

MRI Physics Basics

Nuclear magnetic resonance

Hydrogen protons align in magnetic field

Gradient fields

Spatial encoding of signal

K-space

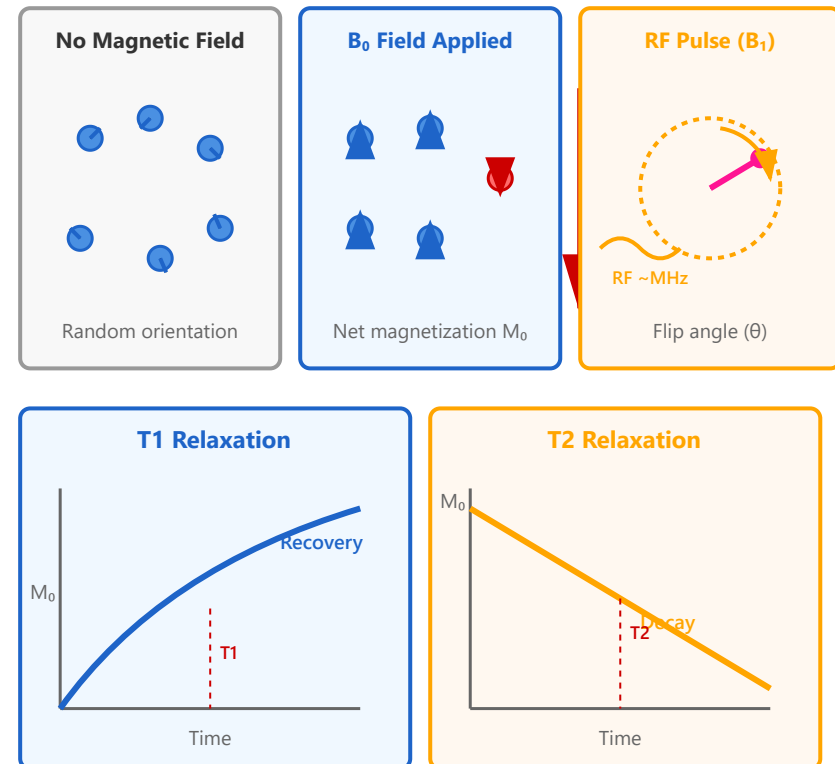
Frequency domain data representation

Relaxation times (T1, T2)

Tissue-specific signal recovery

Signal equation

$$S \propto \rho \cdot (1 - e^{(-TR/T1)}) \cdot e^{(-TE/T2)}$$



1. Nuclear Magnetic Resonance (NMR)

Fundamental Principle

Nuclear Magnetic Resonance is the physical phenomenon where atomic nuclei with an odd number of protons or neutrons possess a magnetic moment and angular momentum (spin). In MRI, we primarily use hydrogen nuclei (^1H) because of their abundance in the human body, particularly in water and fat molecules.

The Process

Step 1 - Random State: Without an external magnetic field, hydrogen protons in tissue are randomly oriented, resulting in no net magnetization.

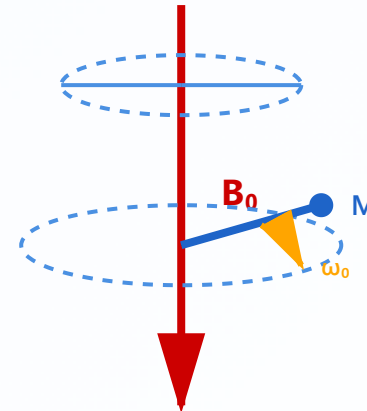
Step 2 - Alignment: When placed in a strong static magnetic field (B_0 , typically 1.5T or 3T), protons align either parallel (low energy) or anti-parallel (high energy) to the field. A slight excess aligns parallel, creating net magnetization (M_0).

Step 3 - Precession: Aligned protons don't simply point along B_0 ; they precess around it at the Larmor frequency: $\omega_0 = \gamma \cdot B_0$, where γ is the gyromagnetic ratio (42.58 MHz/T for hydrogen).

Key Concepts:

- Hydrogen is the most abundant element in human tissue (about 63%)
- Net magnetization is proportional to field strength
- Larmor frequency at 1.5T: 63.87 MHz; at 3T: 127.74 MHz
- The energy difference between spin states is extremely small

Larmor Precession



Larmor Equation

$$\omega_0 = \gamma \cdot B_0$$

Energy Levels



2. Gradient Fields

Purpose and Function

Gradient fields are spatially varying magnetic fields superimposed on the main B_0 field. They create controlled variations in the magnetic field strength across different spatial locations, enabling spatial encoding of the MR signal. Without gradients, we would only detect a signal from the entire imaging volume without any spatial information.

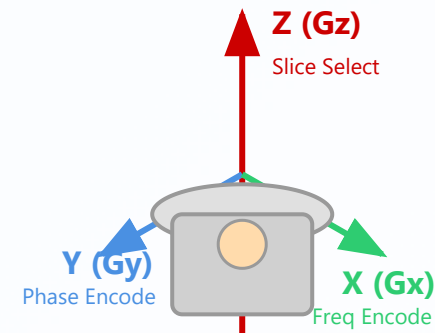
Three Gradient Axes

Slice Selection Gradient (G_z): Applied during RF excitation to selectively excite a specific slice. By varying the magnetic field along the z-axis, different locations have different Larmor frequencies. An RF pulse at a specific frequency will only excite protons at the corresponding location.

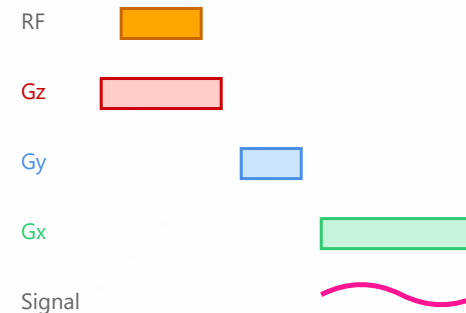
Phase Encoding Gradient (G_y): Applied briefly after excitation to introduce phase differences between rows of spins. Each application encodes one line of k-space. This gradient is stepped through different amplitudes for each phase encoding step.

Frequency Encoding (Readout) Gradient (G_x): Applied during signal acquisition, creating a frequency spread across the field of view. Different positions along the x-axis emit signals at different frequencies, which can be separated by Fourier transformation.

Three Gradient Axes



Gradient Timing



Important Notes:

- Gradient strength is measured in mT/m (millitesla per meter)
- Stronger gradients allow faster imaging and thinner slices

- Gradient switching produces the characteristic MRI "knocking" sound
- The combination of all three gradients determines spatial resolution

3. K-space

Concept and Significance

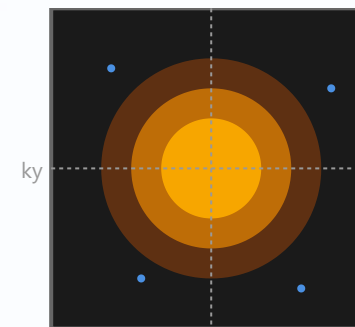
K-space is a mathematical construct representing the spatial frequency domain of MR data. It is not a physical space but rather a data matrix where each point contains raw signal data encoded with specific spatial frequency information. The relationship between k-space and image space is defined by the Fourier transform.

Structure and Properties

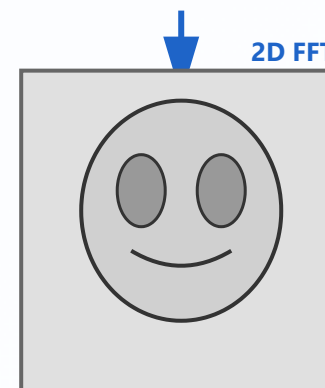
Center of K-space: Contains low spatial frequency information, which determines image contrast, signal-to-noise ratio (SNR), and overall brightness. The center represents the bulk signal from the entire field of view.

Periphery of K-space: Contains high spatial frequency information, which determines image detail, edges, and fine structures. The edges define spatial resolution and sharpness.

Filling Patterns: K-space can be filled in various patterns: line-by-line (Cartesian), radially (radial imaging), or spirally (spiral imaging). Different filling strategies affect imaging speed and artifact patterns.



High freq



Spatial Domain

Center: Contrast & SNR

Edges: Resolution

$$\text{Image}(x, y) = \iint \text{K-space}(k_x, k_y) \cdot e^{i2\pi(k_x \cdot x + k_y \cdot y)} dk_x dk_y$$

Clinical Implications:

- Undersampling k-space periphery reduces scan time but decreases resolution
- Motion during center k-space acquisition causes severe artifacts
- Parallel imaging techniques (SENSE, GRAPPA) skip k-space lines
- Partial Fourier techniques collect only 60-75% of k-space

4. Relaxation Times (T1 and T2)

T1 Relaxation (Longitudinal/Spin-Lattice)

T1 is the time constant for recovery of longitudinal magnetization (M_z) back to its equilibrium value (M_0) after RF excitation. It represents energy transfer from the excited spin system to the surrounding molecular lattice (thermal equilibrium). T1 recovery follows an exponential curve: $M_z(t) = M_0(1 - e^{-(t/T1)})$.

Typical T1 values at 1.5T: Fat: 250ms, White matter: 780ms, Gray matter: 920ms, CSF: 4000ms, Muscle: 870ms. T1 increases with field strength.

T2 Relaxation (Transverse/Spin-Spin)

T2 is the time constant for decay of transverse magnetization (M_{xy}) due to dephasing of spins from interactions with neighboring spins. It represents the loss of phase coherence in the transverse plane. T2 decay follows: $M_{xy}(t) = M_0 \cdot e^{-(t/T2)}$. T2 is always shorter than T1.

Typical T2 values at 1.5T: Fat: 80ms, White matter: 90ms, Gray matter: 100ms, CSF: 2000ms, Muscle: 45ms. T2 is relatively independent of field strength.

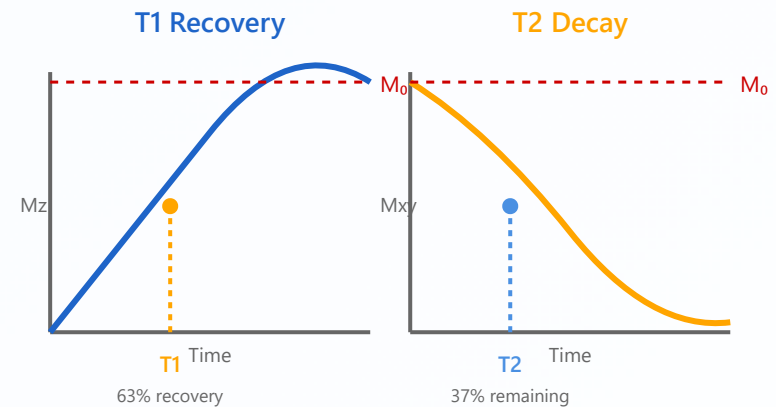
T2* and Susceptibility Effects

T2* includes both T2 relaxation and additional dephasing from magnetic field inhomogeneities. It's always shorter than T2: $1/T2^* = 1/T2 + 1/T2'$. T2* effects are important in gradient echo sequences and functional MRI (BOLD contrast).

Clinical Applications:

- T1-weighted: Good anatomical detail, fat is bright, fluid is dark
- T2-weighted: Sensitive to pathology, fluid is bright (edema, tumors)

Relaxation Processes



T1 Values

Fat: 250 ms	GM: 920 ms
WM: 780 ms	CSF: 4000 ms

T2 Values

Fat: 80 ms	GM: 100 ms
WM: 90 ms	CSF: 2000 ms

Image Contrast Comparison

T1-weighted

Fat: Bright
Gray: Mid
CSF: Dark

T2-weighted

Fat: Mid
Gray: Mid
CSF: Bright

PD-weighted

Fat: Bright
Gray: Mid
CSF: Bright

- FLAIR: T2-weighted with CSF suppression for periventricular lesions
- T2*: Sensitive to hemorrhage, calcification, and iron deposition

5. MRI Signal Equation

The Fundamental Equation

The MRI signal intensity is determined by a combination of tissue properties and imaging parameters. The basic signal equation for a spin echo sequence is:

$$S \propto \rho \cdot (1 - e^{(-TR/T1)}) \cdot e^{(-TE/T2)}$$

Where: **S** = Signal intensity, **ρ** = Proton density, **TR** = Repetition time, **TE** = Echo time, **T1** = Longitudinal relaxation time, **T2** = Transverse relaxation time

Parameter Effects

Proton Density (ρ): The concentration of hydrogen protons in tissue. Higher proton density produces stronger signal. Fat and water have high proton density, while cortical bone has very low density.

TR (Repetition Time): Time between successive RF pulses. Short TR (< 600ms) emphasizes T1 differences, creating T1-weighted images. Long TR (> 2000ms) allows full T1 recovery, minimizing T1 contrast.

TE (Echo Time): Time between RF excitation and signal acquisition. Short TE (< 20ms) minimizes T2 decay. Long TE (> 80ms) emphasizes T2 differences, creating T2-weighted images.

Image Weighting

T1-weighted: Short TR (400-600ms), Short TE (10-20ms). Highlights T1 differences, excellent anatomical detail.

T2-weighted: Long TR (2000-6000ms), Long TE (80-120ms). Highlights T2 differences, sensitive to pathology.

Proton Density (PD): Long TR (2000-6000ms), Short TE (10-20ms). Minimizes T1 and T2 effects, shows proton density

Signal Equation Components

$$S \propto \rho \cdot (1 - e^{(-TR/T1)}) \cdot e^{(-TE/T2)}$$

Proton Density (ρ)

Number of H atoms
per unit volume

T1 Component

$(1 - e^{(-TR/T1)})$
Recovery factor

T2 Component

$e^{(-TE/T2)}$
Decay factor

Flip Angle (α)

$\sin(\alpha)$ for GRE
Excitation efficiency

Image Weighting Matrix

	TR	TE	
T1-W	Short	Short	400-600ms / 10-20ms
T2-W	Long	Long	2000-6000 / 80-120
PD-W	Long	Short	2000-6000 / 10-20

Trade-offs

↑ TR, TE → ↑ Scan Time

↑ TR, TE → ↑ SNR

differences.

Practical Considerations:

- Longer TR and TE increase scan time but improve SNR
- Flip angle also affects signal: $S \propto \sin(\alpha)$ for gradient echo
- Additional factors: receiver gain, coil sensitivity, voxel size
- Modern sequences use multiple echoes and advanced techniques