

Javier Sánchez-González

DWI is currently considered a cancer biomarker and has a role in cancer detection and staging. DWI is also able to depict early posttreatment changes in oncological lesions treated with vascular disruptive drugs and useful for therapies that induce apoptosis. After these treatments, cellular death and vascular changes occur before changes in lesion size can be seen. Successful treatment is reflected by increases in ADC values. Rising ADC values with successful therapy have been noted in several anatomic sites, including breast cancers, primary and metastatic cancers to the liver, primary sarcomas of bone, and in brain malignancies.

These new applications make necessary to control the quality of DWI sequences which must be as accurate as possible for posterior quantitative analysis. In order to analyze the technical issues that can affect the quality of the diffusion images, it is necessary to decompose the typical diffusion acquisition scheme described in Chap. 1. This scheme is made up of two different parts. The first part of the sequence corresponds to the preparation phase of the magnetization, which is called the diffusion preparation part (Fig. 1.1.1). The second part is the readout scheme to acquire the images and will be referred as the acquisition part (Fig. 1.1.2). Although both parts are intimately related, the effect of them in the final image can be separated as well as their related artifacts.

## 2.1 Optimization of Signal to Noise Ratio

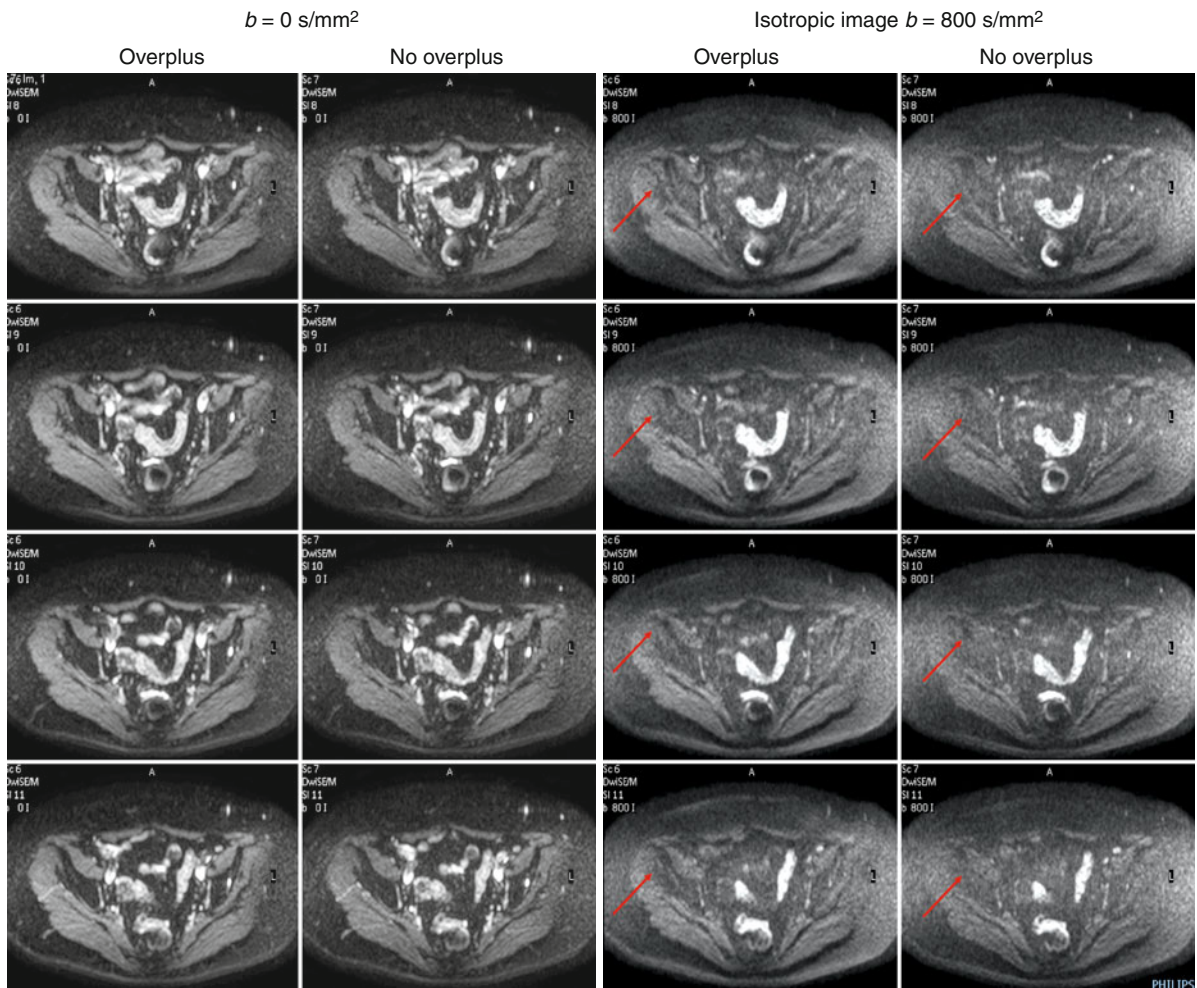
Since DWI is prone to have low SNR, it is necessary to recover as much signal as possible. In order to increase the SNR, it is desirable to reduce the effective TE of the sequence to the minimum. The final TE of the diffusion sequence is affected by the total time of the diffusion preparation and the effective TE of the acquisition part.

As it was commented on in Chap. 1, in order to reduce the signal loss due to T2 effects, it is recommended to reduce the sequence time as much as possible (Fig. 1.3). In this sense, for a given  $b$  factor, it is recommended to use the maximum available gradient strength during the diffusion gradient lobes. In order to obtain the maximum available gradient strength, tetrahedral encoding or other simultaneous applications of gradient schemes (e.g., gradient overplus or three-scan trace) can be also used. These techniques do not use the diffusion-weighted gradients in pure X, Y, and Z direction. On the contrary, new diffusion directions are defined combining the maximum intensity of all the gradients at the same time. This approach allows to obtaining a maximum gradient strength that is the square root of three times higher than the gradient strength in a single X, Y, or Z pure direction. As a result, shorter effective TE can be reached improving the total SNR of the sequence (Fig. 2.1).

DWI normally has a low SNR especially for those anatomies that require high  $b$  values (e.g., prostate). To compensate this signal loss for high  $b$  values, it is desirable to increase the number of averages (Fig. 2.2). In order to reduce the scan time, “state-of-the-art” scanners

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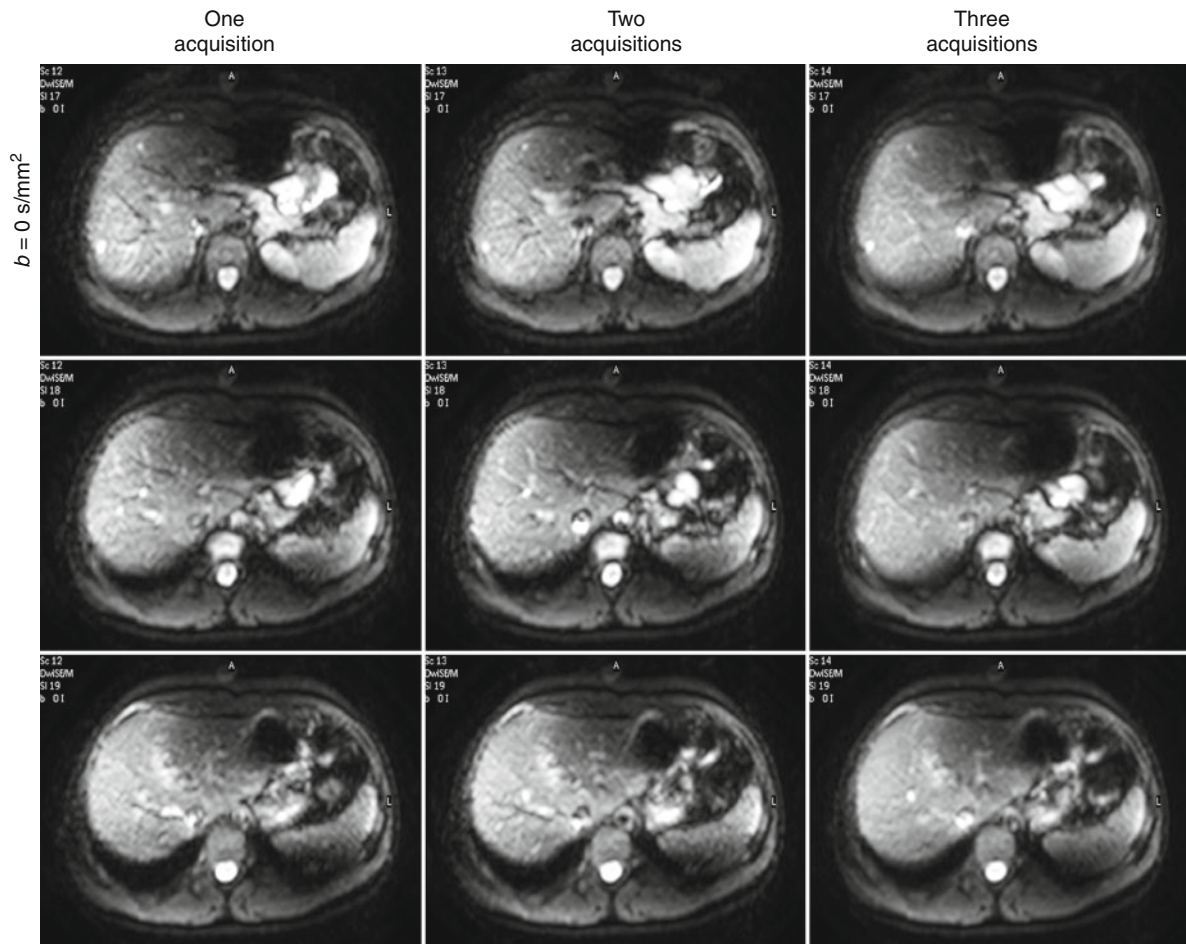
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SSh SE EPI diffusion  
 SENSE factor = 2  
 Acq Reso:  $2.4 \times 2.4 \times 4.5 \text{ mm}^3$   
 Fat suppression: SPIR  
 Scan time = 1:08  
 Effective TE = 56 ms (Overplus gradient strategy)  
 68 ms (Non overplus gradient strategy)

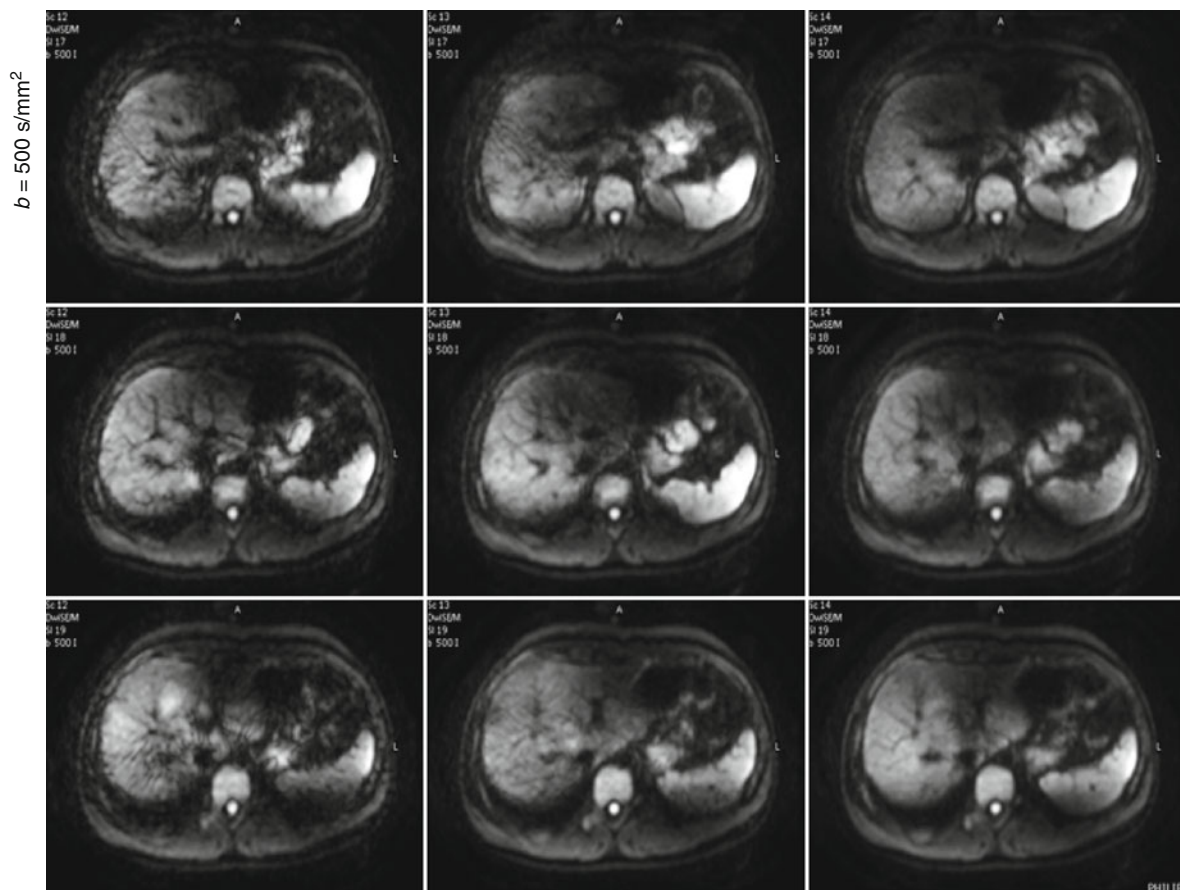
**Fig. 2.1 SNR: gradient overplus use.** In this figure, a pelvic diffusion-weighted SS SE EPI using SPIR sequence with  $b$  values of 0 and  $800 \text{ s/mm}^2$  was acquired with and without gradient overplus strategy in the same volunteer. It can be appreciated that images acquired with overplus strategy have higher signal

compared with those acquired with nonoverplus strategy due to the shorter TE. Notice increased depiction of the borders of the *right acetabulum* in the acquisition with high  $b$  value using gradient overplus compared to the one without gradient overplus (*red arrows*)



**Fig. 2.2 SNR: effect of the number of signal average.** In this figure, three liver DWI acquisitions in the same volunteer with different number of averages are shown. As it is expected, it can be noticed that SNR improves with the number of averages. An improved SNR will provide a better estimation of the ADC. In those cases where the gradients are not powerful enough or the tissue has a very high diffusion coefficient, the best way to increase the SNR is to increase the number of averages. The

main problem of this strategy is that it is very time consuming. In order to reduce the scan time, modern scanners have the option to perform a variable number of averages according to the  $b$  value increasing the SNR just in those images with higher  $b$  values that are more prone to have low signal. This strategy makes feasible to maintain a good SNR in all the acquisitions with different  $b$  values in a reasonable scan time



SSh SE EPI diffusion  
 No triggered acquisition  
 Acq Reso:  $3.0 \times 3.0 \times 7 \text{ mm}^3$   
 Parallel factor = 2  
 Fat suppression: SPIR  
 Scan time = 49 s

**Fig. 2.2** (continued)

has the capability to perform variable number of averages according to the  $b$  value performed, increasing the SNR just in those images with high  $b$  values that are

prone to have low signal. This approach makes it possible to have a reasonable SNR in all the acquisitions with different  $b$  values in a feasible scan time.

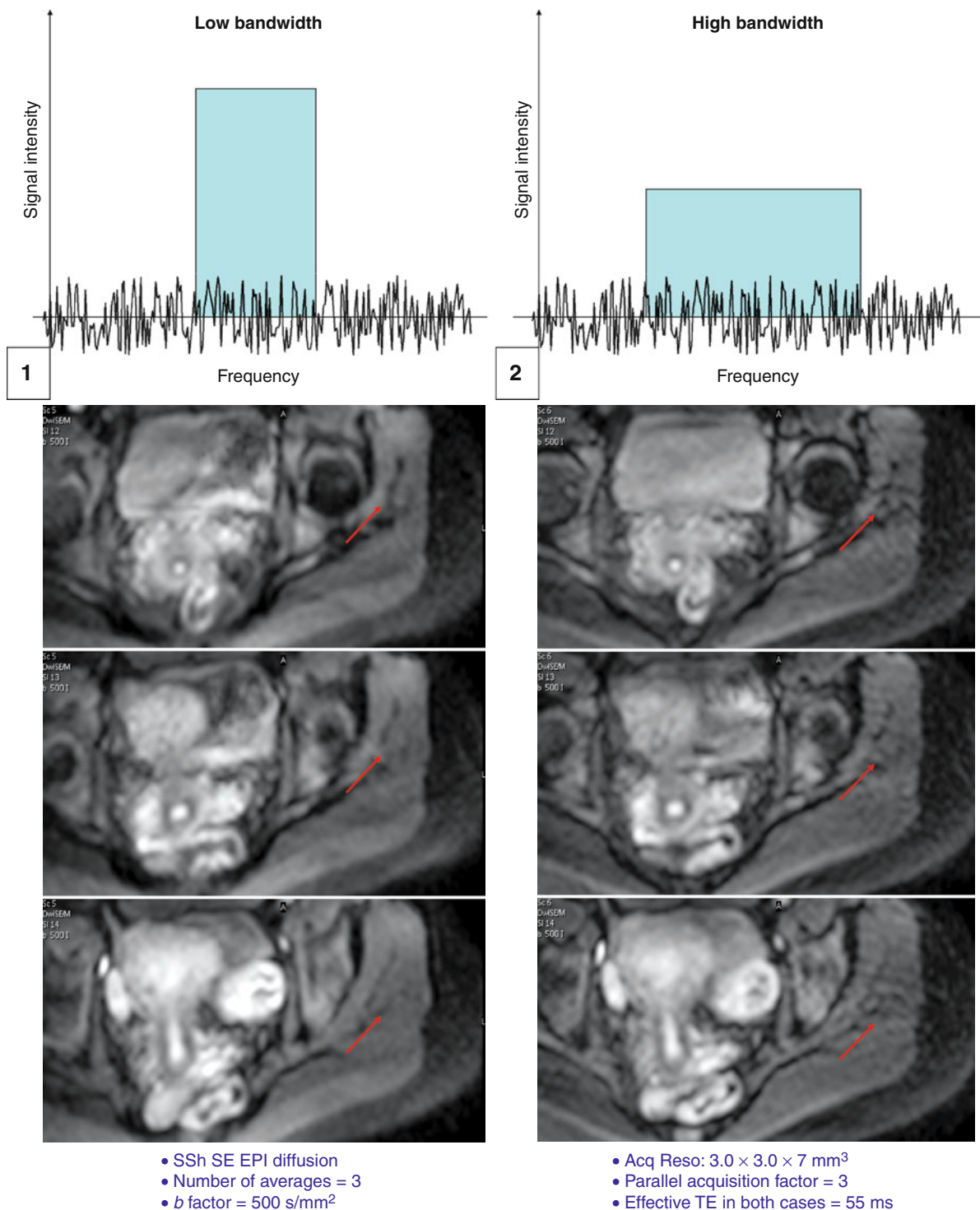
**Fig. 2.3 Bandwidth: noise artifacts.** A tool to improve the SNR is to adjust the acquisition bandwidth of the sequence. Schematic representations of the effect of the use of low and high bandwidth (1 and 2 respectively) on signal intensity are shown in the upper part of the figure. In both schemes, the *blue box* represents the total amount of the acquired image while the *black line* represents the noise level. In both cases, the total signal (area of the *blue box*) as well as the noise contamination is equal. Taking into account the definition of white noise, the same noise contamination is expected for all the bandwidth fre-

quencies. In the low bandwidth acquisition (2.3.1), the *blue box* is spread in a narrow bandwidth obtaining a higher SNR compared with the high bandwidth acquisition (2.3.2). Besides, using high bandwidth, the signal is spread in more frequencies, including more noise in the acquired signal. Two series of a pelvic DWI with a  $b$  value of  $500 \text{ s/mm}^2$  using different bandwidth frequencies in the same volunteer are shown in the lower part of this figure. The images acquired with low bandwidth show a higher SNR than those with high bandwidth (*red arrows*)



Another degree of freedom to improve the SNR is to change the received bandwidth during the acquisition (Fig. 2.3). This sequence parameter has to be treated

carefully as lower bandwidth values increase the SNR of the sequence but produce higher image distortion as will be discussed in the distortion artifact section.



## 2.2 Geometrical Distortion Artifacts

On DWI, some distortion is usually appreciated due to phase error accumulation during the EPI readout, although the MR scanner is perfectly adjusted. These geometric distortions can be reduced by increasing the readout bandwidth. Increasing the readout bandwidth has two effects. The first one is that the frequency difference between two consecutive phase encoding lines is higher, making that the phase error due to magnetic field inhomogeneities has less effect in the total readout. Moreover, the second effect is that a faster readout (equivalent to a higher bandwidth) leaves less time to change the signal phase due to magnetic field inhomogeneities producing less image distortion. For body applications, the difference between two phase encoding lines is typically between 1 and 2 kHz (Fig. 2.4). However, a higher readout bandwidth also increases the noise and Nyquist ghosting. Therefore, bandwidth or echo spacing settings should be optimized (Fig. 2.3). Another way to reduce the accumulation of phase error during the readout is to use parallel acquisition techniques (e.g., SENSE, GRAPPA, etc.). These techniques skip the readout of phase encoding steps that are compensated using the geometrical information of the coil sensitivity maps in the reconstruction process. As a consequence, this acquisition strategy reduces the effective echo spacing and the effect of image distortion (Fig. 2.5).

## 2.3 Motion Artifacts

Another problem involving the diffusion signal is the macroscopic movement produced by respiratory and cardiac motion which is critical in thoracic and abdominal acquisitions. In order to avoid these movements, different strategies have been proposed. These strategies have been carefully studied in the liver. Kandpal et al. demonstrated that respiratory triggered DWI acquisitions showed higher SNR in normal liver and higher CNR between normal liver and focal lesions than breath-hold sequences. Kwee and colleagues studied the effect of the heart motion on DWI of the liver, showing a strong degradation of those images acquired during the heart systole due to the effect of the heart movement (Fig. 2.6). Therefore, motion control mechanisms are necessary to reduce these artifacts (see Chap. 1 and Fig. 13.1).

## 2.4 Eddy Currents Artifacts

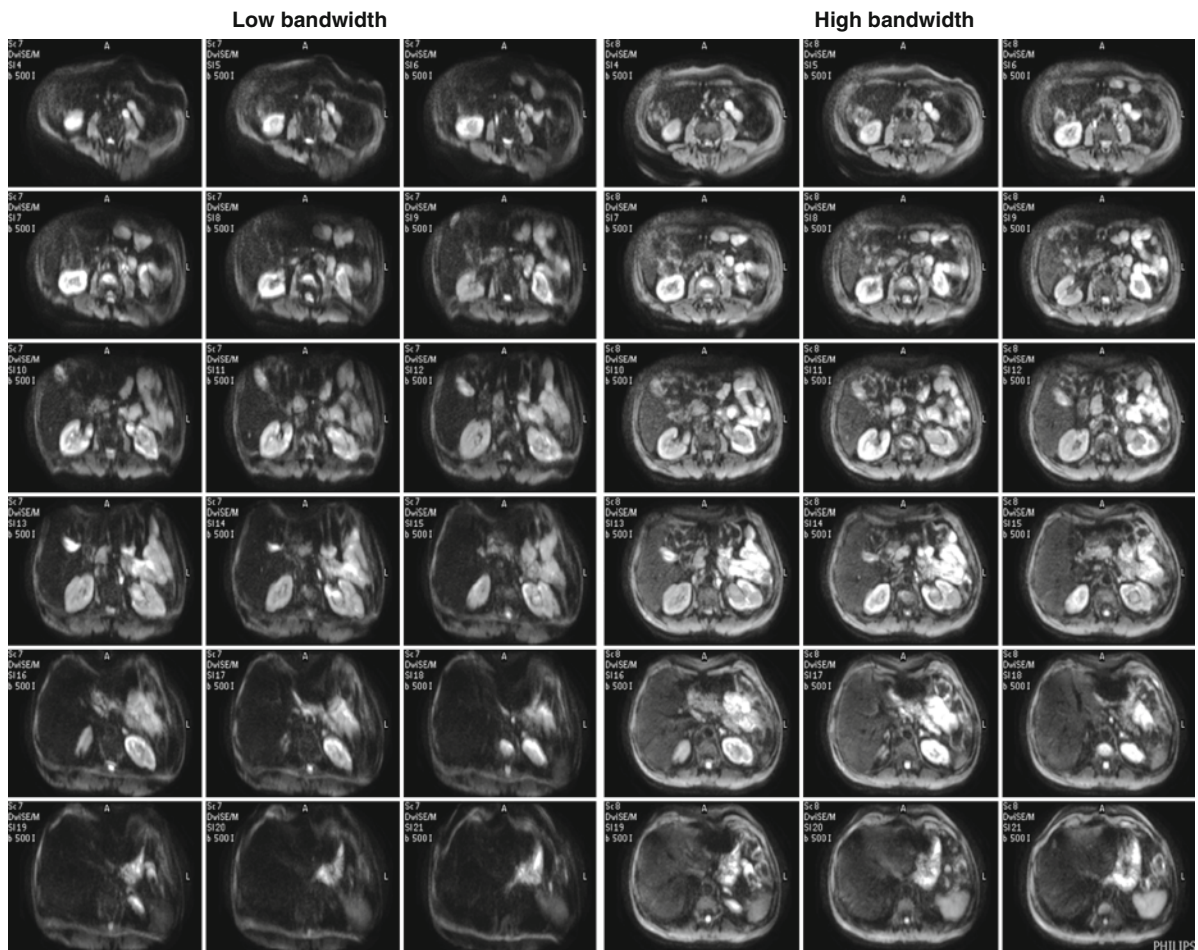
Eddy currents are generated by gradient switching producing changes in the static magnetic field. If the magnetic field variation produced by eddy currents disappears between the time of the applied field gradient and the image readout, a spatially dependent change in image phase with no discernible distortion will result. Diffusion encoding normally relies on the attenuation of the image magnitude rather than in the phase of the image. Therefore, a change in image phase of DWI does not change the diffusion measurement as long as the phase gradient per pixel is small. However, when the eddy currents decay slowly, a residual magnetic field remains during the image readout. This field behaves like an additional spatial encoding gradient field causing distortions or shifting of the image (Fig. 2.7).

From a technical point of view, eddy currents are compensated changing the gradient waveform in such a way that the final result is a very stable gradient on time. This technique is called pre-emphasis.

## 2.5 Fat Suppression Artifacts

Fat signal produces many difficulties in the acquisition of DWI in body applications, which are derived from the 3.4 parts per million shifting of the precession frequency of the fat signal from the water one. This frequency difference produces a water-fat shift in the EPI readout that can make the fat signal overlay in the studied region. Moreover, the contribution of the fat signal to the image is more pronounced for high  $b$  values, due to its very low diffusion coefficient. Under poor fat suppression conditions the combination of both effects produces ghosting artifacts, which can produce an inadequate estimation of the ADC, due to the combination of fat and tissue signal in the same voxel (Fig. 1.6).

Different strategies to reduce the fat contribution in the final diffusion image were reviewed in Chap. 1. When performing DWI over large FOVs on a 1.5-T system, STIR may be more useful than other methods in achieving uniform fat suppression due to its reduced sensitivity to magnetic field inhomogeneities. Unfortunately, diffusion studies based on STIR sequence show low SNR, making it necessary to increase the number of averages to recover signal. For targeted examinations to specific organs or anatomic



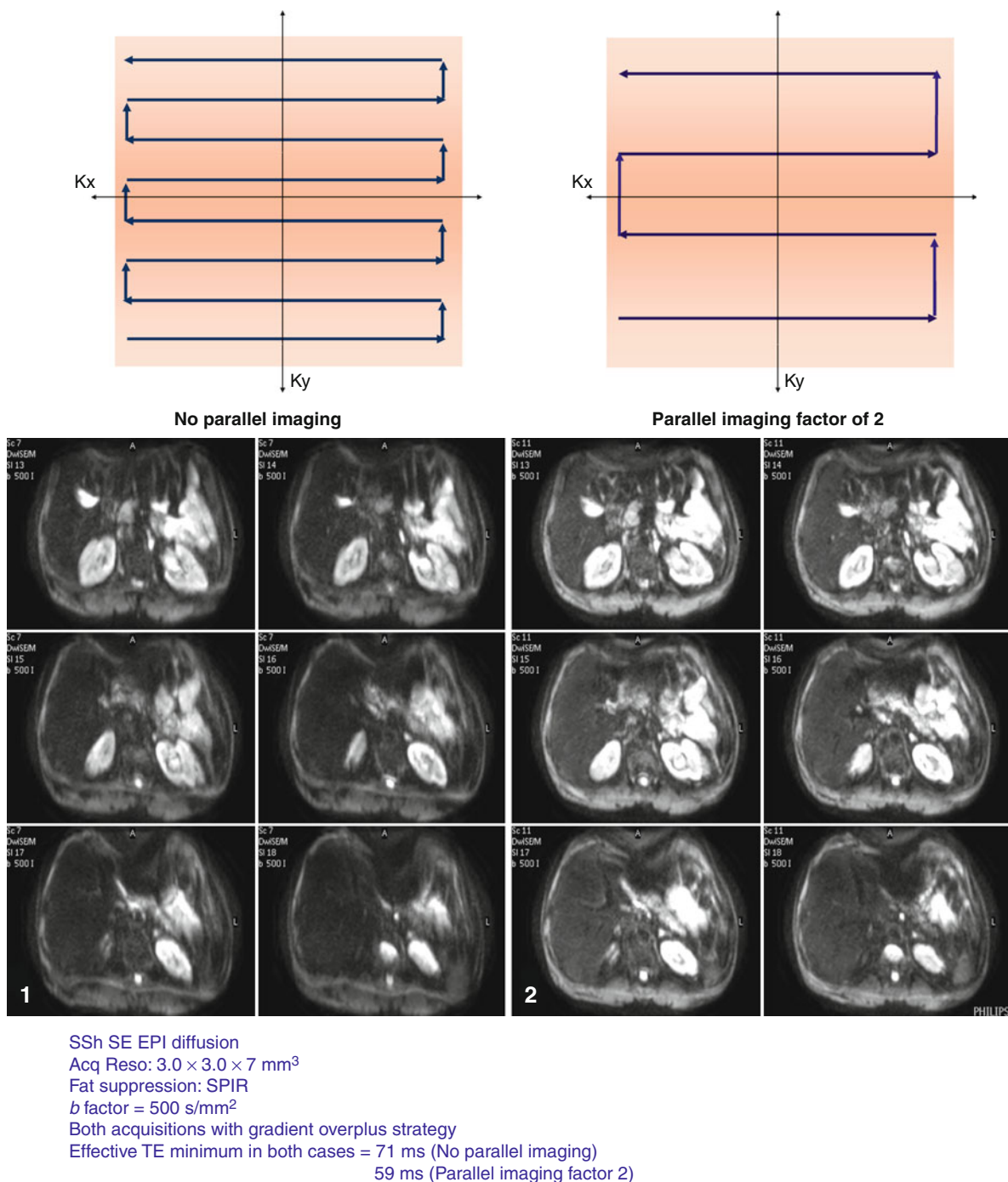
1

2

SSh SE EPI diffusion  
 Acq Reso:  $3.0 \times 3.0 \times 7 \text{ mm}^3$   
 Fat suppression: SPIR  
 Scan time = 44 s  
 Number of averages = 3  
 $b$  factor =  $500 \text{ s/mm}^2$   
 Both acquisitions with gradient overplus strategy  
 Effective minimum TE in both cases = 71 ms (Low bandwidth)  
 49 ms (High bandwidth)

**Fig. 2.4 Bandwidth: distortion artifacts.** Two sets of DWI of the liver in the same volunteer are shown. Series number 1 was acquired with a low bandwidth (1,286 Hz per pixel) and series number 2 with a high bandwidth (3,632 Hz per pixel). The images of the series acquired with low bandwidth show strong distortions, mainly in the anterior aspect of the liver, which were minimized using high bandwidth. The increase of the bandwidth also reduced the effective TE, which helps to compensate the loss in SNR, due to a higher noise contamination proper of higher frequencies. Although reducing the acquisition bandwidth improves the SNR, it can also affect the geometrical dis-

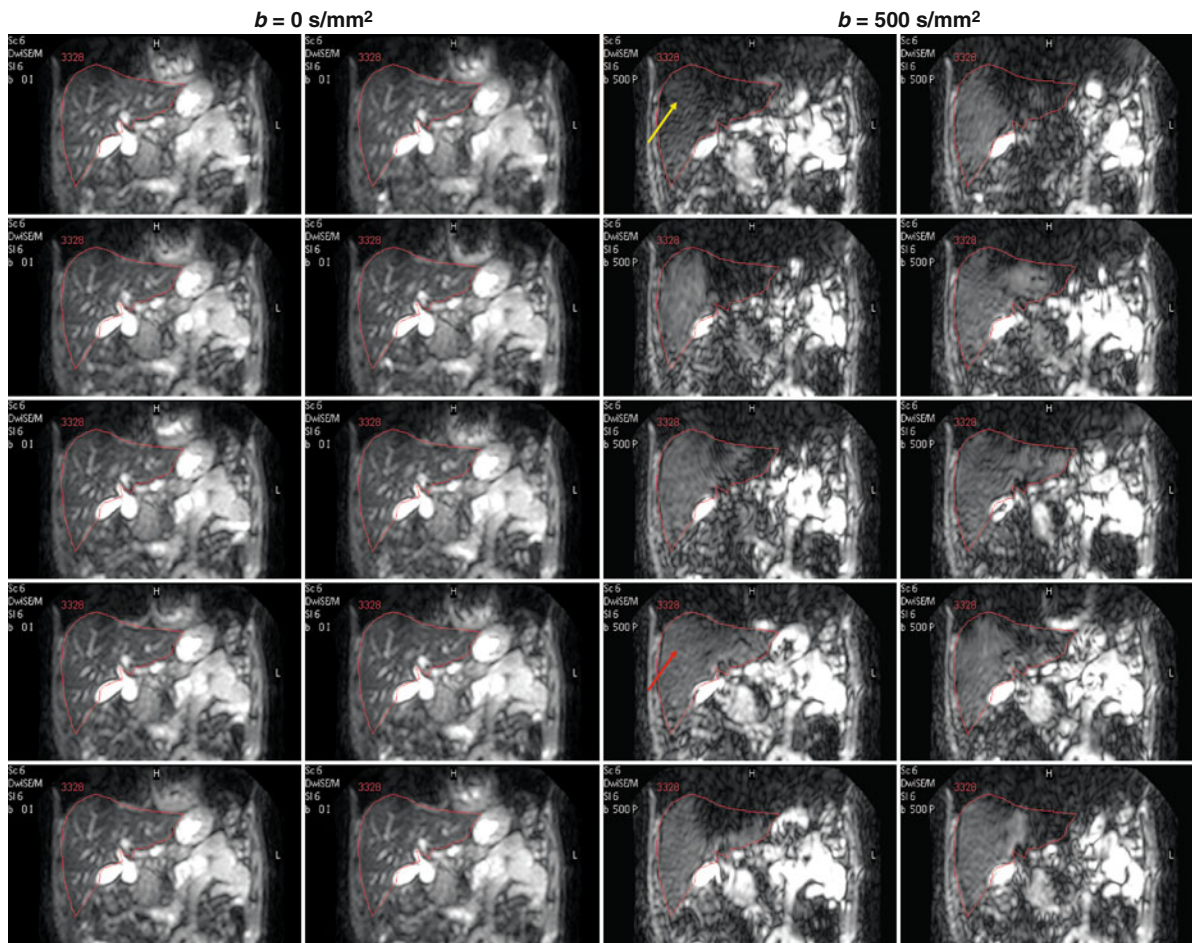
tortion of the images due to the EPI readout. Some distortion can be expected on DWI due to phase error accumulation during the EPI readout. From the acquisition point of view, these geometric distortions can be reduced by increasing readout bandwidths. Increasing the readout bandwidth has two effects: the frequency difference between two consecutive phase encoding lines is higher, making the phase error due to magnetic field inhomogeneities less important in the total readout; and a faster readout (equivalent to a higher bandwidth) leaves a shorter time to change the signal phase due to magnetic field inhomogeneities producing less image distortion.



**Fig. 2.5** Use of parallel imaging for distortion artifacts. Two series of a liver DWI in the same volunteer are shown. Series number 1 was acquired without parallel imaging and number 2 with a parallel factor of 2. This last sequence showed fewer artifacts in the anterior aspect of the liver, obtaining a more accurate geometrical representation of the studied anatomy, than the one without parallel imaging. As it was explained in Chap. 1, these

image acquisition strategies skip some phase encoding lines replacing those non-acquired lines using the spatial information of the sensitivity maps of surface phased array coils (see schemes in the superior part of the figure). To skip some lines during the acquisition means to reduce the phase error accumulation and the associated image distortion





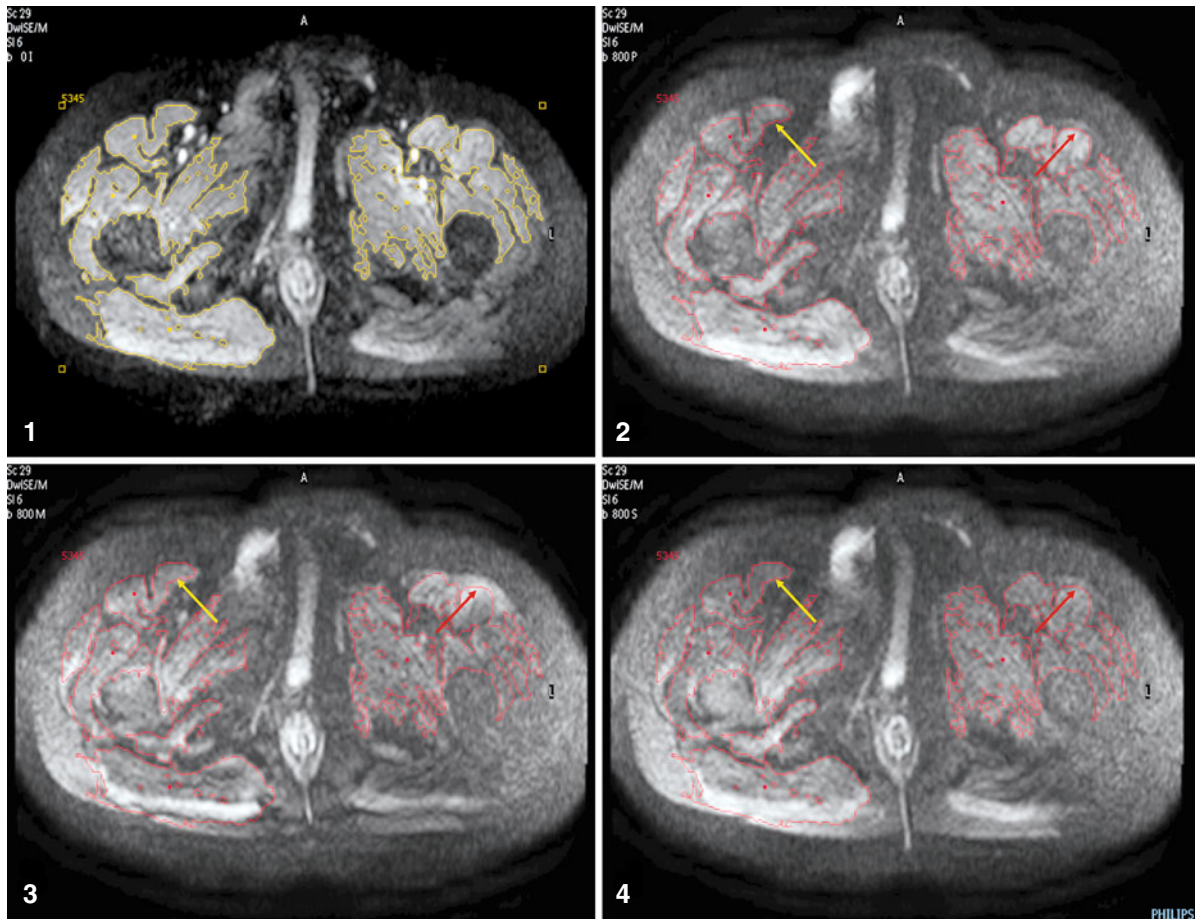
SSH SE EPI diffusion in coronal orientation  
 Acq Reso:  $4.0 \times 4.0 \times 10 \text{ mm}^3$   
 Fat suppression: SPAIR  
 Scan time per dynamic = 1,300 ms  
 Number of dynamics = 10  
 $b$  factor =  $500 \text{ s/mm}^2$  in foot-head direction

**Fig. 2.6 Heart motion effects on DWI.** In this figure, following the work of Kwee and colleagues, a dynamic DWI acquisition was performed in a volunteer under free breathing conditions. The acquisition included ten dynamics obtained in the coronal plane with two  $b$  values, that of 0 and  $500 \text{ s/mm}^2$  (series 1 and 2, respectively). A single diffusion direction was acquired in foot-head direction for better evaluation of the influence of the heart movement. All the dynamics of a central slice of the acquisition using a  $b$  value of  $0 \text{ s/mm}^2$  are shown in series number 1, and those acquired with  $b = 500 \text{ s/mm}^2$  in the

series number 2. In all dynamics of series number 1, images are equivalent. However, in series number 2, there are several artifacts in different dynamics. *Yellow arrow* points an area where respiratory and heart movements completely destroy the signal from the liver. *Red arrow* shows the dynamic with a better signal of the liver as it was acquired during expiration and heart diastole. All the other images with  $b = 500 \text{ s/mm}^2$  show different areas of signal loss due to heart movement. ROIs surrounding the shape of the liver in all dynamics were drawn for easier visualization of the changes in signal

regions, the use of spectral spatial fat saturation techniques (e.g., SPIR or SPAIR) can be advantageous. SPIR produces nice results in a reasonable scan time, especially, on 1.5T magnets, due to the use of  $120^\circ$  pulses in the suppression reducing the required inver-

sion time for zero cross of the fat signal. Conversely, on 3T systems, SPAIR technique has several advantages derived from the more homogenous excitation of the adiabatic pulses that reduce the effect of B1 inhomogeneities (dielectric or quadrupole artifacts).



**Fig. 2.7 Eddy currents artifacts.** Different acquisitions of a pelvic DWI study with a  $b$  value of  $800 \text{ s/mm}^2$  are shown, with the diffusion encoding in the phase, frequency, and slice direction (images 2, 3 and 4 respectively). Several ROIs were drawn in the  $b=0 \text{ s/mm}^2$  acquisition (image 1) and posteriorly overlaid

in the other acquisitions with different diffusion directions in order to simplify the evaluation of image distortion. *Red and yellow arrows mark* those regions where the diffusion images do not perfectly fit with the  $b=0$  image due to geometrical distortion produced by the eddy currents influence during the acquisition

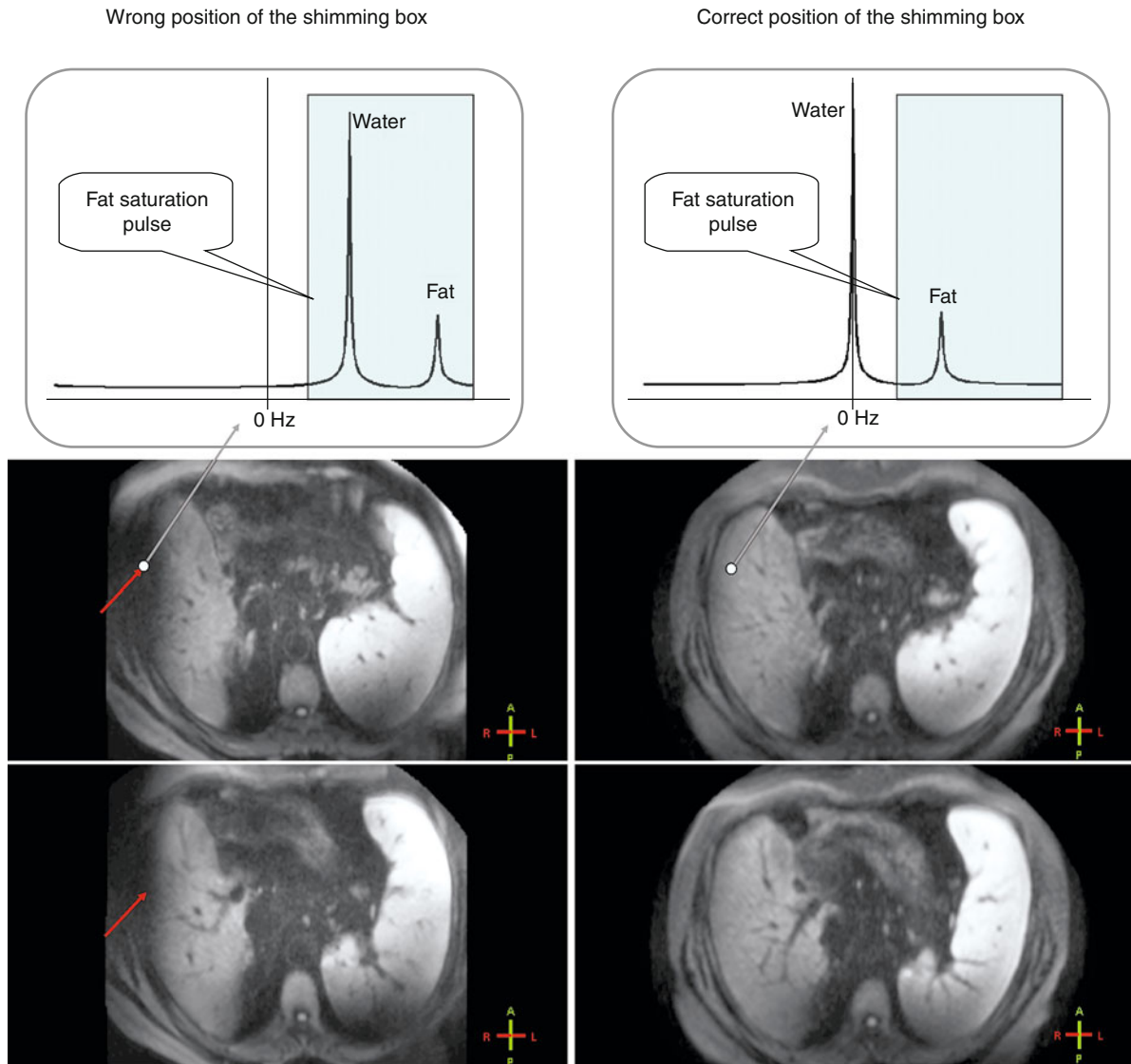
Unfortunately, the adiabatic pulses require a high inversion time as these pulses need to excite a flip angle of  $180^\circ$ . Besides, the SAR of these pulses is also higher than that of the normal excitation pulses, requiring a longer TR in the sequence. Nowadays, the parallel excitation technology (Multi-Transmit) can also provide a homogeneous B1 excitation, allowing a

uniform saturation using SPIR technique even in 3T systems.

Another challenge for spectral fat saturation is the magnetic field inhomogeneities especially in high magnetic fields. In order to compensate this difficulty, modern 3T systems are normally equipped with high order shimming (normally until second order) for

better compensation of magnetic field variation along the high FOV used in body applications. Under poor magnetic field homogeneity conditions, it is possible to obtain two different effects. The first one is to have

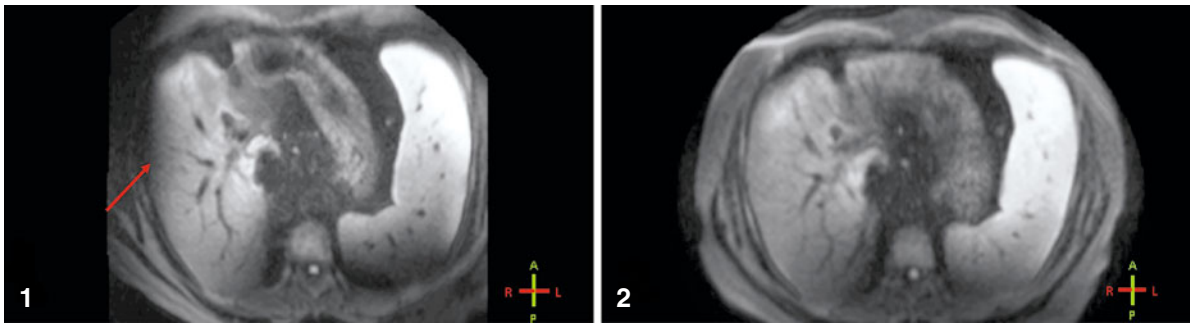
suboptimal fat signal suppression as it was shown in Fig. 1.6. The second effect is that some signal from the studied organ can become saturated losing information from those regions (Fig. 2.8).



**Fig. 2.8 Fat suppression artifacts: undesired tissue suppression.** Fat suppression techniques can be divided into spectral and nonspectral selective techniques. In those techniques like SPIR or SPAIR, a good shimming is required in order to saturate properly just the spectral region of the fat signal. In this sense, a reduced region is selected to improve the magnetic field shimming of the region of interest. This region has to take into account the whole anatomy of interest or otherwise, the saturation pulse can destroy part of the signal from the region of interest. This figure shows a liver DWI acquisi-

tion where the shimming box was placed excluding some part of the right lobe of the liver (*left column*). As a result, the spectrum from a pixel in the right part of the liver is shifted due to magnetic field inhomogeneity and the saturation pulse completely destroys the signal from that region. On the *right column*, the same DWI sequence, but with the shimming box properly placed, shows how the saturation pulse only destroys the fat signal, achieving a homogeneous fat suppression. The spectrum of the same pixel in this case shows how tissue signal is completely preserved





SSh SE EPI diffusion  
SENSE factor = 2  
Acq Reso:  $3.0 \times 3.0 \times 7.0 \text{ mm}^3$   
 $b$  values = 800 acquired with gradient overplus  
Respiratory triggered and SPIR fat suppression

**Fig. 2.8** (continued)

## 2.6 Dielectric Shielding Artifacts

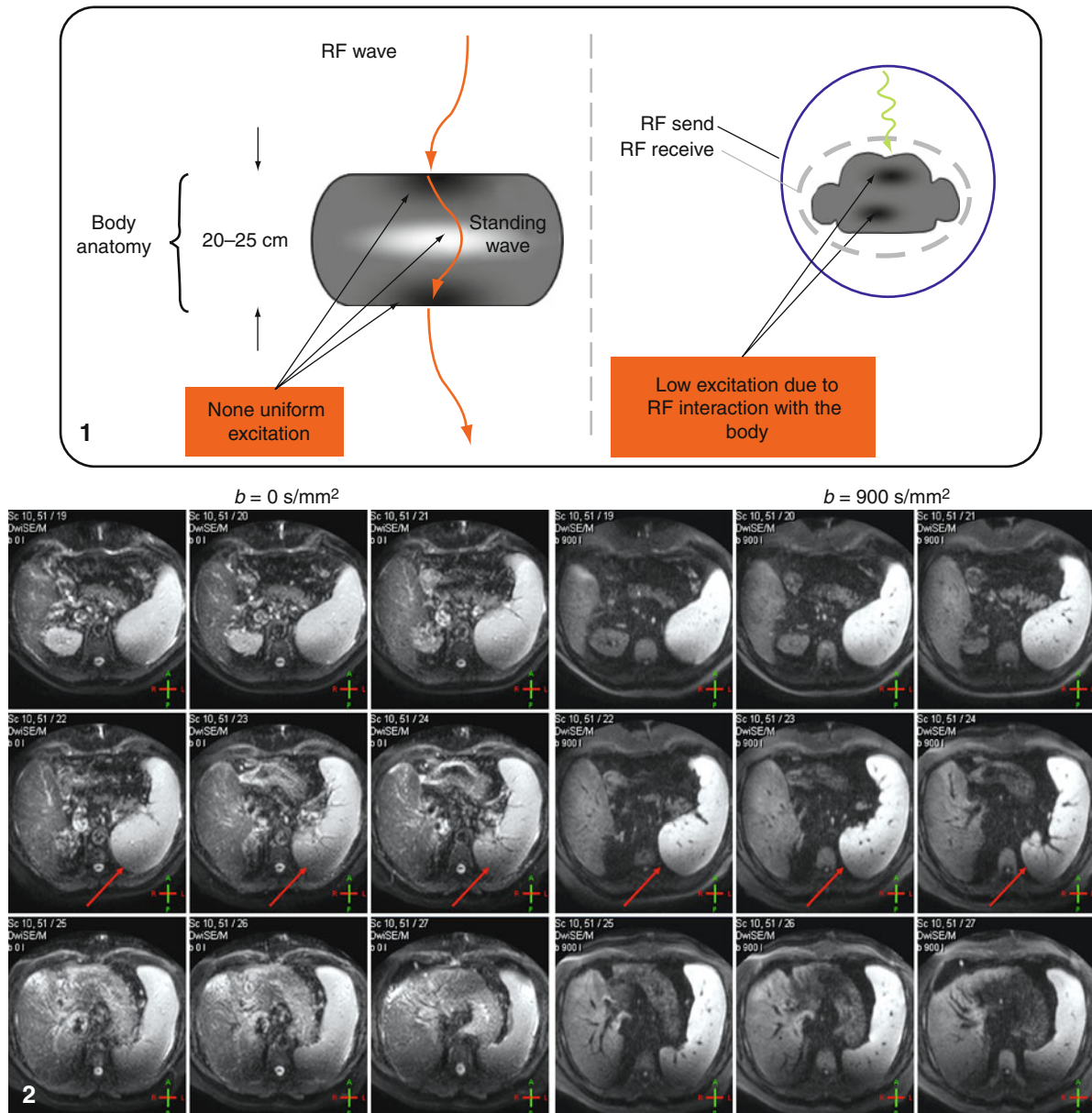
Although very high magnetic field scanners have several advantages, they also present some technical challenges to be overcome. As it was explained in the fat suppression section, there is an inherent artifact associated to the 3T systems called dielectric artifact (Fig. 2.9). This artifact produces a nonuniform excitation of the whole anatomy due to the interaction between the radiofrequency wavelength of 3T systems and the shape of the patient. Therefore, depending on the patient shape, some regions may not be completely excited producing a focal signal loss (Fig. 2.10).

## 2.7 Tips in DWI Sequence Design for Body Applications

The image contrast at DWI relies on intrinsic differences in the water diffusion among tissues. Scanning parameters must be optimized in order to increase SNR and contrast to noise ratio (CNR). As previously described, DWI is prone to motion and magnetic susceptibility artifacts since the majority of DWI are based on EPI sequences. As a general rule, conventional DWI has a limited spatial resolution. Therefore, it is important to find the optimum equilibrium between scan time and spatial resolution. In order to increase the DWI sequence quality, several rules should be followed, which are a short resume of what has been detailed in chapters 1 and 2:

- *Use fat suppression techniques:* The use of fat suppression allows to increasing the dynamic range of the DWI reducing the chemical shift–induced ghosting artifacts. Although inversion-recovery approaches such as STIR are useful for imaging large areas, the use of chemical fat selective saturation is more appropriate for smaller areas of interest due to their better SNR.
- *Minimize T1 saturation:* TR should be long enough to avoid T1 saturation effects, which can result in falsely low ADC values.
- *Use short TE:* This can be done by increasing the gradient intensity in the gradient lobes, increasing the bandwidth and using parallel imaging the bandwidth (up to a maximum of 1,500 MHz) and using parallel imaging.
- *Increase the number of acquisitions (NEX),* because the noise is disruptive and the signal is additive, although it is time consuming.
- *Decrease FOV* to a minimum in the phase encoding direction.
- *Do not increase the resolution in plane* to levels where the noise increases significantly or image quality decreases severely because it will decrease the quality of ADC maps. Enlarging the FOV may have a similar result.
- *Trace approach/gradient overplus:* The use of three orthogonal motion-probing gradients to produce a single diffusion direction allows us to improve the gradient strength by square root of three. Therefore, this approach reduces the effective TE, increases the SNR and minimizes susceptibility, EPI, or motion artifacts.

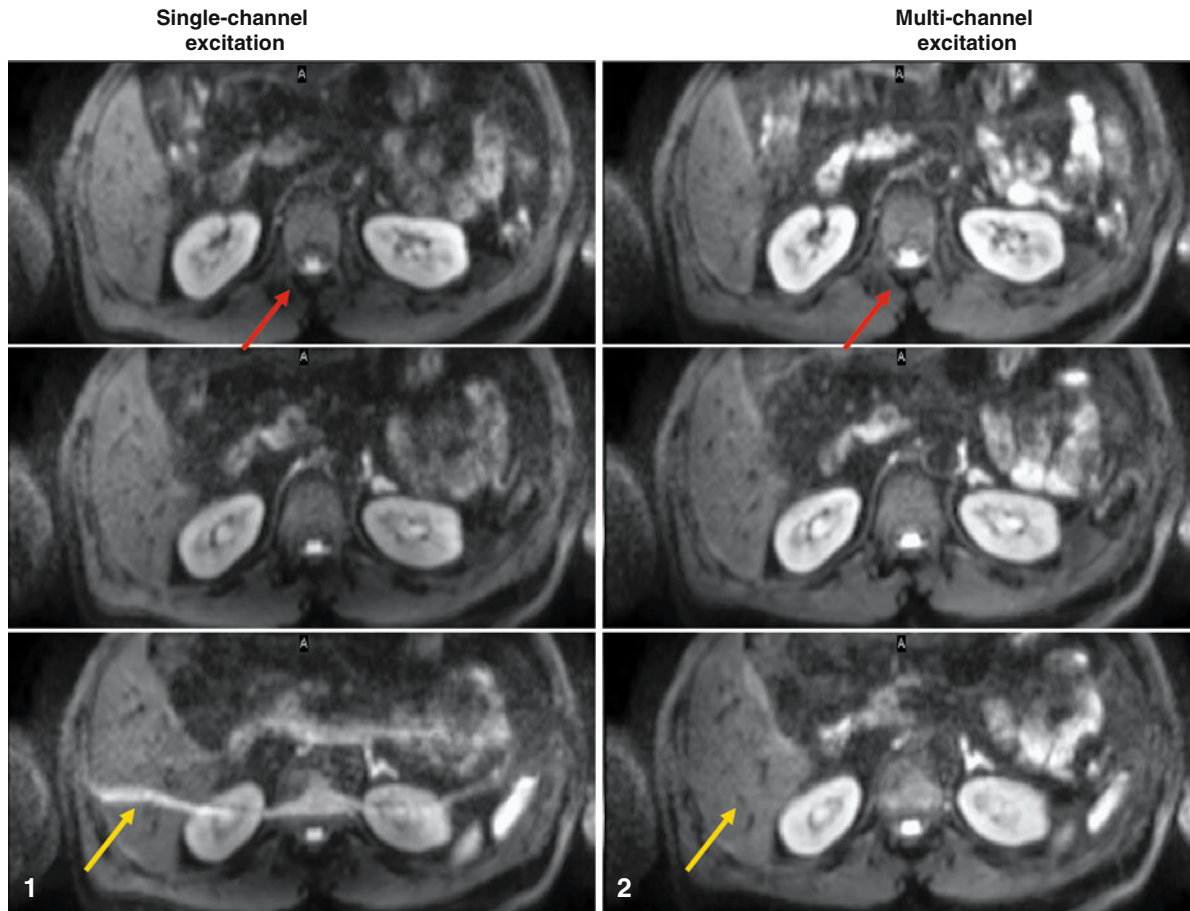




SSH SE EPI diffusion  
 SENSE factor = 2  
 Acq Reso:  $3.0 \times 3.0 \times 7.0 \text{ mm}^3$   
 Effective TE = 59 ms  
 $b$  values =  $0.900 \text{ s/mm}^2$  acquired with gradient overplus  
 Respiratory triggered

**Fig. 2.9 Dielectric artifacts.** Dielectric artifacts are typical of 3T magnets. The schemes in the first part of the figure summarize their origin. These artifacts produce a nonuniform excitation of the whole anatomy due to the interaction between the RF excitation and the shape of the studied region producing a standing wave that can interact in a constructive and destructive manner. These interactions produce a nonuniform excitation of the sample (2.9.1). This effect is particularly relevant in 3T systems where the RF wavelength in the body is around 25 cm that fits

with patient diameter. Therefore, depending on the patient shape, there are some regions that are not completely excited producing signal loss and other regions that are overexcited producing hot spots of signal. In the second part of this figure (2.9.2), a clinical example of a dielectric artifact in a liver DWI sequence is shown. Notice the signal loss in the spine region that reduces the signal in the spleen (red arrows) and in the posterior part of the liver, for both  $b$  values (0 and  $900 \text{ s/mm}^2$ ). This signal loss produces a lower SNR



SSh SE EPI diffusion  
 Acq Reso:  $2.6 \times 2.6 \times 6 \text{ mm}^3$   
 Fat suppression: SPIR  
 Scan time = 20 s  
 Number of averages = 3  
 $b$  factor =  $500 \text{ s/mm}^2$   
 Both acquisitions with gradient overplus strategy  
 Effective TE minimum in all cases = 60 ms (Parallel imaging factor 2.0)

**Fig. 2.10 Single channel versus multichannel excitation.** In order to compensate the nonhomogeneous excitation of the entire FOV due to dielectric artifacts in high magnetic fields, it is necessary to look for new excitation strategies that allow a better RF distribution. The best way to ensure a more homogeneous excitation is to share the excitation between different RF excitation coils that can drive completely independent RF pulses (different amplitude, phase, frequency, and waveform) that allow an accurate excitation over the whole FOV, independently of the patient anatomy. Nowadays, there are 3T systems that allow excitation with completely independent RF excitation sources as well as patient adaptive strategies that ensure a homogeneous excitation over the whole FOV independently of the

patient shape. This figure shows the results of two single breath-hold DWI acquisitions of the same patient using a single-channel (2.10.1) or multichannel acquisition strategies (2.10.2). Both images were acquired with the same acquisition parameters and displayed with the same window level and width for comparison. In these images, *red arrows* showed a dark signal region in the spine in the single-channel excitation acquisition (2.10.1) while a more homogeneous excitation is appreciated in the whole FOV for multichannel excitation (2.10.2). Finally, *yellow arrows* showed some fat artifacts in the single-channel excitation acquisition, that were not present in the multichannel acquisition, due to wrong excitation in the SPIR fat suppression

- *SNR may be increased* by using higher field strength (3T magnets), reducing TE, applying higher gradient power, using a short EPI train, and using phase-array coils with more number of elements.

## Further Reading

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Diffusion MRI Outside the Brain

A Case-Based Review and Clinical Applications

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