Chapter 7

# Improving Your Image: How to Avoid Artefacts

## 7.1 Introduction

As we all know, real life is far from perfect and MRI is just as disappointing in some ways! MR scanners do not have absolutely uniform magnetic fields, the gradients don't produce exactly the pulse shapes programmed by the pulse sequence and patients don't keep still. These problems, and many others, produce artefacts in MR images. An artefact is defined as any feature in an image which misrepresents the object in the field of view (FOV). This could be a bright signal lying outside the body, or lack of signal where there should be something. It might also be a distortion in the image, so that a straight line appears curved, or a certain area is artificially magnified or reduced. A large group of MR artefacts appear as 'ghost' images, where a faint copy of the anatomy appears in the image displaced in one direction or another.

In this chapter we will describe the most common artefacts encountered in MRI, along with ways to avoid or minimize them. The causes of artefacts can be broadly divided into four groups: motion, inhomogeneity, digital imaging artefacts, and hardware-related artefacts:

- *Motion artefacts* appear as ghosts along the phase-encode direction, and are produced by physiological motion or involuntary movement by the patient.
- *Inhomogeneity artefacts* usually cause signal intensity changes and image distortions, and are due to hardware imperfections and to the susceptibility effects within the human body.
- *Digital imaging artefacts* have a variety of appearances, and include phase wrap-around artefacts and problems arising from approximations and errors in the encoding process.
- *Hardware-related artefacts* are less common these days, but RF interference and spike noise are still important to recognize.

## 7.2 Keep Still Please! Motion Artefacts

### 7.2.1 Gross Patient Motion

Probably the commonest cause of artefacts on images is patient motion, resulting in a range of ghosting effects depending on the severity of the motion. Continuous movement during the scan causes a generalized blurring, often making the scan useless (Figure 7.1), while a few twitches or only small movements may only cause a few subtle ghosts which may leave an acceptable image. Patients may move involuntarily if they are suffering from a movement disorder, or they may have difficulty understanding or remembering the instructions to keep still during the scan. In these cases it may be necessary to sedate the patient or even use a general anaesthetic in order to get a diagnostic scan.

More often patients become uncomfortable in the scanner and move to relieve pain or muscle cramps.

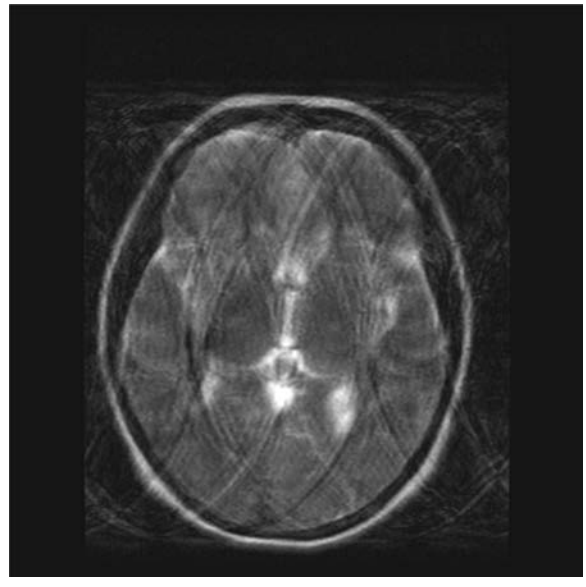

**Figure 7.1** Gross motion artefacts due to the patient moving continuously throughout the scan.

Careful preparation for the scan should minimize the patient's discomfort. This should include a clear description of what they will hear and feel during the imaging, as well as using pads and immobilization straps to help them keep still. Immobilization often works particularly well with young babies who seem to like being well swaddled! Hearing protection is mandatory in many centres, especially for 3 T systems, but you need to make sure patients can still clearly hear you over the intercom – both for instructions and for reassurance during the exam.

If a scan is unacceptably degraded due to motion artefact, the only solution is to repeat the scan. Obviously it makes sense to check that the patient is as comfortable as possible and understands the need to keep still before starting the repeat scan. If the scan is a particularly long one it is also worthwhile trying to reduce the scan time to improve the chances of an image without movement artefact, so long as the reduced SNR or resolution is still acceptable for the radiologist. Even better, choose one of the special motion-control sequences available on all scanners (e.g. BLADE, MultiVane, PROPELLER, JET).

## 7.2.2 Respiratory Motion

Breathing motion causes ghosting on thoracic and abdominal imaging. Sometimes it is easy to recognize the strong ghosts of the chest wall (Figure 7.2).

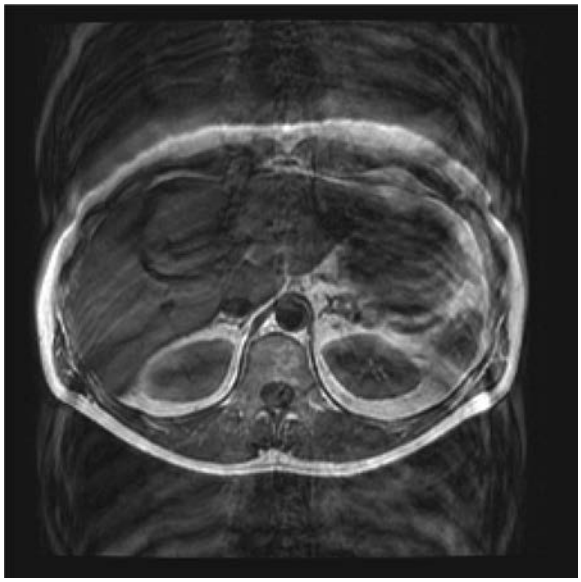

**Figure 7.2** Motion artefacts due to respiration.

The best way of avoiding respiratory artefacts is to reduce the scan time to less than 15 s, so that the scan can be acquired during a single breath-hold. The acquisition can also be split up so that packages of slices are acquired in separate breath-holds. The position of the liver and kidneys is more reproducible if the breath is held in expiration, so if multiple breath-holds are necessary, expiration is preferable. However, inspiration breath-holds can be maintained for longer, up to 25 s, and if you can achieve the whole scan in one breath-hold, it may be better to do it in inspiration. Careful coaching with the patient is needed before the examination starts so that they understand your instructions.

By using a special respiratory monitoring device, often known as a 'bellows' or 'respiratory belt', the MR system can detect the breathing motion. The device has a fixed-volume bellows arrangement strapped around the patient's chest and placed between the chest and the receive coil. The breathing motion causes a change in volume and hence air pressure within the device, which is detected and converted to an electrical signal that tracks respirations (Figure 7.3). This signal can be used either for *respiratory gating* or *triggering*, or respiratory compensation by phase re-ordering

Respiratory gating uses the bellows waveform to start the imaging sequence at a consistent place in each breathing cycle. As a result, each signal is acquired when the chest wall is in the same position, so there are no ghost images. Instead of a regular TR, the sequence now uses the interval between two consecutive triggers, which at a normal breathing rate of about 10–15 breaths per minute gives an effective TR of at least 4000 ms. This means that respiratory gating is only useful for PD or $T_2$-weighted imaging, and scan times may be very long.

Respiratory compensation, also called **R**espiratory-**O**rdered **P**hase **E**ncoding (ROPE), uses the waveform from the bellows to reduce artefacts with breathing. As the name suggests, the phase-encode gradients are re-ordered to match the breathing motion, taking care to maintain the same contrast. To understand this, you need to understand k-space, so this might be a good time to check Box 'An Easy Introduction to k-Space' in Chapter 5. Remember that all the signal and contrast information is in the middle of k-space, which corresponds to the smallest phase-encoding gradients, while the large phase-encoding gradients fill the outer portions

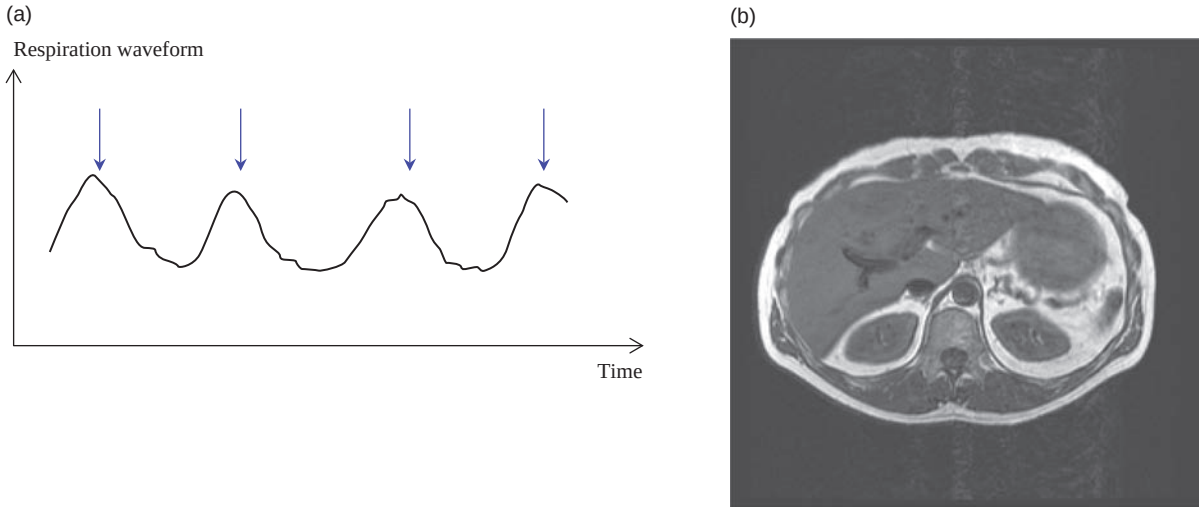


**Figure 7.3** (a) The changing signal produced by respiratory bellows around a patient's chest. Respiratory gating triggers the scans so that all the acquisitions are synchronized to the same point in the cycle (small arrows). (b) Axial image of the abdomen acquired with respiratory-ordered phase encoding (ROPE).

of k-space and control the resolution of the image. In ROPE the order of the gradients is matched to the respiratory cycle (Figure 7.4a) so that neighbouring lines in k-space are acquired close together in the breathing cycle (Figure 7.4b). ROPE is an efficient way of removing the breathing artefacts and it has the advantage that it can be used for any type of image weighting, although it does tend to extend the scan time slightly since it needs to start by measuring and analysing the respiratory waveform for a couple of cycles. However, it is not available with all image sequences.

The last method to avoid respiratory motion is a technique called 'navigator echoes' (see Chapter 16 for full details). This method does not use the bellows; instead, a rapid 1D imaging method is used to monitor the position of the diaphragm. The navigator 'signal' is from a single column of voxels, which you might think is pretty useless. However, the strong contrast between the (dark) lung and the (brighter) liver shows up clearly. When a series of navigators is acquired while the patient breathes normally, the changing position of the diaphragm can be visualized by displaying the 1D images side-by-side (Figure 7.5). The boundary can be automatically detected by the scanner software, and controls the acquisition so that the image data are acquired when the diaphragm is within certain spatial limits, typically 2–4 mm.

So in total there are four different ways of avoiding respiratory artefacts. How do you choose the right one for your application? We can start by saying that respiratory triggering, where the TR is effectively the length of the breathing cycle, has such a long scan time that it is rarely used. For cardiac imaging (see Chapter 16), the methods of choice are breath-hold for morphology and perfusion scans, and navigators for coronary arteries and high-resolution viability. In abdomen imaging, a breath-hold is usually the best choice, but ROPE can also be used with classic SE or GE sequences. Unfortunately ROPE cannot be used with fast spin-echo (TSE) sequences; the only choice is navigator or respiratory gating, which therefore means relatively long scan times for these sequences.

### 7.2.3 Cardiac Motion

The beating heart is a source of artefacts not only when imaging the heart itself, but also in exams like thoracic spine or liver (Figure 7.6). Avoiding the artefacts is achieved by synchronizing the sequence to the cardiac cycle, known as 'gating'. Special electrocardiogram (ECG) electrodes are attached to the patient's chest, in a similar way to normal cardiac monitoring electrodes. These electrodes are usually non-metallic to reduce artefacts on the images, and to avoid local skin heating. The ECG leads typically have high impedance (compared with low impedance in normal ECG monitoring); this reduces the chance of RF burns for the patient. You should never use ordinary ECG electrodes in an MR system.

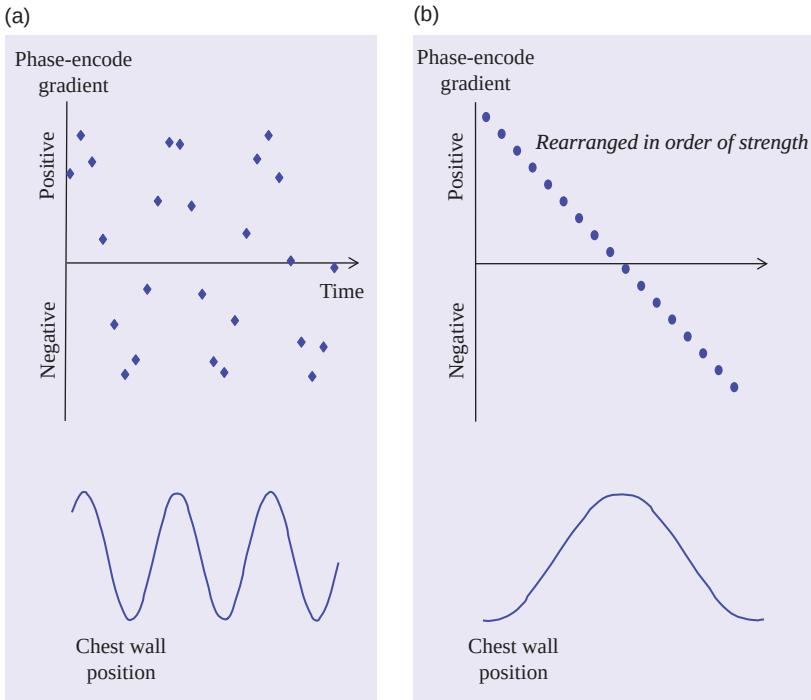


**Figure 7.4** (a) Low-sort ROPE rearranges the phase-encoding order to match the respiratory cycle. (b) When the data are arranged in order of phase-encode gradient strength, neighbouring lines in k-space are close together on the respiratory waveform and it appears that the whole acquisition has taken place over one breathing cycle.

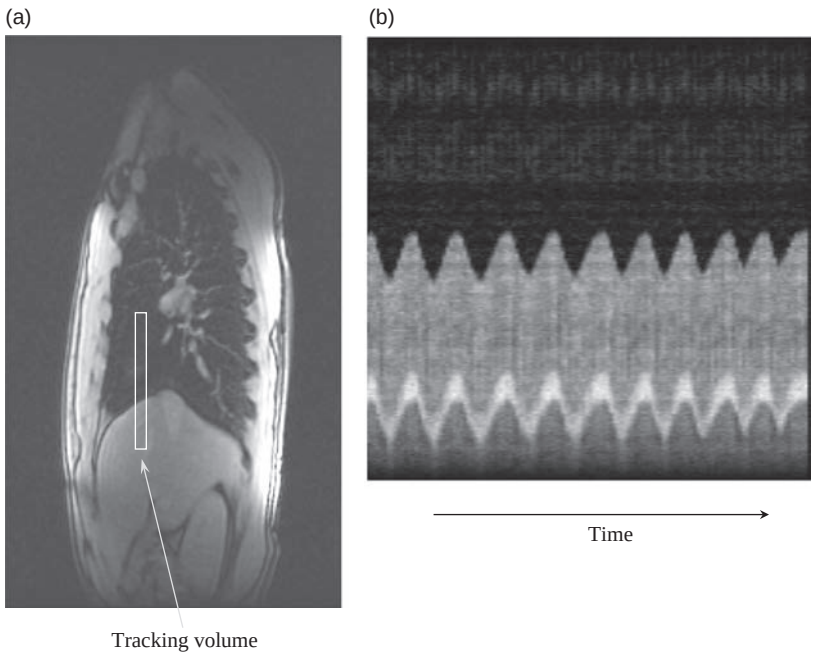


**Figure 7.5** The navigator signal is frequency encoded in one direction, and the other direction becomes time.

The peak of the R wave is detected by the scanner and used to trigger the next phase-encoding acquisition (Figure 7.7a). In this way each line of data is acquired at the same point in the cardiac cycle, and the ghosting is removed. The TR (and therefore the scan time) is determined by the heart rate, so at a typical HR of 75 beats per minute (bpm), TR will be 800 ms. This is an intermediate TR for spin echo,

neither short enough for $T_1$ weighting nor long enough for PD or $T_2$w. It is possible to define the TR as 2 or more R–R intervals, which can extend the usable TR time. However, it is still necessary to avoid acquiring data during the QRS peak, so it slightly reduces the time-efficiency of the sequence.

Cardiac gating can also be achieved by detecting the arterial pulse of blood in the patient's finger or toe. In these areas the blood vessels are very close to the skin, and an infrared light detector can pick up the increased volume of blood as the arterial pulse reaches the extremity. The signal only shows the arterial peak, not the other portions of the cardiac cycle, but this is sufficient to provide a trigger for the MR sequence. This technique is known as peripheral pulse gating or photoplethysmographic gating (rather a technical mouthful!), or simply **P**eripheral **G**ating (PG). Peripheral gating is a useful way of removing pulsatility artefacts in the brain and cervical spine.

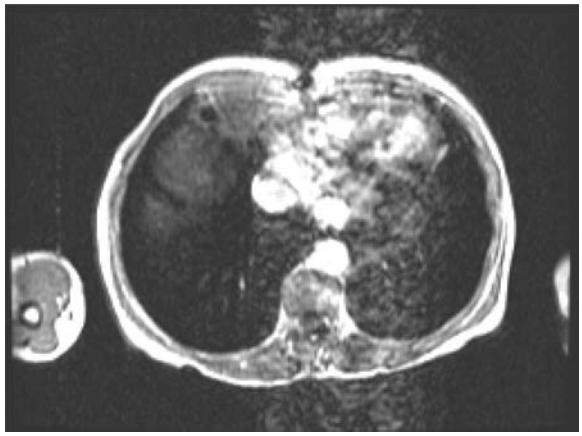

**Figure 7.6** Motion artefacts due to cardiac motion.

### ECG or Peripheral Gating?

Both methods are available on most modern scanners. What are the advantages and disadvantages of the two techniques?

ECG gating is a more accurate gating method, since the R wave is usually sharp and should be easily detected by the scanner. But that does depend on a good connection of the electrodes to the patient. This is particularly important for cardiac imaging since it allows as many multi-slice images as possible to be scanned within the R–R interval. In a multi-slice gated sequence, not only is each slice at a different location, it is also at a different point in the cardiac cycle. During systole the heart moves within the chest and so slices acquired during systole may be spatially mismatched with slices in the rest of the cycle.

In contrast, peripheral gating only detects the arterial pulse and the peak is much broader than the ECG R wave (Figure 7.7c). Thus peripheral gating is not ideal for cardiac imaging because it results in variability of the trigger position. In addition, the trigger is delayed relative to systole due to the time it takes for the arterial blood to arrive in the finger or toe, typically about 500 ms. Bear in mind that the arterial delay for

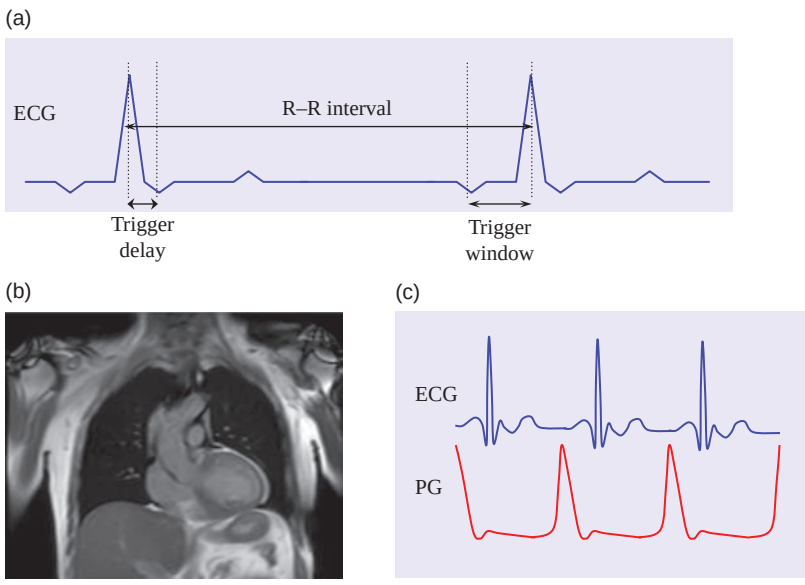


**Figure 7.7** (a) The ECG waveform used to trigger the scan acquisitions to remove cardiac motion artefact, showing the effective TR (the R–R interval), the trigger delay and the trigger window. (b) Coronal view of the chest showing no artefact from cardiac movement. (c) Peripheral gating (PG) signal compared with ECG signal.

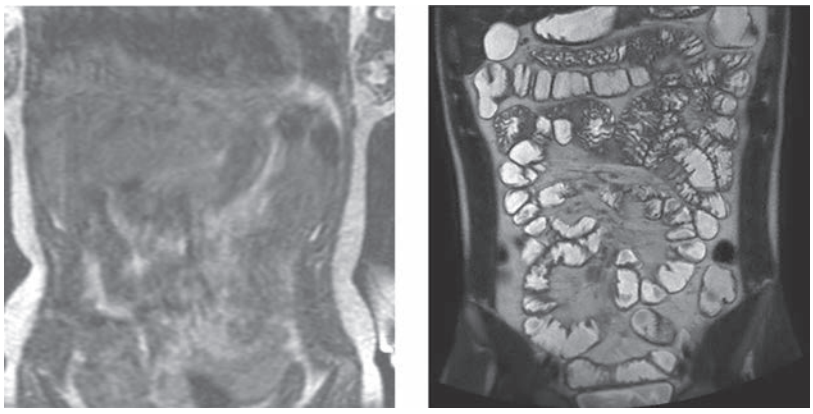


**Figure 7.8** (a) Motion artefacts due to peristalsis. (b) Coronal enterography image acquired using single-shot TSE to 'freeze' the peristalsis motion.

the anatomy you are imaging may be different from that in the finger too, so there may be some residual ghosts. However, for the most common application, that of removing CSF pulsation artefact in neurological imaging, the delay times are actually quite similar, so peripheral gating is usually successful.

The advantages of peripheral gating are in the ease of preparing the patient and for safety. For ECG gating it is necessary for the patient to change into a hospital gown, and for electrodes to be carefully attached to the skin, including hair removal if necessary. In comparison, placing the peripheral trigger detector on one of the fingers is very easy and more comfortable for the patient.

Always follow the manufacturer's instructions when using ECG gating or peripheral gating.

**Clinical Exam: MR Enterography**

Imaging the bowel wall is becoming an important MR exam in some regions, especially for diagnosis of Crohn's disease or irritable bowel syndrome. MR is considered equivalent to CT enterography for sensitivity and specificity, so the choice of technique is largely a matter for local (institutional) preference.

Patient preparation includes drinking a large volume of an enteral contrast agent, such as an aqueous solution of mannitol, or methyl-cellulose. This agent distends the bowel and provides a high-signal lumen, allowing for clear delineation of the bowel walls. Scanning is done with single-shot TSE sequences, or strong $T_2^*$w GE scans. It is also possible to acquire dynamic images with GE scans, to show peristalsis in certain segments of the bowel.

## 7.2.4 Peristaltic Motion

Peristalsis causes a random continuous motion of the abdominal contents, and there is no physiological signal to trigger the MR acquisition. Acquiring multiple averages can reduce the ghost appearances, but for imaging the small or large bowel it is much more effective to use an antiperistalsis drug such as hyoscine butylbromide or glucagon. This has the effect of stopping the peristaltic motion for a short time (usually around 15–20 min) which is just long enough to acquire the required images. If a drug is not possible, then ultrafast pulse sequences such as HASTE (see Section 12.4.3) or **S**ingle-**S**hot **T**urbo **S**pin **E**cho (SS-TSE) can be used to acquire images fast enough to minimize the motion artefacts (Figure 7.8).

## 7.2.5 Motion Artefacts from Flowing Blood

Moving protons in blood vessels or the cerebrospinal fluid (CSF) cause a range of artefacts due to two effects. First, there is an in-flow effect which produces high signal within blood vessels on gradient-echo images, as new protons flow into the imaging slice during the TR. Second, there are velocity-induced phase effects, which reduce the blood signals and create ghost images of arteries or veins in the phase-encode direction (Figure 7.9a). Complete intra-voxel dephasing occurs in areas of turbulent flow, e.g. distal to a bifurcation, leaving a dark appearance. Depending on whether the scan is an MRA or not, these effects may cause problems.

To explain the in-flow effect, we will consider a blood vessel passing through an imaging slice (Figure 7.9b), and assume that the blood flow is

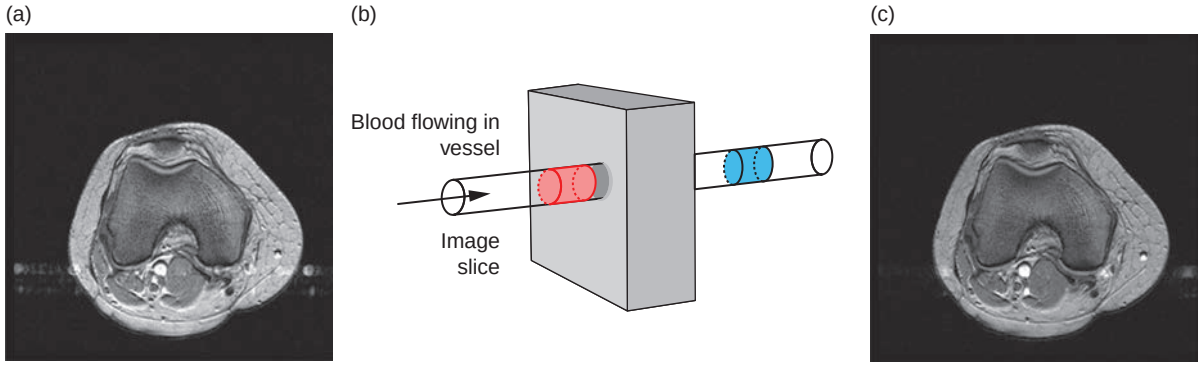


**Figure 7.9** (a) Knee image showing artefact from flowing blood. (b) Blood vessel passing through an imaging slice. In a spin-echo sequence, the blood excited by the 90° pulse (coloured blue) moves on by the time the 180° pulse is applied (red-coloured bolus), and neither bolus produces a signal. In gradient-echo sequences, each bolus has fully relaxed magnetization $M_0$ when the excitation pulse is applied, and its signal is refocused by the gradients so that blood appears bright. (c) Axial knee image acquired with gradient moment nulling (flow compensation).

steady. In SE sequences, we use two RF pulses to create the echo, excitation followed by refocusing. Assuming the blood is moving fast enough, the bolus of blood within the imaging slice will be replaced by a second bolus of fresh, i.e. unsaturated, blood during the gap between the two pulses. The first bolus receives a 90° pulse but not the 180°, while the second receives only the 180° pulse. Since both a 90° and a 180° pulse are required to create a spin echo, neither bolus will produce an echo. There will be a signal loss within the blood vessel, giving spin-echo images a characteristic dark-blood appearance.

GE sequences only have an excitation pulse and the echo is formed using the gradients. So the excited bolus of blood always contributes a signal, provided it is still within the gradient volume. For the next and every subsequent slice there is a fresh bolus of blood within the slice, with fully relaxed magnetization. Thus, on GE images blood vessels have high intensity and can easily produce ghost images in the phase-encode (PE) direction, if the TR is not synchronized with the pulse rate. We can also take advantage of that very bright signal to create MR angiograms with the scan technique called 'time-of-flight' (TOF) – see Chapter 15 for all the details.

Apart from the in-flow or time-of-flight effects, phase-related artefacts arise because the blood protons are moving during the imaging gradients and so their resonant frequencies are continuously changing. When their frequencies change, they acquire phase differences compared with each other and with static tissue. When a voxel contains protons moving at different velocities, for example in a region of turbulent flow, their signals will dephase rapidly and cause a signal dropout. More detail about flow-related phase effects can be found in Chapter 15.

To avoid flow artefacts, we frequently use *spatial saturation* slabs just outside the field of view or in the slice direction. Saturation slabs, also known as **RE**gional **S**aturation **T**echnique (REST) slabs or 'pre-sat', act exactly like slice selection. We apply a 90° pulse to all the tissues within the slab immediately before the RF excitation pulse for the imaging sequence, then apply a large gradient pulse to dephase the protons, leaving no signals from the tissues in the sat band.

Saturation bands can be used in many ways. They can be placed within the FOV, e.g. to saturate signal from the thoracic aorta for sagittal spines or reduce artefact from swallowing (Figure 7.10), or at the edges of the FOV, e.g. to reduce phase wrap on coronal shoulders. Placed above and/or below the image FOV, sat bands can remove arterial and/or venous blood flow; blood flowing from the sat band into the imaging slice will have no time to recover its equilibrium magnetization, so it will give no signal. Be sure to pay attention to the flow direction to make sure you put the sat band in the right place!

It is also possible to correct for velocity-induced flow dephasing effects using a technique called *gradient moment nulling*, also known as *gradient moment rephasing* or *flow compensation*. Extra gradient pulses are inserted into the pulse sequence, so that the velocity-induced phase shifts are corrected. For vessels with simple laminar flow, the vessel will appear on the final image without ghosting artefacts

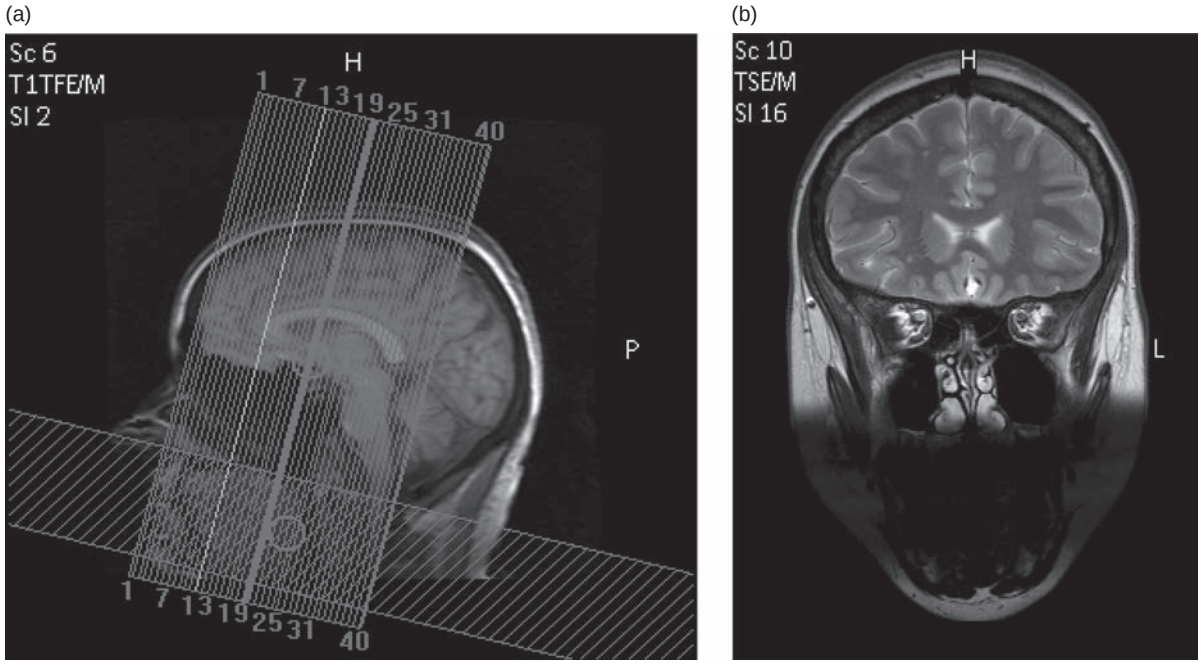


**Figure 7.10** (a) Angled sat band within the FOV to eliminate artefacts from tongue movement and swallowing. (b) Resulting image.

(Figure 7.9c). The details of gradient moment nulling will be fully explained in Chapter 15.

## 7.3 Lose the Fat!

Fat is often a source of problems in MR imaging. It tends to have high signal intensity in all contrast weightings, potentially masking pathology. It also causes two types of artefact, known as chemical shift artefacts.

In Section 3.1 we sub-divided tissues into fat-based and water-based tissues. In all MR images we are detecting signals from protons (the nuclei of hydrogen atoms), but fat and water have very different structures. Water has only two hydrogen atoms and an oxygen atom, so it is a small molecule. Fat is made up of triglyceride chains, long backbones of 10–20 carbon atoms each with two hydrogen atoms on either side. Fat molecules are thus very large and each hydrogen atom is surrounded by many other atoms. The neighbouring electron clouds reduce the effective strength of the external magnetic field $B_0$, so the hydrogen nuclei in fat have a lower Larmor frequency than those in water, which are not shielded. This difference is known as the chemical shift, which is quoted in parts per million (ppm), a unit which is independent of magnetic field strength. We can calculate the actual frequency difference by multiplying the chemical shift in ppm by the resonant frequency in megahertz of protons at a particular magnetic field strength. For example, the chemical shift between fat and water is 3.5 ppm, and at 1.5 T protons have a Larmor frequency of 63.855 MHz, so the frequency difference is approximately 220 Hz; at 3 T it is approximately 440 Hz. If we look at the frequency spectrum from the human body we see two peaks, the larger one from water protons and the smaller one to the right from fat (Figure 7.11a).

### 7.3.1 Chemical Shift Artefact

As we will see in Chapter 8, we use frequency encoding, i.e. we rely on the MR signal's frequency for spatial information. But fat naturally has a lower frequency than water, so the frequency encoding will be fooled into thinking that the fat is in a different position. Due to the frequency difference, the apparent position of fat signals is shifted by a number of pixels, but only in the frequency-encode direction. This appears as light and dark bands on opposite sides of a structure, or as an entire ghost image of the fat distribution in the anatomy (see Figure 7.11b). This is called the *chemical shift artefact* or *chemical shift misregistration artefact*. The severity of the pixel shift depends mainly on the receiver bandwidth used: the

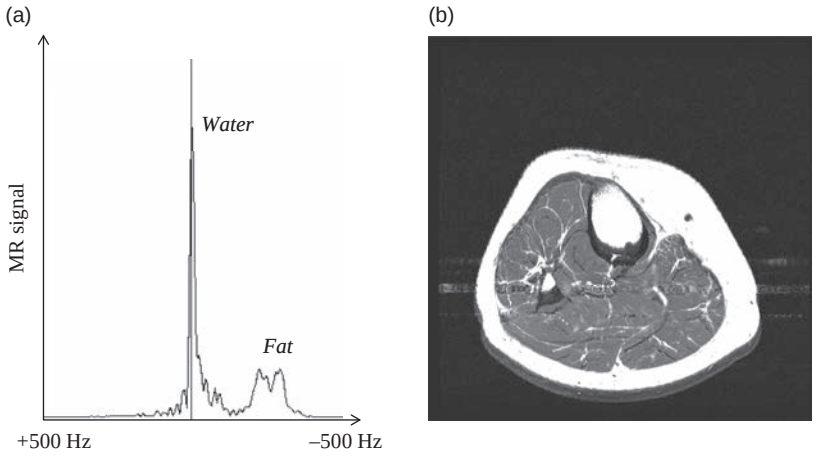


**Figure 7.11** (a) Spectrum from the lower leg showing water and fat peaks with a separation of 3.5 ppm. (b) Chemical shift artefact in the lower leg.

lower the bandwidth the worse the problem. So to minimize the chemical shift artefact you should ideally use a higher bandwidth. However, increasing the bandwidth also reduces the signal-to-noise ratio in the image, so it is not always desirable. The chemical shift artefact occurs with both spin-echo and gradient-echo sequences.

**How Many Pixels is it Shifted?**

You need to know how much the fat signal is shifted with respect to the water in order to know if the chemical shift artefact is a problem. The severity of the chemical shift effect depends on two things: the field strength of the magnet and the receiver bandwidth used for imaging. Some manufacturers (e.g. Philips) quote the receiver bandwidth in terms of the number of pixels by which fat will be shifted; this is the easiest way for operators to minimize the artefact! Others (e.g. Siemens) use 'hertz per pixel' for the receive bandwidth; again this makes life easy as you just have to divide the chemical shift for your magnet field strength by the bandwidth. For example, at 1.5 T the chemical shift is 220 Hz, so if you choose a bandwidth of 100 Hz/pixel the fat signals will be shifted by about two pixels relative to water, whereas a bandwidth of 500 Hz/pixel will give a negligible shift of 220 ÷ 500 ≈ 0.5 pixels.

Lastly, some manufacturers quote the receiver bandwidth directly in kilohertz (e.g. GE Healthcare). Working out the chemical shift is a little more long-winded, so you might like to work it out for a range of bandwidths and either memorize them or have them in a handy notebook when you are at the console. It's important to know that the bandwidth is quoted as '±' the value, which means you need to double it in your calculation, as shown below.

First, work out the bandwidth in hertz per pixel; multiply by 1000 to convert from kilohertz into hertz, then divide by the frequency matrix. At the moment 256 is the commonest frequency matrix (although 512 is becoming much more popular), so in this example we will use 256 and a bandwidth of ±10 kHz:

$$\frac{10 \times 2 \times 1000}{256} = 78.2 \text{ Hz/pixel}$$

Now divide the chemical shift for your scanner's field strength by this number. We will use the example of 1.5 T, so

$$\frac{220}{78.2} = 2.8 \text{ pixels}$$

So at this field strength a ±10 kHz bandwidth with a 256 frequency matrix gives a moderate chemical shift artefact. If the matrix is increased to 512, the shift will be more than five pixels and will be more of a problem.

## 7.3.2 Phase Cancellation Artefact

There is a second type of artefact, also caused by the chemical shift between fat and water, which only occurs with gradient-echo imaging. Some texts call this 'chemical shift of the second kind', 'black line' artefact, 'India ink' or the '*phase cancellation artefact*', which is the term we will use. The phase cancellation artefact appears as a black outline (Figure 7.12b), especially noticeable in the abdomen where water-based tissues are surrounded by peritoneal fat. It occurs in any voxel containing both fat and water, and depends on the fat–water chemical shift and the TE used.

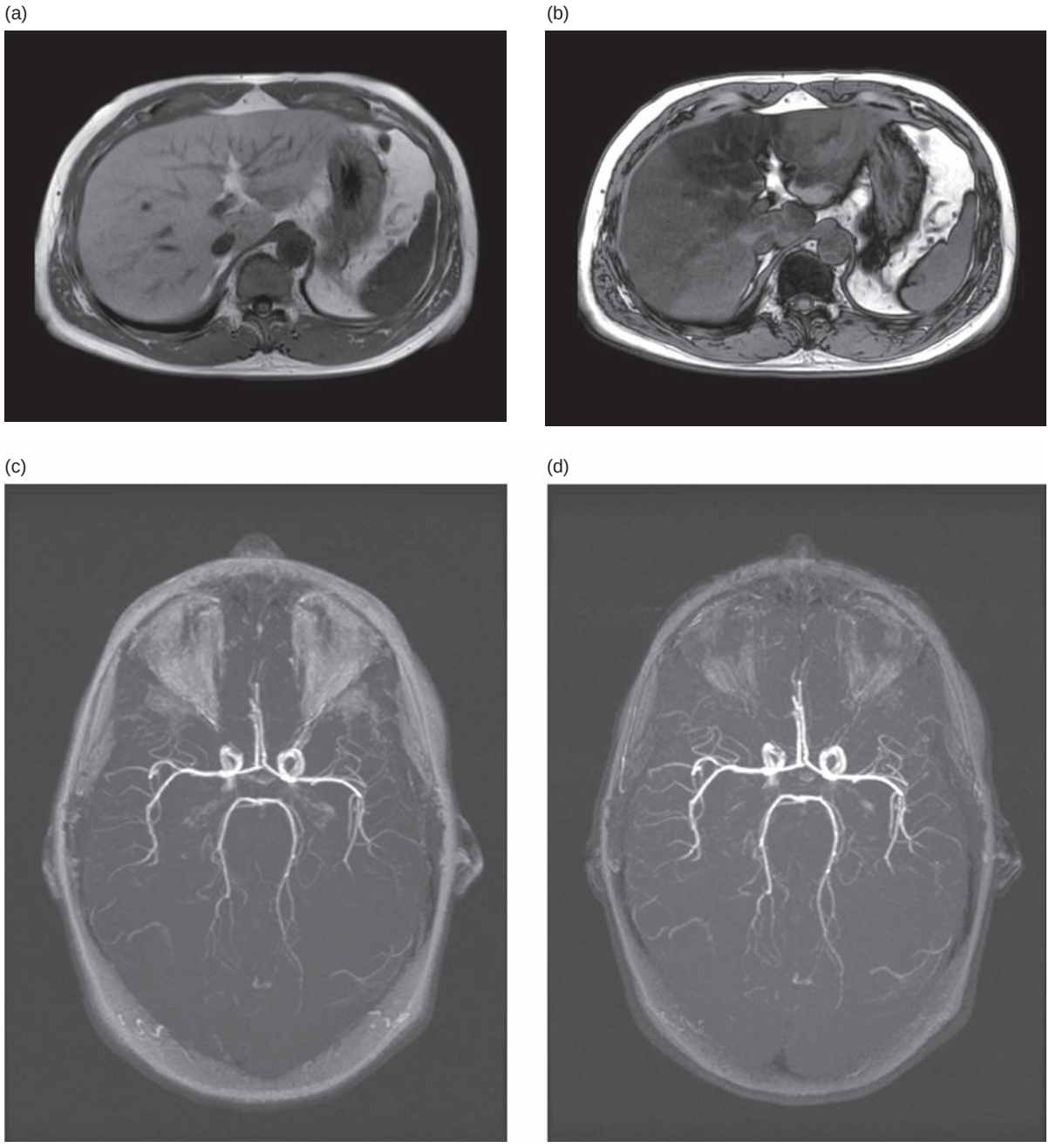


**Figure 7.12** Phase cancellation artefact. (a) At a TE of 4.2 ms the fat and water signals are in phase and no phase cancellation artefact is seen. (b) At a TE of 2.1 ms, a black line appears at boundaries between fat and water, because the fat and water signals within the voxel are out of phase with each other. (c) Cranial time-of-flight (TOF) MR angiography with TE for fat and water in phase, and (d) out of phase. Notice how much the periorbital fat intensity is reduced.

Why does it occur? Immediately after the excitation pulse of the sequence, fat and water signals are in phase with each other, but due to the small difference in their Larmor frequencies they begin to dephase. If the echo is acquired when fat and water are exactly out of phase, i.e. 180° to each other, voxels with a mixture of tissues will have a reduced signal since the fat signal subtracts from the water. This gives the characteristic dark outline at fat–water interfaces, due to all the mixed voxels around the edge. If

the TE is increased so that fat and water are back in phase with each other, the signals will add together instead of subtracting, and no black line occurs (Figure 7.12a).

So there are specific TEs in gradient-echo imaging which produce the phase cancellation artefact, and others which don't. Some manufacturers list 'in-phase' (IP) and 'out-of-phase' (OP) as options for the TE, which makes life easy! Acquiring both IP and OP images is common in liver or spleen imaging, since a signal loss in the OP image can indicate diffuse fatty infiltration in the water-containing organs. Phase cancellation artefact is not a problem in spin-echo imaging because the fat–water phase shift between the 90° and 180° pulses is inverted by the 180° pulse, so that at the echo time they are back in phase.

**In-Phase and Out-of-Phase TEs**

If your scanner doesn't show 'in-phase' and 'out-of-phase' as TE options, you need to work out some appropriate values. As with the chemical shift artefact, you might find it useful to keep this information in a handy notebook. We start with the chemical shift in hertz between fat and water, for example at 1.5 T the shift is 220 Hz. Fat and water are in phase immediately after the excitation pulse, but we can't acquire the signal immediately. The next time they are in phase will be 1/220 s later, i.e. 4.55 ms. So a TE of 4.55 ms or multiples thereof will have fat and water in phase, avoiding the black line artefact. Halfway between these two echo times fat will be exactly 180° out of phase with water, so a TE of 6.9 ms (or 2.3 ms if the gradients allow it) will give the phase cancellation artefact. To get a $T_2^*$-weighted gradient echo with in-phase TE, you may need to go to 22.7 or 27.3 ms, while an out-of-phase $T_2^*$-weighted TE would be halfway between these values at about 25 ms. At 3 T the values will be different, with IP TEs at multiples of 1/440 = 2.27 ms and OP TEs halfway between these values.

## 7.3.3 MRI Liposuction: Removing Fat Signals

There are two easy ways of suppressing fat signals more or less completely. The first we have already met in Section 3.7: the STIR sequence which uses the inversion recovery pulse sequence with the TI set at the null point of fat. This technique depends on the $T_1$ of fat, which is considerably shorter than that of most other tissues. The TI varies slightly with field strength, from approximately 150 ms at 1.5 T to 220 ms at 3 T. The initial ('inversion') 180° pulse inverts all the equilibrium magnetization, which then begins to recover towards the equilibrium value, $M_0$, with $T_1$ recovery (see Figure 3.11). When the 90° pulse is applied at the null point of fat, fat-based tissues produce zero signal as they have nothing to tip into the xy plane. At an appropriate TE a refocusing 180° pulse is applied to generate a spin echo and create an image with no signal from fat. Since we use 'magnitude' reconstruction in MRI, all the other tissues give bright signals, with fluids (with the longest $T_1$s) having the highest signal.

The alternative technique is frequency-selective *fat saturation*, often known simply as 'fat sat' or 'chem sat' (chemical sat). This takes advantage of the chemical shift between fat and water to excite only the fat protons. A narrow range of RF frequencies centred on the fat Larmor frequency is used (see Figure 7.13a) to give a 90° pulse to protons in fat, leaving the water protons unexcited. This is known as a CHESS (**CHE**mical **S**hift **S**elective) pulse. The imaging sequence is started immediately after the CHESS pulse, so that fat has no time to recover its longitudinal magnetization and the image is produced with a suppressed fat signal. Typically *crusher* gradients are applied immediately after the CHESS pulse to dephase the transverse fat magnetization, which otherwise tends to produce an echo due to its rapid relaxation. There has to be some compromise over the fat sat pulse; it needs a bandwidth wide enough to saturate all the fat protons, but shouldn't excite any water protons. Even though the chemical shift of 220 Hz (at 1.5 T) sounds quite large, the variation of main field homogeneity over the field of view can

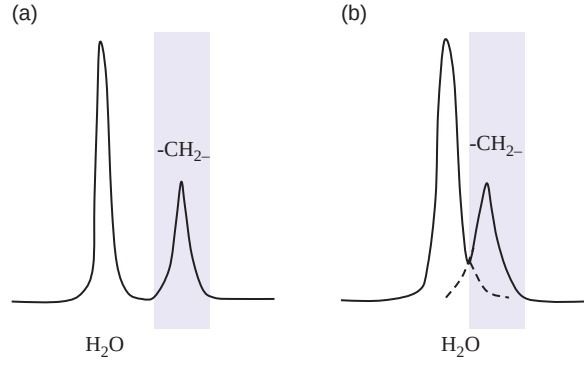


**Figure 7.13** (a) Frequency-selective saturation uses a narrow-bandwidth RF pulse to excite only fat protons. (b) At low field strengths the peaks overlap, making it impossible to saturate all the fat without affecting the water protons.

easily reach this order of magnitude. In areas where the local $B_0$ is reduced by 220 Hz, the CHESS pulse will perfectly saturate the water signal instead of the fat signal. This annoying problem is called fat sat failure and can be found especially with off-centre or very large fields of view.

A slight modification of fat sat combines frequency-selective excitation with STIR. This is known as SPIR (**SP**ectral **I**nversion **R**ecovery) by Philips or SPECIAL (**SPEC**tral **I**nversion **A**t **L**ipid) by General Electric. A frequency-selective pulse is applied to the fat protons, followed by crusher gradients to dephase any signal produced in the transverse plane. At an appropriate TI (depending on field strength) the rest of the imaging sequence is started, producing an image with better fat suppression than the simple CHESS pulse.

Another variation is to use so-called 'adiabatic' RF pulses for fat suppression. These pulses are commercially known as SPAIR (**SP**ectral **A**diabatic **I**nversion **R**ecovery) and are recommended for 3 T use, because they give more uniform fat suppression, especially over larger FOVs. They are tricky to explain, so we leave that for a later chapter (when you are ready, turn to Chapter 12).

**Which is Better, STIR or Fat Sat?**

If your scanner has a field strength of 1.5 T or higher, fat sat or SPIR/SPAIR is almost always better than STIR. At these fields it is possible to apply a good suppression pulse to just the fat, leaving the water protons unexcited, thanks to the higher chemical shift. Frequency-selective fat saturation pulses can be inserted before almost any pulse sequence, so the image contrast can be controlled independently of the fat suppression. In comparison, STIR can only produce a fat-suppressed '$T_2$-weighted' appearance, not fat-suppressed PD.

However, fat sat can become unreliable at the edges of the imaging volume, causing both unsuppressed fat and suppressed water signal. This happens because of the large range of magnetic field non-uniformities which cannot be shimmed as easily as a smaller FOV in the centre of the magnet. Failing fat-sat is particularly a problem for shoulder, elbow and wrist imaging, where the anatomy cannot be brought to the isocentre. Adiabatic fat suppression (e.g. SPAIR) can help to avoid $B_1$ non-uniformities, but cannot help with main field non-uniformities. Under these circumstances STIR is still useful, because it is relatively independent of both $B_1$ and $B_0$ non-uniformities. Dixon-based methods are also useful, since the latest reconstructions offer very robust fat-sat.

**Try It for Yourself 6: Chemical Shift Effects**

The chemical shift artefacts and methods to avoid them are easily shown using a cooking oil and water phantom. Fill a deep plastic container one-third full with water, adding a drop of gadolinium to reduce the $T_1$, then carefully pour on some cooking oil until the container is two-thirds full. The oil will float in a separate layer on top of the water, but you have to handle it carefully to avoid making a salad dressing at the interface! Put the container into the head coil or knee coil, do a localizer scan and start changing parameters one at a time. For instance, use several different receive bandwidths (water–fat shifts on Philips systems) to see the chemical shift artefact, or try gradient-echo scans with echo times for fat and water in and out of phase.

A couple of things to look out for:

- Use a $T_1$-weighted spin-echo sequence to see the chemical shift effect, and make sure the frequency-encode direction is across the fat–water boundary, not parallel to it.
- To compare STIR and fat sat, use the fat sat with a $T_2$-weighted spin-echo scan to get similar contrasts in the final images. Try both techniques at large and small FOVs (you might want to devise a larger phantom for the big FOVs).

## 7.4 Digital Imaging Artefacts

### 7.4.1 Partial Volume Artefact

Partial volume artefacts occur wherever a voxel contains a mixture of tissue types. Considering that a typical voxel is 1 mm × 1 mm × 4 mm, it is easy to see that in a structure as complex as the human body, most voxels in any given slice will have a mixture of tissues. We can consider this as a digital imaging artefact, since we are representing a lot of information in a relatively small number of voxels (yes, 512 × 384 = 196 608 does sound like a lot, but it's not enough!). Look back at Chapter 6 to learn how to optimize the resolution in your scan.

You cannot completely avoid partial volume effects, but you can minimize them by setting the appropriate voxel size for the anatomy and by choosing the correct oblique slice angle. For example, we would use 0.4 mm in-plane resolution for the **I**nternal **A**uditory **M**eatus (IAM), because we know that the VIIIth cranial nerve is only about 2.5 mm in diameter and lesions may be only 1 mm in diameter. We also angulate 'sagittal' slices so that they are perpendicular

to the nerve, and set the slice thickness to 3 mm or less. Conversely 7 or 10 mm slices are appropriate for the liver, because it is a much larger organ and clinically significant pathology is likely to have a diameter of at least 7 mm.

## 7.4.2 Cross-Talk

A similar problem with multi-slice imaging is cross-talk between adjacent slices, also known as cross-excitation or (erroneously) cross-relaxation. Cross-talk appears as a reduced intensity on all but the first slice of a multi-slice set, which is often only detectable when comparing the end slices with their neighbours. It happens because the slices are not straight-edged, like a sliced loaf of bread, but have a curved profile (see Figure 7.14a) due to imperfections in the selective excitation pulse. The slice width is defined by the **F**ull **W**idth at **H**alf **M**aximum (FWHM), so if the slice gap is too small the edges of the slice may overlap with its neighbours (Figure 7.14b). Tissue in the overlapping section is excited by both slices, and experiences a very short effective TR instead of the TR set by the user. It doesn't have time to relax between the pulses, so its signal intensity is reduced. A similar effect occurs with multi-angle oblique acquisitions, e.g. in the lumbar spine, where you may see horizontal black bands across neighbouring slices where the slices intersect.

The best cure for cross-talk is to use an interleaved slice order; excite the odd-numbered slices first, then go back and acquire the even-numbered slices. In this way, tissue in the imperfect slice edges will see a TR much closer to the required TR, instead of a very short TR. This is a default setting on many scanners, so you usually don't need to think about it. If you still see cross-talk effects, you should increase the slice gap, especially in IR sequences since the slice profiles are worse for 180° pulses. However, you should bear in mind that large gaps reduce resolution, as tissues in the gap are not imaged at all. If very thin gaps or contiguous slices are required, it may be better to do a 3D acquisition.

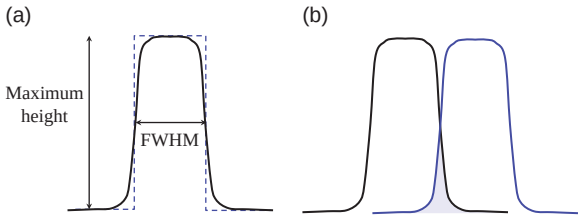


**Figure 7.14** (a) The slice excitation profile, ideally rectangular (dotted line), is in reality a curved shape (solid line) whose full width half maximum (FWHM) defines the slice width. (b) When the slice gap is too small the edges of neighbouring slices overlap.

**Why isn't the Slice Profile Straight-Edged?**

The reason for non-rectangular slice profiles is the nature of selective excitation. The frequency spectrum of the RF pulse, with the strength of the slice-select gradient, defines not only the slice width but also its profile. To get a perfectly rectangular excitation profile, where all protons within the slice receive exactly a 90° pulse and all protons outside the slice are unexcited, the amplitude of the excitation pulse must be a sinc ($\sin(x)/x$) function (Figure 7.15a and Appendix). However, a sinc function is infinitely long in the time domain, and we obviously have to truncate the pulse. In the simplest case, truncating the sinc corresponds to multiplying by a top-hat (see Figure 8.5 and Appendix) in the time domain, and the excitation profile becomes a rect function convolved with a sinc, with significant ripple at the edges of the slice (Figure 7.15b). A better approach is to apodize the sinc, i.e. multiply it by a smoothly varying function such as a Hanning or a Gaussian. The excitation profile then has much less ripple (Figure 7.15c), although its FWHM is slightly greater than the original width.

## 7.4.3 Phase Wrap-Around Artefact

The phase wrap-around artefact happens when the patient's anatomy continues outside the field of view (FOV) in the phase-encode direction. It causes the signal from the tissue outside the FOV to appear at the opposite side of the image in the phase-encode direction (see Figure 7.16a). The wrapped-in tissue can overlay the real anatomy being scanned potentially interfering with the diagnosis. Although it is most commonly seen in the phase-encode direction, it also occurs in the slice direction in 3D imaging (when the slice-select axis is also phase encoded) and causes the end slices of the volume to wrap into each other.

The best way to avoid phase wrap-around is to use *phase oversampling*, also called the 'no phase wrap' or 'foldover suppression' option. This technique increases the FOV in the phase-encode direction and also increases the number of phase-encode steps so that the pixel size remains the same. The simplest

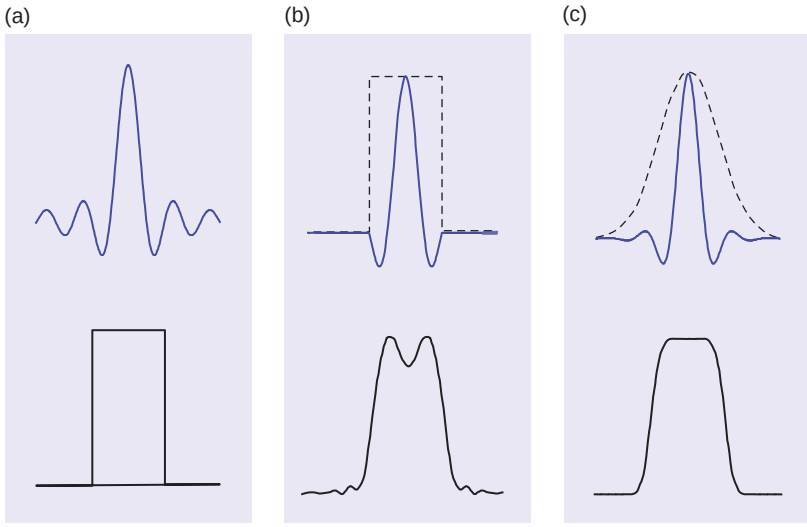


**Figure 7.15** (a) The Fourier transform of a rectangle or top-hat (rect) function is a sinc (sin(*x*)/*x*). (b) Simply truncating the sinc produces large ripples on the slice profile. (c) An apodized sinc RF pulse produces a cleaner excitation profile, although the FWHM is slightly wider.

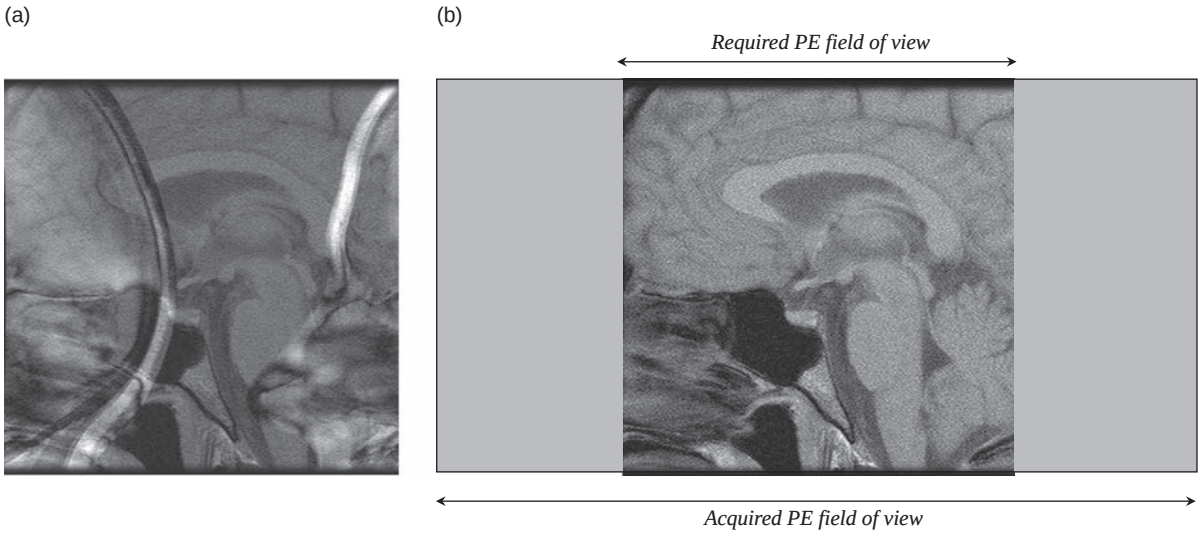


**Figure 7.16** (a) Tissue outside the field of view (FOV) in the phase-encode direction wraps into the image. (b) With phase oversampling the reconstructed image is larger than the required FOV, and the computer just throws away the unwanted regions.

implementation of phase oversampling is to simply double the size of the phase-encode matrix ($N_{PE}$) and also double the acquired FOV, but many scanners allow phase oversampling to be specified as a percentage of the FOV, which optimizes the technique. The anatomy just outside the desired FOV is now properly phase encoded, and the unwanted edges can simply be cut off by the computer, leaving a clean image (Figure 7.16b). Increasing the phase-encode matrix usually extends the scan time, although some scanners adapt the number of signal averages to compensate for this. Suppose we have an FOV of 20 cm with a PE matrix of 256 and two signal averages. When 'no phase wrap' is switched on, the FOV and PE matrix are doubled to 40 cm and 512 respectively. Using one signal average instead of two will maintain the original scan time, and the SNR stays the same. Refer also to Chapter 6 for optimization of scan time and resolution.

## 7.4.4 Gibbs' Artefact

Gibbs' artefact, also known as *truncation* or *ringing* artefact, is another consequence of undersampling most often seen in the phase-encode direction, and

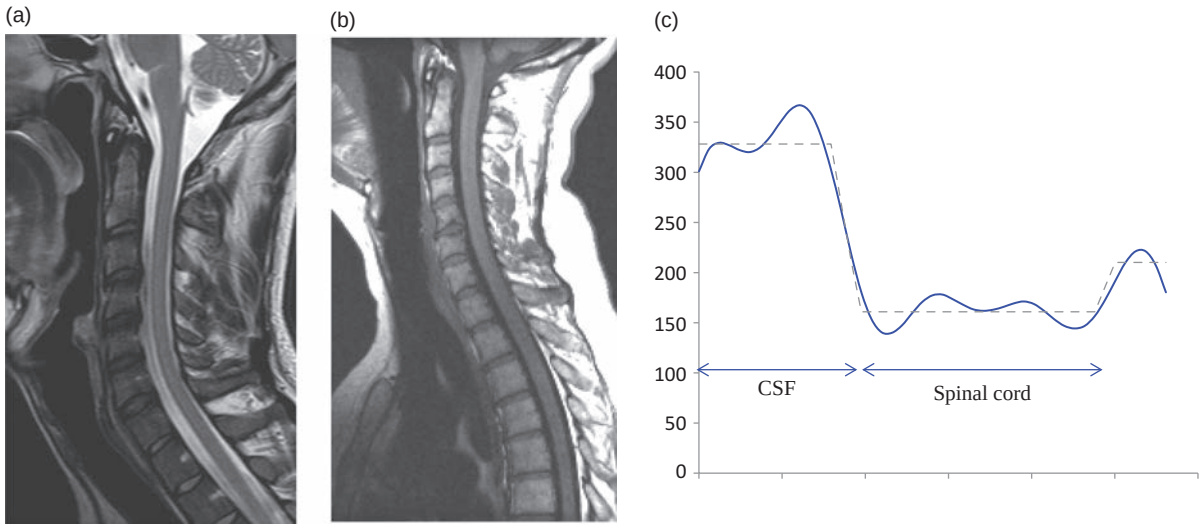


**Figure 7.17** (a) A low phase-encode matrix can cause Gibbs' artefact, alternating light and dark bands near a high-contrast interface. (b) Increasing the phase-encode matrix avoids the artefact. (c) A line profile across the structure reveals the typical pattern of Gibbs' ringing.

it can also be seen in the slice direction in 3D scans. It is visible at high-contrast boundaries, where the intensity changes from bright to dark, and appears as a series of alternating light and dark lines superimposed on the image. The intensity of the lines fades away from the boundary, and they follow the contours of the interface. A common example is on $T_1$w or $T_2$w images of the cervical spine (Figure 7.17a), where it can mimic the appearance of syringomyelia.

Gibbs' artefact is caused by having the acquisition matrix too small, i.e. the pixel size is too large to accurately represent the high-contrast boundary. As a rule of thumb the phase-encode matrix should never be less than half the frequency-encode matrix. If the artefact reduces the diagnostic quality of the image the only solution is to repeat the scan with a larger phase-encode matrix (Figure 7.17b).

**Try It for Yourself 7: Phase Wrap and Gibbs' Artefacts**

You can show the phase-wrap artefact with almost any phantom, although to see the Gibbs' artefact you need something with a sharp high-contrast boundary. Set up the phantom in the head coil or knee coil and scan a localizer. To show phase-wrap, set up a really small FOV in the middle of the phantom, or deliberately move it off-centre in the phase-encode direction. Gibbs' artefact is best seen with a very low matrix of 128–192 on a medium-sized FOV. When you increase the matrix size you probably won't be able to see it visually, but you can measure the pixel intensities across the boundary to see if it is still there: it shows up as 'ringing' on either side of the boundary (Figure 7.17 c).

## 7.4.5 Parallel Imaging Artefacts

Parallel imaging techniques, used to speed up acquisitions, can produce artefacts which are ghost-like or, in severe cases, phase-wrap-around (see also Chapter 14). These can arise when the reference image has uneven signal intensity, for example a bright fat signal, or if the patient has moved a lot between the reference image and the diagnostic scan. They are less likely when **A**uto-**C**alibration **S**cans (ACSs) are used as part of the parallel imaging technique.

It's possible to misinterpret these faint ghosts as 'motion' artefacts. In order to decide if a ghost is due to parallel imaging or motion, you can measure the distance between the ghost and the main image, along the PE direction. In the example shown in Figure 7.18, this distance is 31 mm. Divide the FOV by this distance, e.g. 180 ÷ 31 = 5.8. In this example, a parallel imaging factor of 6 was used, very close to the offset of the ghost within the FOV. So it is most likely that this is a parallel imaging artefact, not patient motion.

Another common artefact with parallel imaging is *noise break-through*, usually seen in the centre of the

image. It is usually found when the selected acceleration factor is too high to be supported by the receive coil. Signals which are the cross-over point of coil sensitivities have an inherent risk of being misplaced in the unfolded image, and this leads to increased noise appearance.

There are no special techniques to avoid parallel imaging artefacts. Most manufacturers impose limits on the maximum reduction factor which can be used with each receive coil, but you still can run into trouble. For example, when imaging the abdomen with a torso coil, a very large patient may have noise break-through artefact on axial images, whereas a thin patient with the same reduction factor would have no artefact.

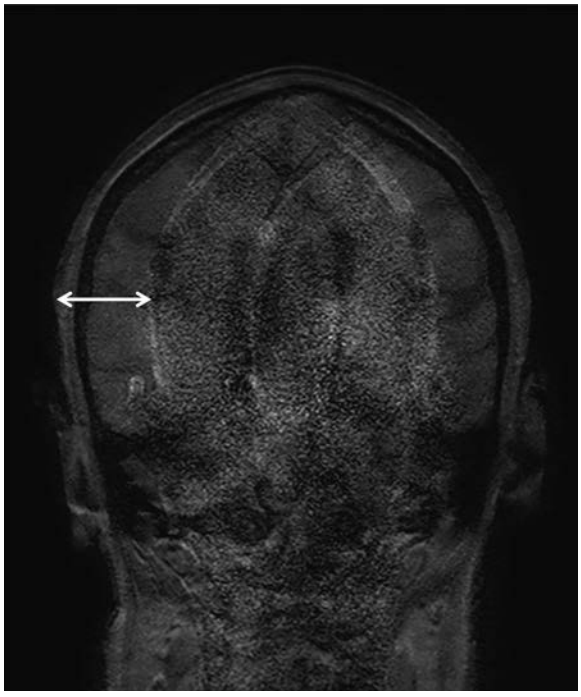

**Figure 7.18** Parallel imaging ghost in coronal brain. The distance between ghosts (white arrow) is 31 mm, while the phase-encode FOV is 180 mm. A very high SENSE factor was used to demonstrate this ghost; at normal SENSE factors it is hardly visible.

## 7.5 Susceptibility and Metal Artefacts

Susceptibility and metal artefacts are closely related, having essentially the same appearance on images, except that susceptibility artefacts are more subtle than metal ones. Typically a metal artefact consists of an area of zero signal, often with a very high intensity rim on one or two edges (Figure 7.19a) and with neighbouring regions showing significant geometric distortion. The high-intensity areas are caused by signal pile-up; the frequencies of these signals are so disturbed by the local ferromagnetic material that they all end up in the same FE location. Susceptibility artefacts may just have reduced, rather than zero, intensity and may not show any geometric distortion (Figure 7.19b).

As described in Chapter 3, different tissues become magnetized to a different extent when placed in the scanner's magnetic field due to their susceptibility differences. These microscopic field changes increase the dephasing of protons around boundaries between these tissues, reducing the signal intensity of voxels in the area. Most metals have much higher susceptibilities than the body tissues, creating large magnetic field non-uniformities around the object. Since metals are good conductors, they also absorb energy from the RF excitation pulses very easily, and can pose a safety hazard if they heat up.

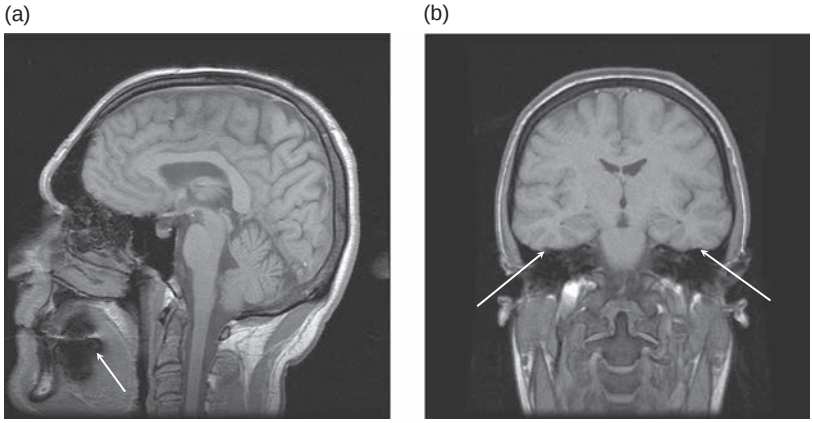


**Figure 7.19** (a) Metal artefact from dental work on a spin-echo image. (b) Susceptibility artefact in the temporal lobes on a gradient-echo image.

### Susceptibility Artefacts and Voxel Size

Magnetic susceptibility is defined as the extent to which any material becomes temporarily magnetized when it is placed in a large magnetic field (see Box 'Magnetization and the Meaning of Life' in Chapter 4). Among the body tissues, bone and air have the lowest susceptibility; most tissues have mid-range susceptibility, while iron-containing molecules such as haemoglobin and blood breakdown products have the highest. At the boundaries between these tissues, the slightly different magnetic fields within the tissues create micro-gradients which speed up the dephasing between protons on either side of the boundary. The phase change caused by susceptibility is given by

$$\Delta\phi = \gamma \cdot G_i \cdot \Delta r \cdot TE$$

where $G_i$ is the internal magnetic field gradient, and $\Delta r$ is the voxel size. This equation shows that susceptibility artefacts are worst with large voxels and at long TEs, and can be minimized by reducing TE or increasing the resolution. Often there is not just one simple boundary but many tiny boundaries on a microscopic level, for example in trabecular bone or the mastoid processes. Thus the $T_2^*$ is reduced over a large area, giving the characteristic low signal of susceptibility artefacts. Also keep in mind that susceptibility acts in all three directions, so you can get artefacts extending over neighbouring slices.

### RF Inhomogeneity Effects

Larger metallic implants also cause distortion of the radiofrequency field generated by the transmit coil. The implants tend to preferentially absorb RF energy, and thus neighbouring tissues don't receive a proper flip angle. The signals will be reduced and the artefacts are very similar to those produced by the susceptibility inhomogeneities. In practice you cannot separate the effects of RF and static field inhomogeneities just by looking at the images. Both gradient-echo and spin-echo sequences are affected by the RF inhomogeneity problem, and there is no way to avoid it.

Another common artefact caused by RF inhomogeneity is known as 'Moiré' fringes or 'zebra stripes' (Figure 7.20a). These effects are usually seen at the edges of large FOVs, especially where the patient's elbow or sides are very close to the transmitting body coil. Be aware that true-FISP-type sequences can also cause alternating stripes, because these scans are very sensitive to static field inhomogeneities $\Delta B_0$, but these tend to be thicker stripes and are not necessarily at the edges of the FOV. Don't confuse Moiré fringes with black-band artefacts!

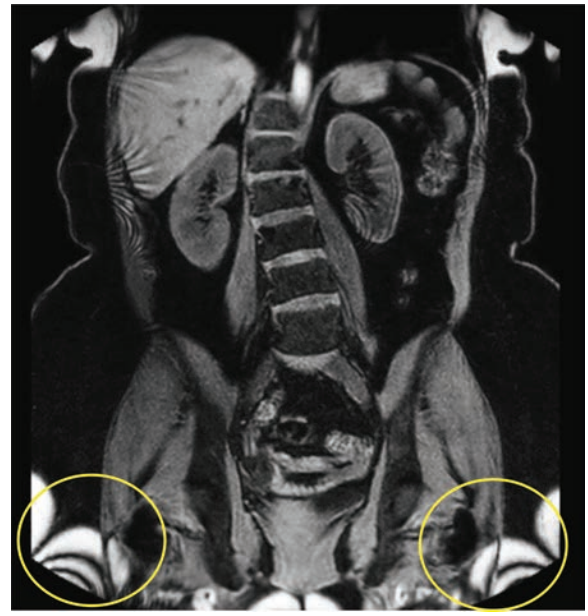

**Figure 7.20** Moiré fringes.

Because they are caused by inhomogeneities, metal and susceptibility artefacts are generally worse on gradient-echo images (Figure 7.19b) than spin-echo images, and they can be particularly marked on echo planar images. Spin-echo or TSE images may not show susceptibility artefacts at all. Susceptibility and metal artefacts can be minimized by using a high receiver bandwidth, or reducing the echo time (if $T_1$w or PDw contrast is required), but they cannot be completely avoided. If the images are severely degraded by metal artefacts, only (fast) spin-echo sequences should be used to acquire the data.

Metal artefacts also raise the question of safety for the patient, since the preferential absorption of RF can cause a local temperature rise. In this book we don't provide a list of MR compatibility for implants, as there are other texts and websites that can be consulted. Rather we hope to give you an understanding of the potential interactions between implants and the various fields used in imaging, which will help you to work out for yourself whether or not a particular implant is safe to scan. Chapters 2 and 20 include other aspects of safety advice.

In the last three years, new pulse sequence techniques have been developed which minimize the artefact around metal implants. They are based on TSE, and use extra phase encoding in the slice direction to collect information about signal displacements due to the metal. SEMAC (**S**lice **E**ncoding for **M**etal **A**rtefact

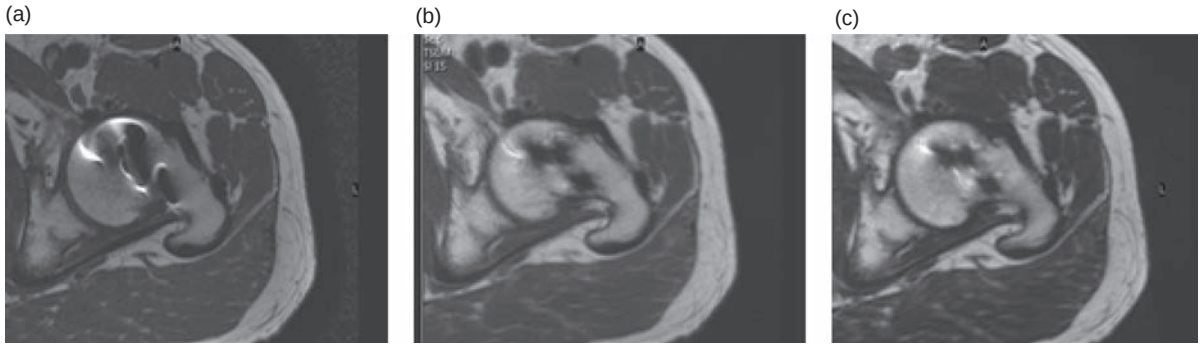


**Figure 7.21** Metal-artefact-reduction scans on an example of hip screws. (**a**) High-bandwidth TSE; (**b**) SEMAC; (**c**) MAVRIC. Note that both SEMAC and MAVRIC offer similar reduction of artefact compared with conventional TSE, in this particular example.

**C**orrection) is a 2D multi-slice technique with a small number of phase encodes per slice. By combining the information from all the slice phase encodes, the SEMAC scan generates an improved slice profile and resolves much of the artefact. MAVRIC (**M**ulti-**A**cquisition **V**ariable-**R**esonance **I**mage **C**ombination) is a 3DTSE method with a limited bandwidth used for excitation. Like SEMAC, MAVRIC is repeated with a small number of different bandwidths, to resolve the frequency displacements of the metal artefact. When these images are combined, the result has much lower artefact around the metal implant (Figure 7.21).

Both SEMAC and MAVRIC are time-consuming scans, because they use extra slice-encoding steps. Although both techniques can provide all types of image contrast, commercial availability is limited. At the time of writing, MAVRIC is only available as PDw, while SEMAC is being introduced with $T_1$w and STIR variants.

## 7.6 Equipment Artefacts

### 7.6.1 Zipper Artefact

The so-called zipper artefact, due to RF breakthrough, is probably the most common equipment artefact. It appears as a line of alternating light and dark pixels, sometimes two or three pixels wide, extending across the image in the phase-encode direction (see Figure 7.22). Occasionally there will be multiple zippers, regularly spaced across the image, but usually there is just one.

The cause of the zipper is external RF radiation finding its way into the magnet room and being picked up by the imaging coils. This may be due to a break in the RF-screened room – the metal shield built into the walls, floor and ceiling of the scan room. In this case the artefact will be present on all images. It may also be a problem in the system itself, either as a result of a component fault, or static electrical discharges caused by dissimilar materials rubbing together (e.g. cable insulation lying next to the metal magnet container). In the latter case, it might only show up on certain images. In all cases, the manufacturer's engineers should be called first to investigate the problem, followed by the RF cabin supplier if needed.

A more common cause is patient monitoring equipment, especially if it relies on metallic leads or mains leads going through the waveguides into the scan room. (This is a guaranteed way to get zippers in your image, but sometimes you have no choice, you have to work with what you've got.) The leads pick up RF waves from the environment and carry them through the Faraday cage, then radiate them into the room where they are picked up by the RF imaging coils. Even if there are no leads going through the waveguides, the power supplies may generate RF noise while charging or when fully charged, which may also be picked up by the imaging coils if the equipment has defective RF shielding. These problems can be particularly difficult to track down. Although it is more expensive, all monitoring equipment used should be specially designed for use in the MR room – it saves a lot of heartache over zipper artefacts!

**More RF Interference**

Generally the zipper artefact is the result of external RF, which is not coherent with the phase-encode gradient and it thus appears across the whole image. In rare circumstances the RF may be coming from

faults within the MR system, in which case it may be coherent and will appear as a very intense spot at the centre of the image. Another rare possibility is RF being carried on the mains electricity, which creates a 50–60 Hz modulation and a regularly spaced series of fairly faint zipper artefacts. Providing the monitoring equipment has been eliminated as a possible cause, any of these problems needs investigation by the system engineers.

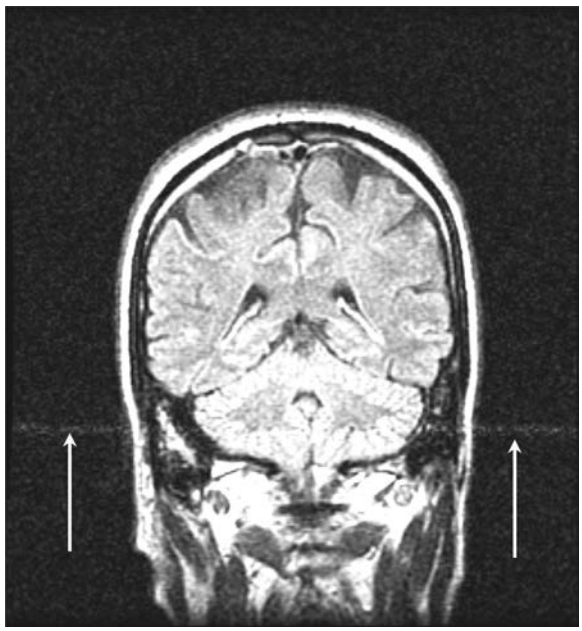

**Figure 7.22** RF break-through artefact.

## 7.6.2 Gradient Non-Linearity

The gradients only produce linear magnetic field gradients over a limited distance, and very large FOVs may include regions of gradient non-linearity; for example, whole-body or spine images. The effect of non-linearities is to distort the image, tending to compress the image information at the edges of the FOV. Many systems apply a correction to the images to stretch out the pixels, and on rectangular FOV a curved edge can be seen (Figure 7.23a). This is quite normal and also unavoidable.

## 7.6.3 Herring-Bone Artefact (Spike Noise)

The 'herring-bone' or 'corduroy' artefact is a regular series of high- and low-intensity stripes extending right across the image (Figure 7.23b). The intensity variation, the angle and the spacing of the stripes are all variable, and it often appears on just one or two images in a multi-slice set. It is caused by spike noise in the raw data (Figure 7.23c), whose Fourier transform (a series of spikes) is then convolved with all the image information. In theory the bad data points can be removed and the image reconstructed again, but usually the only solution is to re-scan the image. A single image with a corduroy artefact is probably just bad luck, but you are more likely to see problems with several different scans, especially diffusion-weighted images. Multiple bad data points in a single image give more severe artefacts which dramatically reduce SNR. This is a symptom of 'spiking' around the system, often caused

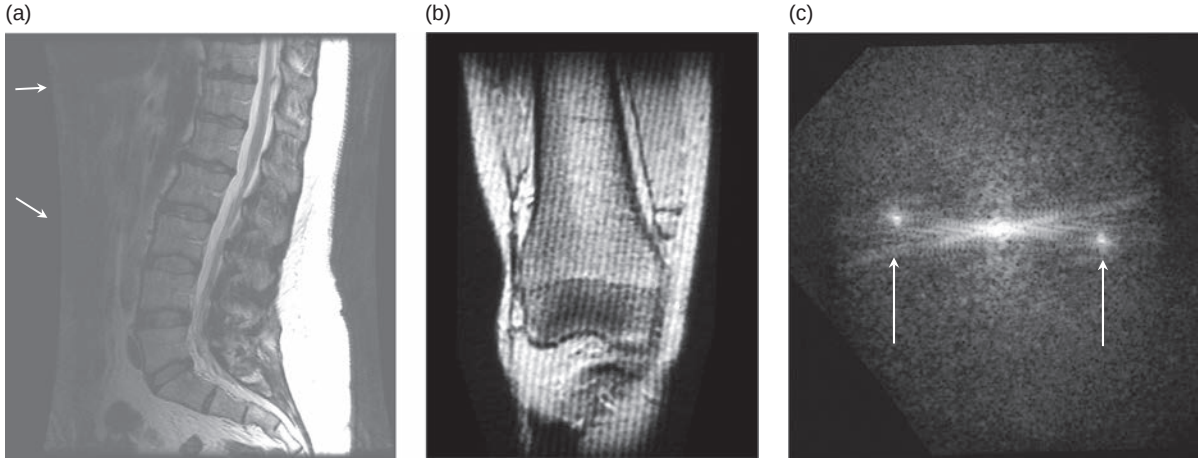


**Figure 7.23** (a) Curved edges on a large FOV after compensation for gradient inhomogeneities. (b) Corduroy or herring-bone artefact is caused by (c) a noise spike in k-space.

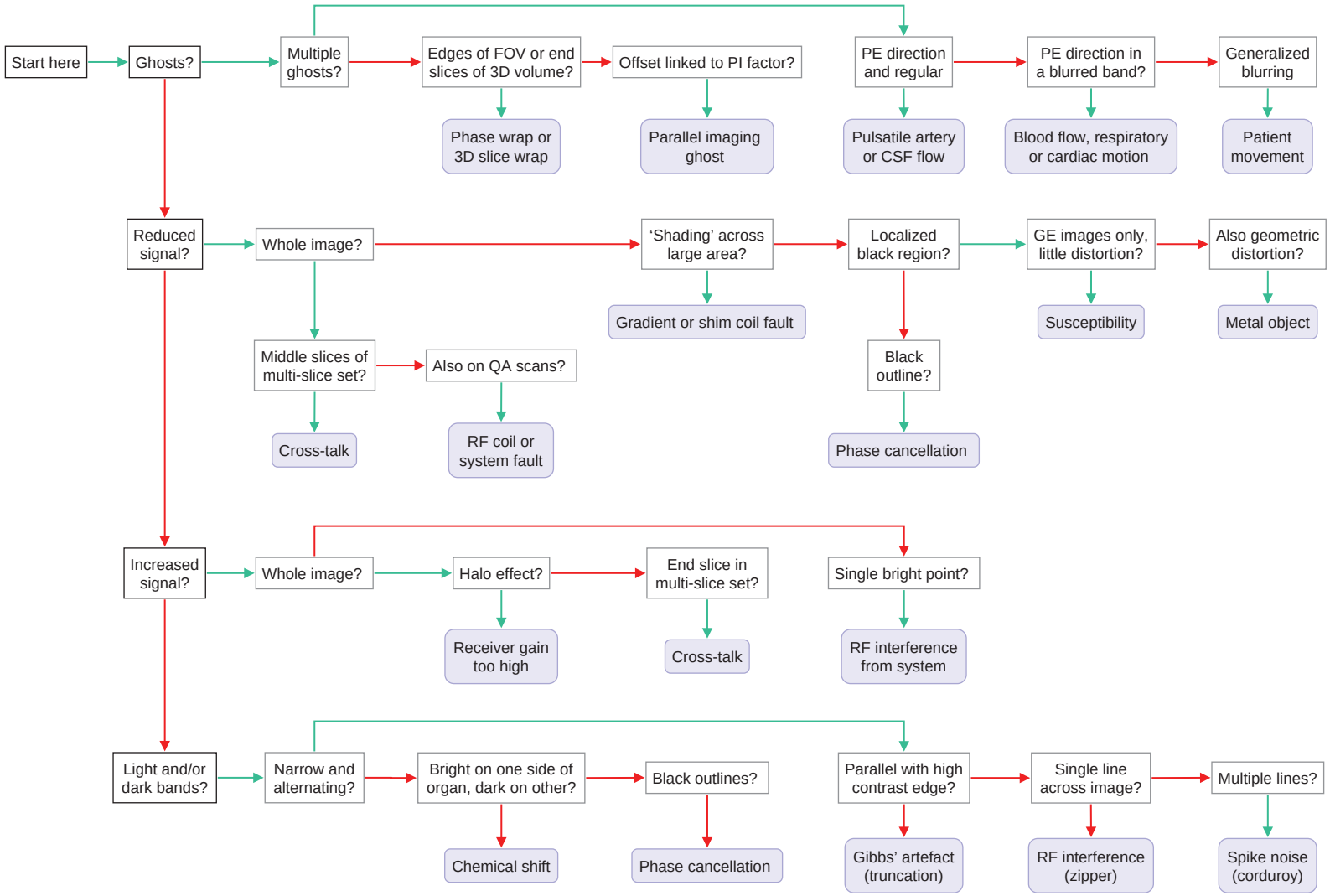


**Figure 7.24** The artefacts flowchart. Start by identifying the symptoms on the degraded image, then ask yourself the questions in the boxes. Follow the green arrows if the answer is 'yes' and the red ones if it's 'no'. You should arrive at one of the blue boxes, which will tell you the most likely cause of your artefact.

by static electrical discharges when components rub against the metal of the magnet, particularly when humidity is low. It is recommended to call the service engineers to investigate the problem.

## 7.7 What's Causing this Artefact?

We hope the flow chart shown in Figure 7.24 will help you to decide the most likely cause of your artefact, but we can't guarantee it! Bear in mind that serious system artefacts are less likely than patient motion or susceptibility effects.

**See also:**

- Flow appearances and MR angiography: Chapter 15
- Cardiac MRI: Chapter 16
- Safety of metal implants: Chapters 2 and 20
- Optimizing SNR and resolution: Chapter 6
- Phase encoding: Section 8.5.2

## Further Reading


Brown RW, Cheng YCN, Haacke EM, Thompson MR and Venkatesan R (2014) *Magnetic Resonance Imaging: Physical Principles and Sequence Design*, 2nd edn. Hoboken, NJ: John Wiley & Sons, various chapters.

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].

McRobbie D and Semple S (eds) (in preparation) *Quality Control and Artefacts in MRI*. York: Institute of Physics and Engineering in Medicine. Available via www.ipem.ac.uk.

MR Technology Information Portal, artefacts section: www.mr-tip.com/serv1.php?type=art [accessed 8 May 2015].

Radiopaedia, http://radiopaedia.org/articles/mri-artifacts [accessed 8 May 2015].