Introduction

The development of magnetic resonance (MR) imaging has revolutionized diagnostic imaging. Prior to the introduction of MR imaging in the late 1970s, medical imaging relied primarily on radiography, which is limited by poor soft tissue contrast and superimposition of multiple structures. The advent of diagnostic ultrasound and x-ray computed tomography (CT) allowed visualization of "slices" of a patient's anatomy, but ultrasound is unable to penetrate gas or mineral, and CT achieves modest improvements in soft tissue contrast compared with radiographs. MR imaging creates images based on how protons (i.e., hydrogen atoms) behave in a magnetic field. Like ultrasound, MR imaging does not require ionizing radiation to generate an image. The ubiquity of water and other molecules containing protons throughout the body makes MR imaging a powerful tool for evaluating parts of a patient's anatomy that are inaccessible or poorly visible by other imaging methods. The sensitivity of MR imaging to changes in tissue water content makes it particularly well suited to detecting pathology, such as edema, infection, and neoplasia. Knowledge of the basic principles of MR imaging will help the dentist or physician to interpret images obtained by commonly used MR imaging protocols, and to know when special protocols may be beneficial for diagnosis.

MR Imaging System Hardware

MR imaging requires sophisticated hardware and software to generate an image, including

- Strong **primary magnetic field**; most clinical MR imaging systems use superconducting magnet to generate magnetic field with strength of 1.5 or 3.0 Tesla (T)
- Series of 3 **gradient coils**; gradient coils add or subtract from primary magnetic field along X-, Y-, or Z-axis of MR imaging system
- **Radiofrequency (RF) coil**; RF coil transmits pulses that "excite" protons within patient and then receives signal emanating from excited protons

Nuclear Magnetic Resonance

MR imaging is based on the phenomenon of nuclear magnetic resonance, in which certain atoms act like tiny bar magnets with positive and negative ends. Protons are the most abundant magnetically active atoms in biological tissues and are the source of signal in conventional MR imaging. When placed in a magnetic field, such as the bore of an MR imaging system, protons will align parallel (a.k.a., "spin up") or antiparallel (a.k.a., "spin down") with the magnetic field. More protons will align in the lower energy, spin-up state, resulting in net magnetization parallel to the external magnetic field, referred to as **longitudinal magnetization**. Quantum mechanics predicts that protons will not perfectly align with the axis of the external magnetic field but at an angle to the magnetic field. The protons then wobble, or "precess," around the axis of the magnetic field with a frequency, known as the **Larmor frequency,** that depends on the external magnetic field strength. At equilibrium, a group of protons will be randomly distributed around their respective precessional orbits, causing no net magnetization perpendicular to the axis of the magnetic field; i.e., there is no net **transverse magnetization**. The equilibrium condition, in which there is positive longitudinal and zero transverse magnetization, can be perturbed by briefly applying a secondary magnetic field (known as the RF pulse) that oscillates at the Larmor frequency. Application of an RF pulse flips some protons from the spin-up state to the spin-down state and aligns the protons' orbits, decreasing the net longitudinal magnetization and inducing transverse magnetization. Transverse magnetization is the source of signal detected by the MR imaging system for constructing an image. A 90° RF pulse results in zero longitudinal and positive transverse magnetization; this effect can be visualized as "tipping" the net magnetization 90° from the longitudinal to the transverse plane. Following excitation by an RF pulse, a group of protons "relaxes" toward the equilibrium state by 2 independent processes: (1) recovery of longitudinal magnetization and (2) decay of transverse magnetization. The rates of relaxation by a tissue depend on inherent properties of the tissue known as the T1 time (longitudinal recovery) and T2 time (transverse decay).

Anatomy of MR Imaging Pulse Sequence

An MR imaging **pulse sequence** consists of a coordinated series of RF pulses, gradient coil application, and data acquisition to create an image with a specific type of contrast. An MR imaging operator can choose from many possible pulse sequences and refine some pulse sequence parameters depending on the prescribed application. The events in a basic pulse sequence serve 1 of the following purposes: (1) excitation, (2) spatial encoding, or (3) refocusing.

Excitation

All pulse sequences employ an excitation RF pulse. Many pulse sequences begin with a 90° excitation RF pulse, but other excitation tip angles are also used. The amount of time between consecutive excitation RF pulses is known as the **repetition time (TR)**. A long TR will allow tissues to recover their longitudinal magnetization before the next excitation, whereas a short TR will only allow tissues with rapid longitudinal recovery to fully recover their longitudinal magnetization.

Spatial Encoding

Spatial encoding involves applying gradient coils so that signal arising from a group of excited protons contains information regarding the 3D location of the protons within the patient. A gradient coil works by adding to or subtracting from the primary magnetic field to create a magnetic field gradient that varies with location along 1 axis of the MR imaging system. Because the Larmor frequency of a proton depends on the strength of the surrounding magnetic field, a magnetic field gradient will result in protons precessing at different frequencies depending on their location in space. Spatial encoding typically consists of 3 steps: slice selection, phase encoding, and frequency encoding, which respectively define the z-, y-, and x-axes of the MR images. Slice selection occurs by turning on a gradient coil(s) to vary the magnetic field strength along the z-axis during the excitation RF pulse. The excitation RF pulse is then applied at a predetermined range of frequencies. Only protons with Larmor frequencies matching the RF pulse will become excited, creating a "slice" of excited protons within the patient. The slice-select gradient strength and frequencies of the RF pulse determine the thickness and location of the slice along the z-axis.

Immediately following slice selection, all protons in the excited slice will be aligned in their precessional orbits and are "in phase." Phase encoding works by applying a gradient coil (known as the phase encode gradient) along the y-axis. Protons located at points along the y-axis with a stronger magnetic field will speed up their precession and advance

their phase. Protons experiencing a weaker magnetic field will precess more slowly, retarding their phase. Once the phase-encode gradient is turned off, the excited protons return to their original Larmor frequency but retain their new phases. Importantly, constructing a typical MR image requires many phase-encoding "steps," but only a single or few phase-encoding steps can be performed per TR interval using basic pulse sequences. The need for phase encoding is 1 of the major rate-limiting steps of MR imaging.

Frequency encoding occurs during data acquisition. The frequency encode gradient is applied during data acquisition to vary the precessional frequencies of protons along the x-axis. Protons at different positions along the x-axis will thus contribute waves with varying frequencies to the signal collected during data acquisition. Many different frequencies can be encoded simultaneously, and frequency encoding does not typically affect image acquisition time. At the time of data acquisition, signal from a tissue will be encoded with spatial information pertaining to all 3 axes. The image position along the z-axis is defined by slice selection, and spatial information in the y- and x-axes is encoded by the phases and frequencies of the signal collected during data acquisition.

Refocusing

Immediately following the excitation RF pulse, excited protons in a tissue generate a signal; however, without spatial encoding, the signal generated by the excitation RF pulse cannot create an image. During basic MR imaging, this initial signal decays prior to and during spatial encoding. Therefore, it is necessary to refocus the decayed signal to form an image.

Transverse magnetization decays much more rapidly than longitudinal magnetization recovers. Imperfections in the primary magnetic field, gradient coil application, and tissue inhomogeneity further hasten the loss of transverse magnetization. By the time image data are acquired, the original transverse magnetization is undetectable. This problem is overcome by inducing an "echo" of the original transverse magnetization using either a 180° refocusing RF pulse for **spin-echo** imaging or by reversing the polarity of a specific gradient coil for **gradient-echo** imaging. Both spin-echo and gradient-echo imaging "rephase" the decaying signal to form a signal echo. The time between the excitation RF pulse and the echo formation is known as the **echo time (TE)**.

With the use of multiple 180° refocusing pulses after an excitation pulse, it is possible to generate and detect multiple signal echoes during a single TR interval. This technique, known as fast spin-echo (FSE) imaging, is often used for clinical pulse sequences to reduce image acquisition time. FSE pulse sequences slightly alter the expected contrast of an image compared with spin-echo pulse sequences and are primarily used for creating images with T2W contrast. Multiple echoes can also be generated per TR interval during gradient-echo imaging, but the rapid decay of transverse magnetization during gradient-echo pulse sequences limits the number of useful echoes that can be obtained.

Mechanisms of Contrast in MR Imaging

Contrast between different tissues in MR images depends on inherent tissue properties and user-defined pulse sequence parameters. Inherent properties that influence the appearance of a tissue in an MR image include the tissue proton density, T1 time, and T2 time. The major pulse sequence parameters that affect tissue appearance are the TR and TE. The TR determines how much time is available to recover longitudinal magnetization between excitation RF pulses. Tissues that incompletely recover their longitudinal magnetization will have less magnetization available to be tipped into the transverse plane by the next excitation pulse, resulting in decreased signal in the final image. A relatively short TR will thus create contrast between tissues with different T1 times, yielding a **T1W image**. The pulse sequence TE determines how sensitive the image will be to differences in transverse decay among tissues. Setting a relatively long TE will allow more time for transverse magnetization to decay, creating contrast between tissues with different T2 times to produce a **T2W image**. A pulse sequence with a long TR and short TE will maximize longitudinal recovery while minimizing transverse decay, resulting in minimal dependence on tissue T1 or T2 times. Contrast in such an image will depend primarily on the density of protons in tissues, creating a **proton density-weighted (PDW or PD) image**. It is important to note that 1 characteristic of a tissue may exert a large influence on the appearance of the tissue in an image, regardless of the contrast intended by the pulse sequence. For example, a tissue with very scarce protons will appear dark on T1W, T2W, and PDW images regardless of the T1 or T2 times of the tissue. Also, gradient-echo pulse sequences are more susceptible to magnetic field heterogeneity and rapidly lose transverse magnetization by a mechanism known as T2* decay. Gradient-echo images with relatively long TE are said to have T2* weighting. Many other mechanisms of generating contrast in MR imaging are possible but beyond the scope of this chapter.

Inversion Recovery and Tissue Nulling

Inversion recovery is a technique that employs an extra RF pulse with a tip angle of 180° to begin the pulse sequence. Immediately after this initial 180° RF pulse, known as the inversion pulse, all excited tissues will have **negative** longitudinal magnetization. The inversion pulse is followed by a delay known as the inversion time (TI), then a 90° excitation RF pulse. During the TI, different tissues experience longitudinal recovery according to their respective T1 times, and each tissue will pass through a zero point when the tissue has no net longitudinal magnetization. Tissues with short T1 times (e.g., fat) will pass through the zero point at relatively short TI and tissues with long T1 times (e.g., water or cerebrospinal fluid) will pass through the zero point at relatively long TI. If a tissue has zero net longitudinal magnetization at the time of the excitation pulse, that tissue will not contribute signal and will appear black in the final image. Commonly used inversion pulse sequences include short tau inversion recovery (STIR), which nulls the signal from fat, and fluid-attenuating inversion recovery (FLAIR), which nulls the signal from low protein fluids.

Contrast Enhancement

Contrast enhancement typically refers to using exogenous contrast media to aid in the detection of pathology or increase visibility of normal anatomic structures. The most common clinically used MR imaging contrast media are based on gadolinium. Gadolinium is a metal ion with magnetic properties that shorten the T1 time of a tissue. Other mechanisms of positive and negative contrast enhancement are used in research and may have specific clinical applications but are not routinely used. Gadolinium contrast medium is typically injected intravenously and taken up by highly vascular tissues and tissues with increased vascular permeability. In the presence of gadolinium, a tissue will have a shortened T1 time and appear brighter in a T1W image compared with a

precontrast image. Gadolinium contrast enhancement can be helpful for identifying inflammation, blood vessels, and some tumors.

Spatial Resolution

Image spatial resolution determines the ability to resolve 2 adjacent structures as separate entities in an image. Greater spatial resolution will allow closer and smaller individual structures to be resolved. In MR imaging, spatial resolution is defined by the image slice thickness, field of view (FOV), and image matrix dimensions. Increasing slice thickness will result in averaging of signal from adjacent tissues in the z-dimension of the patient, blurring the edges of some structures and degrading the image spatial resolution. FOV defines the boundaries of the image to be acquired. If all other parameters are maintained, increasing the FOV will cause poorer spatial resolution; however, FOV must be large enough to contain all of the patient anatomy within the RF coil in the plane of the image. Selecting an FOV smaller than the patient's anatomy can lead to an effect known as "wrap around" in which signal from body parts outside the FOV will be superimposed on other structures in the image. The matrix dimensions of an image define the number of rows and columns of pixels in the image. Together, the slice thickness, FOV, and matrix dimensions determine the size of each piece of tissue that will be represented by individual pixels in an image. Thin slices, small FOV, or a large matrix size will allow resolution of fine details in an image but are all associated with long acquisition time &/or poorer signal:noise ratio (SNR).

MR imaging can obtain images in 2D or 3D modes of acquisition. Both 2D and 3D modes of acquisition produce 2D images representing slices of the patient's anatomy. 2D imaging operates by exciting a relatively thin slice of tissue, then encoding the remaining spatial information as described above. 2D imaging excites and acquires image data from 1 slice of tissue at a time, and typical 2D MR imaging pulse sequences require a gap between adjacent slices to prevent the occurrence of certain artifacts. 3D imaging excites a thick "slab" of tissue at the beginning of the pulse sequence, then uses 2 phase-encoding steps plus 1 frequency-encoding step to encode spatial information within the slab. Slices are then reconstructed from the spatially encoded slab of image data. A 3D approach allows creation of images with very thin, contiguous slices and fine spatial resolution. The disadvantage of 3D MR imaging is that many phase-encode steps are necessary, and image acquisition can be time consuming.

Positioning of the patient and accurately prescribing the desired image orientation are essential for accurate diagnosis. Sagittal, axial, and coronal image planes are all used for visualizing specific anatomic features and abnormalities of the head. In general, an image best portrays anatomic features and abnormalities in the plane of the image and poorly represents features orthogonal to the image plane. Two or more imaging planes are often necessary to completely visualize all surfaces of an anatomical region. Complex imaging planes can be prescribed to capture curved or oblique surfaces. It may also be necessary to acquire images with changes in patient position to characterize a pathologic process. For example, MR imaging of the TMJ is often performed in closed-mouth and open-mouth positions to evaluate for potential changes in position of the TMJ disc.

Signal:Noise Ratio and Acquisition Time

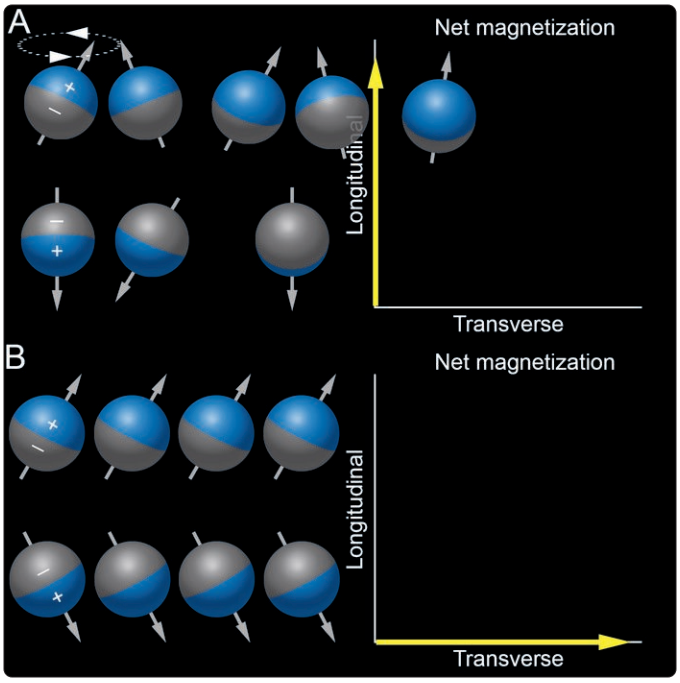
As with any diagnostic imaging modality, MR image quality depends on spatial resolution, contrast, and signal:noise ratio (SNR). A major disadvantage of MR imaging compared with other diagnostic imaging modalities is that an MR imaging study typically requires much longer acquisition time than radiographs or CT. Improving MR imaging spatial resolution &/or SNR generally requires more time for image acquisition. For clinical MR imaging studies, it is necessary to balance image quality with acquisition time.

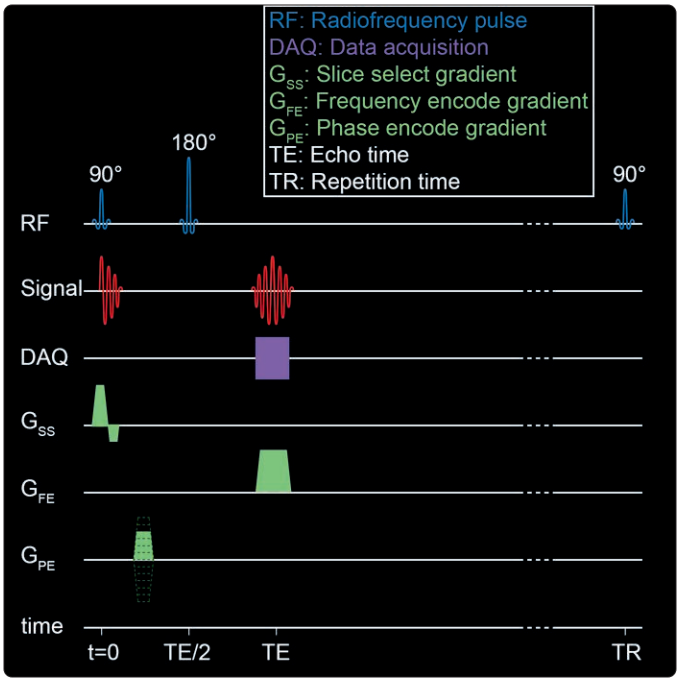
All digital imaging modalities are subject to electronic noise that contributes randomly to the final image and degrades the image quality. For radiography and CT, the number of x-ray photons contributing to image formation (i.e., the signal) far outweighs any electronic noise, and the noise is often imperceptible. Conversely, the signal generated during MR imaging is relatively small, and MR imaging SNR is often the factor that limits image quality. SNR can be improved by using different hardware as well as changing certain pulse sequence parameters. Using an MR imaging system with greater field strength increases the SNR for other factors being equal but is not practical in clinical settings. One of the most effective ways to improve SNR is to use an RF coil that is specially suited to the region of interest because proximity of the RF coil to the region of interest has a major influence on signal. Surface coils are RF coils intended to be placed directly on the surface of the region to be imaged. Surface coils sacrifice the ability to image deep into the body but achieve excellent SNR for tissues close to the surface. Surface coils are frequently used to image the TMJs and spine.

Nearly all pulse sequence parameters have the potential to influence the image SNR. For example, changing the image spatial resolution will also influence the SNR. Decreasing the slice thickness, decreasing FOV, &/or increasing the matrix size will cause a smaller sample of tissue to generate the signal for each pixel in the image. Thus, creating fine spatial resolution will sacrifice SNR. Increasing TR and decreasing TE will tend to increase the amount of signal for image formation but may alter the image contrast. When preservation of contrast is essential, the number of excitations (NEX) can be increased. Using a NEX > 1 simply repeats the pulse sequence 2 or more times and averages the signal from each excitation to create the final image. Importantly, doubling the NEX will only increase the SNR by 41% but double the acquisition time. Other pulse sequence parameters can also influence image SNR but are beyond the scope of this chapter.

Summary

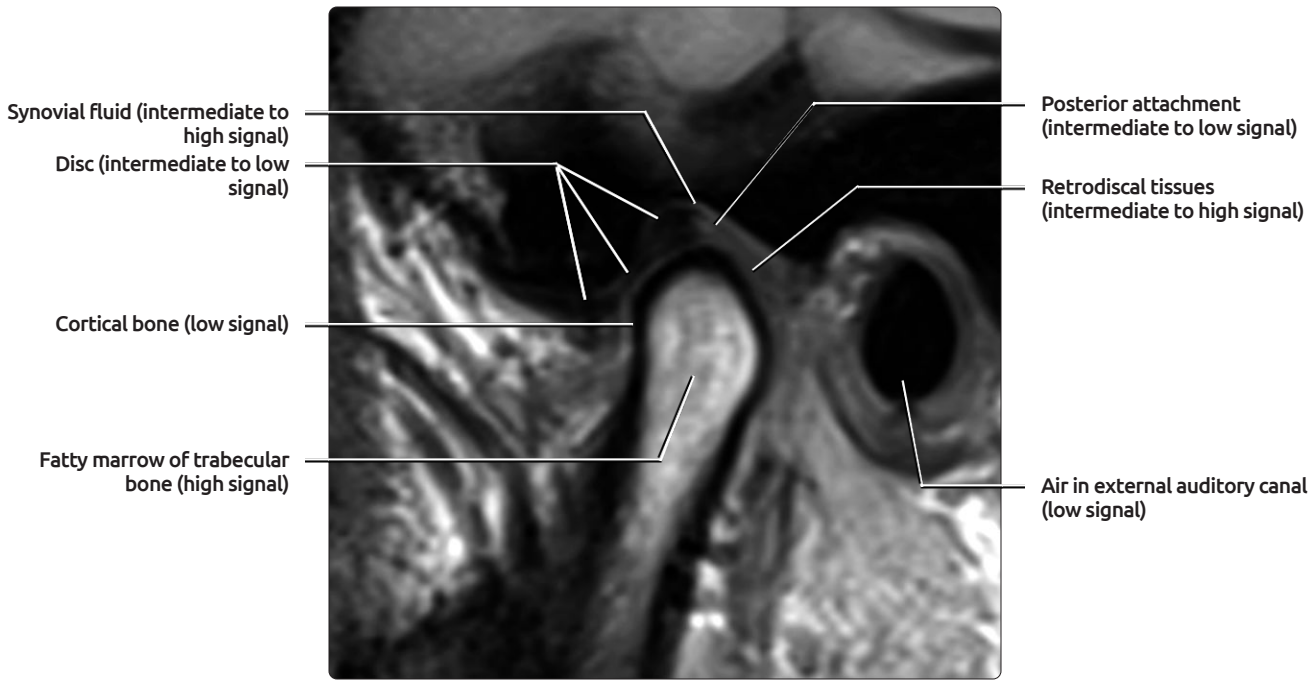
MR imaging system hardware includes a primary magnetic field, gradient coils, and RF coil. A pulse sequence generates an MR image via coordinated gradient coil application, RF pulses, and data acquisition. Inherent tissue properties (e.g., T1 time, T2 time, and proton density) and pulse sequence parameters (e.g., TR and TE) determine the appearance of a tissue in an image. MR imaging can create images with different tissue appearances (e.g., T1W, T2W, PDW, and other images).





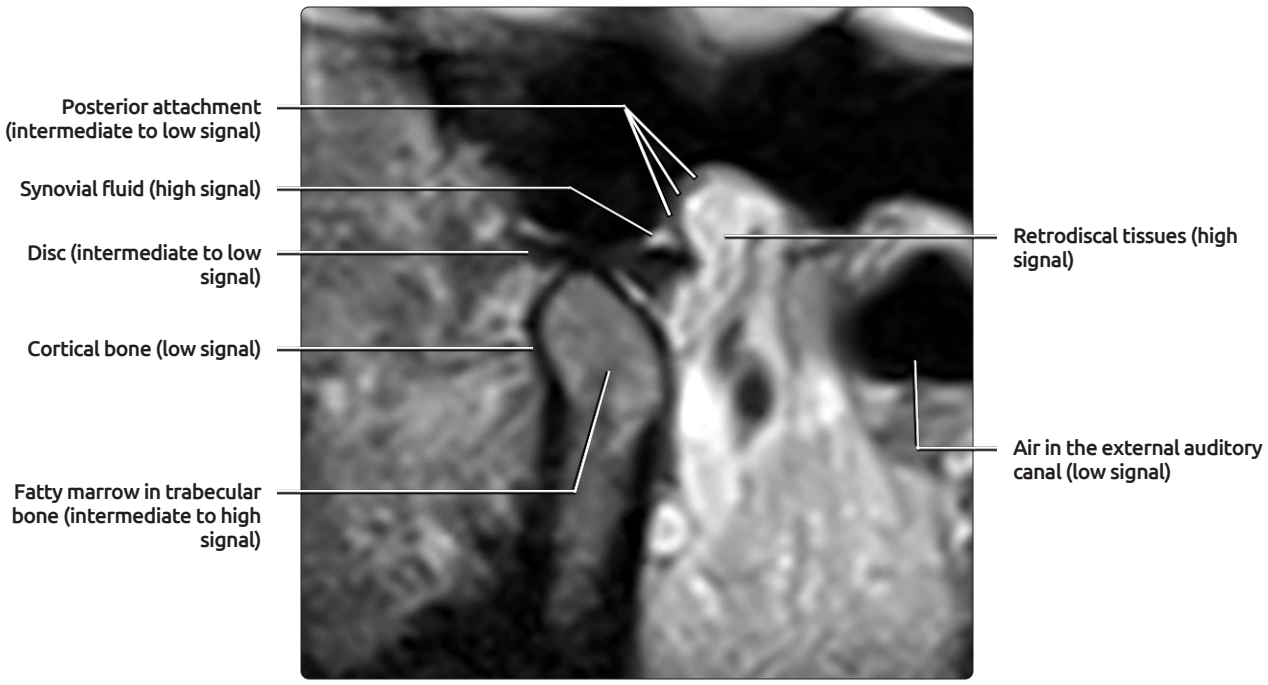
(Top) *(A) In a magnetic field at equilibrium, a small excess of protons will align in the "spin up" orientation, creating a net longitudinal magnetization parallel to the magnetic field. At equilibrium, protons will be randomly distributed around their precessional orbits such that there is no net transverse magnetization. (B) Immediately following a 90° radiofrequency (RF) pulse, equal numbers of protons will be in the spin-up and spin-down orientations, and protons will be aligned in their precessional orbits to create transverse magnetization.* **(Bottom)** *A basic spin-echo pulse sequence consists of coordinated RF pulses, gradient coil application, and data acquisition to achieve proton excitation, spatial encoding, signal refocusing, and image generation. The repetition time (TR) is defined as the time interval between consecutive 90° excitation RF pulses. The echo time (TE) is the time interval between excitation and formation of a signal echo. Data acquisition for image generation occurs at the TE.*

Signal intensity	T1W
	• Air • Mineral/bone • High-velocity blood flow • Chronic hemorrhage
	• High-water content tissues/edema • Free fluid with low cell & protein content • Collagenous tissues • Peracute hemorrhage
	• Protein-rich fluid (e.g., abscess, synovial fluid)
	• Fat/fatty bone marrow • Subacute hemorrhage • Melanin • Paramagnetic contrast media (gadolinium, manganese, copper)



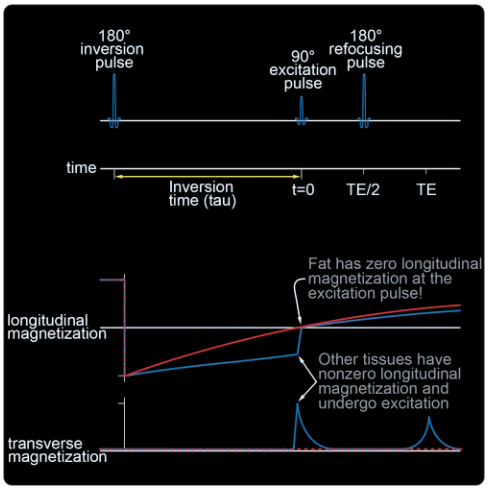
(Top) *T1W images are obtained using pulse sequences with relatively short TR and short TE to create contrast between tissues with different T1 times. Tissues with short T1 times (e.g., fat) will rapidly recover longitudinal magnetization and appear bright in an image. Tissues with long T1 times (e.g., collagenous tissues and tissues with high water content) will appear relatively dark in the image.* **(Bottom)** *T1W image of the TMJ in closed-mouth position shows the signal intensities of normal structures primarily due to differences in T1 relaxation. The cortical bone is low signal (black), and the fatty marrow of the trabecular bone is relatively high signal (bright). The collagenous TMJ disc and attachments are intermediate to low signal (dark gray), and the synovial fluid and retrodiscal tissues are intermediate to high signal (light gray).*

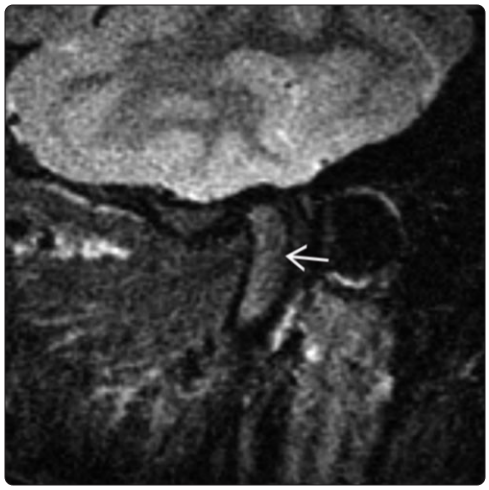
Signal intensity	T2W
	• Air • Mineral/bone • Collagenous tissues (e.g., tendon/ligament) • Chronic hemorrhage • Ferromagnetic contrast media (i.e., iron superoxide)
	• Tissues with high bound water content (e.g., cartilage) • Acute to early subacute hemorrhage
	• Protein-rich fluid (e.g., abscess, synovial fluid)
	• Fluid • Late subacute hemorrhage



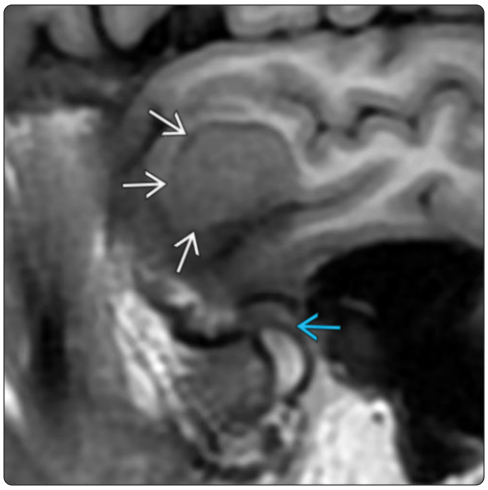
(Top) *T2W images are obtained using pulse sequences with relatively long TR and long TE to create contrast between tissues with different T2 times. Tissues with short T2 times (e.g., chronic hemorrhage and cartilage) will rapidly lose transverse magnetization and appear dark in the image. Tissues with long T2 times (e.g., fluid) will appear bright in the image. Note that, in clinical T2W fast spin-echo images, fat will appear bright although fat has a relatively short T2 time.* **(Bottom)** *T2W fast spin-echo image of a TMJ in the open-mouth position shows the signal intensities of the normal structures due to T2 decay; cortical bone is low signal (black) and the fatty marrow of the trabecular bone is intermediate to high signal (light gray). The collagenous TMJ disc and the posterior attachment are intermediate to low signal (dark gray/black), and the synovial fluid and the retrodiscal tissues filling with blood in the open position are high signal (bright).*

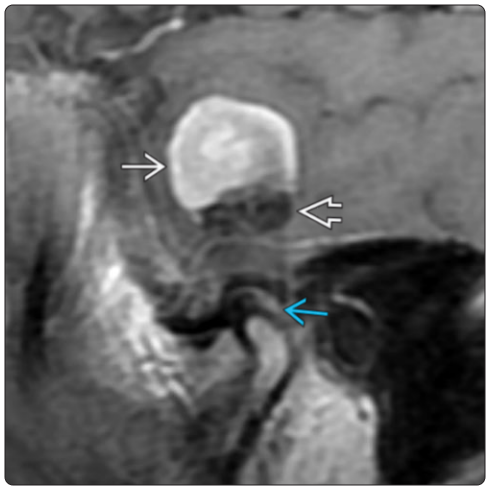
(Left) *Short tau inversion recovery (STIR) pulse sequence is for nulling signal from fat. STIR pulse sequences begin with an inversion RF pulse, causing negative longitudinal magnetization, followed by a delay during which magnetization recovers according to T1 times. A tissue with zero longitudinal magnetization at the excitation pulse will appear black in the image.* **(Right)** *Sagittal STIR image shows the TMJ with decreased signal from fatty tissues, e.g., fatty marrow ➡, compared with a T1W or T2W image.*



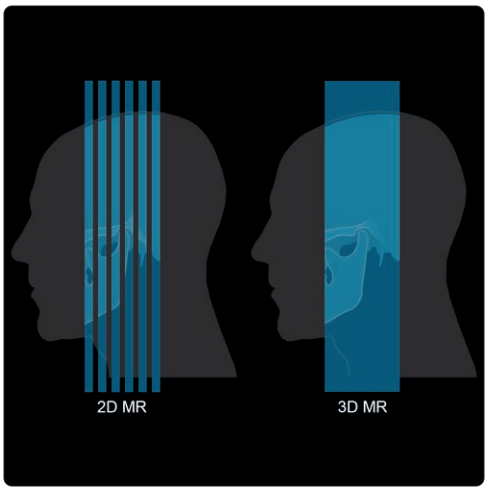


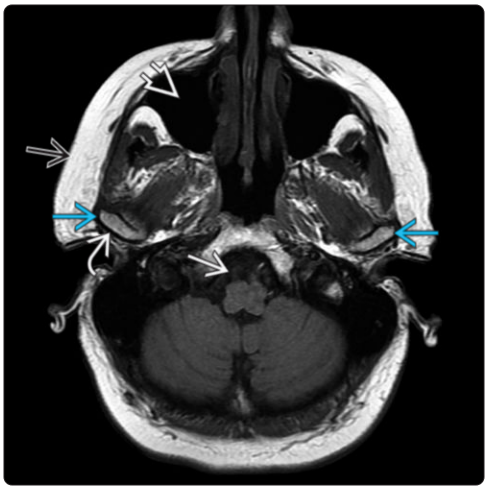
(Left) *Sagittal T1W image is at the level of the TMJ ➡. Note the subtle, hypointense mass compressing the brain superior to the TMJ ➡.* **(Right)** *Sagittal, fat-saturated (FS), postcontrast T1W image is at the level of the TMJ ➡. There is marked contrast enhancement of the mass ➡ in the brain superior to the TMJ. The mass was diagnosed as a meningioma. This was an incidental finding on a TMD patient's CBCT scan. The lesion presented as faint calcification within the middle cranial fossa on CBCT, which appears dark ➡ on this MR.*

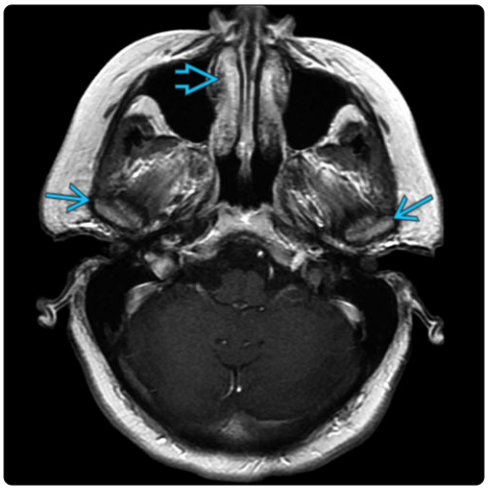




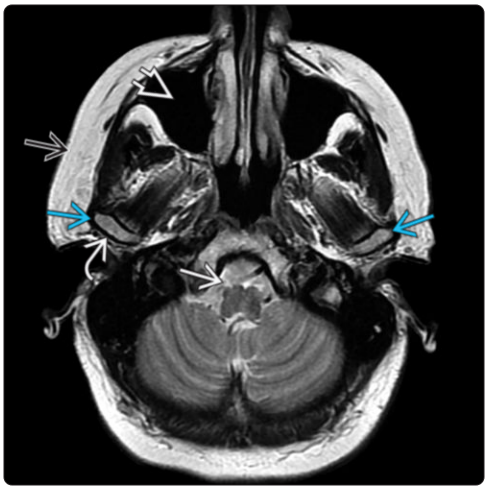
(Left) *MRs can be acquired using 2D or 3D modes of acquisition. During 2D MR, thin slices of anatomy are excited and generate images 1 at a time. During 3D MR, a thick slab of tissue is excited to generate a volume of image data. The 3D volume of data is then used to create 2D images representing very thin (often <1 mm), contiguous slices of anatomy.* **(Right)** *TMJ RF surface coils are typically placed directly against the skin centered on the patient's TMJ to acquire images with high signal:noise ratio.*

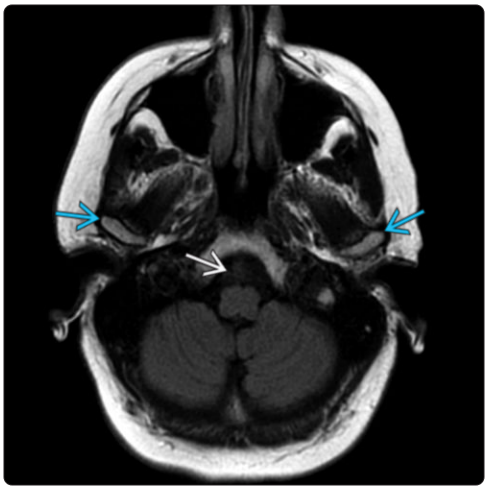




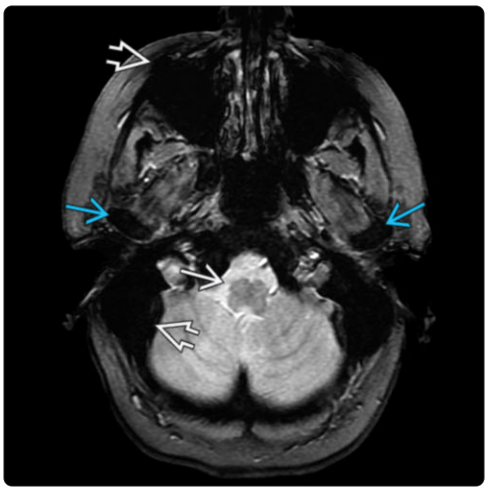


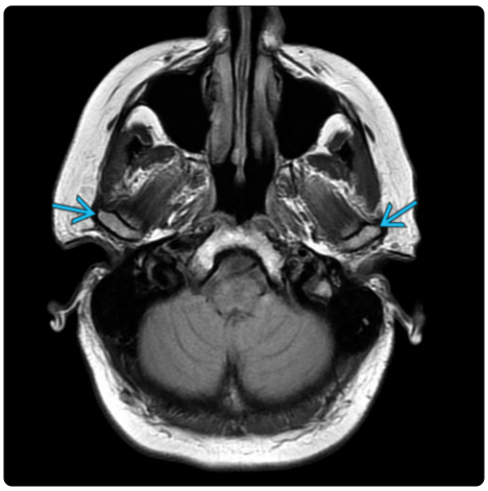
(Left) *Six axial images (same patient) show the appearance of structures at the level of the condyles with different MR sequences: On T1WI MR, fat is bright, and CSF around the brainstem is low to intermediate signal. Air and cortical bone are black.* **(Right)** *Contrast-enhanced (C+) T1WI MR is at the level of the condyles. Gadolinium-based contrast causes increased signal intensity of highly vascular structures, such as the nasal mucosa, and of some pathologic processes, such as malignancies.*





(Left) *Axial T2WI fast spin-echo MR is at the level of the mandibular condyles. Fat and CSF are bright. Air and cortical bone are low signal intensity.* **(Right)** *Axial FLAIR MR at the level of the mandibular condyles shows that, on this sequence, normal CSF and other low-protein fluid will appear of low signal intensity, but edema and other proteinaceous fluids will appear bright (not represented on this image).*





(Left) *Axial T2*WI gradient-echo MR is at the level of the mandibular condyles. Fluid, such as CSF, appears bright, similar to a T2W image. T2*W images are sensitive to tissue inhomogeneity, causing decreased signal or artifact at air-tissue boundaries. T2*W images are useful for detecting hemorrhage (not present in this case).* **(Right)** *Proton density-weighted (PD) MR at the level of the condyles shows that PD images provide excellent signal:noise ratio but may be less sensitive to pathology compared with T2W images.*