5 Factors Affecting the Signal-to-Noise Ratio

In the preceding chapters we have learned how an MR signal is generated and how the collected signal is processed to create an MR image. What we have disregarded so far is that the MR signal can be degraded by noise. Image noise results from a number of different factors:

— Imperfections of the MR system such as magnetic field inhomogeneities, thermal noise from the RF coils, or nonlinearity of signal amplifiers.
— Factors associated with image processing itself.
— Patient-related factors resulting from body movement or respiratory motion.

The relationship between the MR signal and the amount of image noise present is expressed as the signal-to-noise ratio (SNR). Mathematically, the SNR is the quotient of the signal intensity measured in a region of interest (ROI) and the standard deviation of the signal intensity in a region outside the anatomy or object being imaged (i.e. a region from which no tissue signal is obtained).

A high SNR is desirable in MRI. The SNR is dependent on the following parameters:

— Slice thickness and receiver bandwidth
— Field of view
— Size of the (image) matrix
— Number of acquisitions
— Scan parameters (TR, TE, flip angle)
— Magnetic field strength
— Selection of the transmit and receive coil (RF coil)

Before we discuss the effects of each of these parameters, it is first necessary to clarify some concepts.
5.1 Pixel, Voxel, Matrix

An MR image is digital and consists of a matrix of pixels or picture elements. A matrix is a two-dimensional grid of rows and columns. Each square of the grid is a pixel which is assigned a value that corresponds to a signal intensity. Each pixel of an MR image provides information on a corresponding three-dimensional volume element, termed a voxel (Fig. 19). The voxel size determines the spatial resolution of an MR image.

The size of a voxel can be calculated from the field of view, the matrix size, and the slice thickness. In general, the resolution of an MR image increases as the voxel size decreases.

Fig. 19. A voxel is the tissue volume represented by a pixel in the two-dimensional MR image

5.2 Slice Thickness and Receiver Bandwidth

To achieve optimal image resolution, very thin slices with a high SNR are desirable. However, thinner slices are associated with more noise, and so the SNR decreases with the slice thickness. Conversely, thicker slices are associated with other problems such as an increase in partial volume effects.

The poorer SNR of thin slices can be compensated for to some extent by increasing the number of acquisitions or by a longer TR. Yet this is ac-
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accomplished only at the expense of the overall image acquisition time and reduces the cost efficiency of the MR imaging system.

The receiver bandwidth is the range of frequencies collected by an MR system during frequency encoding. The bandwidth is either set automatically or can be changed by the operator. A wide receiver bandwidth enables faster data acquisition and minimizes chemical shift artifacts (Chapter 13.3) but also reduces SNR by about 30%. With a narrow bandwidth, on the other hand, there will be more chemical shift and motion artifacts and the number of slices that can be acquired for a given TR is limited.

An interslice gap is a small space between two adjacent slices. It would be desirable to acquire contiguous slices but interslice gaps are necessary in SE imaging due to imperfections of RF pulses. Because the resultant slice profiles are not perfectly rectangular (Fig. 20), two adjacent slices overlap at their edges when closely spaced. Under these conditions, the RF pulse for one slice also excites protons in adjacent slices. Such interference is known as cross-talk.

Fig. 20. a Ideal slice profile. b Distorted, non-rectangular slice profile in SE imaging with inadvertent excitation of adjacent slices reduces SNR. c With interslice gaps, the drop in SNR is minimized.
Cross-talk produces saturation effects and thus reduces SNR (▶ Fig. 20b).

In selecting an appropriate interslice gap one has to find a compromise between an optimal SNR, which requires a large enough gap to completely eliminate cross-talk, and the desire to reduce the amount of information that is missed when the gap is too large. In most practical applications an interslice gap of 25–50% of the slice thickness is used.

Alternatively, the undesired saturation of protons in adjacent slices can be reduced by *multislice imaging*, which will be discussed in ▶ Chapter 7.3. Scan times are somewhat longer unless a shorter TR is used.

Gradient echo (GRE) sequences are different. They do not require a 180° refocusing pulse and thus allow the acquisition of contiguous slices without interslice gaps.

### 5.3 Field of View and Matrix

There is a close relationship between field of view (FOV) and SNR. When matrix size is held constant, the FOV determines the size of the pixels. *Pixel size in the frequency-encoding direction* is calculated as the FOV in mm divided by the matrix in the frequency-encoding direction and *pixel size in the phase-encoding direction* as the FOV in mm divided by the matrix in the phase-encoding direction.

As illustrated in ▶ Fig. 21, pixel size changes with the FOV. A smaller FOV results in a smaller pixel size as long as the matrix is unchanged. Pixel size is crucial for the spatial resolution of the MR image. With the same FOV, a finer matrix (i.e. a matrix consisting of more pixels) results in an improved spatial resolution (▶ Figs. 22 and 23).

Conversely, a coarser matrix (i.e. one with fewer pixels) results in a poorer spatial resolution when the FOV is held constant (▶ Fig. 23).

From what has been said so far, one might conclude that the matrix should be as large as possible in order to encompass a maximum of picture elements. This is true in terms of image resolution but the minimum pixel size is limited by the fact that, in general, *SNR decreases with the size of the voxel*.

Another limiting factor is image acquisition or scan time, which increases in direct proportion to the matrix size. *Scan time* is the key to the economic efficiency of all MR systems and can be calculated by a simple equation.

\[
\text{Scan time} = TR \times \text{number of phase-encoding steps} \times \text{number of signal averages (NSA)} [\text{echo train length (ETL)}].
\]
A “trick” can be used to achieve a high spatial resolution in a reasonable scan time. This is done by reducing the field of view only in the phase-encoding direction (rectangular field of view) and is possible because spatial resolution is determined by the matrix size in the frequency-encoding direction while scan time is determined by the matrix size in the phase-encoding direction. Reduction of the matrix size in the phase-encoding direction therefore does not reduce spatial resolution. Filling only one-half the normal number of phase-encoding lines in k-space reduces imaging time and the FOV by 50%. However, use of a rectangular FOV may be associated with wraparound artifacts when signal outside the FOV in the phase-encoding direction is mapped back into the image at an incorrect location.
This kind of foldover can be suppressed by specific anti-aliasing options such as “no phase wrap”. Moreover, reduction of the FOV in the phase-encoding direction is associated with a slight drop in SNR. A rectangular FOV is typically used to image the spine and extremities and for MR angiography.

Scan time can be shortened further on state-of-the-art scanners that allow one to use rectangular fields of view in combination with rectangular pixels.

Finally, various techniques of partial k-space acquisition (Figs. 24, 25, and 26) save scan time without one having to change the voxel size. In partial Fourier imaging, only half the lines (or slightly more) in the phase-encoding direction are filled (Fig. 24) while fractional or partial echo imaging (Fig. 25) refers to a technique with incomplete filling of the frequency-
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**Fig. 24.** Complete k-space sampling. Each data point represents one frequency-encoding line and one phase-encoding line.

**Fig. 25.** Partial Fourier imaging. Slightly less than half the k-space lines in the phase-encoding direction are not sampled (gray dots). These lines are interpolated.

**Fig. 26.** Fractional echo imaging. Slightly less than half of the k-space lines in the frequency-encoding direction are not filled directly (gray dots). The unfilled lines represent the echo portions that have not been sampled. The resulting MR image has a similar resolution but poorer SNR compared with an image generated with complete k-space sampling (Fig. 24) (as less “true” data is incorporated).
encoding lines by sampling only part of each echo. Both techniques rely on the inherent symmetry of k-space that allows one to interpolate the unfilled lines and to thus reconstruct an MR image when only half or slightly more than half the lines of k-space have been sampled. Both methods shorten scan time but this is accomplished at the expense of SNR. Partial Fourier and fractional echo imaging are needed for fast imaging techniques (▶ Chapter 8).

In routine 2D Fourier transform or spin-warp imaging, k-space is filled sequentially one line at a time (linear or Cartesian k-space acquisition). More sophisticated sequences use spiral k-space trajectories that fill the lines from the center toward the periphery (elliptical centric ordering of k-space, CENTRA). In MR angiography, for instance, this technique is used to fill the center of k-space with the data important for evaluating contrast enhancement patterns.

5.4 Number of Excitations

The number of excitations (NEX) or number of signal averages (NSA) denotes how many times a signal from a given slice is measured. The SNR, which is proportional to the square root of the NEX, improves as the NEX increases, but scan time also increases linearly with the NEX.

5.5 Imaging Parameters

Other parameters affecting the SNR are the sequence used, echo time (TE), repetition time (TR), and the flip angle. The SNR increases with the TR but the T1 effect is also lost at longer TRs. Conversely, the SNR decreases as the TE increases. With a short TE, the T2 contrast is lost. For this reason, the option of shortening TE to improve SNR is available only for T1-weighted sequences.

5.6 Magnetic Field Strength

Applying a higher magnetic field strength increases longitudinal magnetization because more protons align along the main axis of the magnetic field, resulting in an increase in SNR. The improved SNR achieved with high-field systems (▶ Chapter 14) can be utilized to generate images with an improved spatial resolution or to perform fast imaging.
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5.7 Coils

An effective means to improve SNR, without increasing voxel size or lengthening scan time, is selecting an appropriate radiofrequency (RF) coil. In general, an RF coil should be as close as possible to the anatomy being imaged and surround the target organ. The nearer the coil can be placed to the organ under examination, the better the resulting signal. RF coils can be used either to transmit RF and receive the MR signal or to act as receiver coils only. In the latter case, excitation pulses are delivered by the body coil. The basic coil types that are distinguished are briefly described below.

5.7.1 Volume Coils

Volume coils may be used exclusively as receive coils or as combined transmit/receive coils. Volume coils completely surround the anatomy to be imaged. Two widely used volume coil configurations are the saddle coil and the birdcage coil. Volume coils are characterized by a homogeneous signal quality. Another type of volume coil is the body coil, which is an integral part of an MR scanner and is usually located within the bore of the magnet itself. Head and extremity coils are further examples of volume coils.

5.7.2 Surface Coils

Most surface coils can only receive the MR signal and rely on the body coil for delivery of RF pulses. Combined transmit/receive surface coils are also available. Surface coils are used for spinal MRI and imaging of small anatomic structures.

5.7.3 Intracavity Coils

Intracavity coils are small local receive coils that are inserted into body cavities to improve image quality as a result of the closer vicinity to the target organ. In clinical MRI, endorectal coils are used for imaging of the prostate and the anal sphincter muscle. Experimental applications include endovascular imaging and imaging of hollow organs.
5.7.4 Phased-Array Coils

Phased-array coils serve to receive MR signals. A phased-array system consists of several independent coils connected in parallel or series. Each coil feeds into a separate receiver. The information from the individual receivers is combined to create one image. Phased-array coils yield images with a high spatial resolution and allow imaging with a larger field of view as they improve both SNR and signal homogeneity.

▶ Table 4 summarizes the factors affecting SNR.
▶ Table 5 summarizes the effects of matrix size, slice thickness, and FOV on spatial resolution.
▶ Table 6 summarizes the effects of different sequence parameters on scan time.

**Table 4.** Effects of different imaging and sequence parameters on signal-to-noise ratio (SNR)

<table>
<thead>
<tr>
<th>Change in parameter</th>
<th>SNR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Increasing slice thickness</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing FOV</td>
<td>Increases</td>
</tr>
<tr>
<td>Reducing FOV in phase-encoding direction (rectangular FOV)</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing TR</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing TE</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing matrix size in frequency-encoding direction</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing matrix size in phase-encoding direction</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing NEX</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing magnetic field strength</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing receiver bandwidth</td>
<td>Decreases</td>
</tr>
<tr>
<td>Employing local coils</td>
<td>Increases</td>
</tr>
<tr>
<td>Partial Fourier imaging</td>
<td>Decreases</td>
</tr>
<tr>
<td>Fractional echo imaging</td>
<td>Decreases</td>
</tr>
</tbody>
</table>
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Table 5. Effects of matrix size, slice thickness, and field of view (FOV) on spatial resolution

<table>
<thead>
<tr>
<th>Change in parameter</th>
<th>Spatial resolution</th>
</tr>
</thead>
<tbody>
<tr>
<td>Increasing matrix size</td>
<td>Increases</td>
</tr>
<tr>
<td>Using thicker slices</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing FOV</td>
<td>Decreases</td>
</tr>
</tbody>
</table>

Table 6. Effects of different sequence parameters on scan time

<table>
<thead>
<tr>
<th>Change in parameter</th>
<th>Scan time</th>
</tr>
</thead>
<tbody>
<tr>
<td>Using thicker slices</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing FOV</td>
<td>No direct effect</td>
</tr>
<tr>
<td>Using rectangular FOV (in phase-encoding direction)</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing TR</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing TE</td>
<td>Increases</td>
</tr>
<tr>
<td>Increasing matrix size in frequency-encoding direction</td>
<td>Increases</td>
</tr>
<tr>
<td>Partial Fourier imaging</td>
<td>Decreases</td>
</tr>
<tr>
<td>Fractional echo imaging</td>
<td>Decreases</td>
</tr>
<tr>
<td>Increasing NEX</td>
<td>Increases</td>
</tr>
</tbody>
</table>

References

1. Elster AD, Burdette JH (2001) Questions and answers in magnetic resonance imaging, 2nd ed. Mosby, St. Louis