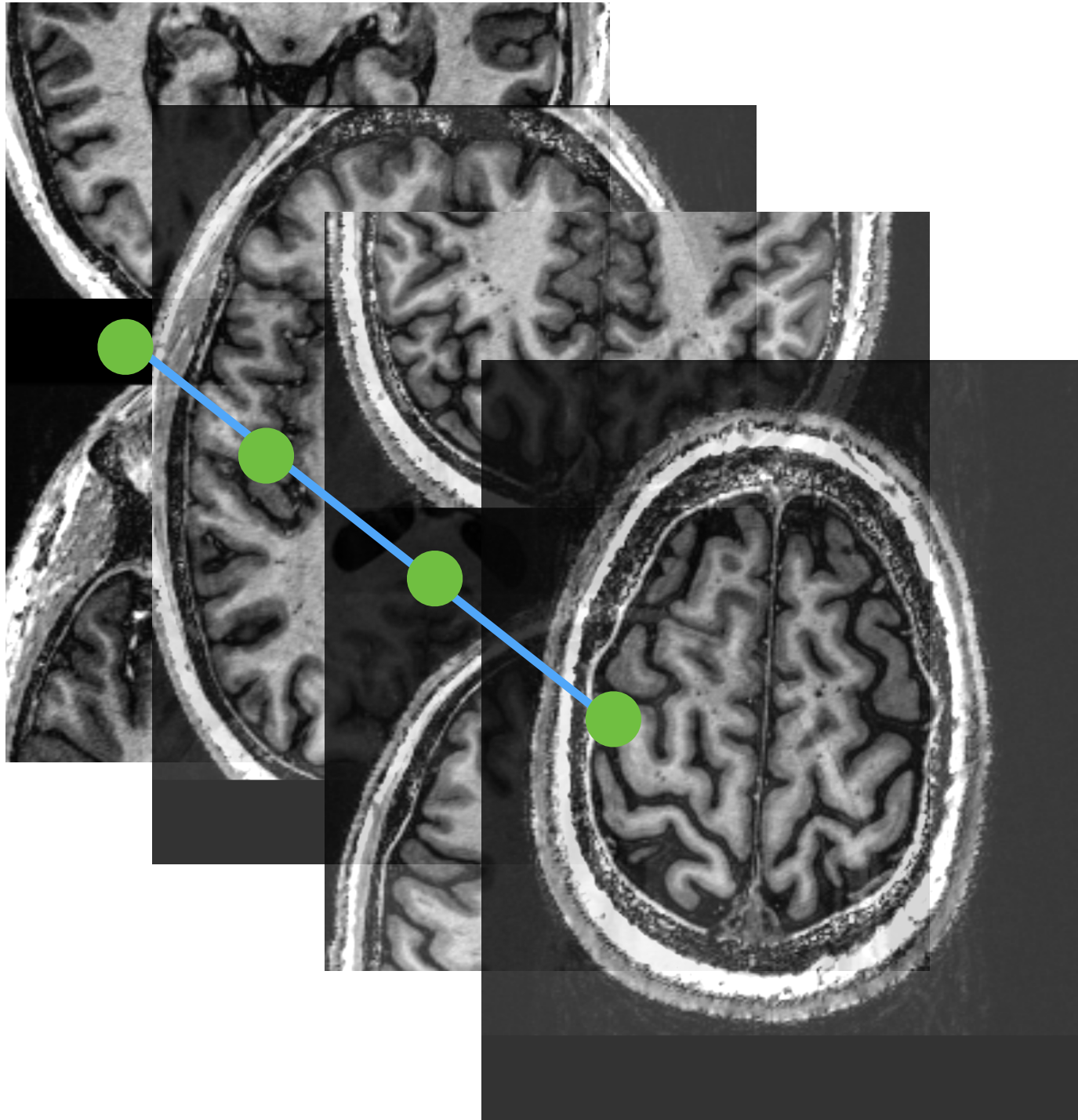
Multiband/SMS imaging

Simultaneous multi-slice imaging

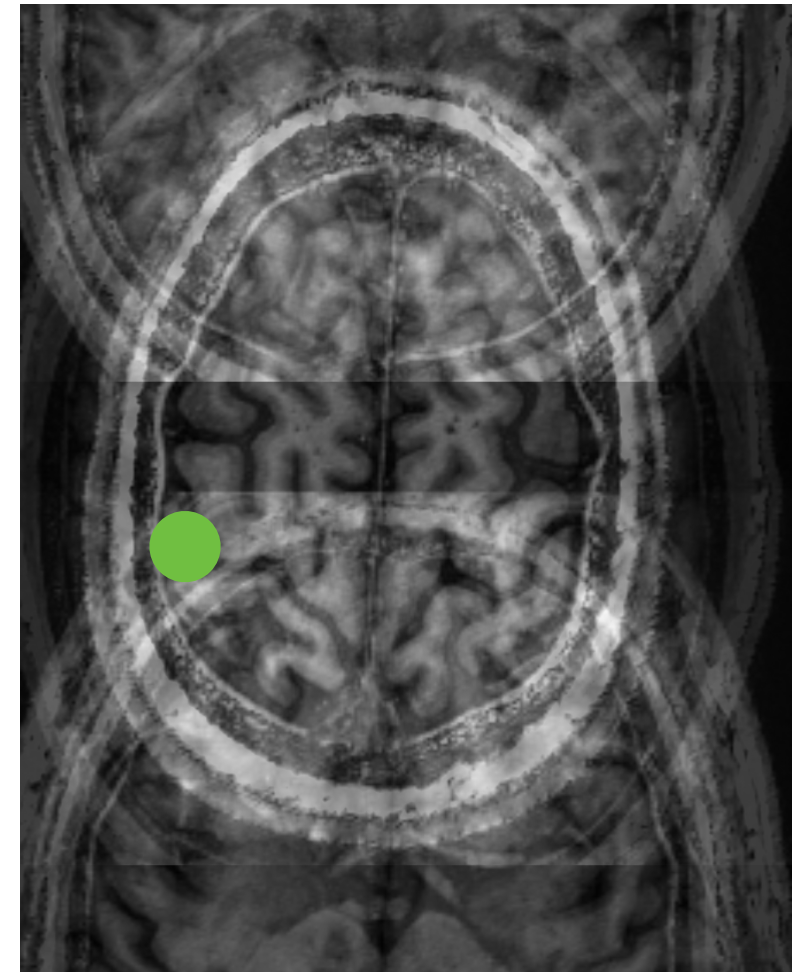


Multiband/SMS imaging

Simultaneous multi-slice imaging



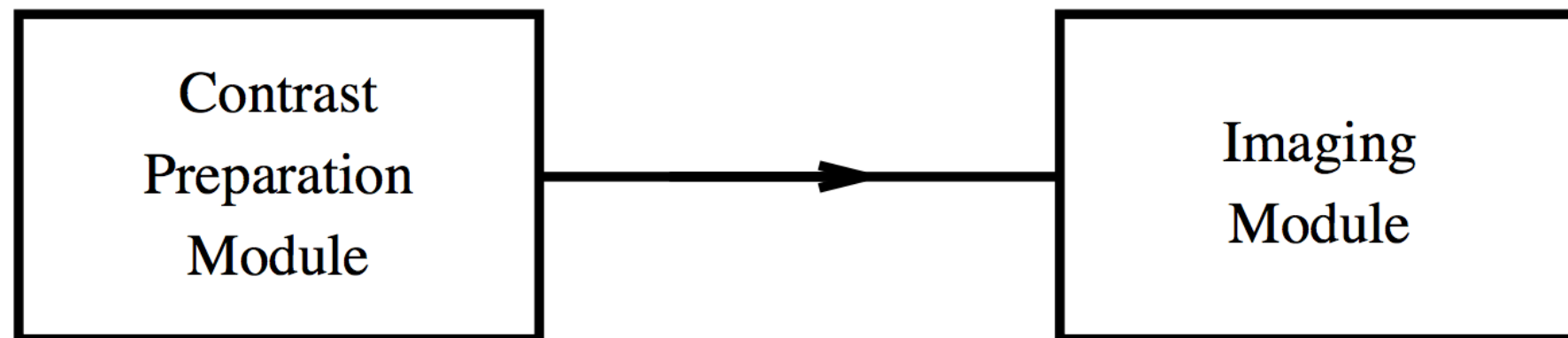
CAIPI



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

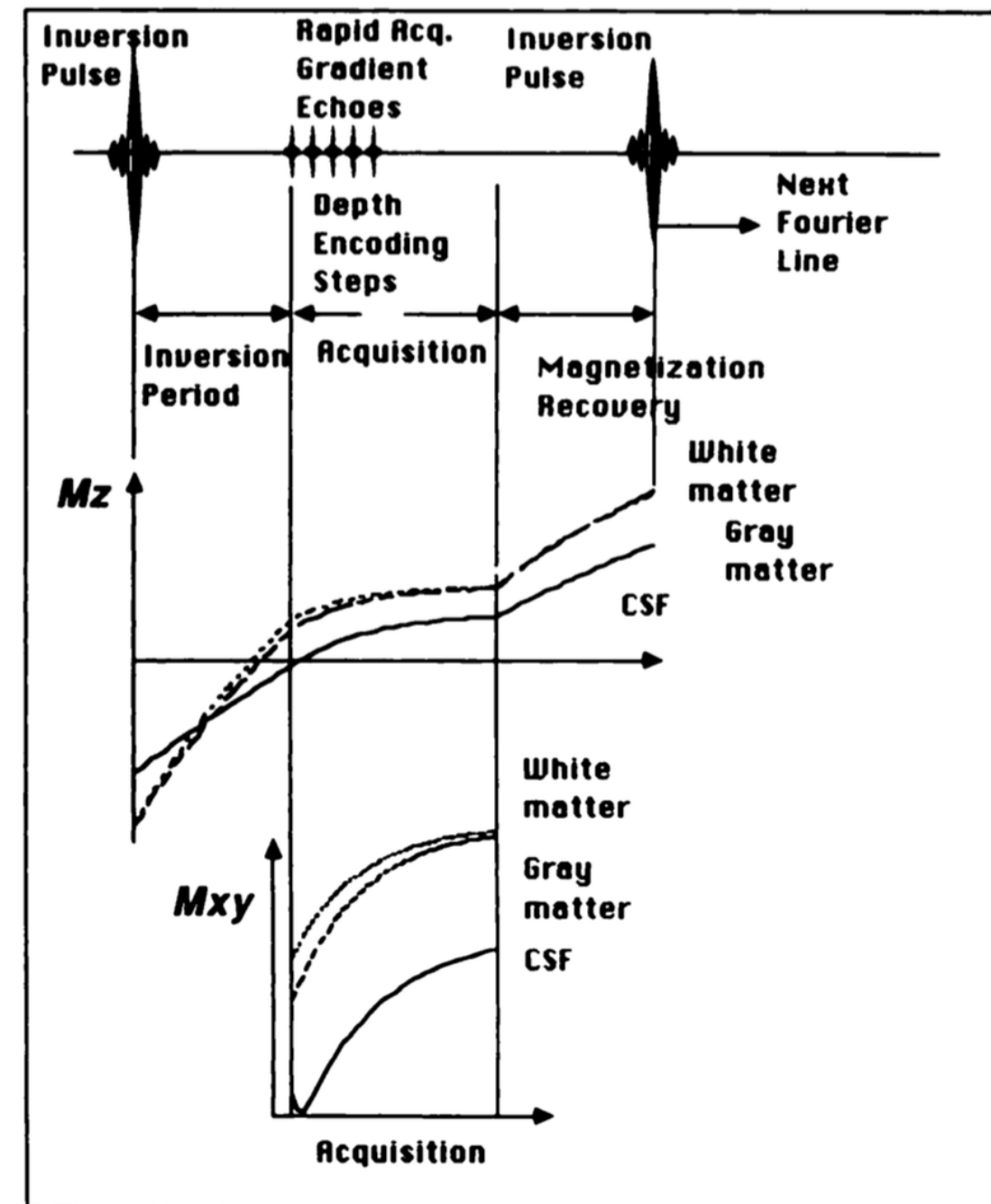
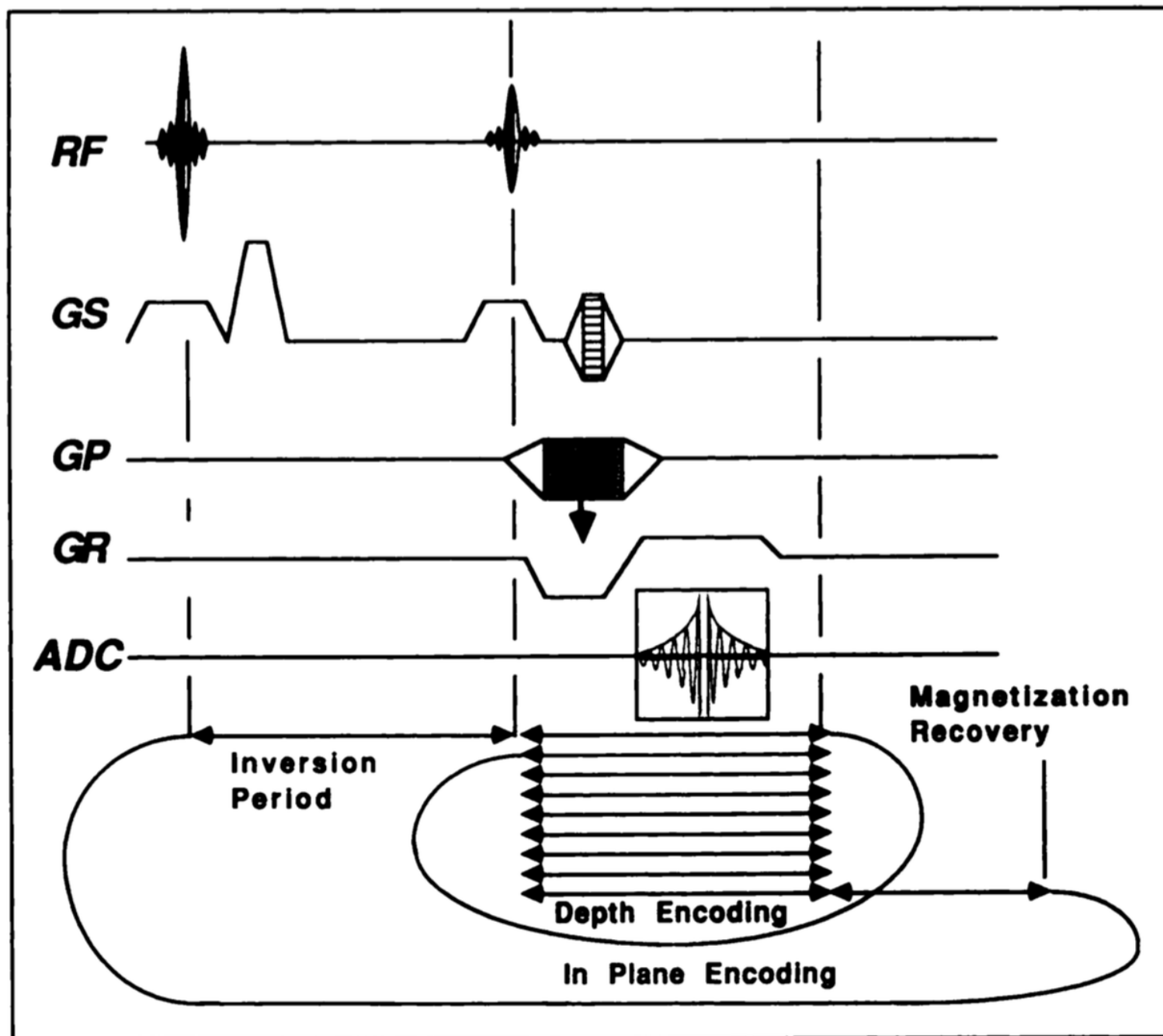
MP-RAGE

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
- Inversion recovery preparation pulse
- Multiple imaging readouts (slice direction phase encoding)

MP-RAGE

Magnetization Prepared Rapid Acquisition with Gradient Echoes



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