CHAPTER 7

Measurement parameters and image contrast

As mentioned in Chapter 6, many of the measurement parameters of a pulse sequence may be modified through the user interface software. The particular parameters will be determined by the scanner manufacturer, based on the template pulse sequence (which parameters are appropriate for modification) and the desired interface design (which parameters should the operator be allowed to modify). There are three general criteria that should be considered when modifying any measurement parameter as described in Chapter 5: acceptable scan time, adequate spatial resolution, and sufficient contrast between tissues relative to the background noise (contrast-to-noise ratio). These criteria are often in conflict in clinical imaging. For example, obtaining images with high spatial resolution (pixel sizes < 0.7 mm) and high contrast-to-noise ratio between tissues will require long scan times due to signal averaging. For each scan, the important criterion must be identified so that appropriate parameter variations can be made. One complication is that, while the scan time and spatial resolution for the final image can be calculated before the scan begins, the contrast-to-noise ratio cannot be determined prior to the measurement. This is because the measured signal amplitude depends on the specific tissue(s) within the imaging volume and their relaxation times. Another complication is that the most important criterion may change, depending on the anatomical region under observation. For example, imaging of the central nervous system will normally have scans with longer measurement times so that higher spatial resolution and contrast-to-noise ratios can be obtained. By comparison, to reduce motion, imaging of the thoracic or abdominal cavity employs scans with short measurement times and will compromise on the spatial resolution and contrast-to-noise ratio to achieve them.

As mentioned above, the specific parameters that are variable within the user interface and their specific definitions will be determined by the manufacturer. However, many parameters are commonly available in most pulse sequences. One approach to categorizing them is by their effect on the final image. Intrinsic parameters modify the inherent signal produced by a volume element of tissue (voxel). These parameters probe the characteristic tissue properties that are

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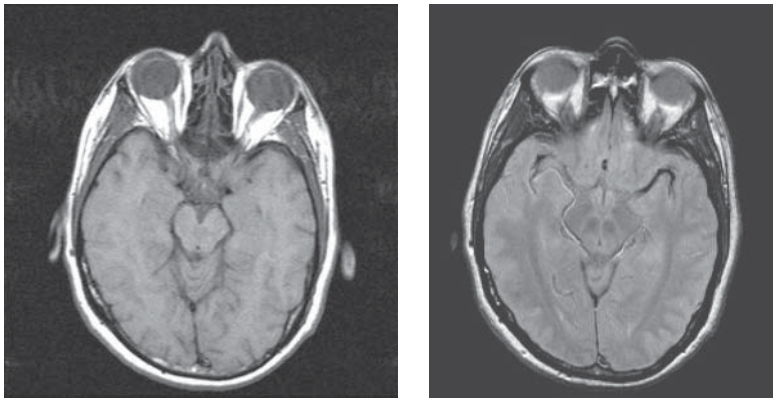
the response to the measurement procedure. Intrinsic parameters affect only the signal-producing portion of the image, which is normally patient anatomy and not background air. Extrinsic parameters influence the mechanics of data collection (e.g., voxel size) or other factors external to the tissue. They typically affect the spatial resolution or general background noise levels in the final image. Many of these parameters are specific to the particular choice of pulse sequence used for the measurement and may not be available in all instances. The definitions below are the common ones for these parameters.

7.1 Intrinsic parameters

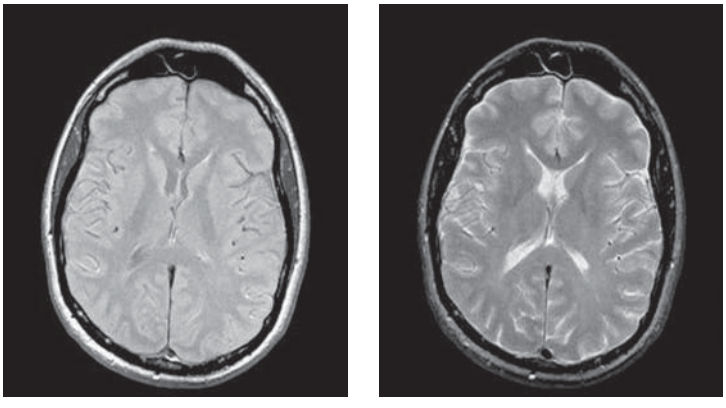
Repetition time, *TR*, measured in ms, is the time between successive RF excitation pulses applied to a given volume of tissue. In conjunction with the excitation angle (see following), *TR* determines the amount of *T1* weighting contributing to the image contrast. If all other factors are equal, longer *TR* allows more time for the RF excitation energy to be dissipated by the protons through spin-lattice relaxation, producing images with less *T1* weighting (Figure 7.1). For a multislice loop, *TR* limits the number of slices that can be acquired during the measurement.



(a) (b)

Figure 7.1 *TR* effects on image contrast. Longer *TR* allows more time for *T1* relaxation and produces more signal from tissues with long *T1* values. Other measurement parameters are: pulse sequence, spin echo; *TE*, 30 ms; acquisition matrix, N_{PE}, 224 and N_{RO}, 256; FOV, 201 mm PE × 230 mm RO; N_{SA}, 1; slice thickness, 5 mm. (a) *TR* of 500 ms; (b) *TR* of 2000 ms. Note reversal of contrast between gray matter and white matter in (b) compared to (a).

Echo time, *TE*, measured in ms, is the time between the excitation pulse and the echo (signal maximum). It determines the amount of *T2* weighting for spin echo images (Figure 7.2). For gradient echo images, *TE* determines the amount of *T2** weighting and the ratio of fat and



(a) (b)

Figure 7.2 *TE* effects on image contrast. Longer *TE* allows more time for *T2* relaxation and produces more signal from tissues with long *T2* values. Other measurement parameters are: pulse sequence, spin echo; *TR*, 2000 ms; acquisition matrix, N_{PE}, 224 and N_{RO}, 256; FOV, 201 mm PE × 230 mm RO; N_{SA}, 1; slice thickness, 5 mm. (a) *TE* of 30 ms; (b) *TE* of 80 ms. Note bright signal from cerebrospinal fluid (CSF) in (b) compared to (a).

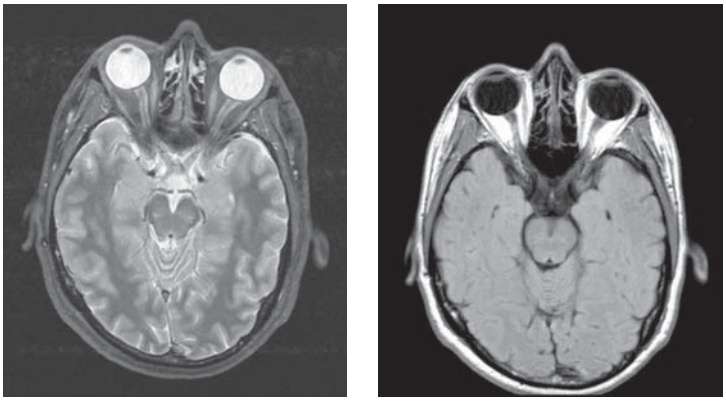
water signal contributions (see Figure 2.6). Longer *TE* allows more time for proton dephasing and produces lower signal amplitudes. In echo train spin echo, echo planar imaging, and MP gradient echo sequences, *TE* is considered to be effective since all echoes used in image reconstruction are not acquired at the same echo time.

Inversion time, *TI*, measured in ms, is the time between the 180° inversion pulse and the imaging excitation pulse. *TI* is used in inversion recovery (IR), echo train IR, and magnetization-prepared gradient echo sequences, and determines the amount of time allowed for *T1* relaxation following the inversion pulse. Short TI times allow minimum *T1* relaxation, while long *TI* times allow significant *T1* relaxation prior to the imaging excitation pulse. Proper choice of *TI* enables signal suppression of tissues based on their *T1* relaxation times (Figure 7.3).

Echo train length (also known as the turbo factor) is the number of echoes (number of phase encoding steps) measured following an excitation pulse that are used to create an image. The echo train length is used in echo train spin echo, echo train IR, and echo planar sequences. For long *TR*, longer echo train lengths allow shorter scan times through more efficient data collection. The sequence kernel time (minimum *TR* per slice) is longer with longer echo train lengths, producing greater signal attenuation in the late echoes through *T2* relaxation as well as requiring a longer minimum *TR* to obtain an equal number of slices. The number of phase encoding steps for the measurement is a multiple of the echo train length.

Echo spacing, measured in ms, is the time between each echo of the echo train. The echo spacing is used in ETSE, echo train IR, and echo planar sequences. Longer echo spacing allows more time for *T2* relaxation between each echo. Shorter echo spacing reduces the sequence kernel time.

The excitation angle (also known as the flip angle), measured in degrees, is the amount of rotation away from the equilibrium axis that ***M*** undergoes through RF absorption. If not variable under the operating software, the excitation angle is usually 90° in order to generate the maximum transverse magnetization. The excitation angle is also proportional to the amount



(a) (b)

Figure 7.3 *TI* effects on image contrast. Longer *TI* allows more time for *T1* relaxation following the inversion pulse. The choice of *TI* can cause signal suppression of different tissues. Other measurement parameters are: pulse sequence, echo train spin echo, five echoes; *TR*, 7000 ms; *TE*, 14 ms; acquisition matrix, N_{PE}, 224 and N_{RO}, 256; FOV, 201 mm PE × 230 mm RO; N_{SA}, 1; slice thickness, 5 mm. (a) *TI* = 140 ms (fat suppression); (b) *TI* = 2100 ms (CSF suppression).

of energy absorbed and the amount of signal produced by the protons. The excitation angle, together with *TR* and the *T1* values for the individual tissues, determines the amount of *T1* weighting present in an image. For gradient echo sequences, the Ernst angle α_E is the excitation angle that produces the maximum signal from a tissue for a particular *TR*:

$$\cos(\alpha_E) = \exp(-TR/T1) \tag{7.1}$$

7.2 Extrinsic parameters

Slice thickness, *TH*, measured in mm, is the volume of tissue in the slice selection direction that absorbs the RF energy during excitation and generates the signal. Variation in slice thickness is usually accomplished through changing the magnitude of G_{SS}. Thicker slices provide more signal per voxel whereas thinner slices produce less partial volume averaging.

Slice gap, measured in mm, is the space between adjacent slices. The slice gap may also be expressed as a fraction of the slice thickness, depending on the operating software. The slice gap allows the user to control the size of the total imaging volume by increasing or decreasing the space between slices. The slice gap also provides a method to compensate for the imperfect RF excitation pulses. If the slices are closely spaced, excitation pulses applied to adjacent slice position partially overlap and excite the same region of tissue because the slice excitation profiles are not uniform. This situation is known as crosstalk (Figure 7.4). Due to the rapid RF pulse application, these regions of overlap become saturated

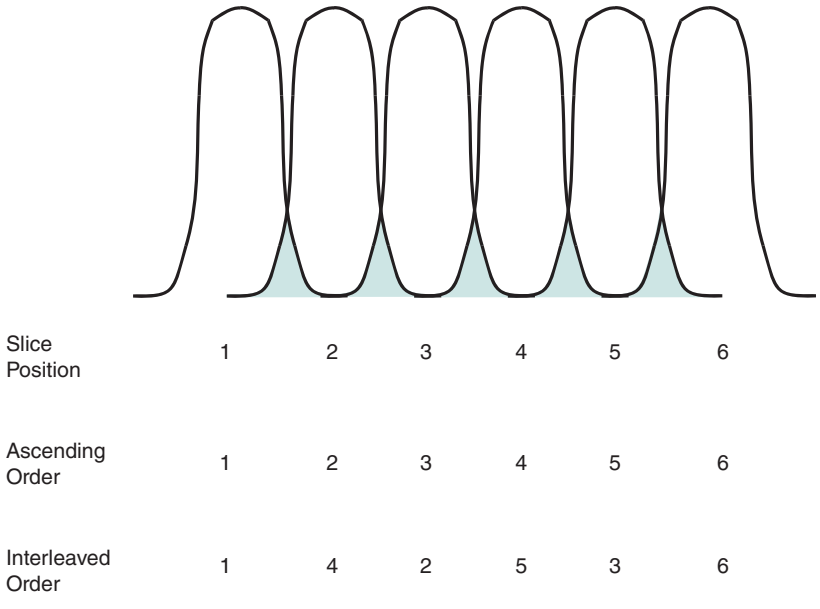


Figure 7.4 Excitation order and crosstalk. If the slices are closely spaced, the bases of adjacent slices overlap (crosshatched regions). Tissues located in this overlap region experience RF pulses from both slices and become saturated. This double excitation called crosstalk, causes reduced signal from these regions. The order of slice excitation also determines the contribution of crosstalk to the image intensity. Ascending order of excitation (second row) acquires data from adjacent slice positions in successive time periods. Interleaved ordering (third row) acquires data from every other slice position first, then acquires data from the intermediate positions. This ordering will minimize the effects of crosstalk between all slices.

and contribute little to the detected signal. The slice gap allows space between adjacent slice positions and reduces the extent of crosstalk for the measurement.

Excitation order refers to the temporal order in which slices are excited during the measurement. Two ordering schemes are typically used (see Figure 7.4). Sequential ordering excites adjacent slice positions in successive time periods. This approach is preferred when relative timing of adjacent slices is critical, such as for electrocardiogram-triggered studies of the heart (see Chapter 10). Interleaved ordering excites alternate slices in successive time periods. Interleaved ordering allows the maximum amount of time for *T1* relaxation of the overlap region prior to the subsequent excitation pulse. The effects of crosstalk are reduced for all slices as much as possible. Arbitrary ordering may also be performed if permitted by the operating software.

The number of partitions, N_{PART}, is used in 3D volume studies and corresponds to the number of slices into which the excited volume is divided. The slices have a signal derived from the total excited volume and are contiguous. The effective slice thickness is the volume excited (thickness) divided by N_{PART}. The scan time for a 3D sequence is linearly proportional to N_{PART} (equation (4.8)).

The field of view, *FOV*, measured in mm^2, specifies the area from which the MR signals are accurately sampled. The FOV may be specified separately for the readout and phase encoding directions (permitting anisotropic or rectangular FOV) or it may be listed as a single number (isotropic or square FOV). Decreasing the FOV is accomplished by increasing the corresponding gradient amplitude. Increasing spatial resolution may be achieved by decreasing the FOV, which decreases the voxel size at the expense of the SNR.

The acquisition matrix (N_{PE}, N_{RO}) defines the raw data sampling grid used for the measurement of the base image. It consists of two numbers: one specifies the number of phase encoding steps (N_{PE}) and the other specifies the number of readout sampling data points (N_{RO}). Different manufacturers have different conventions regarding which number is specified first. The acquisition matrix divides the FOV into individual areas which, together with the slice thickness, define the voxel size. Increased spatial resolution may be obtained by using larger acquisition matrices to produce smaller voxels. Data acquired beyond that necessary for defining the image FOV is referred to as oversampling and is used to reduce the presence of high-frequency aliasing artifacts (see Chapter 9) and increase the SNR (see Chapter 5). Oversampling in the readout direction does not increase the measurement time, while oversampling in the phase encoding direction increases the scan time proportional to the amount of oversampling.

The number of signal averages, N_{SA} (also known as N_{EX}, the number of excitation pulses), is the number of times the signal from a given slice for a given phase encoding amplitude is measured and added together. Signal averaging is performed in order to increase the SNR ratio. Depending on the operating software, all acquisitions may be performed at each phase encoding amplitude for a slice, or the entire set of phase encoding amplitudes may be measured for each slice before performing the second acquisition for any slice. The SNR is proportional to the square root of N_{SA} while the measurement time is proportional to N_{SA}.

The receiver bandwidth, BW_{REC}, measured in Hz, is the maximum frequency (Nyquist frequency) that can be accurately digitized. The Nyquist frequency depends on the sampling time and N_{RO}. The receiver bandwidth may also be expressed as the total bandwidth over the entire readout FOV or as the bandwidth per pixel (frequency resolution), depending on the particular convention. A lower BW_{REC} improves the SNR at the expense of potentially larger chemical shift artifacts (see Chapter 9).

7.3 Parameter tradeoffs

As mentioned previously, the three criteria used for determining a "good" measurement protocol are sufficient spatial resolution to resolve the underlying anatomy, reasonable signal-difference-to-noise (termed contrast-to-noise) ratio between tissues, and an acceptable measurement time. In general, these three criteria conflict with one another,

and the difficulty in protocol optimization is to obtain the proper balance between them. In addition, optimal parameters for one set of tissues may or may not be optimal for another set of tissues. Finally, while each parameter can be specified separately, they are not completely independent. For example, *TE* must be less than *TR*. The following formulas can provide guidance on the tradeoff of one parameter versus another for SNR or image intensity. Tables 7.1 and 7.2 also summarize the parameter changes and their effects on spatial resolution, SNR, and measurement time.

Table 7.1 Measurement effects – extrinsic parameters.

Parameter	Direction of change	Effect on spatial resolution	Effect on S/N ratio	Effect on scan time
TH	Increase	Linear decrease	Linear decrease	None
N_{PART}	Increase	Linear increase	Square root increase	Linear increase
FOV_{RO}	Increase	Linear decrease	Linear decrease	None
FOV_{PE}	Increase	Linear decrease	Linear decrease	None
N_{RO}	Increase	Linear increase	Square root increase	None
N_{PE}	Increase	Linear increase	Square root increase	Linear increase
N_{SA}	Increase	None	Square root increase	Linear increase
BW_{REC}	Increase	None	Square root decrease	None

Table 7.2 Measurement effects – intrinsic parameters.

Parameter	Direction of change	Effect on spatial resolution	Effect on S/N ratio	Effect on scan time
TR	Increase	None	Increase	Linear increase
TE	Increase	None	Decrease	None
Excitation angle, α	Increase	None	Increase for long *TR* Decrease for short *TR*	None

7.3.1 Intrinsic variables

Standard single spin echo signal intensity

$$I_{SE} = \exp(-TE/T2) * [1-2\ \exp(-[TR-TE/2]/T1)\ + \exp(-TR/T1)] \quad (7.2)$$

Standard inversion recovery signal intensity

$$I_{IR} = \exp(-TE/T2) * [1 - 2\ \exp(-Tl/T1) + 2\ \exp(-[TR - TE/2]/T1) - \exp(-TR/T1)] \quad (7.3)$$

Standard spoiled gradient echo signal intensity

$$I_{SGE} = \exp(-TE/T2^*) * \sin\alpha * (1 - \exp(-TR/T1))/[1 - \cos\alpha * \exp(-TR/T1)] \quad (7.4)$$

Equations (7.2), (7.3), and (7.4) assume exact RF pulses of the desired excitation angle. For the spin echo and inversion recovery equations, the excitation angle is assumed to be 90° and the refocusing pulses are assumed to be 180°. For the spoiled gradient echo equation, the excitation angle is α.

7.3.2 Extrinsic variables

Standard 2D imaging acquisition

$$SNR_{2D} = R * I * (TH) * (FOV_{RO}/N_{RO}) * (FOV_{PE}/N_{PE}) (N_{SA}N_{RO}N_{PE}/BW_{REC})^{1/2} \quad (7.5)$$

Standard 3D imaging acquisition

$$SNR_{3D} = R * I * (TH/N_{PART}) * (FOV_{RO}/N_{RO}) * (FOV_{PE}/N_{PE}) * (N_{SA}N_{RO}N_{PE}N_{PART}/BW_{REC})^{1/2} \quad (7.6)$$

Equations (7.5) and (7.6) assume uniform tissue content and relaxation behavior throughout the excited volume. *R* and *I* contain constants and terms based on the intrinsic parameters as described previously and in equation (5.1). The terms in parentheses represent voxel dimensions and the term inside the square root is the time spent measuring each voxel's signal, so equations 7.5 and 7.6 simplify to equation (5.1). These equations may be used to estimate SNR changes for combinations of parameter changes. For example, the SNR changes in a linear manner with variation in slice thickness, but changes linearly with the square root of N_{SA}; therefore a twofold reduction in slice thickness may be offset by a fourfold increase in N_{SA}.