MR Physics and Techniques for Clinicians - High Field Imaging

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Introduction

Recent advances in engineering and technology in magnetic resonance imaging (MRI) have resulted in a broad availability of clinical MRI scanners operating at 3T and even higher magnetic fields. The main motivation for scientists and clinicians to perform high field imaging is the improved signal-to-noise ratio (SNR). In combination with dedicated coil arrays (Roemer et al., 1990; Wiggins et al., 2006), modern commercial 3 Tesla (T) scanners provide substantial SNR gains (of a factor of 5 or more) compared to 1.5T scanners of a decade ago.

Typically, the higher SNR is spend to improve sensitivity and temporal-spatial resolution, often in combination with parallel imaging methods for accelerated clinical protocols (Sodickson and Manning, 1997; Pruessmann et al., 1999; Bernstein et al., 2001; Kruger et al., 2001; Griswold et al., 2002; Bammer et al., 2005). Noteworthy, higher magnetic fields also alter a number of physical parameters such as relaxation parameters, RF power deposition, B1 homogeneity, chemical shift, susceptibility effects and physiological noise. Consequently not all MRI techniques and protocols only benefit from the stronger SNR at higher fields. In recent years, engineers, scientists and clinicians were challenged by moving high field MRI toward a fully accepted clinical investigational tool. In the following, high field related changes of the most relevant physical and technical parameters and the implications on the MRI experiment are discussed.

Contrast (T1, T2, T2*-Relaxation Times).

Compared to 1.5 Tesla the T1 relaxation times for semi-solid tissues are significantly longer at 3T and increases further at 7T (Breger et al., 1989; Fischer et al., 1989; Jezzard et al., 1996; Deichmann et al., 1999; Wansapura et al., 1999; Deoni et al., 2005; Deoni, 2007; Marques et al., 2009). For example, the

T1s in some gray matter tissue in the human brain account for roughly 900 ms, 1300 ms and 1900 ms at 1.5T, 3T and 7T, respectively. To obtain the same contrast as at lower fields, this implies for numerous imaging applications protocol adaptations that may even result in longer total scan time at the higher field. To counteract such adverse effect, parallel imaging may be the technique of choice. Fortunately, the performance of parallel imaging is considerably improving at the higher fields (Wiesinger et al., 2006).

On the other hand, many MR applications directly benefit from the longer T1, techniques such as MR angiography (e.g. Time-of-Flight (ToF) techniques with and without contrast agent) and more research oriented arterial spin labeling (ASL) techniques. With adequately short repetition times (TR), the longer T1 leads to an improved suppression of background signals and provides a stronger signal enhancement from unsaturated spins moving into the imaging slice (Willinek et al., 2003).

In contrast to the longer T1 at higher fields, T2 stays roughly constant when comparing 1.5T and 3.0 T. Conversely, T2* is reduced at higher fields, which has particular impact on the choice of sequence parameters in order to reduce image artifacts and degradation of the point-spread-function (PSF) as a result of significant signal decay during the readout duration. T2*-weighting is increasingly used in clinical research with functional magnetic resonance imaging (fMRI) and susceptibility weighted imaging (SWI). The shorter T2* at higher fields leads to an optimal echo time (TE) at shorter echo times and has to be adapted accordingly in the protocol.

Susceptibility Effects

Local susceptibility gradients in the imaging volume cause phase dispersion of the spins and reduction of the received MR signal. Indeed, the phase effects scale with the magnetic field and echo time.

Some techniques, such as functional imaging make use and nicely benefit from the stronger susceptibility and signal de-phasing at higher fields. In the case of fMRI, changes in the local concentration of the deoxy-hemoglobin complex in the blood modulate the field dependent phase dispersion and give rise to the blood oxygenation-level dependent (BOLD) contrast (Ogawa et al., 1990). Similarly, novel methods such as susceptibility weighted imaging (Mittal et al., 2009) and phase imaging (Duyn et al., 2007) use the same de-phasing contrast mechanism.

In many applications, however, the same mechanism leads to undesired degradation of the image quality. In particular, air-tissue interfaces and iron deposits in the tissue may cause severe susceptibility artifacts that facilitate stronger distortion or even signal voids in high field imaging experiments. In brain MRI, susceptibility artifacts are particularly present in the frontal lobe, posterior fossa, and auditory cortex at the air-tissue interfaces. Susceptibility artifacts can be modulated by a number of sequence parameter such as echo time, readout time and voxel size. As an example, higher spatial resolution and methods that reduce the readout duration, such as multi-shot techniques, parallel imaging, and stronger gradients hardware systems provide means to address this high field challenge.

Physiological Noise

The noise in MRI at clinical field strengths is typically dominated by the thermal noise (Macovski, 1996). When scanning a subject, however, the so-called physiological noise increases the measurable image noise (Kruger and Glover, 2001). In particular in BOLD-imaging strong physiological noise components can be observed that arise from respiration and cardiac cycles, but also from resting brain activity and events causing local changes in blood flow, blood volume, and metabolism. In contrast to thermal noise, the physiological noise scales with the signal intensity and therefore with the magnetic field strength. In an in-vivo fMRI scan at 3T and even more at 7T, the physiological noise easily dominates the

total image noise and thereby reduces the expected gain in SNR when going to higher magnetic fields. Fortunately, a careful choice of sequence parameters allows addressing this issue. For example, imaging at higher spatial resolution reduces the signal strength and minimizes the respective physiological noise components.

In resting state fMRI experiments, however, the same effect is even used as contrast. Here, temporal signal fluctuations arising from changes in blood flow are used to identify brain regions with correlating signal responses to investigate functional connectivity (Raichle et al., 2001). Here, the contrast is boosted by the high field properties.

RF-Power Deposition

The Radio Frequency (RF)-power needed to generate a certain B1 amplitude increases quadratic with the field strength. Comparing 1.5T and 3T, an excitation pulse will require four times the power at the higher field strength. Simultaneously with the B1 field, electrical fields are generated that result in undesired tissue heating. To keep tissue heating within safe limits, the RF-power is monitored carefully during the entire scanning time. The Specific Absorption Rate (SAR, or the energy deposited in the body per unit of mass) describes the distribution and amount of absorbed RF power per kilogram sample weight and is limited to 3.2 Watt/kg and 4 Watt/kg over a 6 minutes averaging period for head and body applications, respectively. RF-intense sequences, i.e. protocols that frequently apply 180º pulses (e.g. Turbo Spin Echo sequences) are prone to exceed the SAR limits at high fields. Common strategies to address this issue are conventional parallel imaging and RF-pulse design. In some sequences, parallel imaging reduces the required RF-power by factor of 2 or more, depending on the acceleration factor used. Specific RF-pulse designs may also help to overcome this problem. Specifically, the Hyperecho (Hennig and Scheffler, 2001) and Variable Flip Angle techniques (Mugler JP, 2004) as well as specifically designed VERSE-pulses (Conolly et al., 1991). allow for significant SAR reduction, often with negligible impact on imaging time and quality.

B1 Uniformity

At high magnetic fields, the RF wavelength is reduced and, within tissue, becomes close to the physical dimensions of the human body. This causes a reduced B1-uniformity, i.e. a spatial distribution of flip angles in the object. In more detail, both sample properties (dielectric constant and electric conductivity) and dimensions determine local B1-focusing (signal pile) or B1-shielding (signal loss) effects, that might appear as brightness or contrast variations. Furthermore, the signal reception may be affected by an enhanced distribution of the spatial sensitivity leading to additional brightness variations. Therefore, the received MR signal at high fields exhibits an increased dependency on the sample properties and dimension. The spatial variation in signal intensity appears to be of rather complex nature and is difficult to control. Recently, techniques have been proposed to reduce B1 non-uniformity through the use of optimized sequences (Deichmann et al., 2000). A particular promising, but also hardware intense, approach to better control the B1 non-uniformities are parallel transmit techniques (Katscher and Bornert, 2006), that become increasingly available on high field research platforms.

Spectral Resolution

The chemical shift δB_0 in units of Hertz between different metabolites increases proportionally with the magnetic field. At a first glance, this results in a better separation of metabolites in MR spectroscopy. On the other hand, the stronger sensitivity to susceptibility effects at higher fields leads to an increased spectral line-width. Therefore, an excellent magnet homogeneity and improved shimming is mandatory for high quality spectroscopy at high magnetic fields.

The improved spectral resolution at high fields also works in one's favor when performing spectral fat saturation. As before, susceptibility effects and inhomogeneity at high-field may compensate this advantage. Therefore, it is not only for MR spectroscopy but also for MR imaging of great importance to have a good shimming hardware.

Acoustic Noise

Acoustic noise also represents a topic to be considered when moving to higher magnetic fields. The acoustic noise is closely related to the force on the gradient coil that arises when field gradients are switched in the strong B0 field. The force scales with the magnetic field resulting in increased acoustic noise levels at higher fields when not compensated.

For patient safety, the acoustic noise level in the MRI scanner is limited to < 100 dB, which in most high field scanners is satisfied with additional ear protection. Ongoing research is optimizing insulating materials, hardware modifications and applications, such as parallel imaging capable to greatly reduce noise. To some extent gradient waveforms can be designed in a way that reduces the dB/dt of both the attack and decay ramps and, hence, minimizing the acoustic burden.

Summary

During the past few years, high field MRI around 3T has evolved to a clinical tool. Initially, image quality in various applications had been compromised by many challenges, such as SAR, susceptibility artifacts, and B1 non-uniformity. It is fascinating, however, to see that these challenges were similar to the challenges faced at 1.5T field strength at the end of the 80's. A strong focus on new developments and optimizations of high field applications from vendors and research has resulted in overcoming many of these challenges. Particular important tools for the control of high field effects are parallel imaging and RF-pulse technology. They allow addressing image artifacts from susceptibility effects, reducing energy deposition in the body and acoustic noise reductions. Particular high expectations for remaining high and ultra-high field challenges are arising from the recent developments in the field of parallel transmission that is capable to address the issue of high-field B1 non-uniformity, but will also open new perspectives to imaging and spectroscopic applications.

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