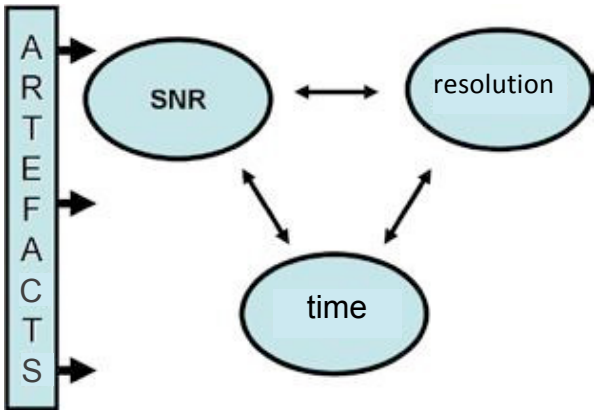
Module 6
Image quality and artefacts

After reading this module, you will:

- Be familiar with the factors influencing the image quality.
- Be familiar with the causes of artefacts and strategies for their avoidance / reduction.

6.1 Image quality



A clinical MRI examination always involves reaching a compromise between the image quality produced and the imaging time required. Each image should aim to reach an optimum balance between:

- A sufficient signal-to-noise ratio.
- An acceptable scan time.
- Sufficient resolution.
- Imaging is often disrupted unexpectedly by artefacts.

Every sequence stored in the scanner standard protocols has been adapted to the characteristics (T1, T2, pathology) of the organ regions to be examined; these often require only a minor adaptation of the parameters.

6.1.1 Spatial resolution

The spatial resolution is determined by the size of the cubes into which the examination volume is divided. The volume of these cubes (voxels) is determined by the size of the matrix (512 x 512, 256 x 256, etc.), the Field of View (FOV) (10 cm, 20 cm, etc.) and the thickness of slice (3 mm, 5mm, ...).

The smaller the FOV selected (with a fixed matrix size), the greater the resolution in the level: Whilst the number of pixels per unit area increases, the pixels themselves become smaller. Not only do the FOV, matrix size and slice thickness influence the resolution of the image, but also the measurement time and the signal-to-noise ratio.

Altering one of these measurement parameters has a number of effects. The optimum setting always involves a compromise, primarily between image quality and measurement time.

6.1.2 Signal and noises

An MRI image can be likened to an ancient relief picture, on which every spatial point is assigned to a specific height (intensity). In this analogy, the noise represents the result of a sandstorm which has rushed across the picture and left its mark.

These random variations in the signal intensity reduce the amount of information provided by the image. This interference comes primarily from the body of the patient (thermal noise and the associated HF emissions). The entire scanner measurement system (coils, leads, computer...) makes a contribution.

This noise impairs the signal which the spins receive from the slice subject to examination. Mathematically, the signal-to-noise ratio (SNR) represents a measurement of the relationship of the average signal from a slice and the fluctuation (standard deviation) of the noise.

Figure 6.1:
The rock reliefs of Bishapur. Parts in various states of weathering

The SNR is subject to a number of factors which cannot be changed: The scanner specifications (field strength), the nature of the sequence, the patient BMI, tissue characteristics. Other factors can be altered:

- The coils used for imaging
- The sequence parameters (voxel size, averages, receiver bands)

Receiver coils

The coils covering the length of the patient body integrated in the scanner "illuminate" the entire scan area. Initially, this appears to be useful.

Nevertheless, the smaller the sensitivity volume (the "illumination") of a coil, the lower is the noise-reduction of the surrounding structures. This improves the SNR. A local coil or a surface coil has a better SNR than the coil covering the entire body.

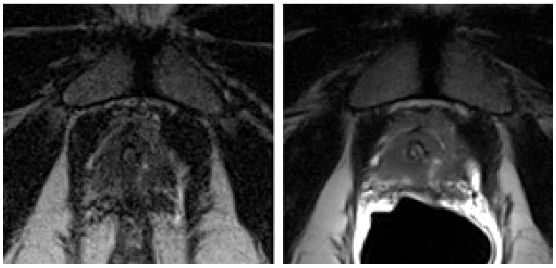


Figure 6.2:
Surface vs. endorectal coils

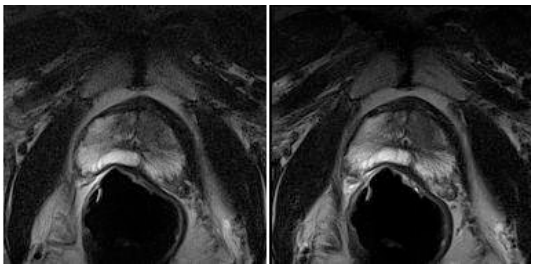
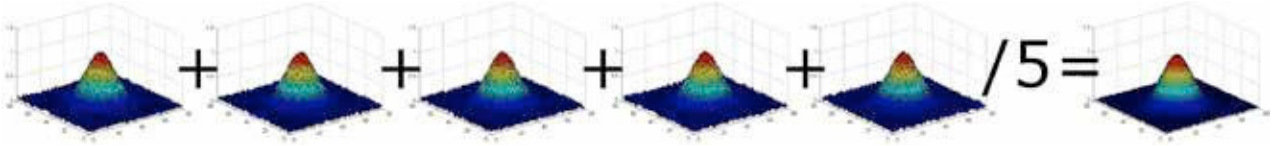


Figure 6.3:
3 and 5 mm slice thickness

The signal comes from the stimulated hydrogen nuclei in the selected slice. As outlined in chapter 2, at 1.5 T, the number of the stimulated spins in ratio to the total number of spins is exceptionally low (six in a million nuclei). Moreover, the imperfect nature of the gradients means that a proportion of the spins in the imaging slice do not contribute to the signal, as the resonance conditions are not met.

We lose the signal, but still measure the same level of noise. Even without the external source of interference, when halving a voxel in all three dimensions, only an eighth of the spins make a contribution to the signal whilst the noise remains constant. One compromise involves selecting a high resolution in the plane but a larger slice thickness. In practice, voxels present a greater resemblance to "image pins" than cubes.



The number of averages

It is possible to compensate for a poor SNR by repeating an imaging procedure and taking an average. Every repetition involves the reception of an identical signal from the voxel, whilst the noise remains random and is not the same with every measurement.

Sometimes it hits one voxel, sometimes another. Adding the signals from all images pointwise and dividing them by their number shows that the influence of noise decreases with every additional acquisition. Take care: two averages do not mean a double improvement of the SNR, but an improvement of c. 40 %.

Doubling the SNR requires four averages. The time:benefit ratio worsens with every average due to dependence on the root. Despite the unacceptable lengthening of the measurement time, eight averages brings only a 2.8 times greater SNR.

The receiver bandwidth

The reception of a MRI signal can be performed quickly or slowly, depending on the "reception bandwidth.

A large (frequency) bandwidth corresponds to a fast reception (a strong gradient); the measurement at a low bandwidth benefits from a slow read-out gradient and takes longer.

As the background provides a constant noise contribution on all frequencies, a large bandwidth records noise from a number of frequencies around the actual signal. A reduction of the bandwidth thus enables a better SNR, but is much more time-intensive.

6.2 Artefacts

The quality of MR images is often impaired by artefacts. They can only be avoided to a limited extent. It is necessary to know their cause in order to deal with them.

In general, artefacts are caused by:

- The scanner system
- Movement
- Magnetic susceptibility
- Infolding
- Layers of signal intensity
- Chemical displacement

6.2.1 HF artefacts

HF artefacts have a range of causes: Adulterated image contrasts or fish bone patterns indicate a disruption of the transmitter and receiver system for the HF waves.

Noise bands along the phase encoding direction or "zip artefacts" are often caused by external disruption frequencies, such as a loose cable in the MRI room.

Figure 6.4:
Artefacts caused by disruptions of the HF transmitter and reception system and exterior disruption frequencies.

6.2.2 Motion artefacts

Patient movements represent the most common cause of artefacts.

We differentiate between two types of motion artefacts:

- Random movements resulting in blurred or noisy images (along the phase encoding direction).
- Periodical movements of the ghost images along the phase encoding direction.

Motion artefacts appear when the spins are no longer located in the same position during signal reception as they were during phase encoding. They also develop if the spins change position during the repeat of the sequence (scanning a different k-space line) and then contribute to the signal again.

In both cases, the spatial encoding of the voxels has been falsified; this has an impact on the artefacts along the phase encoding direction. The encoding in the frequency direction is performed so fast that the motion does not have any influence on the images.

Periodical movements (heart beat, blood pulses, breathing) can cause ghost images. Ghost images and their intensity depend on the strength of the movement (depth of breathing) and the periodical repeat rate and the TR selected.

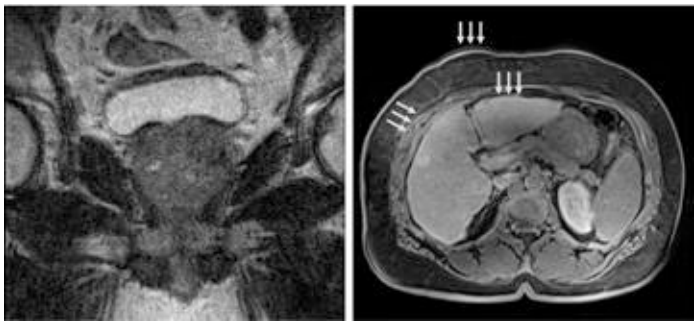


Figure 6.5:
Left: Patient movement Right: Breathing

Avoiding motion artefacts

The strength of the artefacts resulting from random movements depends on the preparation of the patient (location, sedatives, explanations given) and their co-operation (children). Abdomen imaging is often supported by the addition of butylscopolamine or glucagon in order to avoid uncontrolled intestine movements.

Special sequences using breath holding phases enable minimization of the influence of periodical movements. Alternatively, the periodicity of respiration can be measured; the image acquisition can be performed in breaths of the same length. This is highly time-consuming but the slowed T1 relaxation and the longer TR does not permit the application of breath holding techniques with an abdominal examination at 3T.

Cardiac imaging usually uses an approach which measures the heart frequency to adapt the length of TR.

The scanning of the k-space is triggered. This makes the image-acquisition time-consuming and sensitive in examinations of patients with irregular heart frequencies. Motion artefacts from blood flow (especially in the aorta) can be "kept out" of the area of interest by a simple inversion of the phase and frequency encoding directions from the region of interest (e.g. the spine).

Activating saturation bands in front and behind the investigation volume in the direction of inflow of the blood enables its saturation.

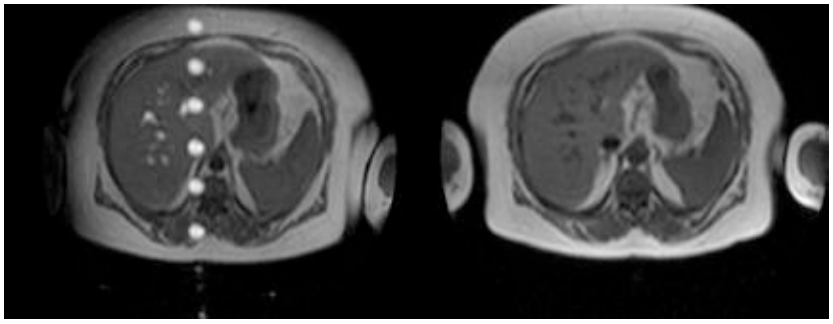


Figure 6.6:
Pulsation artefact and its avoidance

6.2.3 Susceptibility artefacts

Magnetic susceptibility is associated with the inner magnetization of a tissue and indicates the strength of the reaction with other magnetic fields (static field, gradients).

If two materials exhibiting very different susceptibility are located close to each other, the magnetic field is subject to local perturbation. In particular, this exerts an influence on the transversal relaxation, as the inhomogeneity of the spins dephases faster than expected. Natural causes can include for example, the differences of susceptibility between air and tissue or bones and soft tissue.

The image artefacts often appear in the form of local signal losses (poor contrast) or image distortion. Every metal (whether ferromagnetic or not) generates strong artefacts and distortions in the images.

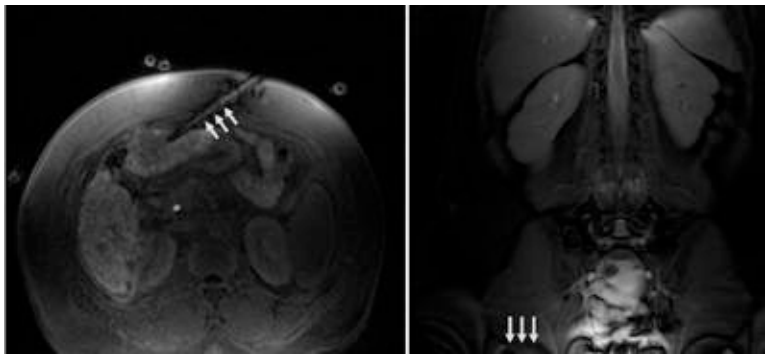


Figure 6.7:
Left: Artefact from a titanium needle.
Right: the artefact of a belt.

Avoiding susceptibility artefacts

Should they be accessible, all disruptive objects (dental prosthetics, belts etc.) are to be removed. Where this is not possible, the pulse sequences should be chosen in such a way as to minimize the extent of their influence.

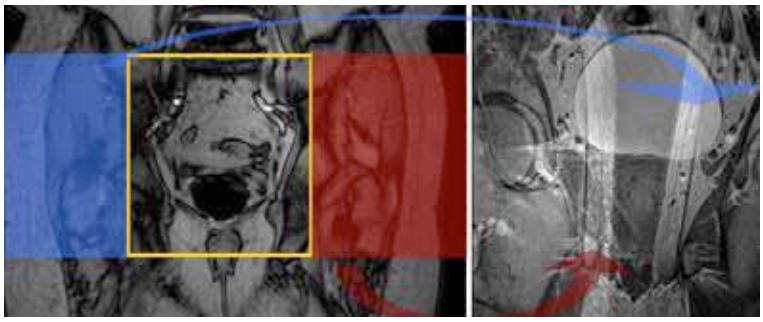
In general, SE sequences are less prone to susceptibility artefacts, as the reception of a spin echo partially intercepts the inhomogeneities (see the information on rephasing provided above).

The targeted use of susceptibility differentiation

Susceptibility-Weighted Imaging (SWI) uses spatially high-resolved 3D gradient echo sequences to reveal the different magnetic susceptibilities of the various tissues.

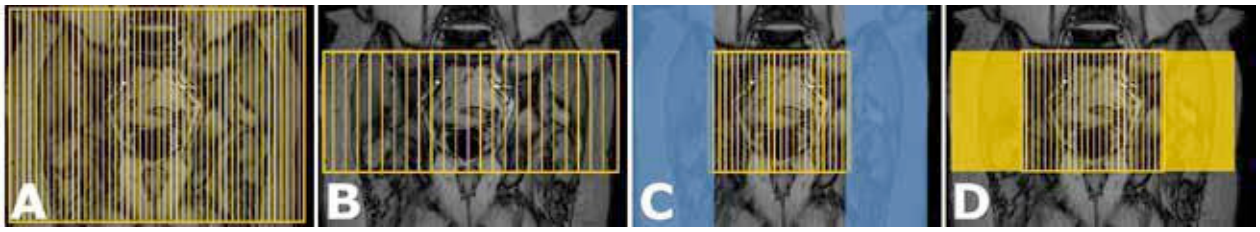
Oxygen-rich arterial blood (short T*,2) produces e.g. a stronger signal loss than venous, oxygen-poor blood. Devices with higher magnetic field strengths produce e.g. good images of venous vascular systems and local iron deposits.

6.2.4 Infolding



Infolding is nothing less than signal contributions from anatomic structures outside the Field of View and their "infolding" as a mirror image of the opposite side of the image.

They are caused by disruption of the spatial encoding under the influence of the objects outside the FOV and the incorrect spatial assignment in the reconstruction of the k-space information. The effect manifests itself along the phase encoding direction.



Avoiding infolding

A number of instruments help avoid infolding: if the FOV covers the body part to be examined during phase encoding, nothing is infolded. This is accompanied either by inferior resolution (the same matrix with a larger FOV) or a significantly longer measurement time with an identical resolution (A).

Alternatively, it is possible to reverse the direction of the frequency and phase coding - not practical in our example, as the head-foot direction would be folded in. It is also possible to "inflate" the voxels in the phase encoding direction until the entire pelvis has been covered. Nevertheless, the spatial resolution falls in a direction and the voxels are no longer isotropic (B).

Alternatively, saturators can be placed around the investigation volume, which target and destroy the magnetization in this area, thereby removing the contribution to the signal (C). This does not always work perfectly. The phase oversampling / fold-over suppression / no phase wrap can be used to increase the number of phase encoding steps until the pelvis area has been covered entirely (D). The strips outside the FOV are "discarded"; only the region of interest is depicted.

This method is not associated with any disadvantages in terms of the image quality. On the contrary, the signal also grows stronger, although lengthening the acquisition time.

6.2.5 Stair-step artefacts

If an area of very high-intensity borders on an area of low intensity, it is difficult to express this in the frequency space. This can be conceived of as a sharp edge which needs to be approached by a curve.

This is possible given a high spatial resolution (the composite of a large number of frequencies). With poor resolution, the resulting image appears as though a wave had broken on a coast and then moved into the land's interior in parallel to this line.

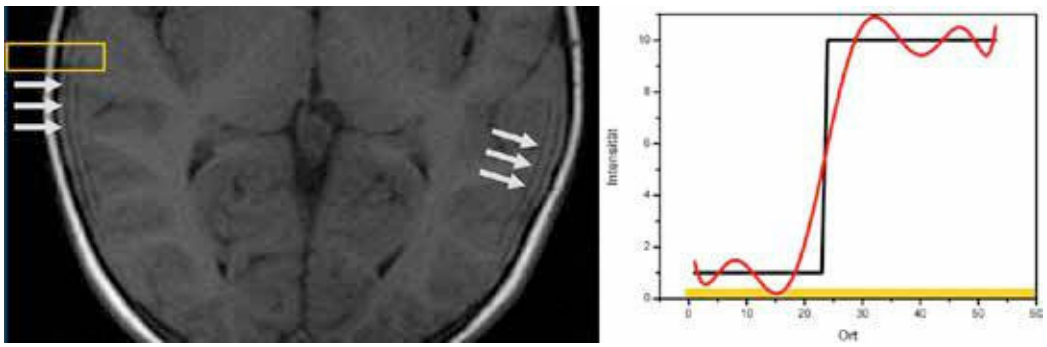


Figure 6.8:
Stair-step artefacts (arrows)

Avoiding stair-step artefacts

Stair-step artefacts can be avoided by using a higher spatial resolution and through the selection of special filters used in image calculation.

6.2.6 Chemical displacement

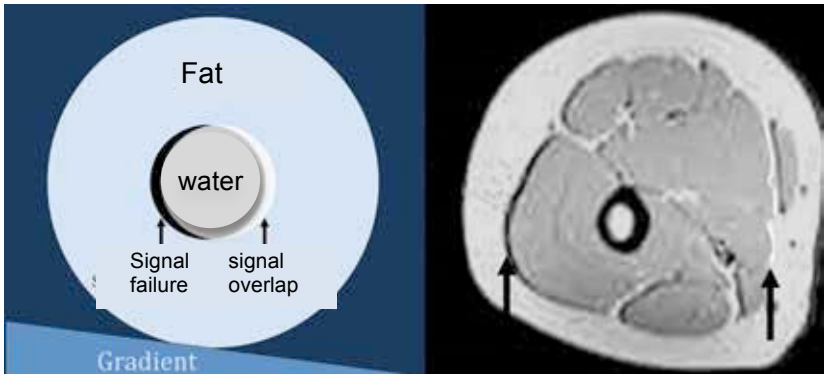


Figure 6.9:
Fat-water displacement

Every hydrogen nucleus precesses in the static magnet field with the Larmor frequency. With a 1.5 T MRT, this amounts to c. 63.8 MHz. This frequency is dependent not only on the strength of the static field, but the composition of the surrounding tissue. Every other type of nucleus, e.g. chemical bonds in metabolism intermediate products effects a minimal change to the resonance frequency of the hydrogen spins.

The strength of the displacement depends on the nature of the chemical element, but not its concentration. The changes to the resonance frequency are specified in the unit ppm (parts per million). This is NOT a measurement of the concentration of a metabolite in tissue, rather it indicates the number of HZ per MHz (Hertz per million Hertz) by which the frequency was displaced. This brings the advantage of comparability independently of the MRI static field.

The fat-water displacement is a special case of this general chemical displacement. At 1.5 Tesla, the resonance is displaced by some 220 Hz – compared to some 64 million Hz, this represents just 0.0003% – but the impact can be considerable:

As already described, frequency encoding uses a local change in the Larmor frequency with gradients as an encoding method; every location should accord to a specific frequency and is assigned accordingly upon the image reconstruction. Fatty tissue also displaces the Larmor frequency. During image reconstruction, voxels high in fat are subject to systematic displacement in a single direction, as the frequency displacement resulted in their being assigned to the wrong location. Basically, we receive a fat and a water image, both slightly displaced towards each other.

Were we to restrict ourselves to an investigation of fatty tissue, all the voxels would be displaced; the effect would be invisible. As things are, every transition between fatty tissue and watery tissue presents a jump: Everything high in fat is located a little to the left, resulting in the one-sided absence of the signal on the transition. On the other side of the image, it appears as though the fat presses up onto the water-rich tissue and produces a signal overlap. Tissue with direct transitions between fat and water such as that presented in the spine and the intervertebral discs or between the spleen and kidneys and the surrounding fat are especially susceptible to these artefacts.

Avoiding chemical displacement

As described, the chemical displacement displaces the frequency of the spins. At 1.5 T for fat, this is some 220 Hz. The selection of a bandwidth of > 220 Hz / pixel reduces the influence of the chemical displacement; nevertheless, this is achieved at the cost of the SNR (see above).

An alternative response is to invert the phase and frequency encoding direction or use fat / water suppressing sequences.

The advantages of chemical displacement

As every tissue is characterized by specific processes of metabolism, specific concentrations of chemical elements (metabolites) in tissue influence the resonance frequency of the hydrogen.

These influences are so low that they are usually unable to exert any influence on normal imaging. We call a depiction of these influences a spectrum. Within such a spectrum, free hydrogen would only issue a single clear line (peak) at 0 ppm. On the other hand, water bound in hydrogen is displaced by 4.7 ppm. The presence of other chemical elements in the surroundings of the hydrogen exerts an influence on the frequency of a certain number of spins (those lying close to these foreign molecules). This produces a line division.

This generates an additional peak in the typical frequencies for this chemical element (characteristic ppm value). The concentration of this element influences the height of the additional peaks. Knowledge of the ppm value of at least two metabolites and the height of the measured peaks enables calculation of their concentration ratio in the tissue.

Spectroscopy permits the identification of atypical concentrations of metabolites in tissue, permitting conclusions to be drawn regarding various illnesses. Spectroscopy is highly sensitive and in comparison to conventional imaging techniques, more susceptible to exterior disruptive influences as the measured effects are extremely small.

Metabolite		Frequency [ppm]	Function	Anomalies
Myo-Inositol	mI	3.6	Signal transmission	↑ Glioma, MS
				↓ Herpesviral encephalitis
Choline	Cho	3.2	Metabolism marker	↑ Tumours, MS
			of the cell membrane	(demyelinating diseases)
Creatine	Cr	3.0	Energy metabolism	Reference marker
				(constant)
Citrate	Ci	2.6	Prostate metabolism	↓PCA
GABA, Glutamine	Glx	2.1-2.5	Intracellular Neuro transmitter	↑ Hepatic Encephalitis
N-acetylaspartate	NAA	2.0	Brain metabolism	↑ Canavan disease
				↓ Neuron damage
Succinic acid	Succ	2.4	Suppurative abscess	
Acetate	Ac	1.9	Abscess	
Alanine	Ala	1.5	Meningioma, abscess	
Lactate	Lac	1.3	Ischemia, tumour	↑ Anaerobic metabolism
Free lipids	Lip	0.9	High-grade	
		1.4	necrotic tumour	

Table 6.1:
Typical metabolites, their function and illnesses with anomalous concentrations.