2.1 Introduction

Over the past decade, DW-MRI has been a great success in neurological imaging, particularly for the assessment of acute cerebral vascular events (Bammer et al. 2001; Marks et al. 2008) and in mapping the anatomical cerebral pathways using DTI (Chen et al. 2001; Bammer et al. 2002). DW-MRI has become practical in extracranial applications following the introduction of parallel imaging techniques in the...
late 1990s (Pruessmann et al. 1999; Blaimer et al. 2004; Larkman et al. 2007) and the continued improvements in MR hardware. The use of parallel imaging in combination with diffusion-weighted single-shot echo-planar imaging (EPI) has overcome many of the challenges that have limited its earlier implementation in the body. Although using EPI DW-MRI acquisition and enhanced hardware advances have enabled high-quality DW-MR images in the body to be acquired within a reasonable time frame, a number of challenges still remain.

As is well known, EPI is highly sensitive to static magnetic field inhomogeneity, chemical shift artefacts and eddy currents resulting from the application of the diffusion encoding gradients (Le Bihan et al. 2006). Physiological motions within the body are an additional challenge and the number and range of diffusion weightings (b-values) used within a measurement also require consideration for DW-MRI in the body, which could be tailored for disease evaluation at specific anatomical sites. Methods to overcome, or at least reduce, the impact of a range of artefacts in DW-MRI (e.g. geometric distortion, chemical shift, N/2 ghosting, susceptibility artefacts, G-noise) are required to ensure consistent high-quality images. Many post-processing methods for correcting and eliminating DW-MRI artefacts have been developed for neurological applications (Le Bihan et al. 2006). However, many of these techniques have failed to deal with the challenges in body DW-MRI.

In this chapter, we discuss the factors that affect the image quality of echo-planar DW-MRI acquisitions. An imaging optimization strategy that can be conducted at any institution to optimize the MR data acquisition to an acceptable level of image artefact and image signal-to-noise is presented. As the process of imaging optimization can be quite complex, it is recommended that this should be done in collaboration with an experienced physicist or vendor application specialist. Clearly, improvements in image quality can also be achieved through post-processing of the DW-MRI data (Bodammer et al. 2004). However, in clinical practice, DW-MR images enter the radiological workflow immediately after acquisition and therefore acquisition schemes should be as robust as possible. Image post-processing can be very useful but these techniques are largely being developed and used on a research basis. However, such developments may be integrated onto future image-viewing platforms for wider access.

Imaging phantoms are useful devices that can help to test and improve the image quality from DW-MRI acquisition protocols on clinical MR scanners and we will describe how these could be applied for this purpose. The key factors that affect image signal-to-noise and image artefacts using single-shot echo-planar DW-MRI acquisitions, are discussed alongside how these may be optimized. Breath-hold, free-breathing and physiologically gated DW-MR imaging techniques in the body are reviewed, highlighting the advantages and disadvantages of each method.

2.2 Factors that Affect Image Quality Using EPI DW-MRI Acquisition

There are many factors that affect image quality in an EPI DW-MRI image acquisition scheme. These can be divided into factors that influence image signal-to-noise and artefacts.

2.2.1 Image Signal-to-Noise (SNR)

DW-MR images are inherently noisy because of the EPI technique and range of signal attenuating gradients used for image acquisition, and hence the image signal-to-noise should be optimized. The following factors can be adjusted to improve image signal-to-noise: matrix size (smaller), echo-time (shorter), number of signal average (higher), section thickness (increase), receiver bandwidth (optimized), parallel imaging (optimized) and choice of fat-suppression scheme.

2.2.2 Image Artefacts

These include motion, chemical shift artefacts, eddy currents effects, N/2 ghosting, susceptibility artefacts and G-noise. Motion can cause image blurring, spin dephasing and signal loss on DW-MRI. Chemical shift artefact can result from the misregistration of the fat and water signal when the protons imaged in the same voxel result in a “banding” artefact. Eddy currents (residual gradient fields) are formed in the presence of a changing magnetic field, i.e. when gradients are rapidly turned on and off as in EPI. This results in geometric distortion and misaligned images with the appearance of “shearing”, “translational” and “scaling” (magnification) artefacts. N/2 ghosting
artefact results from phase errors due to a phase difference between each echo acquisition during read-out, and is seen as a repeated image "ghost" throughout the image in the phase-encode direction. Susceptibility artefact occurs due to T2' relaxation as a result of inhomogeneities in the main magnetic field, which is accentuated in EPI because of the small bandwidth (BW) per pixel. Last but not least, G-noise results from the use of parallel imaging, which results from noise already inherent in the data being amplified during parallel imaging reconstruction.

2.3 Bulk Diffusion Phantoms

Appropriate phantoms are useful for DW-MRI optimization as they enable quantitative assessment of artefacts and distortions. Phantoms are also being widely applied for evaluating the accuracy of the diffusion encoding b-factor on clinical and research MR scanners.

A range of materials including water, alkanes, sucrose and copper sulphate solutions, are suitable for use as diffusion phantoms on clinical MR systems (Tofts et al. 2000; Padhani et al. 2009). At our institution, we have made extensive use of different phantoms containing polydimethylsiloxane (PDMS), ice water and sucrose solutions to evaluate the performance of MR systems.

PDMS is a viscoelastic material that can be considered non-diffusing because of its high molecular weight and viscosity (Price 1998). This enables DW-MRI protocols to be evaluated using a phantom containing PDMS in a quantitative manner without having to account for the effects of diffusion. Water is non-toxic and readily available but its diffusion properties are temperature-dependent, its diffusion rates being higher than in tissues at body temperature and it also has a longer relaxation time compared with tissues. One way of minimizing variations is to use ice water that is held at a constant temperature of about 0°C such that the diffusion measurement data are constant (insensitive to the ambient temperatures in the scanner) over repeat measurements (Padhani et al. 2009). In addition, the diffusivity of water at 0°C is about 1.12 × 10^-3 mm^2/s, which is in the range found in human tissues. For these reasons, a phantom made out of ice water is highly suitable for quality assurance and b-value calibrations. Another material that can be used to measure diffusivity is sucrose, which is more viscous than water. Sucrose phantoms are highly stable and can be manufactured, depending on their solution concentration, to obtain a wide range of diffusivities and are thus also suitable for quality-assurance procedures. Sucrose phantoms are usually scanned at room temperature. However, sucrose phantoms are temperature-sensitive and the results may thus differ if the ambient room temperature varies between measurements.

2.3.1 Optimizing Image Quality Using a Phantom

EPI data acquisitions are highly sensitive to static field inhomogeneity, causing geometric distortions in the phase-encoding direction. These distortions are proportional to the ratio of the field of view and the image acquisition receiver BW. This relationship would suggest that the DW-MR images should be acquired with the minimum field of view and the highest BW possible. However, in practice it is found that increasing the imaging BW (reducing the inter-echo spacing of the EPI readout) introduces and increases an artefact known as N/2 ghosting (formally Nyquist ghosting), as demonstrated in Fig. 2.1. N/2 ghosts arise from a number of factors including gradient switching timing errors, residual eddy currents which are b-value-dependent (increases with b-values) and non-linearity in the low pass filters. These factors are highly scanner-dependent and hence need to be individualized for each scanner and acquisition protocol used, as factors such as imaging field of view, the choice of b-values and the receiver BW all contribute to the magnitude of both the geometric distortion and N/2 ghosting.

In practice the magnitude of these effects can be quantified using a suitable phantom. PDMS is well suited for these evaluations as subtraction images obtained from a number of images obtained with and without diffusion weighting can be used to establish the magnitude of N/2 ghosts and eddy current induced geometric distortions over a range of different experimental conditions including alterations in the receiver BWs (Fig. 2.1). One possible optimization strategy would be to establish a balance between the magnitude of N/2 ghosting and the geometric distortions. To do this, a number of measurements are performed with a fixed b-value (typically the largest b-value employed in the measurement protocol) on the PDMS phantom with varying BWs, whilst measuring the magnitude of the distortions by recording...
displacements in the difference images of the phantom. Figure 2.1 illustrates that as the BW is increased, the magnitude of the displacements observed on the subtraction images is reduced. Measurements in a fixed region of interest outside of the phantom can be used to measure the magnitude of the N/2 ghosting. In general, a readout BW value can be determined, which reduces the magnitude of the distortion to an acceptable level of ±1 px with a ghosting magnitude of less than 10%. If this cannot be achieved, it may be necessary to reduce the magnitude of the $b$-value until this is attained. This process can be repeated by independently applying the $b$-factor in all three orthogonal magnet axes, to determine whether there is any directional dependency in the gradient performance. It is not uncommon for one gradient direction to perform less well compared to the other encoding directions, and this axis may thus limit the maximum $b$-factors that can be employed in an imaging protocol using orthogonal diffusion gradient encoding. A similar optimization process can be performed for other diffusion gradient encoding methods (e.g., tetrahedral encoding or three-scan trace approach), parallel imaging factors or the types of parallel imaging applied (Fig. 2.2).

A major advantage of applying parallel imaging to single-shot DW-MRI is the reduction in the echo-train length of the EPI readout. This in turn reduces the number of phase encoding steps which brings substantial benefits to the DW-MRI measurement as it reduces the magnitude of ghosting in the phase encoding direction and enables a reduction in the minimum echo-time achievable by the measurement. This, in turn, helps to maintain image quality and image signal-to-noise. The advantages of applying parallel imaging (e.g., using GeneRalized Autocalibrating Partially Parallel Acquisitions or GRAPPA) are shown in Fig. 2.2. We have evaluated a number of parallel imaging

Fig. 2.1. Top: $b = 0$ s/mm$^2$ ($b_0$) images of a polydimethylsiloxane (PDMS) phantom taken at different readout bandwidths (BWs) (displayed at bottom in Hertz per pixel, Hz/px) using a Stejskal–Tanner DW-MRI pulse sequence. Middle: Difference of $b_0$ and $b = 1000$ s/mm$^2$ ($b_{1000}$) images at the respective BWs. Bottom: Background $b_0$ images of the phantom. It is clear from the difference images that distortions due to eddy currents at high $b$-values are reduced at higher BWs but N/2 ghosting on the background images also increases as BW increases.
imaging techniques (Sensitivity Encoding SENSE, mSENSE and GRAPPA) employed in DW-MRI and all methods have demonstrated improvements in the DW-MR image quality.

In clinical studies, it is usually advantageous to apply reduced phased encoding steps in the direction requiring the least volume coverage. For example, in axial acquisition of the body, this would be in the anterior–posterior (AP) direction. In general, a successful strategy to high-quality DW-MRI using EPI technique is to reduce both the echo-time and the number of phase-encoding steps to the minimum for the clinical studies. However, it must also be remembered that parallel imaging itself does reduce the signal-to-noise by a factor proportional to $\sqrt{\text{parallel imaging acceleration factor}}$ and also changes the distribution of both noise and ghosting artefacts within the image. However, as the application of parallel imaging allows shortening of the echo-time and echo-train length, these help to improve the overall image quality. Nevertheless, a poor choice of imaging field of view and parallel imaging acceleration factor can result in additional noise and artefacts within the image. Detailed discussions of the various implementations of parallel imaging are beyond the scope of this chapter although the interested reader is directed to an excellent review by BLAIMER et al. 2004. It is important to mention that when applying parallel imaging techniques (e.g. GRAPPA) that allow the user to specify the number of reference scan lines, this should be checked to establish that sufficient reference lines are specified to minimize image ghosting.

A number of measurement-specific methods have been developed to reduce the effect of eddy-current-induced distortions, the most effective of which have been the following: (a) simultaneous application of diffusion encoding gradients in more than one direction, implemented on clinical scanners as tetrahedral encoding or three-scan trace approaches and (b) the use of the twice refocused spin-echo with diffusion weighting (ALEXANDER et al. 1997; KOCH et al. 2000; REESE et al. 2003).

The application of simultaneous diffusion-weighted encoding gradients results in a reduction in both the magnitude and rate of change of diffusion encoding gradients as the vector sum of the applied gradients creates the required $b$-value. This can significantly improve image quality by reducing the magnitude of eddy currents. An additional benefit of using a simultaneous gradient application scheme is the ability to reduce the measurement echo-time to a minimum. The minimum echo-time achievable at clinical DW-MRI may be proportional to the magnitude of the diffusion-weighting $b$-value applied, as most gradient

---

**Fig. 2.2.** Top: $b = 0 \text{s/mm}^2$ ($b_0$) images of a PDMS phantom acquired using a readout BW of 2,416 Hz/pixel and utilizing parallel imaging technique known as generalized autocalibrating partially parallel acquisitions (GRAPPA) at various acceleration factors. Bottom: Background $b_0$ images of the phantom. It can be seen that using GRAPPA reduces the magnitude of $N/2$ ghosting observed on the background image. There does not seem to be any additional advantage in this case of using acceleration factors greater than 2 and it can be seen that the signal-to-noise and noise heterogeneity worsen when acceleration factors of 3 or 4 are used.
systems operate at the maximum possible slew rate. Thus, an increase in \( b \)-values may necessitate increasing either the duration of the applied gradient or the time delay between diffusion encoding gradients; both of these will lead to longer echo times (Kingsley 2006). A reduction in the echo-time at DW-MRI can be highly advantageous in tissues with short T2 relaxation times such as the liver by maintaining good image signal-to-noise.

The twice-refocused spin-echo technique is very effective in cancelling the effects of eddy currents (Figs. 2.3 and 2.4). Briefly, the diffusion encoding gradients normally employed in a standard Stejskal-Tanner DW-MRI measurement scheme are split by the introduction of a 180° refocusing pulse. In the twice-refocused spin echo implementation, the diffusion encoding gradients are applied in reverse polarity around each RF pulse thereby cancelling the eddy currents which are set up by the diffusion encoding gradients applied in the opposite direction. The net result of the application of these four diffusion-encoding gradients is substantial reduction of eddy currents during the EPI readout (Alexander et al. 1997; Koch et al. 2000; Reese et al. 2003). The twice-refocused spin-echo method can be combined with the three-scan trace or tetrahedral diffusion-weighted encoding scheme to further reduce the effects of eddy currents in the measurement. By using such a combination of approaches, diffusion-weighting factors employing \( b \)-values >1,000 s/mm\(^2\) can be applied over relatively large field of view for DW-MRI studies with minimal geometric distortion. In our experience, a maximum pixel displacement of 2 mm or less can be achieved when imaging is performed using a relatively large field of view (400 mm) with a well-optimized imaging protocol.

As the information gained from phantom studies can be very helpful towards the understanding and optimization of a DW-MRI technique, we have found it advantageous to initially test new imaging protocols on phantoms prior to volunteer or patient studies.

### 2.4 Optimizing Image Quality on Volunteer or Patient Studies

Phantom studies are useful to check diffusion gradient performance on MR scanners and to quantify image signal-to-noise, image ghosting and geometric...
distortion. However, there are other factors that come into play that have a major effect on the quality of DW-MRI studies that cannot be adequately or reliably addressed using a phantom. For these, volunteer and/or patient studies are required.

2.4.1 Fat Suppression

Eliminating signals from the methylene resonances in fat is an essential requirement in DW-MRI body applications. Fat signals produce significant artefacts in EPI in general but can be particularly pronounced in EPI DW-MRI as a result of a lower rate of signal attenuation in fat on DW-MR imaging compared to other tissues. Signal arising from protons associated with fat typically has a resonant frequency difference of 220 Hz from protons associated with water at 1.5 T and significantly lower diffusion rates than tissue water. Water and fat signals arising from the same pixel location will appear within the same pixel provided that the pixel BW is greater than the fat–water chemical shift of 220 Hz. In EPI DW-MRI experiments, the BW per pixel in the phase-encoding direction is substantially less than the chemical shift difference of fat and water: the typical pixel BWs for single-shot EPI are 10–30 Hz in the phase-encoding direction. As a result, unsuppressed fat signals in EPI DW-MRI appear as high signal intensity ghosting through the image. Ghosting artefacts arising from fat–water chemical shift can be displaced by 10–15 pixels from their source location. The typical pixel dimensions for DW-MRI in the body are 2–4 mm, and the chemical shift ghosting artefact observed can therefore be distant from the source of the artefact by distances ranging from 2 to 6 cm. These chemical shift artefacts can propagate further ghosting artefacts as a result of N/2 induced ghosting. It is therefore of great importance to evaluate and optimize

Fig. 2.4. Top: \( b = 0 \text{s/mm}^2 \) (\( b_0 \)) images of a PDMS phantom acquired using both DSE and parallel imaging GRAPPA at acceleration factor 2. Middle: Difference of \( b_0 \) and \( b = 1,000 \text{s/mm}^2 \) (\( b_{1000} \)) images. Bottom: Background \( b_0 \) images of the phantom. Using both GRAPPA and DSE techniques, both N/2 ghosting and eddy current distortion artefacts have been reduced over the range of imaging BWs (displayed at bottom in Hertz per pixel, Hz/px)
the performance of fat suppression techniques for DW-MRI applications in the body.

It is a great credit to the equipment designers and vendors that modern clinical MR machines are able to suppress fat signals very effectively over very large fields of view. Nevertheless, it is still essential that the quality of fat suppression is checked and optimized on a per-protocol and per-scanner basis. The choice of fat suppression technique used in part depends on the ability of the scanner to optimize the B0 field homogeneity over relatively large fields of view. This in turn, is influenced by the magnitude of the static field strength, the presence or absence of adjustable higher-order shimming currents and the shimming technique employed. Other factors that are important include the BW of the fat suppression pulses, as well as the accuracy and efficiency of the inversion pulses. Fat suppression techniques that are routinely employed on clinical scanners include water only excitation, chemical fat suppression (frequency selective fat excitation and magnetization spoiling), STIR (short-tau inversion recovery) and SPAIR (spectral selection attenuated inversion recovery). STIR and SPAIR are inversion-recovery-based techniques.

In our experience, the SPAIR method works well in nearly all circumstances across different vendors’ platforms except when the field of view is very large (>25 cm in major magnet axis) or when there is a significant transition in the normal anatomy (e.g. at the interphase of the shoulders/neck and pelvis/legs). In these cases, conventional inversion recovery fat suppression techniques are recommended, and may be preferred in whole-body DW-MRI examinations. For whole-body DW-MRI, a further advantage of using the inversion recovery techniques is that imaging of multiple image stations can be conducted at the same resonant frequency following initial station specific shimming, which helps to overcome station-to-station misregistrations that frequently occur in whole-body DW-MRI. A 20–30 Hz difference of the water resonance following shimming at each imaging station for whole-body DW-MRI can result in a geometric shift of at least one or two pixels in the phase-encoding direction. Resonant frequency shifts of 1–2 ppm following shimming are not uncommon particularly in the head and neck region. However, the disadvantages of using inversion recovery prepared DW-MRI protocols are the increased measurement time, altered T1 contrast on the b0 image and reduced signal-to-noise as the full longitudinal magnetization is no longer available.

2.4.2 Single-Shot Acquisition vs. Multiple Signals Averaging

Clearly, a major problem with imaging in the body is physiological motion. The major advantage of single-shot DW-MRI is that the image data is acquired within 60–100 ms, thereby effectively freezing motion in any given image. The challenge to performing DW-MRI in the body is to acquire DW-MR images of different b-values sufficient for ADC quantification from the same image location within a single measurement time. There are three acquisition methods that are in use to address this challenge.

The first is to acquire all the DW-MRI data required for an ADC calculation in a breath-hold using a single-shot EPI technique. Clearly this limits the number of b-values and/or the number of signal averages that can be accommodated in a breath-hold for ADC calculation and the anatomical coverage that can be achieved. A significant limitation is that the DW-MRI data are inherently noisy and the signal-to-noise is reduced exponentially with increasing b-values. Breath-hold data are therefore signal-to-noise-limited and the ADC estimates will have larger uncertainty.

The second method is to acquire DW-MRI data with multiple averaging with some form of physiological triggering. For respiratory motion, a navigator is typically used at the diaphragm, and this may be combined with cardiac triggering. The two main problems with this approach are poor measurement efficiency and that the navigator does not entirely eliminate motion-induced artefacts as it normally has an acceptance window of the order of the slice thickness. In addition, the liver is an elastic organ and will not necessarily deform in a consistent manner during the course of the DW-MRI measurement. Despite these issues, navigator-controlled DW-MRI with multiple averaging is an effective and robust technique when well-implemented.

The final option is to acquire DW-MRI data in free-breathing with multiple b-values and averages. The advantage of this approach is the high efficiency of data acquisition and increased coverage that can be achieved in a similar imaging time compared with DW-MRI measurements with physiological gating. DW-MR images are averaged in image space where only the magnitude of the image data is used. Motions occurring during imaging results in a phase shift in the complex image data and by discarding the phase information during averaging, the effects of motion on the resultant image are reduced. The important
question is how much averaged data is required to obtain an ADC estimate with comparable accuracy and repeatability as a breath-hold study. Based on our observations, a factor of six provides similar ADC estimates with improved repeatability over breath-hold studies. Suggested clinical DW-MRI protocols are presented in Table 2.1.

2.5 Optimization Strategies for the Selection of \( b \)-Values

While DW-MRI is being applied in the body to evaluate non-oncological and oncological conditions, it is the oncological application of the technique that is being widely investigated. In particular, there is growing interest in the development of DW-MRI as a potential biomarker for the assessment of tumour response to treatment.

When developing DW-MRI protocols for evaluating cancers, it is important to consider the choice of \( b \)-values so that the resulting images provide the most accurate information. The optimization processes depend in part on whether DW-MRI is used as a qualitative tool to provide image contrast between diseased and normal tissue, or the technique is applied primarily to quantify the diffusivity of the tissue by means of calculating the ADC. Hence, we will discuss \( b \)-value optimization along two lines: quantitative (ADC maps) and qualitative (anatomical diffusion-weighted images). This section briefly outlines the optimization strategies that may be applied in each case, taking into account the fundamental considerations. We will also illustrate these with clinical examples.

In all cases, a clear knowledge of the relationship between image signal intensity and \( b \)-value (Stejskal et al. 1965) is essential:

\[
S(b) = S(0)e^{-bD}.
\]  

(2.1)

When DW-MRI is performed at a particular \( b \)-value, the relation of the measured signal intensity \( S(b) \) with other parameters is as shown in the equation where \( D \) is the apparent diffusion coefficient of the tissue (\( \text{mm}^2/\text{s} \)) and \( b \) is the \( b \)-value (\( \text{s/mm}^2 \)). \( S(0) \) is the signal intensity of the tissue when \( b = 0 \text{s/mm}^2 \). The equation has two unknown quantities, namely \( D \) and \( S(0) \), such that at least two DW-MRI measurements using different \( b \)-values will be required to estimate \( D \).

2.5.1 Quantitative Data

Statistical noise is an inherent characteristic in all MR images and efforts should be made to reduce its presence as far as possible. In the calculation of ADC maps, image noise causes errors in the ADC estimates and so it is important to choose \( b \)-values which will minimize the effects of noise. A simple approach is to use error propagation to derive an expression for the standard deviation in ADC calculations from \( N \) independent \( b \)-values (Bito et al. 1995);

\[
\sigma_D = \frac{D_0}{\text{SNR}_b} \times \sqrt{\frac{\sum_n (b_n N - \sum_n b_n)^2 e^{2b_n D_0}}{N \sum_n b_n^2 - (\sum_n b_n)^2}}.
\]  

(2.2)

Here, \( D_0 \) represents the true ADC of the tissue and \( \text{SNR}_b \) is the signal-to-noise ratio of a single \( b = 0 \text{s/mm}^2 \) image. Minimization of this function with respect to \( b_n \) then provides the optimum set of \( b \)-values. It may be shown (Bito et al. 1995) that for the case of a single \( D_0 \), the ideal sequence consists of \( n \) image acquisitions at \( b = 0 \) and \( m \) acquisitions at some other \( b \)-value which may be found by numerically solving;

\[
e^{2b_n D_0} (bD_0 - 1) = \frac{m}{n} (m + n = N).
\]  

(2.3)

Furthermore, if the function is optimized with respect to \( m/n \) then;

\[
\frac{m}{n} = 3 \quad \text{and} \quad bD_0 = 1.25.
\]  

(2.4)

An important consideration is that prior knowledge of the \( D_0 \) values is required in all calculations. Such information could be obtained from previous reports or if direct measurement is needed then a range of \( b \)-values (e.g. \( 0-1,000 \text{s/mm}^2 \)) should be used so that at least one is near optimum. When a distribution of \( D_0 \) is encountered, the mean value should provide a reasonable approximation.

To demonstrate the processes involved in this optimization scheme consider the test case shown in Fig. 2.5. A patient with liver metastases from colorectal cancer has been scanned using six \( b \)-values from 0 to 750 \text{s/mm}^2 and regions of interest have been hand-drawn around four index lesions, each measuring \( >2 \text{cm} \) in diameter, on the \( b-750 \text{s/mm}^2 \) images. Using all of the \( b \)-values, the ADC map was generated.
and voxel-wise ADC values from all ROIs were exported for analysis in a statistics software package. The mean ADC value from all voxel ADC values was calculated to be $1.54 \times 10^{-3}$ mm$^2$/s suggesting that the optimum single $b$-value for use in this study was:

$$b = \frac{1.25}{D_0} \Rightarrow b = 814 \text{ s/mm}^2$$

In practice, however, it would be rare for a scanner to achieve any arbitrary $b$-value and so this value should be rounded to the nearest whole value possible. As an example we would say that $b$-values of 800 or 850 s/mm$^2$ would be optimal in this test case. It is noticed in Fig. 2.1 that one of the lesions (outlined in red) is necrotic and has a higher ADC than the others. The implications of this are that the mean ADC is increased when including the necrotic tumour and so the optimum $b$-value calculated may be too low for the other non-necrotic metastases (which have a mean ADC in this example of $1.25 \times 10^{-3}$ mm$^2$/s). It may be worthwhile considering whether a tailored optimization to a specific tissue is needed or whether a slightly less optimized but global solution is preferred as has been presented here.

Once the optimization calculations are complete, tests can be conducted to check that the optimum $b$-value has been approximated. A useful approach is to plot frequency histograms of the voxel ADC data calculated using the $b = 0$ s/mm$^2$ image and each individual $b$-value in the imaging range, as well as all $b$-values within the imaging range. It is important to normalize the histograms from each ROI (i.e. divide by the number of voxels) before consolidating them into a single population histogram. The plots in our example are as shown in Fig. 2.6. From these data it is clear that as the optimum $b$-value is approached (~900 s/mm$^2$) the ADC histogram more closely resembles the ADC profile with the narrowest distribution obtained using all $b$-values in the range between 0 and 750 s/mm$^2$ suggesting an improvement in the accuracy of ADC measurement.

In some circumstances it may be necessary to consider including T2 dependence into the optimization (JONES et al. 1999) and also the form of the noise distribution in the optimization calculations (SARITAS et al. 2007). The mathematics governing these methods is beyond the scope of this chapter but may need to be taken into account when the tissue T2 values are short (roughly <60 ms) (SARITAS et al. 2008) or the SNR$_0$ is low (roughly <6) (GUDBJARTSSON et al. 1995). It should also be mentioned that parallel imaging, which is now routinely applied in diffusion imaging to improve spatial integrity, introduces a problem in the above discussion as SNR$_0$ is no longer homogeneous throughout the image. However, signal-to-noise estimation based on the ratio between mean lesion signal and the standard deviation of pixel values within a region of interest drawn over a large background area may be adopted when parallel imaging is employed.

### 2.5.2 Qualitative Data

The acquisition of high $b$-value images in the body is now possible due to the advances in gradient technology and the development of strategies to improve imaging artefacts. DW-MRI in the body, using $b$-values as high as 1,000 s/mm$^2$ is now being applied to provide contrast between lesions and background tissue for tumour localization (KWEE et al. 2008). However, at these high values the signal-to-noise ratio decreases exponentially and it is thus important to find optimum $b$-values that provide a compromise between tissue contrast and signal-to-noise.

A relationship has been described for finding the $b$-value which maximizes the contrast between two homogeneous tissues with apparent diffusion coefficients of $D_1$ and $D_2$ at $b = 0$ s/mm$^2$. The definition of
The contrast that can be applied is known as the Michelson contrast and is defined as:

$$C_{1,2} = \frac{L_1 - L_2}{L_{\text{max}} + L_{\text{min}}}$$

(2.5)

where $L_{\text{max}}$ and $L_{\text{min}}$ represent the maximum and minimum “lightness” in the image respectively as perceived by the observer. We will assume that the minimum lightness of the image is black ($L_{\text{min}} = 0$), the image contrast is stretched such that $L_{\text{max}}$ is as high as can be for a given viewing monitor and that there is a linear relationship between the lightness of tissue $i$ and signal intensity as defined by (2.1):

$$L_i(b) = k S_i(0)e^{-b D_i} \quad k = \text{constant}$$

(2.6)

The Michelson contrast may then be written as:

$$C_{1,2} = \frac{k S_1(0)e^{-b D_1} - k S_2(0)e^{-b D_2}}{L_{\text{max}}}$$

(2.7)

The standard approach is then to assume that $L_{\text{max}}$ remains constant in all images (i.e. it is independent of $b$-value) and that the maximization of the contrast is achieved by differentiation with respect to “$b$” and equating to 0:

$$b_{\text{opt}} = \frac{\ln(D_1 / D_2) + \ln(S_1(0)/S_2(0))}{D_1 - D_2}$$

(2.8)

If $S_1(0)$ is approximately equal to $S_2(0)$ then we have (Kingsley 2006)
which may be applied as a reasonable approximation in most circumstances.

As an example of this optimization procedure, the mean ADC for a region of background liver from the previous section was found to be \(D_2 = 1.38 \times 10^{-3}\) mm\(^2\)/s. Applying this to Equation 7 along with the ADC value for the liver metastases excluding necrotic regions \((D_1 = 1.25 \times 10^{-3}\) mm\(^2\)/s), gives us \(b_{opt} = 761\) s/mm\(^2\). Figure 2.5 shows an image acquired using \(b = 750\) s/mm\(^2\) where the metastases are clearly visible against the background liver tissue. In this way, the choice of \(b\)-value can be optimized for the detection of tissue within a certain ADC value range.

We finish this section with some thoughts regarding the choice of whether to acquire data using \(b\)-values that optimize ADC accuracy or \(b\)-values that optimize image contrast. From a diagnostic point of view it is clearly advantageous to have superior contrast between diseased and healthy tissue. It may be argued, therefore, that the total scanning time should be used to acquire the maximum possible number of signal averages at a single optimized \(b\)-value to improve the signal-to-noise ratio. However, it is our belief that wherever possible, images should be acquired with the intent of calculating ADC values. The estimated ADC may then be substituted into the Stejskal–Tanner relationship enabling the computation of images at any desired \(b\)-value providing the clinician with an increased degree of freedom to adjust image contrast. Furthermore, the signal-to-noise ratio is improved this way as noise decreases with increasing \(b\)-value, and high \(b\)-value images which are difficult to achieve using commercial scanners due to large image distortions may be simulated with ease (BLACKLEDGE 2009). Figure 2.7 shows an example where extrapolating a computed image out to \(b = 2,000\) s/mm\(^2\) revealed the presence of a cancer within the prostate that was not observed on the acquired \(b = 1,000\) s/mm\(^2\) image. The use of computed \(b\)-value images will be a welcomed development in the future.

2.6 Summary of Technique Optimization for DW-MRI in the Body

In summary, DW-MR imaging in the body can be optimized to improve image quality and minimize artefacts using phantoms and patient/volunteer studies. DW-MRI phantoms can provide quality assurance and also be used for sequence optimization. For sequence optimization, it is advantageous to reduce the phase-encoding field of view to the minimum required for a given measurement acquisition. The receiver BW should be adjusted to minimize geometric distortion and image ghosting. Parallel imaging should be applied to reduce the length of the EPI readout. Where possible, the lowest possible echo-time achievable for any given DW-MRI measurement protocol should be used. Where
available, simultaneous diffusion gradient application schemes or double spin-echo (DSE) acquisition techniques should be considered, to reduce eddy-current-induced geometric distortion.

The use of at least two $b$-values is recommended for DW-MRI measurements, the higher $b$-value could be optimized for image contrast or ADC accuracy as discussed in the chapter. Multiple $b$-values, which include lower ($\leq 100 \text{ s/mm}^2$) and higher $b$-values ($>100 \text{ s/mm}^2$) could be applied if ADC calculation is the primary purpose for the DW-MRI study, especially when fractionation of the ADC calculation (perfusion sensitive vs. perfusion insensitive) or bi-exponential fitting of the data is desired (see Chap. 1).

When performing DW-MRI in the body, multiple averaging techniques provide better signal-to-noise. Navigator-triggered acquisitions can be applied in the thorax or upper abdomen. If physiological triggering is not possible or practical, multiple signal averaging with multiple $b$-values in shallow free-breathing are also robust. Breath-hold DW-MRI studies, although quick to perform, have lower signal-to-noise and offer limited coverage in terms of range of $b$-values employed and the volume of coverage.

When applying fat-suppression schemes, SPAIR is the fat saturation method of choice over small fields of view (less than 20 cm in the magnet z-orientation). The use of STIR for fat suppression is usually applied over larger fields of view such as in whole-body imaging (DWIBS). STIR fat suppression is also preferred in head and neck region and in the upper thorax.

### References


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