Chapter 6

What You Set is What You Get: Basic Image Optimization

6.1 Introduction

We have seen that MRI is a digital, truly three-dimensional imaging modality of great flexibility with respect to image contrast and geometry. However, one of the downsides of this flexibility is a greater complexity in terms of the choice of scanning parameters. This aside, does MR have any other weaknesses? Yes it does: in general scan times are not negligible and there is a certain tendency towards artefacts (which we will investigate in the next chapter). However, most MR people would probably agree that the fundamental limitation in MRI is the signal-to-noise ratio (SNR). This is dependent on many factors, but unlike X-ray imaging, there is no radiation dose or milliampere-seconds (mAs) that can be increased to improve image quality. Good image quality depends upon making good scanner parameter choices. There are many parameters to tweak, buttons to press and dialogue boxes to click, and one way to learn about how each affects image quality would be to spend a lifetime tweaking! Alternatively, you could read this chapter which investigates the influence of various acquisition parameters and the practical trade-offs between SNR, contrast-to-noise, spatial resolution and scan time. You will see that:

- signal intensities and contrast are determined by the timing parameters TR and TE (also inversion time (TI) and flip angle (α) where appropriate);
- SNR is proportional to the voxel volume;
- signal scales with size parameters (field of view, slice width), and noise scales with averaging parameters (number of signal averages, phase-encode matrix size, frequency-encode matrix size) with an 'inverse square root relationship' (some parameters affect both);
- an appropriate choice of receive coil helps SNR;
- resolution is not usually the limiting factor;
- good **C**ontrast-to-**N**oise **R**atio (CNR) is essential for diagnostic-quality images;
- parameter juggling is often required to get a suitable scan time.

Knowledge of this chapter should enable you to predict the effect of changing the basic scan parameters: TR, TE, bandwidth, matrix, FOV, slice thickness and number of excitations (NSA) or **N**umber of signal **EX**citations (NEX). Fundamental aspects of image quality will be discussed in Chapter 11.

6.2 Looking on the Bright Side: What are we Trying to Optimize?

This section introduces the basic parameters: contrast, SNR, contrast-to-noise ratio (CNR) and resolution. These are illustrated in Figure 6.1. Simple mathematical definitions are given in Box 'Here's the Maths Bit'.

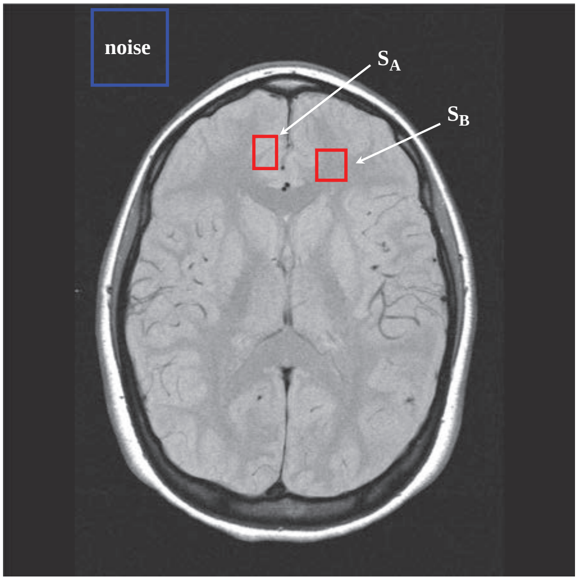

Figure 6.1 Definitions of contrast, signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR). See Box 'Here's the Maths Bit' for mathematical definitions.

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Here's the Maths Bit

Mathematically we can define contrast as

$$C = \frac{S_A - S_B}{S_A + S_B}$$

where S_A and S_B are signal intensities for tissues A and B. Refer to Figure 6.1. Signal-to-noise ratio (SNR) is defined as

$$\mathrm{SNR} = \frac{\mathrm{signal}}{\mathrm{noise}}$$

Contrast-to-noise ratio (CNR) is defined for tissues A and B as

$$\mathrm{CNR}_{AB} = \frac{S_A - S_B}{\mathrm{noise}}$$

In the simplest terms, spatial resolution of the voxels is related to the field of view (FOV) and matrix thus

$$\Delta x = \frac{\mathrm{FOV}}{N_{FE}} \qquad \Delta y = \frac{\mathrm{FOV}}{N_{PE}} \qquad \Delta z = \text{slice width}$$

6.2.1 Contrast

Contrast was introduced in Chapter 3 in terms of the image appearance, or relative brightness of different tissues and pathology. Image contrast arises (or doesn't) when tissues generate MR signals which have different intensities because of their physical properties, i.e. T_1 and T_2 relaxation times and proton density. You can refer to Figure 6.2 or Box 'Try It for Yourself 3: Predicting Contrast Appearance' to estimate the relative MR signals for a given tissue (if you know its T_1 and T_2). Mathematical expressions are given in Box 'Signal Calculator'.

Try It for Yourself 3: Predicting Contrast Appearance

Let's take two easily obtained substances: a saline bag and some cooking oil. The T_1 and T_2 of the saline are probably about 2000 ms and 1500 ms, and for the oil 200 ms and 180 ms. If we use a spin-echo (SE) sequence with TR = 600 ms and TE = 20 ms, what will we see?

Looking at Figure 6.2a, we have a ratio TR/T_1 of 0.3 and 2.5 for the saline and oil. For the oil the ratio of TE/T_2 is 0.1 so we should use the darkest of the grey curves. For saline TE/T_2 is 0.02. The blue curve is for TE/T_2 of zero, and as this is the closest value to the calculated ratio for saline, we should use the blue curve.

To calculate the oil signal, read off the value on the MR signal axis corresponding to TR/T_1 of 2.5 using the top grey curve. This gives about 0.83. For the saline, using the blue curve and the x axis value of 0.4, we get 0.28.

So we predict that oil will be brighter than water and that the contrast will be

$$C = \frac{0.83 - 0.28}{0.83 + 0.28} = 0.48$$

Now do the experiment on your scanner. Choose a single 5 mm slice which includes both substances and use a spin echo (not turbo or fast spin echo) with TR = 600 ms and TE = 20 ms. It doesn't matter what resolution you choose (256 × 256 will do fine). Check that the oil is indeed brighter and use regions of interest to measure the mean signal intensities in each container and calculate the contrast.

As a further test, using Figure 6.2a, how should you change TR to make the saline and oil closer to the same signal intensity? Looking at Figure 6.2b, for inversion recovery, what value of TI should you use to get zero signal from the oil? How will this look in the image? Try it for yourself to check your predictions.

One point to note using these graphs and the maths is that although they let you predict the changes to image appearance when you vary the parameters (TR, TE, TI), they do not include any tissue differences in proton density. Nor do they give you absolute values, but they do serve as a guide for predicting the image contrast.

Signal Calculator

To calculate the relative signal strength in terms of relaxation effects, use the following equations for the sequence-dependent relaxation factor *F*. If you don't like the look of the maths, you can use Figure 6.2 instead.

Spin Echo

$$F_{SE} \propto \left[1 - \exp\left(\frac{-TR}{T_1}\right)\right] \cdot \exp\left(\frac{-TE}{T_2}\right)$$

provided TE ≪ TR.

Inversion Recovery

$$F_{IR} \propto \left[1 - 2\exp\left(\frac{-TI}{T_1}\right) + \exp\left(\frac{-TR}{T_1}\right)\right] \cdot \exp\left(\frac{-TE}{T_2}\right)$$

also provided TE ≪ TR; or if TR > 5 × T_1 this simplifies to

$$F_{IR} \propto \left[1 - 2\exp\left(\frac{-TI}{T_1}\right)\right] \cdot \exp\left(\frac{-TE}{T_2}\right)$$

Gradient Echo

$$F_{GE} \propto \frac{\sin\alpha \cdot (1 - \exp(-TR/T_1)) \cdot \exp(-TE/T_2^*)}{1 - \cos\alpha \exp(-TR/T_1)}$$

for a 'spoiled' gradient echo (possibly called 'SPGR', 'FLASH' or 'T1-FFE' on your scanner). This will probably be the gradient-echo (GE) sequence you encounter most often. Other types of GE contrast are considered in Chapter 13.

6.2.2 SNR and CNR

In using the term 'signal' in this chapter we mean the pixel or voxel brightness in the image. This is related to the MR signal (i.e. what we measure from the coils). In any acquisition there is a finite amount of signal available dependent upon the MR characteristics of the tissue and the pulse sequence chosen. In Chapter 5 we considered the image as being made up of a number of voxels, each with a particular volume.

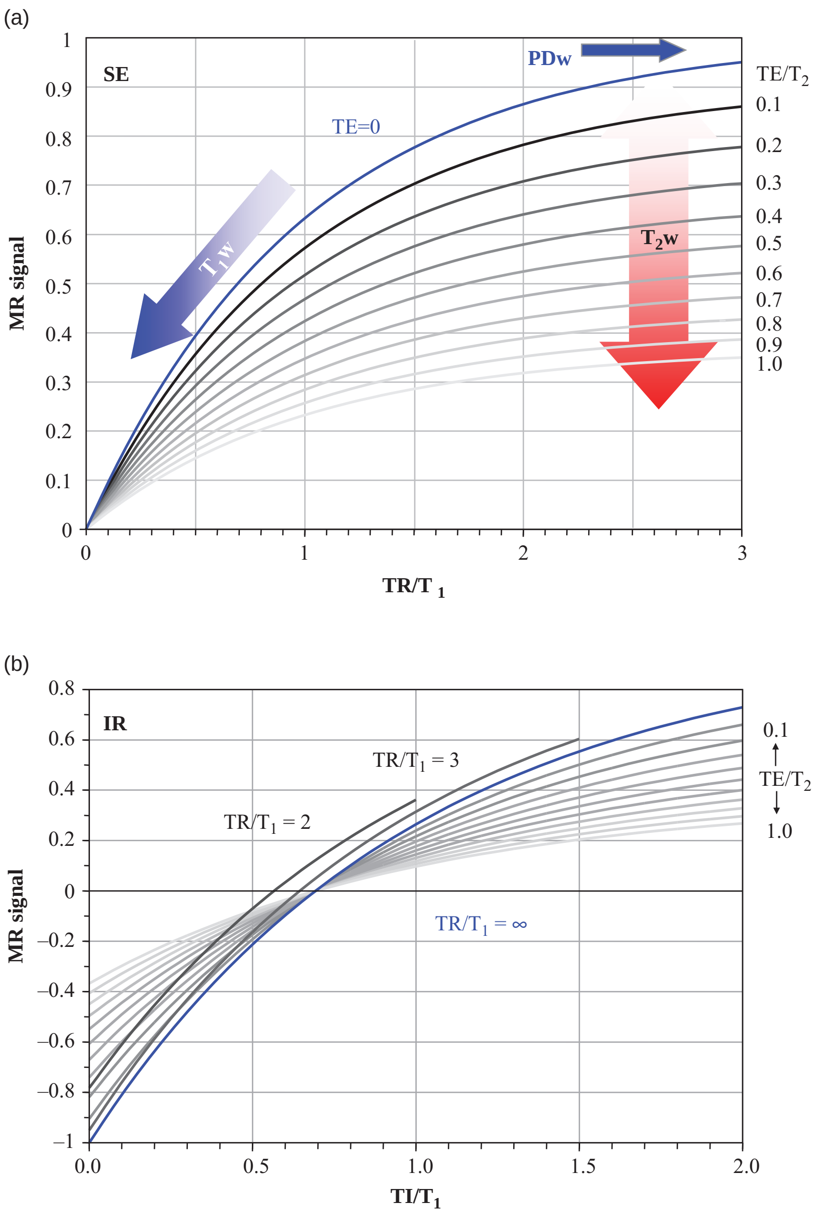

Figure 6.2 Normalized contrast behaviour. To use these graphs work out the ratio of tissue TR/T_1 and TE/T_2 or TI/T_1. Choose the nearest curve and read off the notional signal value. (**a**) Spin echo – blue line is for TE = 0. (**b**) Inversion recovery – blue line is for TE = 0, TR = ∞. The curves for TR/T_1 = 2 and 3 assume TE = 0 or that T_2 is very long. (**c**) Gradient echo (spoiled) all assuming TE = 0 or a long T_2^*.

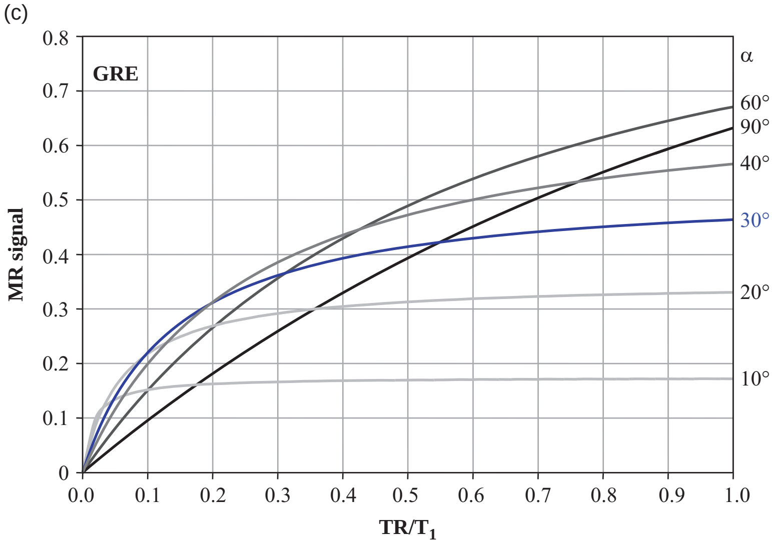

Figure 6.2 *(cont.)*

Since the MR signal that is returned from the patient during the scan has to be divided amongst the voxels that make up the image, the fundamental factor influencing the size of the signal is the number of protons within each voxel.

By 'noise' we don't mean the banging of the gradients (acoustic noise), but random differences in pixel values which give images a grainy, mottled look (like quantum mottle in a radiograph). Usually this noise originates mainly from the patient's tissues (see Box 'Who's Making All That Noise?').

In an MR image the individual voxels that make up the image will contain a mixture of signal and noise. The ratio of signal intensity in the image to noise level is the SNR. Images with a poor SNR will appear fuzzy. An important aspect of image optimization is to ensure that there is a high enough SNR for the images to be diagnostically useful. Low SNR may result in missing small details or the obscuring of subtle contrast changes. For this reason we often speak of a *contrast-to-noise ratio* (CNR). CNR is arguably the most important aspect of image quality. Ways of measuring image quality are considered in Chapter 11.

Who's Making All That Noise?

The noise comes from random fluctuations in electrical current. It therefore is called electronic noise, and exists in all electrical conductors. This obviously includes the MR coils with which we measure the signal, but it also includes the electrically conducting tissues of the patient. Human tissue contains many ions such as sodium, potassium and chloride which are electrically charged atomic particles carrying electrical currents within the body, e.g. in nerve conduction. These currents generate fluctuating magnetic fields which induce a noise voltage in the coil. The most effective way to reduce this noise is to use a small or dedicated anatomy coil. Where large fields of view are essential, array or matrix coils are usually best (see Section 10.5.2).

In the example of Figure 6.3 we see how the contrast and CNR are affected by the choice of slice thickness. If the slice is too thick, we get a good SNR but the contrast is reduced by a partial volume effect (mixing of the signals of the lesion and background). If the slice is too thin, the CNR may be too low to visualize the detail clearly.

Which is More Important, Resolution or SNR/CNR?

Certain applications such as MR angiography work best with higher resolution, but in general you need a certain SNR whatever the resolution. How much? As a rule-of-thumb an SNR higher than 20:1 offers little image quality advantage to the observer and excess

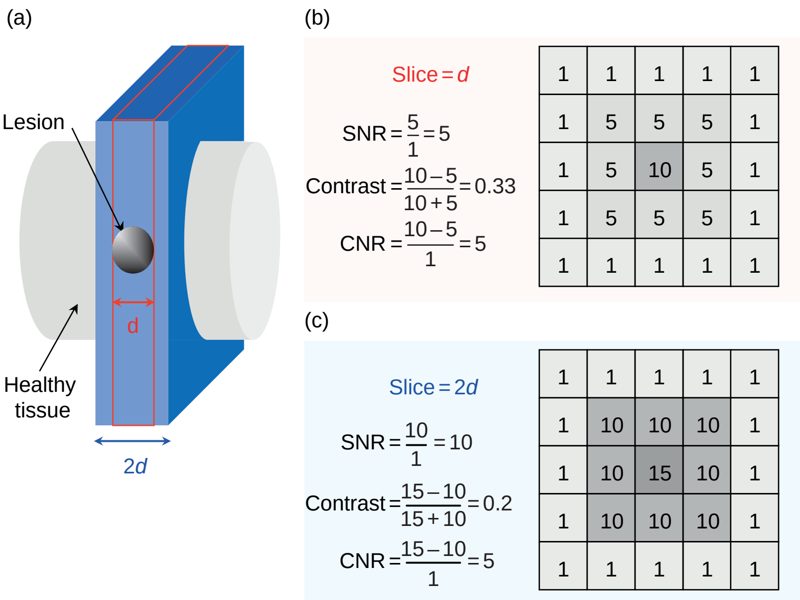

Figure 6.3 Contrast and CNR (a) example of changing slice thickness for a small lesion. The optimal contrast is obtained for a slice width less than or equal to the lesion diameter d. (b) Simulated pixel values for a small lesion with slice thickness d and calculated SNR, contrast and CNR. The lesion has a signal value of 10, surrounding tissue 5 and background noise 1. (c) Pixel values and image quality calculated values when the slice thickness is $2 \times d$. SNR is better, but contrast is down.

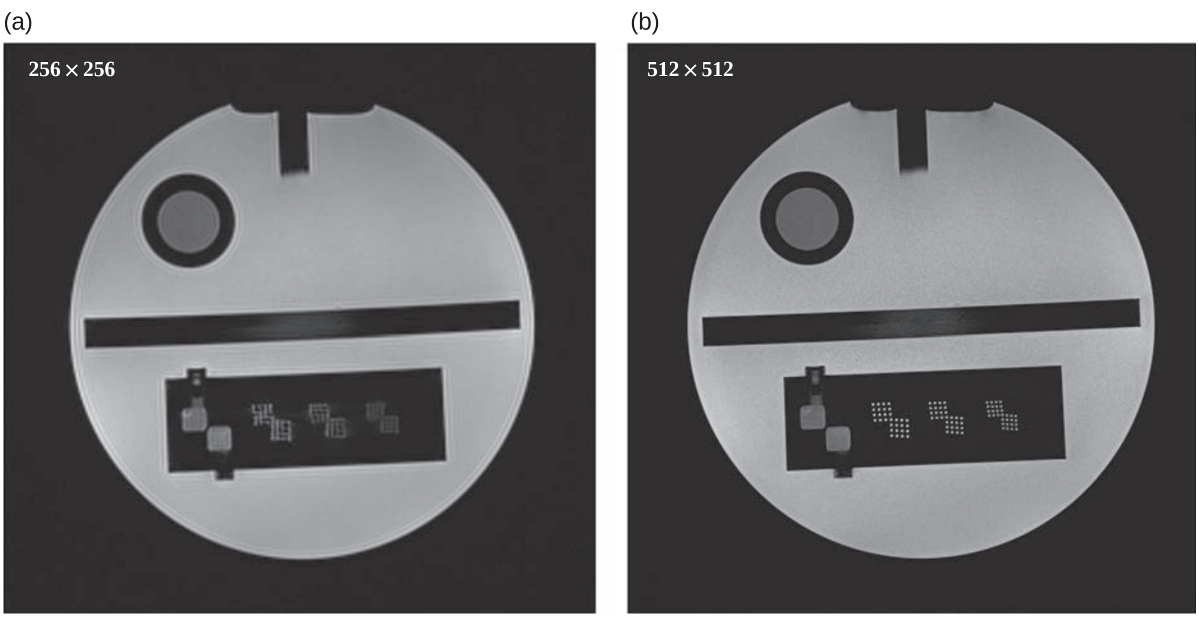

Figure 6.4 Effect of doubling the matrix, for the same FOV on the ACR phantom. The image in (b) has double the scan time of (a).

SNR would be best converted to either a larger matrix or reduced scan time. If SNR is adequate, high-resolution images will always look better but the diagnostic advantage of say 1024 matrix over 512 has not yet been established. Figure 6.4 shows the effect of different matrix sizes on image quality.

6.2.3 Resolution

The other important property of the images is *spatial resolution*. In MR we need to consider both the in-plane resolution, which may be different in each axis, and the through-plane resolution or slice width. Generally the latter is the largest dimension and is usually the more critical in visualizing a lesion.

6.3 Trading Places: Resolution, SNR and Scan Time

Just as compromises are common in real life, so too in MRI. Here we look at the trade-offs between the image quality parameters. For the mathematically minded, it's all in Box 'A Complicated Relationship: Resolution and SNR'.

A Complicated Relationship: Resolution and SNR

The signal is proportional to the voxel volume and the appropriate sequence-specific relaxation factor (given in Box 'Signal Calculator').

$$\text{signal} \propto \Delta x \cdot \Delta y \cdot \Delta z \cdot F_{sequence}$$

where Δx and Δy are the in-plane pixel dimensions and Δz is the slice width. F_{sequence} is the appropriate sequence-dependent factor from Box 'Signal Calculator'.

The noise is related to the receive bandwidth and 'averaging parameters':

$$\text{noise} \propto \frac{\sqrt{BW}}{\sqrt{NSA \cdot N_{PE} \cdot N_{FE}}}$$

where BW is the bandwidth across the whole image. In terms of the 'bandwidth per pixel' (bw) we can write

$$\text{noise} \propto \frac{\sqrt{bw}}{\sqrt{NSA \cdot N_{PE}}}$$

Putting this together with the signal equation we get

$$\text{SNR} \propto \frac{\Delta x \cdot \Delta y \cdot \Delta z \cdot F_{sequence} \cdot \sqrt{NSA \cdot N_{PE} \cdot N_{FE}}}{\sqrt{BW}}$$

where BW is the total receiver bandwidth. In terms of the field of view (FOV), we can say

$$\text{SNR} \propto \frac{\text{FOV}_{FE} \cdot \text{FOV}_{PE} \cdot \Delta z \cdot F_{sequence}\sqrt{NSA}}{\sqrt{BW \cdot N_{FE} \cdot N_{PE}}}$$

For systems which utilize the 'bandwidth per pixel' concept we get the following equation:

$$\text{SNR} \propto \frac{\Delta x \cdot \Delta y \cdot \Delta z \cdot F_{sequence} \cdot \sqrt{NSA \cdot N_{PE}}}{\sqrt{bw}}$$

A good rule of thumb is that if the scanning time is held constant, the achievable SNR is directly proportional to the voxel volume. So reducing the matrix from 256 × 256 to 128 × 128 and doubling NSA (to keep the same scan time) while maintaining the bandwidth per pixel will quadruple the SNR.

6.3.1 Resolution and SNR

Generally MRI resolution is pixel limited. That means that the smallest object or detail you can visualize in the image has the dimensions of a single pixel. So for a 256 matrix and a 25 cm FOV details of the order of 1 mm should be visible. In this, MR is quite distinct from digital radiography, computed tomography (CT) and ultrasound, where other processes (focal spot size, blurring, detector aperture, etc.) affect the ultimate resolution.

Three factors determine whether a particular detail or structure can be visualized in the image. Clearly there needs to be contrast between the structure and its surroundings. Second, if the resolution is insufficient, information about the object will not be transferred into the image by the image-formation process. Third, if the SNR or CNR is too low, the details of the structure may be obscured by image noise. You can get a feel for the effect of these parameter changes in Box 'Try It for Yourself 4: FOV and Matrix Size'.

Try It for Yourself 4: FOV and Matrix Size

To see the effect of field of view (FOV) and matrix size, you need a phantom with some fine structure, preferably with a range of sizes between 0.2 and 2 mm. These are commercially available, but you can make one of your own with a selection of plastic hair combs with different size 'teeth', including one designed to remove head lice (don't be embarrassed to buy this, you're going to use it for a scientific experiment!). Use a fairly deep plastic container and fill it with water, adding 1–2 ml of old gadolinium contrast to reduce T_1, and a few drops of detergent to break down the surface tension. Put the combs in the bottom, making sure there are no air bubbles trapped between them.

Put the container in the head or knee coil, and perform a localizer scan. Use a T_1-weighted spin-echo sequence and keep the receiver gain constant in order to be able to measure changes in SNR related to the other parameters. Select a coronal slice at the bottom of the container, with a slice width of about 5 mm so that you can see the effect of various matrix sizes. Then start changing FOV and matrix size. Be sure to change only one parameter at a time, keeping all others constant.

An important stage in image optimization therefore is to decide on the trade-off required between the voxel size required for an adequate SNR and the

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requirement for the voxel size to be small enough to permit the visualization of small anatomical or pathological details. Figure 6.4 shows images acquired on the ACR phantom with different resolution – 256 × 256 and 512 × 512. Clearly the 512 image (b) has greater spatial detail, but at the cost of more noise. Which do you prefer? Which is optimized?

6.3.2 Resolution and Scan Time

Spatial resolution in the frequency-encoding (FE) direction comes 'free' in terms of scan time (but not in terms of SNR) if the matrix is increased while keeping the FOV constant. To change the phase-encoding (PE) matrix, we have to acquire more lines of data, which takes time.

$$\text{Scan time} = \text{NSA} \times \text{TR} \times \text{N}_{\text{PE}}$$

In terms of the image slice, reducing the slice width will reduce the anatomical coverage unless slice gaps are increased. It is important to realize that changing the ratio of slice–slice increment to slice width in MRI is not like the 'pitch' in spiral CT. In MR the gaps are real gaps – and small lesions occurring exactly in a gap will be missed. The number of slices obtainable, in standard 2D mode, will be determined by the sequence timing parameters, particularly TR.

High-Resolution Brain Scan

Your radiologists have requested more spatial resolution from your brain scans in order to see smaller lesions, e.g. infarcts, MS plaques and micro haemorrhages. What do you change? Let's start from your existing 256 × 256 matrix. The simplest thing to do is to double the matrix to 512 × 512. Depending upon your MR system, this will give you either 50% or 35% of the SNR of your original protocol. Images are shown in Figure 6.5. The improved spatial resolution is evident, but is the increased level of noise acceptable (they are sure to complain about it)? The easiest way to improve SNR is to increase NSA. However this would take a four- to eight-fold increase, leading to a clinically unacceptable scan time – and you have already doubled the scan time by increasing N_{PE}.

To reduce the scan time you can increase the parallel imaging reduction factor or increase the turbo factor. The former will further erode SNR – so we don't recommend it, except in small, fractional amounts (e.g. a few tenths) if your scanner allows this. The latter is feasible provided the longer echo train length does not introduce blurring or hit SAR limits. In practice you may have to settle for an intermediate resolution of e.g. 512 × 384, reducing the PE matrix slightly.

This example is also considered in Box 'Try It for Yourself 5: Predicting SNR'.

6.3.3 Predicting the Effect on Image Quality

The relationships between SNR, CNR and spatial resolution are quite complicated, not the least in that many user-controllable scanner parameters affect them. How can we get our heads round what is going on sufficiently to predict the effect of parameter changes on image quality and, indirectly, diagnostic potential? It is clear that SNR and resolution are related. Throw in contrast and CNR and you have a recipe for confusion. One way is to understand the maths! Alternatively you could consider parameters as falling into two categories: *size parameters* determine how much signal is produced, *averaging parameters* reduce noise. Some parameters are a combination of both. Size parameters bear a linear relationship with SNR; averaging parameters have an inverse square-root relationship.

Field of view and slice width are size parameters only. They only affect the signal. Increasing them increases the signal and the SNR. Of course they affect resolution but fortunately that is intuitive. So doubling the slice width doubles the SNR. Halving the FOV while keeping the same matrix will quarter SNR, as we have changed two dimensions, the FOV in the phase-encoding direction and in the frequency-encoding direction.

Signal averaging (NSA or NEX) is an averaging parameter. It does not affect resolution and reduces the noise. Going from NSA = 1 to NSA = 4 will double SNR. NSA obviously affects scan time. The effect of signal averaging is illustrated in Figure 6.6. The image in (c) took four times longer to acquire than the one in (a) and twice that in (b). You have to ask yourself, is averaging worth the time?

N_{PE} and N_{FE} are combination parameters. They affect resolution and hence voxel volume, with a linear effect on signal. With standard two-dimensional Fourier transform (2D FT) MRI the acquisition of multiple 'lines' of data can be considered as a kind of averaging; therefore, they also affect the noise. So if we double N_{PE} we halve the

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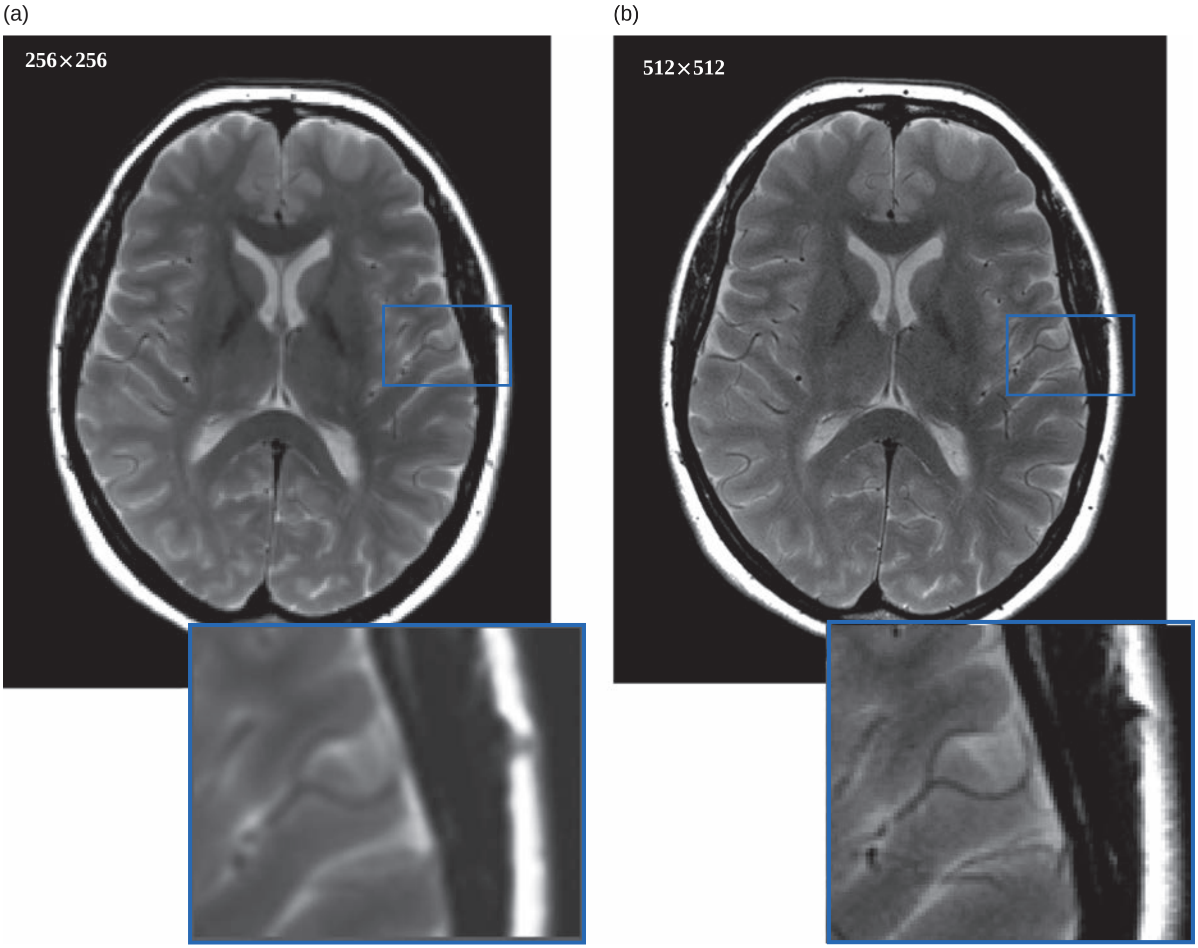

Figure 6.5 Effect of resolution on image quality. (a) 256 matrix (b) 512 matrix. The higher spatial resolution results in greater image noise (lower SNR).

number of protons in a voxel, producing half the signal, but we reduce the noise by √2 – so the net effect on SNR is a reduction of approximately 30% (1 ÷ √2).

Changing N_{FE} affects the noise, but the effect depends upon what happens to the 'bandwidth' and field of view. It is a classic result of electronic theory that noise is proportional to the square root of total bandwidth (√BW). If the total bandwidth does not change, increasing N_{FE} has the same effect as increasing N_{PE}: doubling the FE matrix while maintaining FOV gives a two-fold reduction in signal (because the voxel size has halved), but a √2 reduction in noise, and hence a √2 reduction in SNR. However, on systems which define bandwidth in hertz per pixel, the noise is unaffected but the signal and therefore the SNR will be reduced by half.

These inter-parameter dependencies are not obvious, so in Box 'Try It for Yourself 5: Predicting SNR' we present a 'paper app' for predicting the effect of parameter changes on SNR – the 'SNR abacus', illustrated in Figure 6.7.

Try It for Yourself 5: Predicting SNR

You can look at the maths in Box 'A Complicated Relationship: Resolution and SNR' to predict the effect of changing parameters on image quality. Alternatively you can use the 'SNR abacus' in Figure 6.7. Working from left to right, each arrow indicates the effect of doubling or halving the

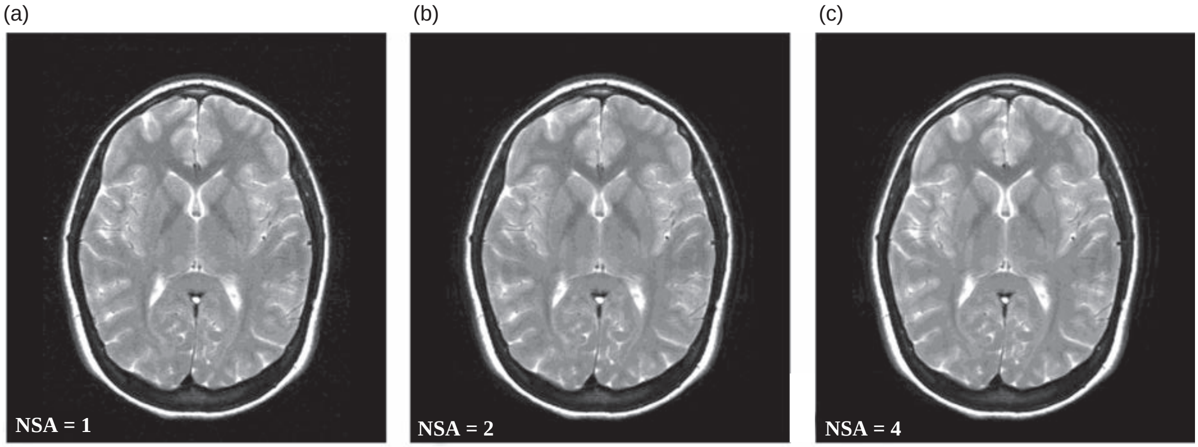

Figure 6.6 Effect of signal averaging (**a**) NSA = 1; (**b**) NSA = 2; (**c**) NSA = 4. Scan times increased proportionately. Image SNR improves with increasing NSA.

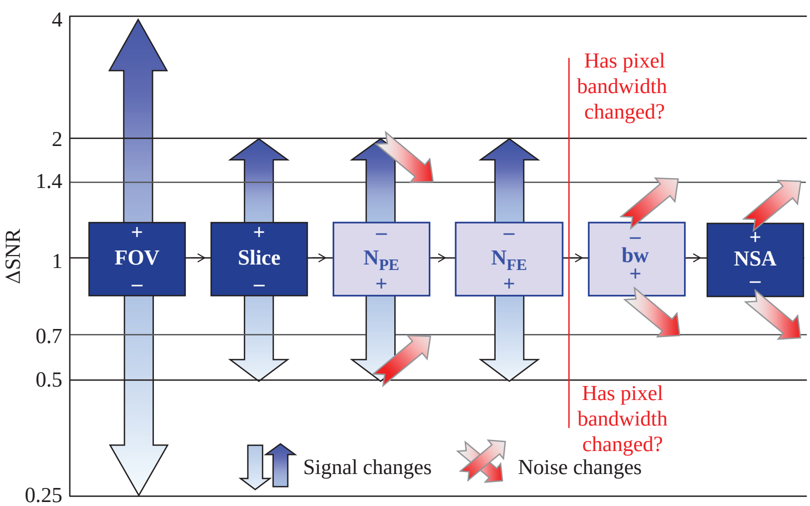

Figure 6.7 'SNR abacus' showing how parameter changes affect SNR, when each parameter value is either doubled or halved. To use the abacus, start at the left and work towards the right. For each parameter change follow the arrows to calculate the effect on SNR. To predict the overall effect on SNR multiply all the SNR changes together. Note that the arrow directions indicate the effect on SNR. An increase in either signal or noise is represented by the shading in the arrows, with a deeper colour indicating an increase in either signal or noise.

parameter in each box. Read off the values on the left for the effect on SNR. Then multiply up for each parameter change.

Here is an example. Suppose you double the matrix size, keeping FOV and slice width unchanged. The first two boxes have no effect (FOV and slice are unchanged). Doubling N_{PE} halves the signal but lowers the noise – with an overall SNR reduction of 0.7.

Doubling N_{FE} will halve the signal and the SNR. However, you need to consider what happens to the bandwidth. If the total bandwidth (BW) does not change, then the effect of doubling N_{FE} will be to halve the bandwidth per pixel (bw) and SNR will improve by $\sqrt{2}$. This will not affect systems which use a constant bandwidth per pixel.

As N_{PE} has increased, let's keep scan time the same by halving NSA, reducing SNR by a further $\sqrt{2}$ (0.7). Table 6.1 summarizes what has happened.

Starting from the same original image, let's try halving the FOV and doubling the NSA, but keep all other parameters unchanged. What do you get? You can now try it for yourself on the scanner using any phantom. Actually most scanners will do this calculation for you while you change the parameters, so you don't even have to do the scans – but do it anyway, it's good for your soul!

Table 6.1 Example of doubling the matrix for the same FOV on SNR

Step	SNR (constant BW)	SNR (constant bw)
FOV unchanged	× 1	× 1
Slice unchanged	× 1	× 1
Double N_{PE}	× 0.7	× 0.7
Double N_{FE}	× 0.5	× 0.5
Check bandwidth per pixel. . . .	× 1.4 (bw halved)	× 1.0 (bw unchanged)
NSA unchanged	× 1	× 1
TOTAL (multiplied)	**0.5**	**0.35**

6.3.4 2D Versus 3D

We have seen from the simple example in Section 6.2.2 that too thin a slice can ruin the CNR. What if we really want very thin slices, e.g. for multi-planar reformatting? The solution is to acquire a 3D volume instead of multiple 2D slices. How this works is described in Section 8.8. The scan time is increased to

$$\text{Scan time} = \text{NSA} \times \text{TR} \times N_{PE1} \times N_{PE2}$$

where N_{PE2} is the number of 'slices' or partitions. Since these are acquired sequentially, TR must be reduced using one of the sequences described in Chapter 13. The signal is reduced since the slice or 'partition' thickness is often very small. However, a reduction in the noise is achieved by using the extra dimension of phase encoding. Box '3D Maths' contains . . . er, the maths.

3D Maths

The maths for 3D is the same as for 2D, except that we have an extra 'combination' term N_{PE2}.

$$\text{SNR} \propto \frac{\Delta x \Delta y \Delta z \cdot F_{sequence} \cdot \sqrt{NSA \cdot N_{PE1} \cdot N_{PE2}}}{\sqrt{bw}}$$

or on systems which use the total bandwidth instead of bandwidth per pixel:

$$\text{SNR} \propto \frac{\Delta x \Delta y \Delta z \cdot F_{sequence} \cdot \sqrt{NSA \cdot N_{FE} \cdot N_{PE1} \cdot N_{PE2}}}{\sqrt{BW}}$$

6.4 Ever the Optimist: Practical Steps to Optimization

Optimization is a complicated subject. Our golden rules for image optimization are:

1. Set the required image contrast by choice of pulse sequence and basic timing parameters TR and TE (and α for gradient echo). In general for T_1 contrast using spin echo, a TR with a value intermediate to the tissue T_1 values of interest will produce the optimum contrast. The same is true for the choice of TE in terms of T_2 contrast.
2. Adjust for the desired geometry – slice, FOV, resolution (pixel size), remembering that this will involve a compromise with SNR.
3. Adjust for acceptable SNR. The chances are that your radiologist will be demanding more spatial resolution, higher matrix, etc., so your SNR will probably suffer. Some of the desired geometric changes may have to be scaled back or NSA increased.
4. Check your scan time. This may have exploded from a couple of minutes to hours if you are not careful! Do what it takes to get a clinically acceptable scan time.

Optimization will invariably lead to some compromises. However, other unforeseen pitfalls may also arise. Figure 6.8 summarizes which parameters affect others (but refer to Figure 6.7 for how they do so). Usually SNR ultimately limits what you can do. Some further practical advice is given below.

6.4.1 Check Your Slices

If you cannot get enough slices you may have to run the scan as two batches or *concatenations*, i.e. the slices are split into two or more blocks and the scans are run consecutively. To get sufficient coverage without increasing the number of slices, you could increase the slice thickness (at a cost of resolution loss and possible reduced contrast due to partial volumes), or increase the slice separation or gap (at a risk of missing a small lesion). If you are only one or two slices short, an increase in TR may help, but you will be changing the contrast which may have undesirable consequences. Bear in mind that any extra sequence options such as saturation bands (Section 7.2.5) and fat suppression (Section 7.3.3) will all add to the time required to acquire each slice and should be used sparingly if scan time is a major constraint.

Adapting a Protocol for Paediatric Scanning

In this example let's suppose that you are taking an adult brain protocol and modifying it for use in neonates. Let's suppose that the adult FOV is 23 cm and we need to reduce this to 17 cm. It doesn't seem that much but for the same matrix will only have about half the SNR. However, there is no effect on scan time. Assuming that the baby is asleep, or the child under general anaesthetic, we can get away with longer scan times. The only way to restore the SNR in this case is to increase NSA, and consequently the scan duration, by up to four times. In the case of a neonatal brain, the long relaxation times (longer than for adults) will be less susceptible to T_2 blurring than for adults and so the turbo factor (ETL) in TSE can be increased. Figure 6.9 shows adult and paediatric T_2-weighted brain scans from the protocols shown in Table 6.2. Note that the immature brain requires longer TR to give reasonable contrast on account of the longer T_1.

Table 6.2 Example scan parameters for the adult and paediatric brain examinations shown in Figure 6.9

	Unit	Adult	Paediatric
TR/TE/ETL	ms	2850/105/10	4650/118/15
FOV	cm	23	17
Matrix		320 × 256	320 × 288
Slice thickness	mm	5	3
Pixel bandwidth	Hz/pixel	150	190
Scan time	minutes: seconds	2:25	3:20

6.4.2 How to Boost SNR

Above all, make sure you *choose the best coil*. The simple rule of thumb is that ideally the receive coil should encompass the whole of the anatomical region of interest and no more. Smaller coils 'see' less noise. Array coils allow the MR receiver to see more useful anatomy without picking up more noise.

The easiest way to improve SNR is to increase the number of signal averages (NSA). This also increases the scan time, so this may not be the most time-efficient way of improving the SNR. Five methods to improve SNR that do not affect scan time are listed below:

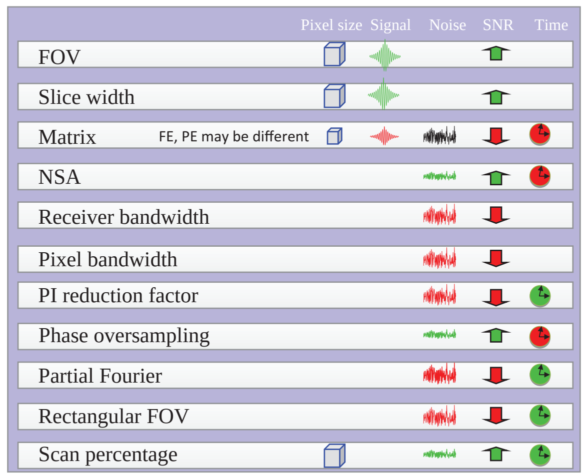

Figure 6.8 Interdependency of parameter changes. For an increase in the value of the listed parameter, beneficial image quality improvements are shown in green, detrimental in red.

(a) (b)

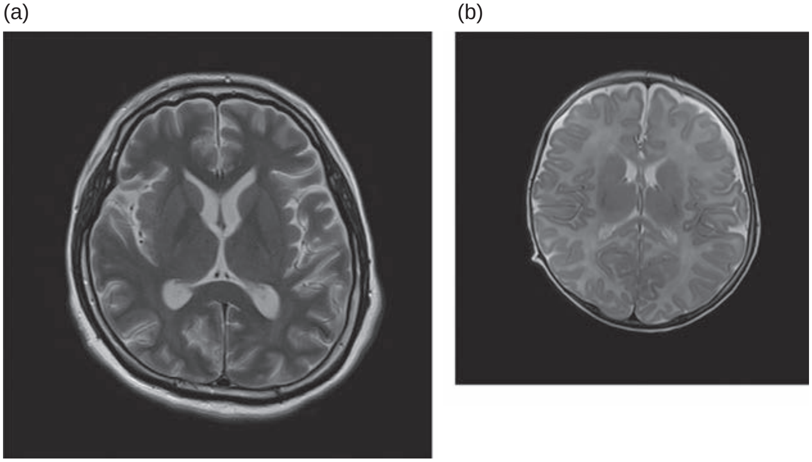

Figure 6.9 (a) Adult and (b) paediatric brain scans.

Increase the slice thickness

This improves the SNR in direct proportion to the slice thickness with no time penalty at all. But be aware of the effect on image contrast from partial volumes. If thin slices are essential, you should consider a 3D acquisition.

Increase the FOV

Increasing the FOV without changing the matrix size makes the in-plane pixels bigger, giving more signal without changing the noise. Spatial resolution is, of course, reduced and the desired part of the image will appear smaller, surrounded by more empty background. However, as you would normally match the FOV to the anatomy, this option may not be a terribly useful one.

Reduce the bandwidth

Reducing the bandwidth reduces the noise by a factor proportional to the square root of the reduction, e.g. halving the bandwidth reduces the noise by a factor of 1.4. The way this is done in practice depends on your scanner (see Box 'Adjusting the Bandwidth: A Scanner Guide'). The side-effect of reducing the bandwidth is an increase in the chemical shift, which may create an unwelcome artefact (see Section 7.3.1). Where chemical shift artefact is not a concern, the lowest bandwidth achievable is usually a good starting point. Your TE will determine the limit on how low you can set the bandwidth.

Adjusting the Bandwidth: A Scanner Guide

Some systems (e.g. General Electric) have the total MR receiver bandwidth in kilohertz (kHz) as a user-adjustable parameter. So to improve SNR you simply reduce the receiver bandwidth. For others (e.g. Siemens with Syngo software) the bandwidth per pixel is selectable.

For older Siemens scanners bandwidth is not adjustable. Instead, each pulse sequence has a bandwidth per pixel in hertz, given as the last number in the sequence name following the letter 'b', e.g. se15_b130 has a pixel bandwidth of 130 Hz. So you must select the appropriate pulse sequence from the list.

For other scanners (e.g. Philips) the pixel bandwidth is presented as the 'water–fat shift' (WFS). The meaning of this is explained in Section 7.3.1. To improve SNR on these systems you must increase the water–fat shift.

Select a pre-processing filter

Filtering of the MR signals prior to reconstruction improves SNR at a cost of reducing spatial resolution. Effectively we reduce the magnitude of the high spatial frequencies, where the noise is most apparent (because the signal here is low); in doing so, we attenuate genuine high spatial frequency information and thus reduce the resolution. Some manufacturers may apply filtering by default, in which case you are probably not told that it is happening, or as a user-selectable option. When should you filter? Basically, never, if you can avoid it. You would do better to reduce the number of phase-encode steps, which would improve SNR and save scan time. An exception to this is for certain so-called segmented sequences, e.g. fast or turbo spin echo where filtering reduces ringing artefacts (Chapter 12).

6.4.3 Check Your Scan Time Again

Even where scan time is not a major issue, shorter scans improve patient cooperation and reduce the opportunity for movement-related artefacts. They also improve your throughput. If the scan time is too long, a few tricks are available to help reduce it. One of the easiest ways is to reduce TR, but this will affect the contrast. The next easiest way is to reduce the number of phase-encode steps. Three ways of doing this are:

1 rectangular FOV;
2 partial Fourier;
3 reduced matrix.

How they work will be considered in Chapter 8, but they are illustrated in Figure 6.10. Using a higher reduction factor in parallel imaging will also help, but remember all time-saving techniques will reduce your SNR.

Scan Time Reduction in Abdominal Scanning

Let's suppose that your sequence lasts 20 s – and that's too long for your patient. You don't want to change either the matrix or TR as that will affect the diagnostic quality of the images. If you just need a minor reduction – say 10–20% in breath-hold time – then consider partial Fourier of 6/8 (0.75) or 7/8 (if available) which will maintain resolution but with a 5–10% loss in SNR; this is usually acceptable. Alternatively, using 80–90% scan reduction (phase

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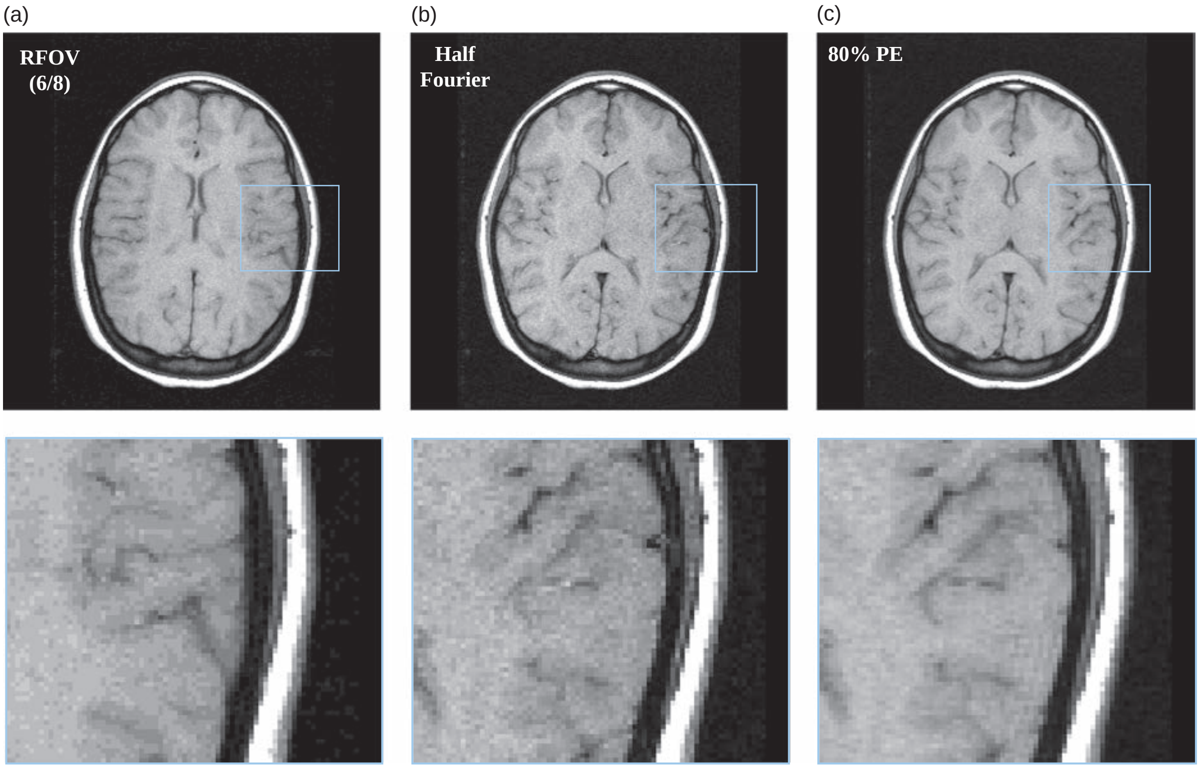

Figure 6.10 Images showing (a) rectangular field of view (RFOV): scan time 1 min 39 s, (b) half Fourier: scan time 55 s, (c) reduced matrix (80%), scan time 1 min 20 s. Full acquisition scan time would have been 2 min 8 s (spin echo, TR = 500 ms, TE = 15 ms). The half Fourier image is noisier, while the reduced matrix has less noise but reduced spatial resolution.

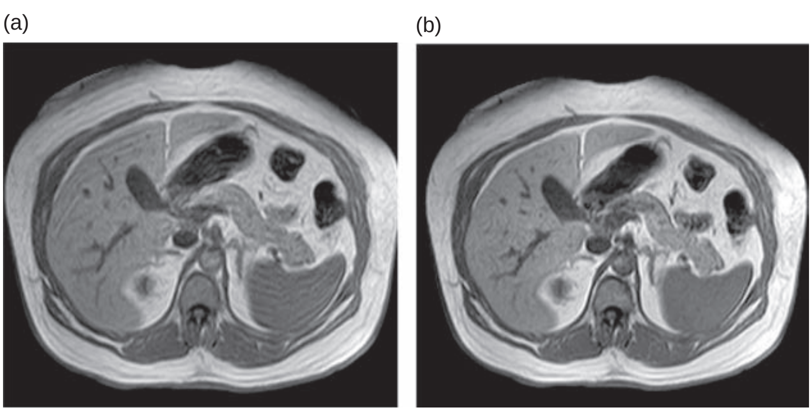

Figure 6.11 (a) Image from patient unable to maintain breath-hold. (b) Parallel imaging is used to reduce breath-hold time and respiratory movement artefact.

resolution) will maintain SNR but with a minor loss of resolution. A combination of 7/8 Fourier and 90% scan percentage would also work. Alternatively, on a Philips scanner you could use parallel imaging (SENSE) with a reduction factor of 1.1 to 1.2 without changing any of the other sequence parameters. This would give the desired reduction in scan time with a small but negligible loss in SNR but no loss in resolution.

However, let's say your patient is really sick and can barely manage a 10 s breath-hold. If NSA is greater than 1 then reduce it. If NSA = 1 already, then parallel imaging with a reduction factor of two or slightly more will get your scan time down to the required 10 s but with an SNR loss of about 30%. Figure 6.11 shows how this helps the overall image quality where the loss in SNR is more than compensated by the reduction in movement artefact.

6.4.4 Tweaker's Charter

Usually your scanner's protocols will be well set-up and optimized so off-the-cuff 'tweaking' will not be required. Remember that making seemingly innocuous parameter changes can have an adverse effect on the diagnostic quality of the images. This chapter is not a licence to tweak, but instead provides the theoretical and practical knowledge for when you do have to optimize a protocol, or when the specifics of the patient requires that some minor parameter modifications are made.

See also:

- Image artefacts: Chapter 7
- Spatial encoding using gradients: Chapter 8
- Parallel imaging: Chapter 14.

Further Reading

Brown MA and Semelka RC (2010) *MRI: Basic Principles and Applications*, 4th edn. Hoboken, NJ: Wiley-Blackwell, chapter 6.

Elster AD and Burdette JH (2001) *Questions and Answers in Magnetic Resonance Imaging*, 2nd edn. London: Mosby-Yearbook, chapter 4. Also on the web at http://mri-q.com [accessed 23 March 2015].

Hashemi RH and Bradley WG Jr (2010) *MRI The Basics*, 3rd edn. Baltimore, MD: Lippincott, Williams & Wilkins, chapter 17.