Educational course: Imaging Acquisition & Reconstruction

Magnetization-Preparation Modules (Saturation, Inversion and T_2-preparation)

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Objective: Understand the applications and implementation of magnetization-preparation modules involving saturation, inversion and T_2-preparation.

Key points:

- Saturation modules are used for fat suppression, T_1 quantification and saturation bands (to minimize artifacts and signal from inflowing blood)

- Inversion modules are used to achieve strong T_1 weighting, to null signal from selected tissues, and to manipulate signal from inflowing blood (for dark-blood imaging, non-contrast MRA and arterial spin labeling)

- T_2-preparation provides flow-insensitive T_2 weighting

- These modules can be combined to achieve an enormous variety of contrast mechanisms

Introduction:

There are many ways to acquire data in MRI, each of which offers its own range of signal contrast and its own balance between speed and image quality. Gradient echo, for example, can generate T_1- or T_2*-weighting, depending on the choice of imaging parameters, while spin echo and fast spin echo offer T_1- or T_2-weighting. Balanced steady-state free precession (bSSFP) provides a fast acquisition with T_2/T_1 weighting and insensitivity to flow, making it ideal for cardiovascular imaging, while echo-planar imaging (EPI) is the fastest option of all, but is vulnerable to magnetic susceptibility artifacts.

Each of these data acquisition mechanisms can be preceded by one or more magnetization-preparation modules. These are combinations of RF pulses, magnetic field gradients and time delays that are executed before data acquisition begins. They differentially manipulate the magnetization of tissues based on relaxation time, location or chemical composition. They vastly expand the range of signal contrast mechanisms available in MRI and offer a variety of tools for improving image quality. Five important (but overlapping) applications of magnetization-preparation modules are the following:

- Diversifying image contrast
- Suppressing unwanted signal
- Minimizing artifacts
- Manipulating inflow
- Quantifying tissue relaxation times

Inversion modules, for example, can be used to obtain strong T_1 weighting or to null signal from selected tissues based on their T_1 times. T_2-preparation (or 'T_2-prep') provides a flow-insensitive means of T_2 weighting. Saturation bands can be used to prevent aliasing (a.k.a. 'wrap') from signal outside the field of view, or to suppress ghosting from moving tissue or inflowing blood. Inflow can alternatively be exploited as a contrast mechanism itself, with the aid of magnetization-preparation modules, and this underlies many of the techniques used for non-contrast MR angiography (for visualizing arteries) and arterial spin labeling (for measuring tissue perfusion). Saturation and T_2-preparation modules also offer a time-efficient means of quantifying T_1 and T_2 respectively.

Saturation modules

In its most basic form, a saturation module consists of a 90° RF pulse followed immediately by a strong crusher gradient. The RF pulse tips the magnetization away from the longitudinal axis into the transverse plane, and the crusher gradient then dephases the resulting transverse magnetization. This leaves zero net magnetization within the imaging voxels; the longitudinal component is zero due to the RF pulse and the transverse component is zero due to the crusher gradient.

Saturation bands

Applying the RF pulse in the presence of a magnetic field gradient makes it slice selective, and shaping the envelope of the pulse controls its slice profile. Ideally, signal should be perfectly saturated everywhere inside the saturation band, but unaffected outside the saturation band. The saturation module is executed immediately before data acquisition to minimize any recovery of longitudinal magnetization. Saturation bands can be placed over tissue outside the field of view to prevent its signal from aliasing or 'wrapping' into the image. They can also be placed parallel to the imaging slice to avoid ghost artifacts from inflowing pulsatile blood. Arterial inflow can alternatively be exploited as a contrast mechanism to visualize arteries, and a saturation band is then placed distal to the imaging slab to suppress signal from veins.

Fat saturation

Protons in fat resonate at a slightly different frequency from those in water due to their different chemical environment. This property, known as 'chemical shift', can be exploited to saturate the signal from fat while leaving the magnetization of water undisturbed. To saturate the fat signal, the RF pulse of the saturation module must be applied with a frequency offset equal to the chemical shift, and it must be executed in the absence of a slice-select gradient (although it will still be followed by a crusher gradient). The pulse is then said to be 'chemically selective'. The waveform of the RF pulse is typically Gaussian, which produces a Gaussian saturation profile in the frequency domain. The saturation profile is centered on the fat peak and the bandwidth is chosen to ensure that protons in fat are saturated while those in water are unaffected. Since the T_1 of fat is very short, the flip angle of the saturation pulse is often chosen to be slightly higher than 90°, to compensate for longitudinal relaxation that occurs during the time taken to play out the crusher gradients.

Fat saturation is a commonly used method (but not the only method) to suppress signal from fat. Without the aid of magnetization-preparation modules, fat would usually be the brightest tissue on the image, since it has high signal on T_1-weighted acquisitions, T_2-weighted fast spin echo and T_2/ T_1-

weighted bSSFP. Suppressing signal from fat can improve the conspicuity of tissues with inherently lower signal. It is also helpful in minimizing motion artifacts, particularly ghosting of subcutaneous fat due to breathing motion.

Saturation recovery

If a time delay (TD) is inserted between the saturation pulse and the data acquisition, longitudinal magnetization begins to recover. The rate of recovery depends on the T_1 of the tissue. Tissues with short T_1 undergo faster recovery than those with long T_1, and appear brighter on the image. Saturation recovery with a single-shot gradient echo acquisition is the method of choice for contrast-enhanced myocardial perfusion imaging [1]. Images are acquired every cardiac cycle throughout the first pass and washout of the contrast agent. As blood containing the contrast material perfuses the tissue, it shortens the T_1 time and increases the signal. The T_1 of the tissue at each time point can be calculated from its signal intensity, given knowledge of the time delay TD and the signal M_0 that would be obtained in the absence of the saturation module.

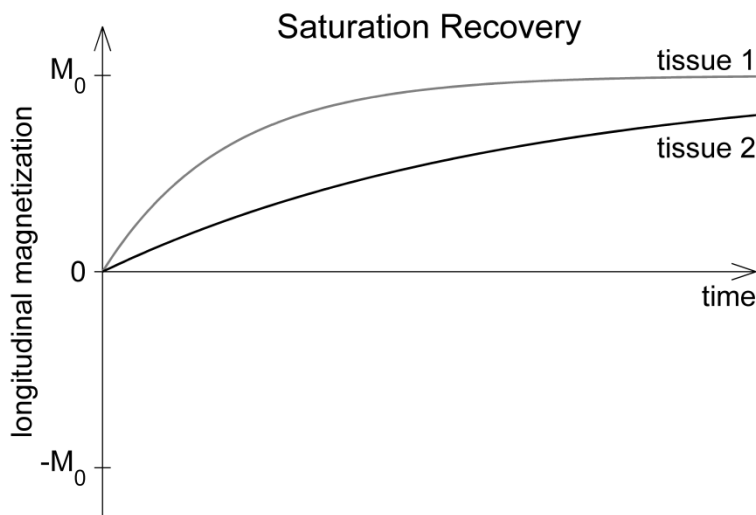
Figure 1: Saturation-recovery curves for tissues with relatively short
T1 (tissue 1) and relatively long T1 (tissue 2)

Accurate quantification of T_1 relies on the magnetization being perfectly saturated by the saturation module. If a simple amplitude-modulated RF pulse is used, however, this is not necessarily true, since the flip angle in that case is proportional to the magnitude of the RF (B_1^+) field. Any inhomogeneity in B_1^+ will result in a flip angle that is not exactly 90° and signal that is not perfectly saturated. There are several methods to optimize the saturation, but the simplest conceptually is the use of a composite pulse consisting of three saturation modules applied in rapid succession with different crusher gradients [2]. Any longitudinal magnetization left after the first saturation module, due to inaccuracy in the flip angle, will be tipped again into the transverse plane by the second RF pulse, leaving proportionally less longitudinal magnetization the second time. After repeating the process a third time, the magnetization is almost perfectly nulled. The use of different crusher gradients and slightly different intervals between the three RF pulses avoids the buildup of coherences (i.e. 'echoes'), which could compromise the quality of the saturation.

Inversion modules

Inversion refers to tipping the longitudinal magnetization through 180° so that it lies along the direction opposite that of the static magnetic field B_0. Like saturation pulses, inversion pulses can be slice selective, chemically selective, or non-selective. To reduce sensitivity to B_1^+ inhomogeneity, the inversion is usually implemented using an adiabatic pulse [3]. Any residual transverse magnetization is dephased using a crusher gradient. The main applications of inversion modules are to provide strong T_1 weighting and to null the signal from selected tissues based on their T_1 values. In both cases, a delay, known as the inversion time (TI), is required between the inversion pulse and the subsequent data acquisition to allow for differential recovery of magnetization from tissues with different T_1 times. Inflowing blood can be specifically targeted using slice-selective inversion pulses, and the delay in that case allows time for blood to flow into the imaging slice.

T_1 weighting

Inversion recovery provides stronger T_1 weighting than saturation recovery. In particular, the maximum signal difference between tissues with different T_1 values but the same proton density is twice as large for inversion recovery as for saturation recovery. One technique that takes advantage of this strong T_1 weighting is MP-RAGE (magnetization-prepared rapid acquisition gradient echo). MP-RAGE is an inversion-recovery fast gradient echo sequence with an inversion time of around $900 - 1100$ ms. It provides high resolution 3D images of the brain with excellent signal contrast between gray and white matter, and is one of the workhorse sequences in neuroimaging, used for morphology and volumetric assessment [4].
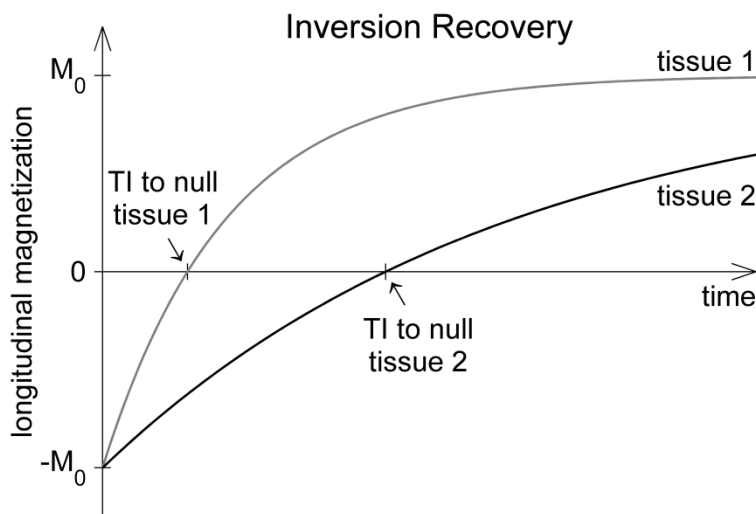

Figure 2: Inversion-recovery curves for tissues with relatively short
T1 (tissue 1) and relatively long T1 (tissue 2)

Tissue nulling

If the inversion time is chosen to match the point at which the longitudinal magnetization of a particular tissue passes through zero, then the signal from that tissue will be nulled. An important application of this is FLuid-Attenuated Inversion Recovery (FLAIR), in which the inversion time is chosen to suppress

signal from cerebrospinal fluid (CSF). FLAIR is an inversion-recovery fast spin echo acquisition, which provides images of the brain that are T_2-weighted but have dark CSF. Nulling the signal from CSF helps to improve the conspicuity of periventricular hyperintense lesions, such as plaques associated with multiple sclerosis (MS) [5].

Since fat has a very short T_1 compared to other tissue types, its signal can be selectively suppressed using Short TI Inversion Recovery (STIR). STIR provides more uniform fat suppression than frequency-selective methods such as fat saturation in regions where it is difficult to achieve a sufficiently uniform B_0 field. However, it also alters the signal contrast of other tissues in the image.

Simultaneous suppression of signal from multiple tissue types can be achieved by concatenating inversion modules with appropriately chosen inversion times.

Inflow

The delay between the inversion pulse and the data acquisition can be exploited to produce differential effects between inflowing blood and stationary tissue. An example is 'dark-blood' (or 'black-blood') imaging, which employs a double inversion preparation. Two inversion modules are executed in rapid succession, one of which is non-selective and the other of which is applied selectively over the imaging slice. Since tissue in the imaging slice experiences both inversion pulses, its magnetization will undergo no net change. By contrast, blood that is initially outside the imaging slice will experience the non-selective pulse but not the slice-selective pulse, and its magnetization will be inverted. As that blood flows into the imaging slice, its magnetization will begin to recover. Choosing the inversion time to match the null-point of blood ensures that inflowing blood will have zero magnetization at the time of acquisition, and will therefore appear dark on the image. The dark-blood preparation is used in fast spin echo imaging of the heart to improve depiction of the endocardial border. By adjusting the parameters of the fast spin echo acquisition, the sequence can be optimized for visualizing morphology or inflammation [6].

Combinations of slice-selective and non-selective inversion pulses can also be designed to produce MR angiograms of the renal arteries [7] and to measure tissue perfusion via arterial spin labeling [8].

T_2-preparation

Although fast spin echo sequences can be used to provide T_2 weighting of stationary tissues, they produce variable signal in flowing blood. This is because flowing spins may not experience all the refocusing pulses if they move out of the imaging slice, or they may be dephased due to their motion with respect to the imaging gradients. An alternative way to achieve T_2 weighting, which is largely insensitive to flow, is by means of a T_2-preparation module. This consists of a 90° excitation pulse, followed by one or more 180° refocusing pulses, and terminated by a -90° tip-back pulse. Between the first and last pulses, the magnetization lies in the transverse plane, where it undergoes T_2 relaxation. The final pulse tips the remaining magnetization back onto the longitudinal axis, resulting in T_2-weighted longitudinal magnetization that can be read out using any of the standard data acquisition strategies described earlier. The absence of magnetic field gradients makes the module largely insensitive to flow. To minimize sensitivity to B_1^+ and B_0 inhomogeneity, the excitation, refocusing and tip-back pulses are usually implemented using composite or adiabatic pulses [9].

Applications of T_2-prep modules include improvement of contrast between vessel lumen and myocardium in coronary MRA [6] and measurement of venous oxygenation [10].

Conclusion

Magnetization-preparation modules, including the saturation, inversion and T_2-preparation schemes described above, can be combined in multiple ways to generate a huge variety of contrast mechanisms, and are responsible in large part for the enormous versatility of MRI as an imaging modality.

References

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