Nuclear Magnetic Resonance Imaging

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NSS-MIC: Fundamentals of Medical Imaging

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These annotated slides were prepared by Jeff Fessler for attendees of the NSS-MIC short course on medical imaging.

Outline

- Background
- Basic physics
- 4 magnetic fields
- Bloch equation
- Excitation
- Signal equation and k-space
- Pulse sequences
- Image reconstruction
- Summary

This presentation is a very brief introduction to the basics of NMR imaging. The main goals ... todo

History

- NMR discovered independently by Bloch and Purcell in 1946
- Discovery led to Nobel Prize in physics
- First MR images in early 1970's by Lauterbur

F. Bloch, Phys. Rev. 70:460, 1946.

Advantages of NMR Imaging

- Nonionizing radiation
- Good soft-tissue contrast
- High spatial resolution
- Flexible user control - many acquisition parameters
- Minimal attenuation
- Image in arbitrary plane (or volume)
- Potential for chemically specified imaging
- Flow imaging

The variety of degrees of freedom in acquisition parameters has led to many different imaging protocols (pulse sequences) optimized for specific imaging situations.

Disadvantages of NMR Imaging

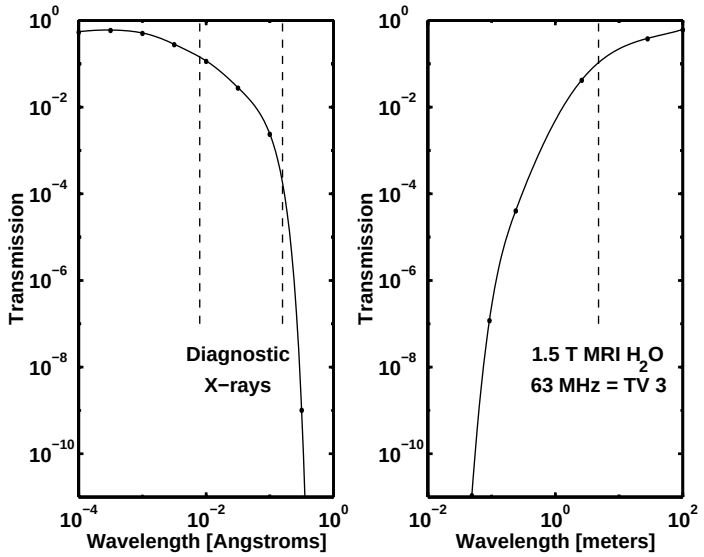
- High cost
- Complicated “siting” due to large magnetic fields
- Low sensitivity
- Little signal from bone
- Scan durations

NMR-4

Naturally there is considerable ongoing effort to reduce or minimize the disadvantages of MRI (especially scan duration).
In fact, recently “real-time” NMR images have been produced, *e.g.*:
Cox RW, Jesmanowicz A, Hyde JS, Real-time functional magnetic resonance imaging.
Magnetic Resonance in Medicine, 33(2):230-6, Feb. 1995.

Attenuation Considerations

Transmission of EM Waves Through 25cm of Soft Tissue



NMR-5

Nuclear Spins

The NMR phenomena is present in nuclei having an odd number of protons or neutrons and hence *nuclear spin angular momentum*.

Such nuclei often just called “spins”

Most abundant spin is ^{1}H, a single proton, in water molecules.

NMR concerns the interaction of these spins with magnetic fields:

1. Main field $\vec{B}_0$ (static)
2. Local field effects
   - Effect of local orbiting electrons (chemical shift)
   - Differences in magnetic susceptibility
3. RF field $\vec{B}_1(t)$ (user-controlled, amplitude-modulated pulse)
4. Gradient fields $\vec{G}(t)$ (user-controlled, time-varying)

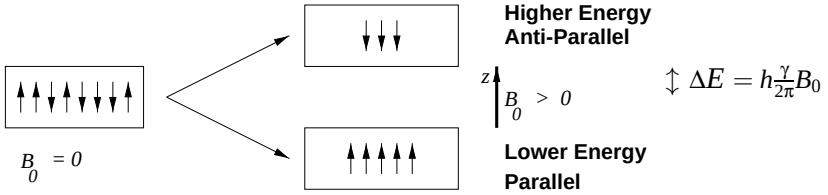
There are over 100 stable nuclei that have an odd number of protons or neutrons.
Particularly important ones: ^{2}H, ^{7}Li, ^{13}C, ^{19}F, ^{23}Na, ^{31}P, ^{127}I.

NMR images reflect properties of the local environment of the spins.

1. Main Field $\vec{B}_0$

^{1}H has two energy states: parallel and anti-parallel to main field.

Zeeman Diagram:



Thermal agitation causes random transitions between states.
Equilibrium ratio of anti-parallel (n_-) to parallel (n_+) nuclei:

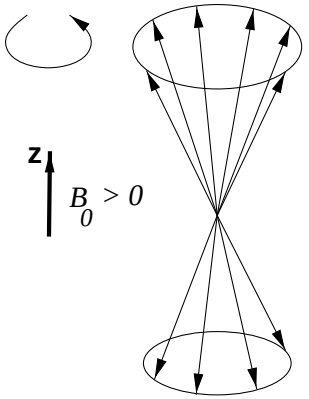
$$\frac{n_-}{n_+} = \mathrm{e}^{-\Delta E/(k_b T)} \quad \text{(Boltzmann distribution)}$$

- γ is called the *gyromagnetic ratio* ($\gamma/2\pi = 42.48$ MHz/Tesla)
- $h = 6.6{\cdot}10^{-34}$ J sec (Planck’s constant)
- $k_b = 1.38 \cdot 10^{-23}$ J/ $^\circ$K (Boltzmann constant)
- T temperature in degrees Kelvin
- Typically B_0 is about 1 Tesla = 10^4 Gauss (earth’s magnetic field is about 0.5 Gauss).

With no field applied ($\vec{B}_0 = 0$) the two energy states are equally likely.
At absolute zero, all nuclei would occupy the lower energy (parallel) state.
At room temperature, the difference in occupancy is only a few parts per million.
Thus NMR is a fairly insensitive modality (fortunately there is a lot of ^{1}H).
To increase sensitivity (signal), one can either decrease temperature (not clinically feasible, but fine for traditional spectroscopy) or increase B_0 (which has tradeoffs w.r.t. T_1).

Precession (Classical Description)

Spins do not align exactly parallel or anti-parallel to z.
A spin's magnetic moment experiences a torque, causing precession.



Lower Energy State (Parallel)

M_0 **Net Magnetization of Material**

Higher Energy State (Anti-Parallel)

Larmor Equation

Precession frequency is proportional to (local) field strength:

$$\boxed{\omega = \gamma |\vec{B}|}$$

- γ is called the *gyromagnetic ratio*
- $\gamma/2\pi = 42.48$ MHz/Tesla for ^{1}H
- At $B_0 = 1.5$T, $f_0 = \frac{\gamma}{2\pi} B_0 \approx 63$ MHz (TV Channel 3)

At 37° C = 310° K, M_0 for protons in H_2O is 3.25 $\cdot 10^{-3}$ B_0 A / m.

Prior to applying RF excitation pulses, the spatial distribution of magnetization is essentially proportional to the local density $\rho(x,y,z)$ of ^{1}H, *i.e.*:

$$\vec{M}_0(x,y,z) = \begin{bmatrix} 0 \\ 0 \\ M_0 \end{bmatrix} \rho(x,y,z) = M_0 \rho(x,y,z) \vec{k}.$$

An image proportional to $\rho(x,y,z)$ would be of some interest, but would barely scratch the surface of the potential of NMR imaging.

2.1 Local Field Effect: Chemical Shift

Orbital electrons surrounding a nuclei perturb the local magnetic field:

$$B_{\text{eff}}(\vec{r}) = B_0(1 - \sigma(\vec{r})),$$

where $\sigma(\vec{r})$ depends on local electron environment.

Because of the Larmor relationship, this field shift causes a shift in the local resonant frequency:

$$\omega_{\text{eff}} = \omega_0(1 - \sigma).$$

Example:
the resonant frequency of ^{1}H in fat is about 3.5 ppm lower than in water.

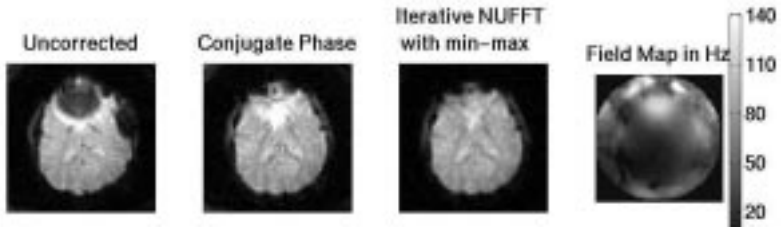
Chemical shift causes artifacts in usual “frequency encoded” imaging.

Later we discuss how spatial location is encoded in temporal frequency by applied gradients. So any unrelated shift in resonant frequency, such as chemical shift, causes a corresponding undesired apparent shift in spatial location.

2.2 Local Field Effect: Susceptibility

Differences in the magnetic susceptibility of different materials, particularly air and water, cause local perturbations of the magnetic field that also cause local shifts in the resonance frequency.

This effect can cause undesirable signal loss.



It can also be a source of contrast, such as the BOLD effect in fMRI.

See [1] for our correction method and a review of previous correction methods.

3. RF Field $\vec{B}_1(t)$

The first step in any NMR experiment is to perturb the spins away from equilibrium using an RF pulse.

Amplitude-modulated RF pulse circularly polarized in x,y plane:

$$\vec{B}_1(t) = B_1 \begin{bmatrix} \cos\omega_0 t \\ -\sin\omega_0 t \\ 0 \end{bmatrix} p(t).$$

Spins now precess about the *total* time-varying magnetic field:

$$\vec{B}(t) = \vec{B}_0 + \vec{B}_1(t).$$

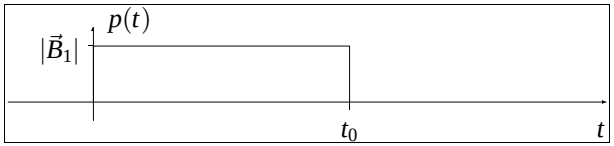
$|\vec{B}_1| \ll |\vec{B}_0|$

"RF excitation" puts energy into the system.

RF amplitude B_1 is fraction of a Gauss.
Pulse $p(t)$ is a few msec long, and often is approximately a truncated sinc pulse.
Note that $\Delta E = h f_0$, which is exactly the formula for the energy in the quanta of an EM field with frequency f_0. So RF tuned to f_0 will resonate with spins.
In the rotating frame, the magnetization precesses about the x' axis at a rate γB_1.
For $B_1 = 0.1$G, this rate is about 426 Hz, so a pulse of duration about 0.6 msec tips the magnetization into the $x'y'$ plane.

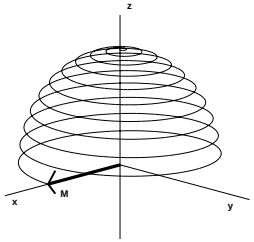
RF Excitation Example

Non-selective excitation:

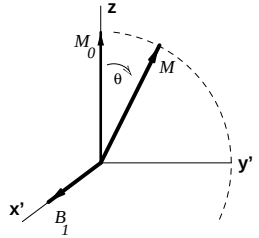


Time evolution of local magnetization $\vec{M}(t)$

Laboratory Frame

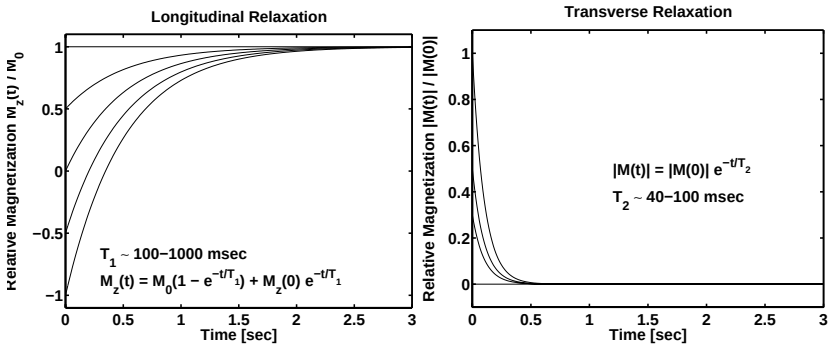


Rotating Frame



Relaxation

After RF excitation, the magnetization $\vec{M}(t)$ returns to its equilibrium exponentially, releasing (some of) the energy put in by the excitation.



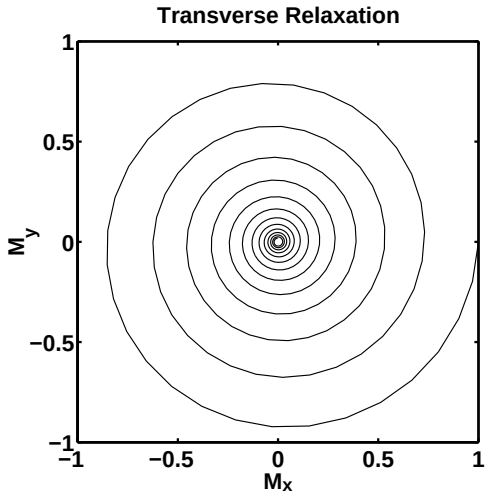
- T_1: spin-lattice time constant. Long T_1 slows imaging :-(
- T_2: spin-spin time constant
- $T_1 \neq T_2 \Rightarrow |\vec{M}(t)|$ not constant

Longitudinal relaxation is due to exchange of energy from spins to surrounding lattice.
T_1 values lengthen with increasing B_0

Transverse relaxation caused by loss of phase coherence of the spins due to interactions between nearby microscopic dipoles causing a broadening of the spins' resonant frequencies.
T_2 values largely independent of B_0

These relaxation parameters are material dependent, and are a primary source of the exquisite image contrast in MR.

Relaxation of Transverse Magnetization



Signal!

As spins return to equilibrium, measurable energy is released.

By Faraday's law, precessing magnetic moment induces current in RF receive coil:

$$s_r(t) = \int_{\text{vol}} b_{\text{coil}}(\vec{r}) \frac{\partial}{\partial t} M(\vec{r},t)\, \mathrm{d}\vec{r},$$

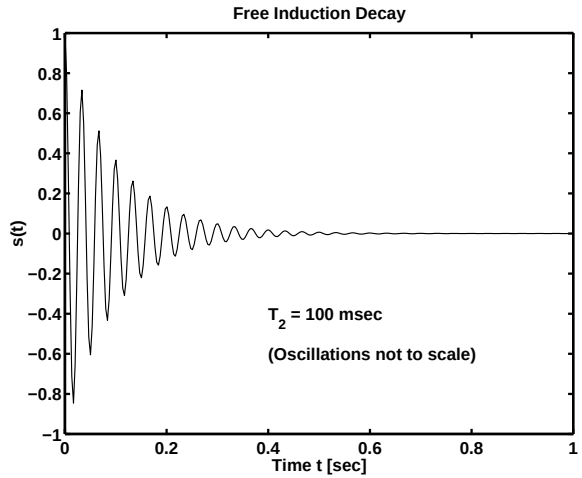
- *transverse* magnetization: $M(\vec{r},t) \triangleq M_{\text{x}}(\vec{r},t) + \imath M_{\text{y}}(\vec{r},t)$
- Transverse RF coil sensitivity pattern: $b_{\text{coil}}(\vec{r})$.
- Higher spin density or larger $\omega_0 \Rightarrow$ more signal.

Spatial localization? Not much!

Small coils have a *somewhat* localized spatial sensitivity pattern. This property has been exploited recently in "sensitivity encoded" (SENSE) imaging, to reduce scan times.

Think of each spin as a transmitting oscillator; RF coil measures superposition of signals from each.
In practice the RF sensitivity is nonuniform so a (spatially slowly varying) weighting term should be included in the above integral.

Free-Induction Decay



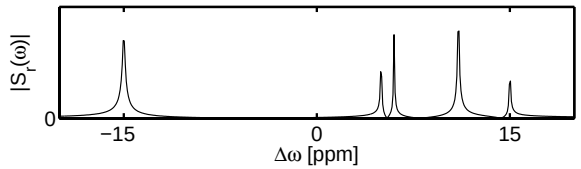
NMR Spectroscopy

Material with ^{1}H having several relaxation rates and chemical shifts.

Received FID signal: $s_r(t) = \sum_{n=1}^{N} \rho_n \, e^{-t/T_{2,n}} e^{\imath \omega_n t}, \qquad t \geq 0$

- ρ_n: spin density of nth species
- $T_{2,n}$: relaxation time of nth species
- ω_n: resonant frequency of nth species

Fourier transform: $S_r(\omega) = \sum_{n=1}^{N} \rho_n \frac{1}{1 + \imath(\omega - \omega_n) T_{2,n}}$



3. Gradient Fields

For a uniform main field $\vec{B}_0$, all spins have (almost) the same resonant frequency. For *imaging* we must "encode" spin spatial position.

Frequency encoding relates spatial location to (temporal) frequency by using gradient coils to induce a field gradient, *e.g.*, along x: a gradient along the x direction:

$$\vec{B}(x,y,z) = \begin{bmatrix} 0 \\ 0 \\ B_0 + x G_{\mathrm{x}} \end{bmatrix} = (B_0 + x G_{\mathrm{x}}) \vec{k}.$$

Here, x location becomes "encoded" through the Larmor relationship:

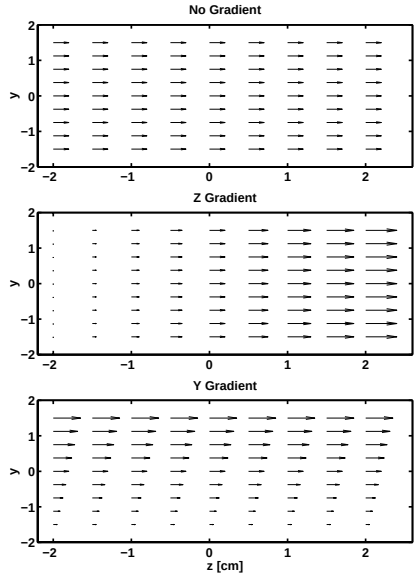
$$\omega(x,y,z) = \gamma |\vec{B}(x,y,z)| = \gamma (B_0 + x G_{\mathrm{x}}) = 2\pi \left(f_0 + x \frac{\partial f}{\partial x} \right), \quad \frac{\partial f}{\partial x} = \frac{\gamma}{2\pi} G_{\mathrm{x}}.$$

- $B_0 \approx 10^4$ Gauss $\Rightarrow$ $f_0 \approx 60$ MHz
- $G_{\mathrm{x}} \approx 1$ Gauss/cm $\Rightarrow \frac{\partial f}{\partial x} \approx 6$ KHz / cm

The local resonant frequency is proportional to the *strength* of the local magnetic field. A field gradient means that some spins experience a stronger field than others, hence have different resonant frequencies.

Now one can use Fourier decomposition of the received signal to determine the magnetization in different spatial locations.

Effect of gradient fields: $\vec{B}(0,y,z)$



This figure is greatly exaggerated since $G_x x_{max}$ is a factor of about 10^3 smaller than $\vec{B}_0$.

Bloch Equation

Phenomenological description of time evolution of local magnetization $\vec{M}(\vec{r},t)$:

$$\frac{d\vec{M}}{dt} = \vec{M} \times \gamma \vec{B} - \frac{M_x \vec{i} + M_y \vec{j}}{T_2} - \frac{(M_z - M_0)\vec{k}}{T_1}$$

Precession	↑		
Relaxation		↑	↑
Equilibrium			↑

Driving term $\vec{B}(\vec{r},t)$ includes

- Main field $\vec{B}_0$
- RF field $\vec{B}_1(t)$
- Field gradients $\vec{r} \cdot \vec{G}(t) = xG_x(t) + yG_y(t) + zG_z(t)$

$$\vec{B}(\vec{r},t) = \vec{B}_0 + \vec{B}_1(t) + \vec{r} \cdot \vec{G}(t)\vec{k}$$

$\vec{i}$, $\vec{j}$, and $\vec{k}$ are unit vectors in x, y, and z directions respectively.

Although not explicitly notated above, everything is a function of spatial coordinates and time, *i.e.*, $\vec{M}(\vec{r},t)$.

No general solution to the Bloch equation is known for an arbitrary applied field $\vec{B}(t)$.

Excitation Considerations

Excitation

- Selective (for slice selection)
- Non-selective (affects entire volume)

Design parameters (functions!):

- RF amplitude signal $B_1(t)$
- Gradient strengths $\vec{G}(t)$

Non-Selective Excitation

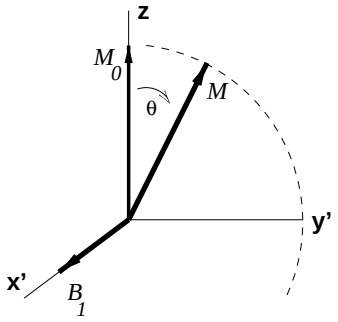
- No gradients: excite entire volume
- $\vec{B}_1(t) = B_1(t)(\vec{i}\cos\omega_0 t - \vec{j}\sin\omega_0 t)$ (circularly polarized)
- Ignore relaxation

Solution to Bloch equation:

$$\vec{M}_{\text{rot}}(t) = R_{\text{x}}\left(\int_0^t \omega_1(s)\,\text{ds}\right)\vec{M}_{\text{rot}}(0)$$

$$\omega_1(t) = \gamma B_1(t)$$

$$R_{\text{x}}(\theta) = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\theta & \sin\theta \\ 0 & -\sin\theta & \cos\theta \end{bmatrix}$$

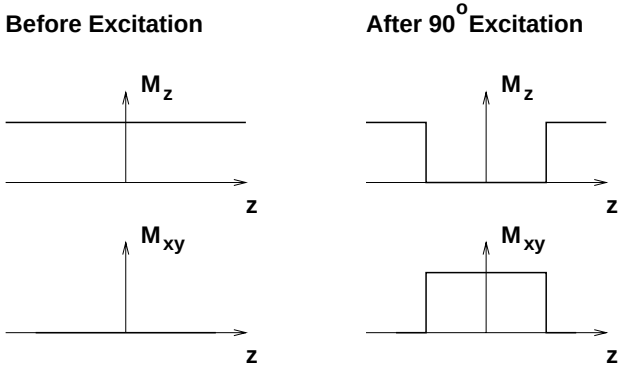


Typically $\theta = 90^\circ$ or $\theta = 180^\circ$.

Ignoring relaxation is reasonable since RF pulse duration $\approx$ 1 msec $<< T_1, T_2$.

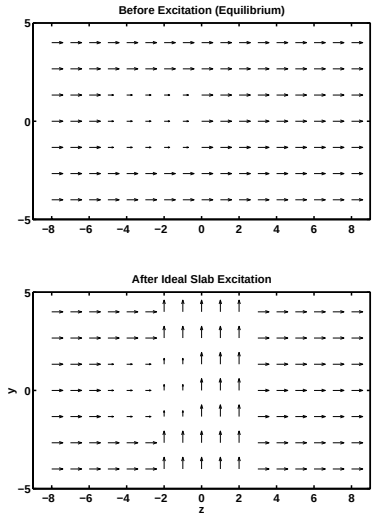
Selective Excitation

Ideal slice profile:



Requires non-zero gradient so the resonant frequencies vary with z.

90° Selective Excitation



This figure shows magnetization vector $M(\vec{r})$ before and after excitation.

RF Pulse Design: Small Tip-Angle Approximation

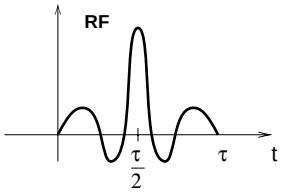
Assumptions:

- $\theta < 30^\circ$ (weak RF pulse)
- $M_z(t) \approx M_0$
- $d/dt M_z(t) \approx 0$
- Constant gradient in z direction

(Approximate) solution to Bloch equation:

$$M(z) \propto F_1\{B_1(t)\}\Big|_{f=z\frac{\gamma}{2\pi}G_z}$$

For a rectangular slice profile
we a need sinc-like RF pulse:



Can use this relationship to design $B_1(t)$ to achieve desired slice profile. Ideally both $B_1(t)$ and slice profile have finite support, which is impossible. Thus choosing $B_1(t)$ is like an FIR (since $[0,\tau]$) filter design problem.

This Fourier-transform relationship between slice profile and RF spectrum can also be understood qualitatively. The z gradient maps spins in different slices to different resonant frequencies. The spins in a given slice will be excited by the part of the RF spectrum corresponding to the resonant frequency for that slice.

By using a combination of gradients instead of just a z gradient, one can excite arbitrary oblique slices.

Remarkably, even though the mathematical derivation holds only for small tip angles, the above relationship yields good pulse designs even for tips of up to 90°. For 180° tip angles however, more sophisticated design methods are needed to provide good slice profiles [2–4].

Receive Considerations

Fundamental Equation of MR

After excitation, magnetization continues to evolve via Bloch equation.

Solve Bloch equation for RF=0 to find evolution of transverse magnetization: $M(\vec{r},t) = M_{\mathrm{X}}(\vec{r},t) + \imath M_{\mathrm{Y}}(\vec{r},t)$.

Fundamental Equation of MR Imaging

$$M(\vec{r},t) \;=\; m_0(\vec{r})\; \mathrm{e}^{-\imath\omega_0 t}\; \mathrm{e}^{-t/T_2(\vec{r})}\; \mathrm{e}^{-\imath\phi(\vec{r},t)}$$

Post-excitation ↑
Precession ↑
Relaxation ↑
User-controlled phase term ↑

$$\phi(\vec{r},t) = \gamma \int_0^t \vec{r}\cdot\vec{G}(\tau)\,\mathrm{d}\tau.$$

- Key design parameters are gradients: $\vec{G}(t)$.
- How to manipulate the gradients to make an image of $m_0(\vec{r})$?

$M = M_{\mathrm{X}} + \imath M_{\mathrm{Y}}$ is a complex representation of the transverse component of the magnetization.

$m_0(\vec{r})$ denotes the transverse magnetization immediately after excitation, *i.e.*, $m_0(\vec{r}) = M(\vec{r},0^+)$.

If no gradients were used, then the phase ϕ would be zero, so (ideally) all spins in the volume would precess at the same frequency.

In practice, the phase ϕ is not entirely user-controlled, since off-resonance effects like chemical shift and susceptibility also influence $\phi(\vec{r},t)$.

Received Signal

From Faraday's law (assuming uniform receive coil sensitivity):

$$s_r(t) = \int_{\text{vol}} \frac{\partial}{\partial t} M(\vec{r},t)\,\mathrm{d}\vec{r}.$$

Using the Fundamental Equation of MR and converting to baseband:

$$s(t) = s_r(t)\,\mathrm{e}^{-\imath\omega_0 t} = \int_{\text{vol}} m_0(\vec{r})\,\mathrm{e}^{-t/T_2(\vec{r})}\,\mathrm{e}^{-\imath\phi(\vec{r},t)}\,\mathrm{d}\vec{r}.$$

Disregarding T_2 decay:

$$\boxed{s(t) = \int_{\text{vol}} m_0(\vec{r})\,\mathrm{e}^{-\imath\phi(\vec{r},t)}\,\mathrm{d}\vec{r},}$$

where

$$\phi(\vec{r},t) = \gamma \int_0^t \vec{r}\cdot\vec{G}(\tau)\,\mathrm{d}\tau.$$

$\therefore$ The baseband signal is a phase-weighted volume integral of $m_0(\vec{r})$.

The received signal is proportional to the volume integral of the electromotive force (EMF) induced by the time-changing magnetic field of the spins.

Assume ideal RF receiver
- Uniformly sensitive over volume of interest (B_{1xy})
- Detects flux changes in transverse direction

$$s_r(t) = \int_{\text{vol}} B_{1xy}\frac{\partial}{\partial t}M(\vec{r},t)\,\mathrm{d}\vec{r} = \iiint B_{1xy}\frac{\partial}{\partial t}M(x,y,z,t)\,\mathrm{d}x\,\mathrm{d}y\,\mathrm{d}z$$

$$= \int_{\text{vol}} B_{1xy}\frac{\partial}{\partial t}\left[m_0(\vec{r})\,\mathrm{e}^{-t/T_2(\vec{r})}\,\mathrm{e}^{-\imath\omega_0 t}\,\mathrm{e}^{\imath\phi(\vec{r},t)}\right]\mathrm{d}\vec{r} \approx -\imath\omega_0 B_{1xy}\,\mathrm{e}^{-\imath\omega_0 t}\int_{\text{vol}} m_0(\vec{r})\,\mathrm{e}^{-t/T_2(\vec{r})}\,\mathrm{e}^{\imath\phi(\vec{r},t)}\,\mathrm{d}\vec{r}$$

Baseband signal:

$$s(t) = \frac{\mathrm{e}^{\imath\omega_0 t}}{-\imath\omega_0 B_{1xy}} s_r(t) = \int_{\text{vol}} m_0(\vec{r})\,\mathrm{e}^{-t/T_2(\vec{r})}\,\mathrm{e}^{\imath\phi(\vec{r},t)}\,\mathrm{d}\vec{r}$$

Signal Equation

Rewriting baseband signal:

$$\boxed{s(t) = \iiint m(x,y,z)\,\mathrm{e}^{-\imath 2\pi[xk_{\mathrm{X}}(t)+yk_{\mathrm{Y}}(t)+zk_{\mathrm{Z}}(t)]}\,\mathrm{d}x\,\mathrm{d}y\,\mathrm{d}z}$$

where the "k-space trajectory" is defined in terms of the gradients:

$$\begin{aligned} k_{\mathrm{X}}(t) &= \frac{\gamma}{2\pi}\int_0^t G_{\mathrm{X}}(\tau)\,\mathrm{d}\tau \\ k_{\mathrm{Y}}(t) &= \frac{\gamma}{2\pi}\int_0^t G_{\mathrm{Y}}(\tau)\,\mathrm{d}\tau \\ k_{\mathrm{Z}}(t) &= \frac{\gamma}{2\pi}\int_0^t G_{\mathrm{Z}}(\tau)\,\mathrm{d}\tau. \end{aligned}$$

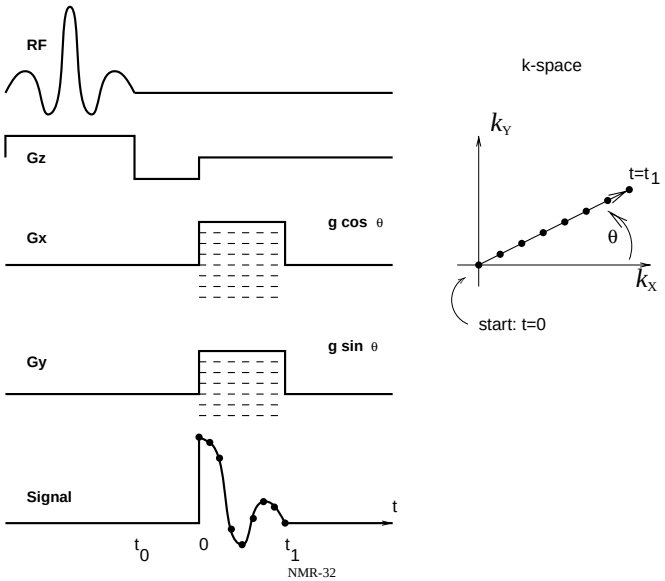
$\Rightarrow$ MRI measures samples of the FT of the object's magnetization.

(Ignoring relaxation and off-resonance effects.)

At time t, the value of the baseband signal $s(t)$ is proportional to a particular component, namely $k_{\mathrm{X}}(t), k_{\mathrm{Y}}(t), k_{\mathrm{Z}}(t)$ of the 3DFT of the transverse magnetization of the object.

To make an image, we must choose the gradient functions to "scan" through enough of k-space to collect a sufficient number of samples of the object's spectrum from which the object can be (approximately) reconstructed.

2D Projection-Reconstruction Pulse Sequence



Using selective excitation reduces the problem from 3D to 2D. Now we just need to choose the x and y gradients $G_X(t)$ and $G_Y(t)$ to scan through 2D k-space. Perhaps the simplest way to do this is to use constant gradient functions as pictured above. Then

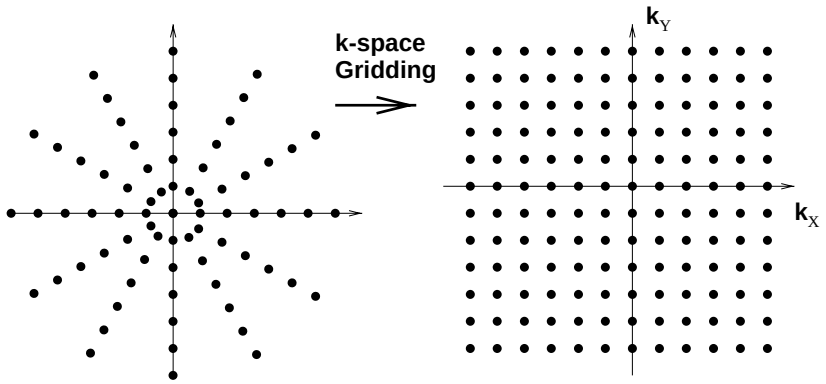
$$k_X(t) = \frac{\gamma}{2\pi}\int_0^t G_X(\tau)\,d\tau = t\frac{\gamma}{2\pi}g\cos\theta$$
$$k_Y(t) = \frac{\gamma}{2\pi}\int_0^t G_Y(\tau)\,d\tau = t\frac{\gamma}{2\pi}g\sin\theta$$

so the trajectory in k-space is just radial lines emanating from the origin.

By repeating the pictured pulse sequence for several values of θ, a "complete" polar sampling of the 2DFT of the object is collected. These samples can be gridded onto a rectangular coordinate system and then the object can be reconstructed by taking an inverse 2D FFT.

Between the acquisition of each radial line in k-space, one must wait for T_1 relaxation to equilibrium (100s of msec). This is the scan-time limitation with such an approach.

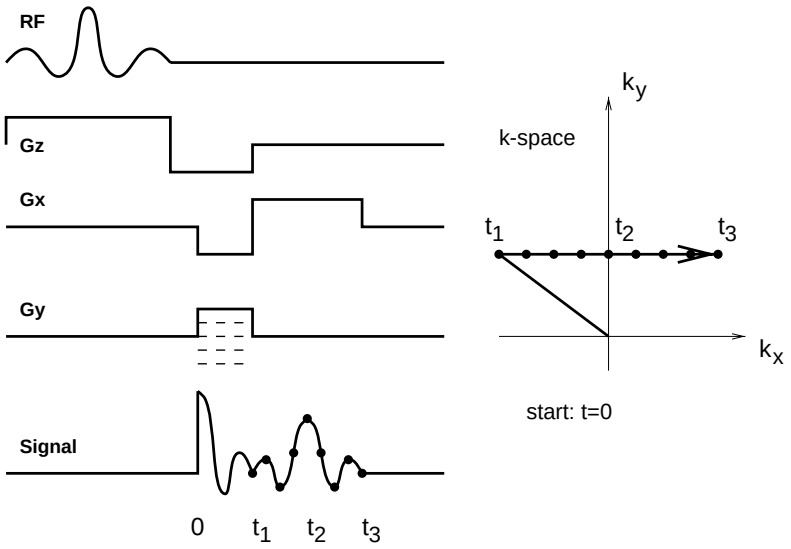
Gridding



Followed by inverse 2D FFT to reconstruct image of magnetization.

Alternatively, one can apply the filtered backprojection algorithm.

Spin-Warp Imaging (Cartesian k-space)



Now the raw data consists of samples of the 2DFT of the object on a Cartesian grid, so one can reconstruct directly using an inverse 2D FFT.
This reconstruction "problem" is quite simple compared to computed tomography!

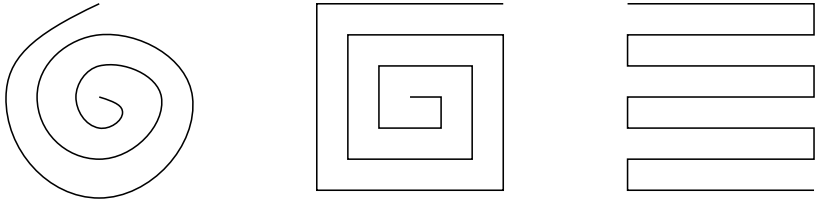
Since the measured data consists of only a finite number of samples of the 2DFT, there will be "aliasing" in object space. However, since (most!) imaging subjects are space-limited, aliasing wrap-around can be avoided by proper choice of the sample spacing in k-space: (FOV=$1/\Delta k$).

Collecting samples further from the origin of k-space leads to improved spatial resolution, but generally at a price of increased scan time, noise variance, and processing time.

In this example, G_Y is called the "phase encoding" gradient.

Fast Imaging Methods

Spiral **Square Spiral** **Echo Planar**

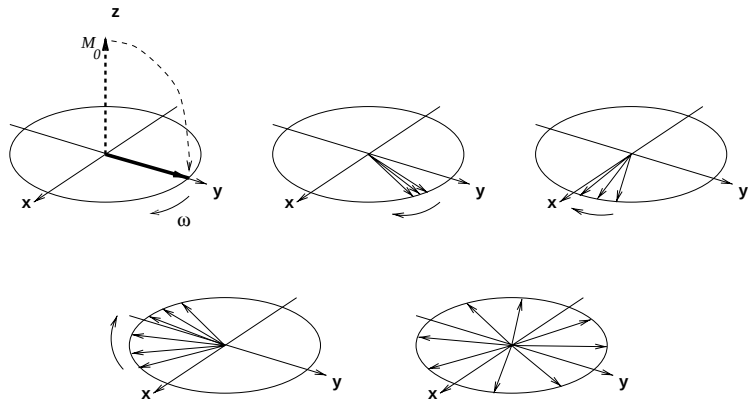


Problems
- irregular sampling in k-space so hard "gridding"
- T_2^* decay (ignored in derivation) means need small t so large gradients and slew rate.

One compromise: interleaved spirals

Off-Resonance Effects

Main field inhomogeneity, susceptibility variations, chemical shift.

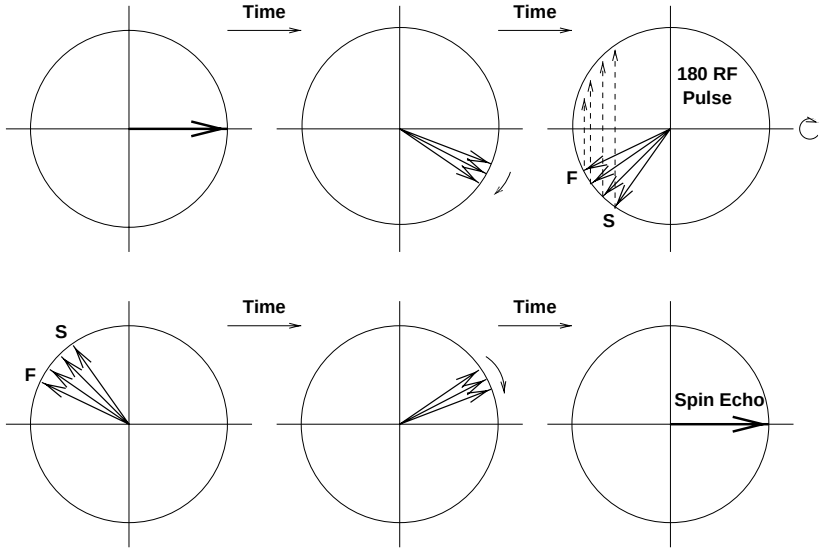


Dephasing causes destructive interference and signal loss.

Susceptibility variations are especially prevalent at air/tissue boundaries.

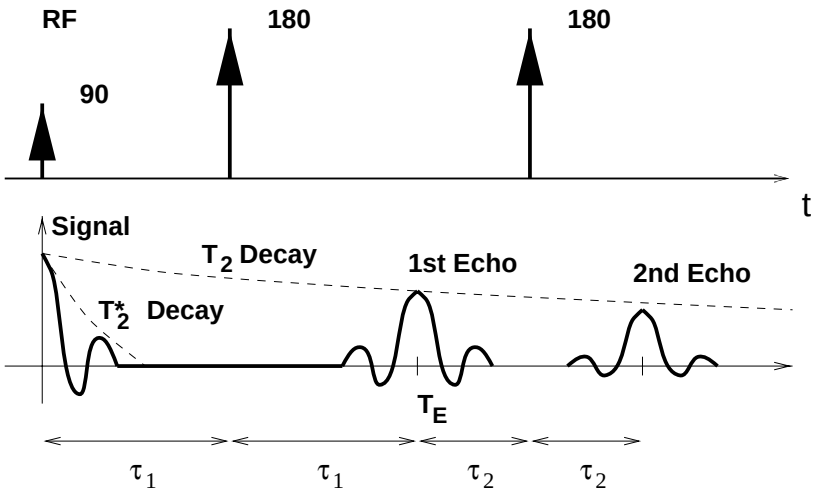
The inhomogeneous magnetic field causes spatial variations in resonant frequency. Thus some spins precess faster than others.

Spin Echo



If the 180° pulse is applied at time τ, then the spin echo occurs at time $T_E = 2\tau$.

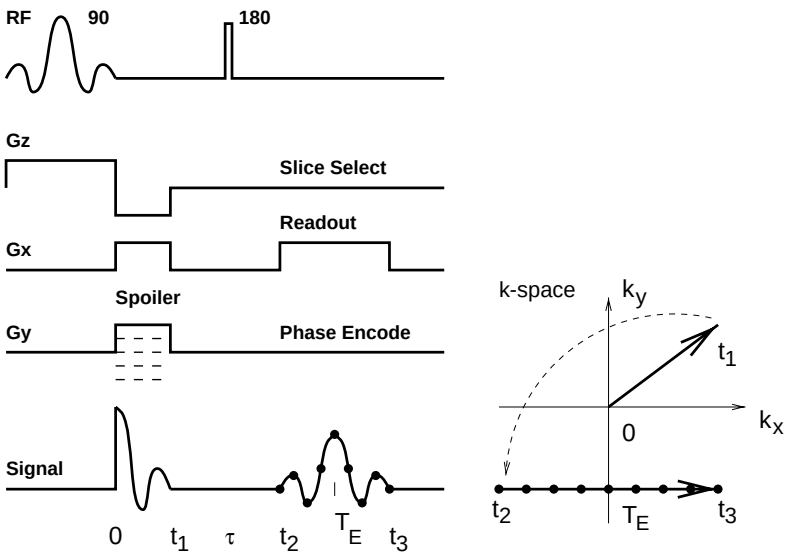
Spin Echo and Signal



The signal is now weighted by T_2, which can be desirable since T_2 can contain diagnostic information.

For more "T_2 weighting" use longer T_E.

2DFT Spin-Echo Spin-Warp Pulse Sequence



Only change to previous 2DFT sequence is addition of 180° pulse and *positive* G_x spoiler.

Image Reconstruction Options

- Cartesian sampling patterns: inverse FFT
- Non-Cartesian sampling patterns:
  - gridding then inverse FFT
  - iterative reconstruction (regularized least squares)
- Iterative reconstruction with field inhomogeneity compensation:

$$s(t) = \int m_0(\vec{r})\, \mathrm{e}^{-\imath\omega(\vec{r})t}\, \mathrm{e}^{-\imath 2\pi \vec{k}(t)\cdot\vec{r}}\, \mathrm{d}\vec{r}$$

Not a Fourier transform!

To apply iterative reconstruction to non-Cartesian samples, an NUFFT is needed [5]

Summary

NMR images represent the local magnetic environment of ^{1}H protons

- Large main field establishes resonance
- RF pulses and gradient fields perturb magnetization and phases
- Measured signal related to Fourier transform of object
- Reconstruction can be simple: inverse FFT
- Time-varying gradients control k-space trajectory
- Timing parameters control T_1 and T_2 contrast

Slides and lecture notes available from:

`http://www.eecs.umich.edu/~fessler`

References

[1] B. P. Sutton, D. C. Noll, and J. A. Fessler. Fast, iterative image reconstruction for MRI with in the presence of field inhomogeneities. *IEEE Tr. Med. Im.*, 22(2):178–88, February 2003.

[2] J. Pauly, D. Nishimura, and A. Macovski. A k-space analysis of small-tip-angle excitation. *J. Mag. Res.*, 81:43–56, 1989.

[3] J. Pauly, D. Nishimura, and A. Macovski. A linear class of large-tip-angle selective excitation pulses. *J. Mag. Res.*, 82:571–87, 1989.

[4] J. Pauly, P. L. Roux, D. Nishimura, and A. Macovski. Parameter relations for the Shinnar-le roux selective excitation pulse design algorithm. *IEEE Tr. Med. Im.*, 10(1):53, March 1991.

[5] J. A. Fessler and B. P. Sutton. Nonuniform fast Fourier transforms using min-max interpolation. *IEEE Tr. Sig. Proc.*, 51(2):560–74, February 2003.

Some of Many Omitted Topics

- Spectroscopic imaging
- Dynamic imaging
- Functional imaging
- Blood flow imaging
- Multi-slice imaging
- Main-field hardware (shimming, open magnets, ...)
- RF coil design
- Gradient nonlinearities
- Signal to noise considerations
- MR contrast agents
- Interventional MR
- Gating (cardiac and respiratory)
- ...

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