

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

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

No Magnetic Field



B₀ Field Applied

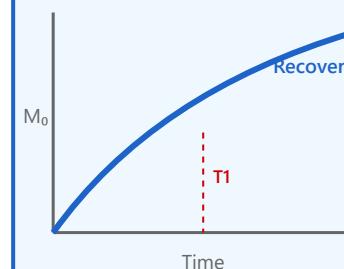


RF Pulse (B₁)

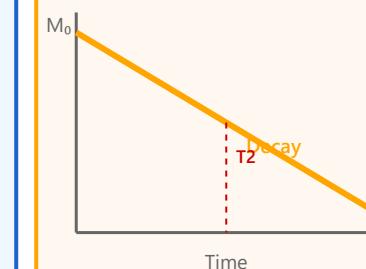


Flip angle (θ)

T1 Relaxation



T2 Relaxation



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.

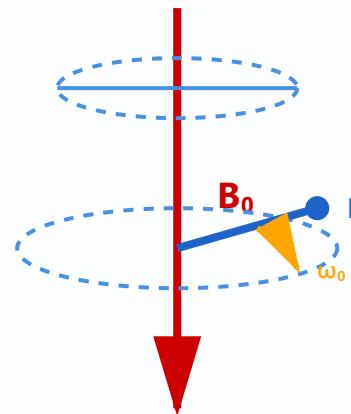
Step 2 - Alignment: When placed in a strong static magnetic field (B_0 , typically 1.5T or 3T), protons align either parallel (low energy) or anti-parallel (high energy) to the field. A slight excess aligns parallel, creating net magnetization (M_0).

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.

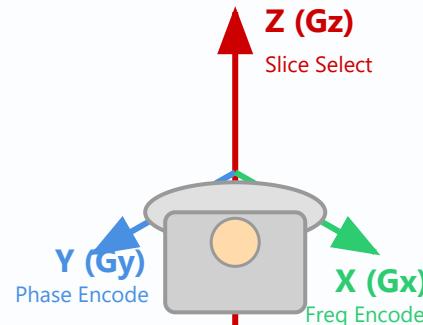
Three Gradient Axes

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.

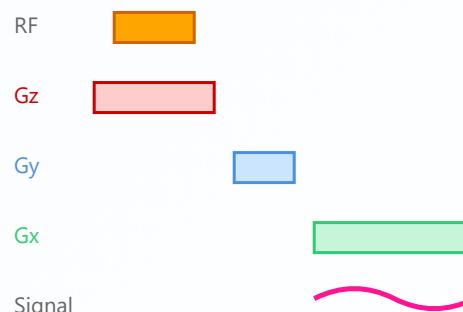
Phase Encoding Gradient (G_y): Applied briefly after excitation to introduce phase differences between rows of spins. Each application encodes one line of k-space. This gradient is stepped through different amplitudes for each phase encoding step.

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

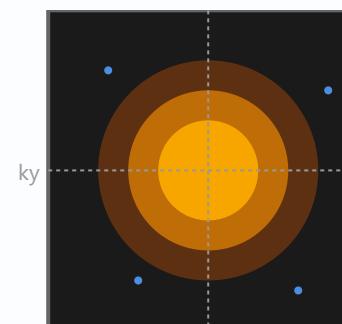
K-space is a mathematical construct representing the spatial frequency domain of MR data. It is not a physical space but rather a data matrix where each point contains raw signal data encoded with specific spatial frequency information. The relationship between k-space and image space is defined by the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

Structure and Properties

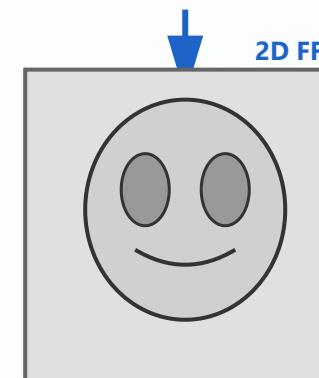
Center of K-space: Contains low spatial frequency information, which determines image contrast, signal-to-noise ratio (SNR), and overall brightness. The center represents the bulk signal from the entire field of view.

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



2D FFT

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: $M_z(t) = M_0(1 - e^{-t/T_1})$.

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)

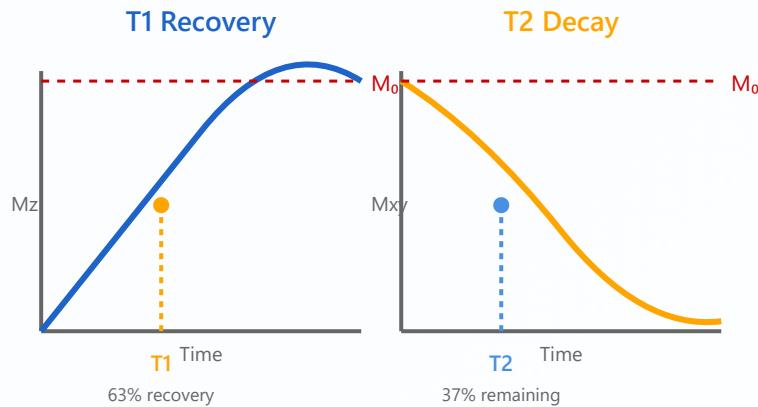
T2 is the time constant for decay of transverse magnetization (M_{xy}) due to dephasing of spins from interactions with neighboring spins. It represents the loss of phase coherence in the transverse plane. T2 decay follows: $M_{xy}(t) = M_0 \cdot e^{-t/T_2}$. T2 is always shorter than T1.

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

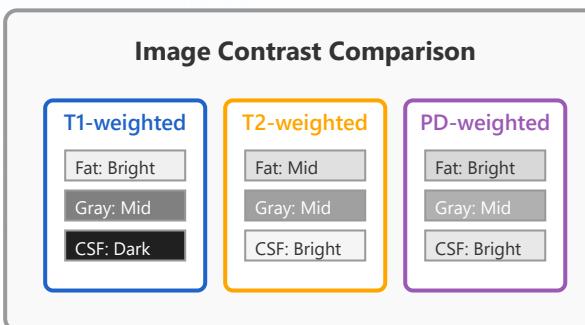
$T2^*$ includes both T2 relaxation and additional dephasing from magnetic field inhomogeneities. It's always shorter than T2: $1/T2^* = 1/T2 + 1/T2'$. $T2^*$ effects are important in gradient echo sequences and functional MRI (BOLD contrast).

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



Clinical Applications:

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

- 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:

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

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

Parameter Effects

Proton Density (p): 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

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

Proton Density (ρ)

Number of H atoms per unit volume

T1 Component

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

T2 Component

$e^{-\frac{TE}{T2}}$
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

MRI Physics Basics - Comprehensive Educational Material

Understanding the fundamental principles of magnetic resonance imaging