Cardiac MRI at 3 Tesla
- Challenges and Technical Solutions

Tobias Schaeffter
Division of Imaging Sciences and Biomedical Engineering
King’s College London

Educational Objectives
• To describe the physical limitations and technical challenges when performing cardiac MRI at 3 Tesla.
• To describe technical solutions for successful cardiac MRI at 3 Tesla with respect to image quality and patient safety.

Introduction
Cardiac magnetic resonance (CMR) has rapidly become the imaging method of choice for the diagnosis of cardiovascular disease, and its clinical indications have expanded greatly in the last decade [1]. Although the majority of CMR examinations is still performed at a field strength of 1.5 Tesla, the growing availability of 3 Tesla scanners has resulted in an increasing number of CMR examinations at higher field strengths [2,3]. One of the driving forces to migrate CMR to higher field strengths is the better signal-to-noise ratio, which can be used to improve spatial and/or temporal resolution. Over the last years it has been demonstrated that CMR at 3 Tesla has advantages in comparison to 1.5 Tesla in a number of applications, such as perfusion imaging [4-6]; delayed enhancement [7] myocardial tagging [8] and MR coronary angiography [9]. However, while clinical imaging at 3 Tesla has been rapidly adapted for neuro and musculoskeletal applications, the migration of CMR to higher field strengths has been slower. This is due to image artifacts [10] and issues arising from limitations such as specific absorption rate and field inhomogeneity. This presentation will describe these issues and the technical solutions for successful CMR at 3 Tesla.

Physical Limitations
One of the main drivers for imaging at higher field strength is the higher signal-to-noise ration (SNR) allowing MR-scans at higher spatial or temporal resolution. Ignoring differences of the noise performance of RF-coils and assuming fix scan parameters, the SNR is essentially linearly proportional to field strength (and thus the resonance frequency) [11], i.e.

\[ SNR \propto \omega_0 = \gamma B_0 \]

\[ \frac{SNR_{3.0T}}{SNR_{1.5T}} = \frac{2}{1} \]

(1)
The increased SNR can be exploited towards higher spatial resolution or faster imaging. Since SNR is proportional to volume and to the square-root of the acquisition time, the exploitation of SNR is different for the voxel size and acquisition time

\[ SNR \propto V \sqrt{T_{AQ}} = \Delta x \cdot \Delta y \cdot \Delta z \cdot \sqrt{T_{AQ}} \tag{2} \]

For instance the migration from 1.5 to 3.0 Tesla allows the reduction of the isotropic voxel size by a factor of \( \frac{1}{\sqrt{2}} = 0.79 \) while keeping the SNR constant. On the other hand images can be obtained 4-times faster at 3 Tesla while keeping the same SNR as at 1.5 Tesla. However, this theoretical numbers can often not be realized because of several limitations, like specific absorption rate limitations, field inhomogeneity and prolonged relaxation times of the tissues, which will be discussed in the following.

**Specific Absorption Rate (SAR)**

The Specific Absorption Rate (SAR) is a measure of the absorbed power in the human tissue when exposed to a time varying \( B_1 \)-field, and is measured in Watts per kilogram (W/kg):

\[ SAR = \frac{Power}{Mass} \tag{3} \]

The SAR describes the potential for heating of the patient's tissue due to the application of the RF energy necessary to produce the MR signal. The SAR is proportional to the square of \( B_1 \)-amplitude and frequency:

\[ SAR \propto B_1^2 \cdot \omega_0^2 \cdot \frac{1}{T_R} \tag{4} \]

The rate at which SAR related power deposition increases with field strength quickly outpaces the rate of SNR gain. At higher field strength it is essential to find ways to circumvent the fast increase of SAR related power deposition before it exceeds the regulatory limits. Furthermore, inhomogeneities of the \( B_1 \)-field can lead to a local exposure where most of the absorbed energy is applied to a certain body region, which is described by the concept of a local SAR. Thus, at 3.0T, the situation can arise where the whole-body SAR is at or below the regulatory limit (e.g. 4 W/kg) but the local SAR is considerably beyond the limit [12]. In order to reduce SAR value either the \( B_1 \)-amplitude (flip-angle) needs to be reduced or the repetition time TR has to be increased. However, larger TR values result in longer acquisition times and or can induce banding artifact in SSFP-sequences.

**Main Field Inhomogeneity**

The bare magnet has usually a very high field homogeneity, which is in the order of 1-2 ppm over a spherical volume of 20-30 cm. However, this homogeneity is significantly distorted by the patient’s body, since tissue has a different magnetic susceptibility than air resulting in field distortions. These
susceptibility effects are greater at higher field strengths due to the linear relationship between magnetization and field strength. Usually the magnetic susceptibility difference between air and tissue is given in terms of parts per million (ppm) which is a fixed ratio. However, in terms of absolute magnetic field the inhomogeneity becomes larger with increasing field strength. Furthermore, there are global and local field inhomogeneities. The global field inhomogeneity is induced by the bulk susceptibility of the body volume, whereas local field inhomogeneities are a result of local tissue interface between the heart, lung and liver. Such local field inhomogeneities result in artifacts when using frequency selective RF-pulses (as applied in fat suppression techniques). For instance, local field distortions can result in suppression of water signal instead of fat suppression and banding artifacts can occur in SSFP-imaging. These banding artifacts are even more prominent if a larger repetition times is required due to SAR limitations.

**RF-Field inhomogeneity**

The wavelength of RF pulse decreases with frequency. When the wavelength of the RF is in the order of the object, wave effects can spatially modulate the image intensity. Already at 3 Tesla complex wave effects can occur inside an object with conductivity and permittivity. The primary effect of conductivity results in a shielding effect, which reduces the B1 amplitude in the center of the object. On the other hand, the permittivity of an object can cause a partial focusing effect. In the human body both effects (conductivity and permittivity) can occur resulting in a local attenuation and amplification of the B1-Field. Consequently, intensity variation in the image can occur. Furthermore, local B1-amplification might result in a high local SAR values.

**Relaxation times**

Compared with 1.5 T, the longitudinal relaxation times (T1) are longer at 3T, while T2 values are similar [13]. The T1 of many tissues increases on the order of 20% to 50% as the field strength is increased from 1.5T to 3T. This has impact on the image contrast. In particular, in SSFP the decreased T2/T1 ratio at 3 T can result in lower image contrast. Furthermore in some sequences the increased T1 times can prolong waiting times that are required for signal recovery. The effect of most contrast agents (given by the property relaxivity r) is usually only slightly reduced (5 -10%) when going from 1.5T to 3T [14]. At a given concentration C of the contrast agent, its influence on the resulting relaxation the $T_1$ is given by

$$\frac{1}{T_1} = \frac{1}{T_{10}} + C \cdot r \quad (5)$$

where $T_{10}$ is the native longitudinal relaxation time of the tissue without contrast agent. Since the native $T_1$ typically increases more than the enhancing tissue, equivalent doses of Gd chelate often produce a greater contrast effect at 3T than at 1.5T. This is greatly advantageous for MR angiography (MRA). In fact, lower concentrations of contrast agent may be preferable at higher field strengths, because high concentrations may result in $T_2^*$ related artifacts.
Magnetohydrodynamic Effect

The magnetohydrodynamic effect results in an additional electrical charge generated by ions in blood moving perpendicular to the main magnetic field. Since the blood flow is time-varying the magnetohydrodynamic effect induces an additional time-varying electrical signal that is a function of the magnitude and direction of flow. This additional electrical signal superimposes to the ECG waveform and lead to artifacts in the ECG/VCG trace. In particular, the additional electrical voltage can mimic a T-wave elevation that might be misinterpreted as R-waves resulting in erroneous triggering. The effect increases with higher magnetic field strengths making cardiac synchronization sometimes more difficult.

Technical Solutions

In the previous section a number of the physical limitations have been described that make CMR at high field more challenging. The next section concentrates on some technical solutions that become available on current high field systems.

SAR reduction

Since the SAR is proportional to square of the B1-amplitude (see Eq. 4), one way to reduce the SAR is the use RF-pulses with lower B1-amplitude. For instance it has been shown that the average SAR of an MR scan can be reduced by lowering the flip angle in the outer k-space for SAR critical protocols such as steady-state free precession [15] or hyperecho [16] sequences. Furthermore, advanced RF-pulse design can be used to reduce the peak amplitude [17,18].

One recent development that can counteract for high local SAR values or harness these effects is the use of parallel RF-transmission [19,20]. This approach uses arrays of independent transmit coil elements in which the amplitude, phase and frequency of the RF-field can be independently controlled for excitation of the MR signal. Commercial systems with two integrated transmit channels are available and commercial systems with up to eight channels are on the horizon.

B0-Shimming

At 3 Tesla the main field inhomogeneity must be addressed more carefully. In particular, SSFP sequences are particularly sensitive to field inhomogeneity at 3 T and frequency mismatches can cause signal loss, banding and pulsatile flow artifacts. In order to reduce these artifact short repetition times are required. However, regulatory limits on SAR deposition sometimes dictate the minimally allowed TR. Furthermore, B0-inhomogeneity related artifacts can be reduced by adjustment of the resonance frequency, summing frequency modulated acquisitions or application of localized linear or second-order shimming corrections [21-23].
**B1-shimming**

Several data-adaptive methods have been used to try to estimate a non-uniformity field from the image itself. Unfortunately, all of these methods cannot handle the issue of contrast changes due to influence of B1 non-uniformity on the applied imaging sequence. Adiabatic RF pulses can provide more homogenous flip angles [24] but very often result in high SAR values. Recently, parallel transmission has been proposed to improve RF excitation, in particular, multidimensional, spatially selective RF excitation allowing the compensation of patient-induced B1 inhomogeneities. In particular instance B1-shimming the amplitude and phase in each transmit channel have been chosen to minimize the B1 non-uniformity and thus to enhance the tissue contrast for target region in the body. Furthermore, advanced strategies of parallel transmission take the reduction of B1-inhomogeneities as well as the local SAR into consideration [25,26].

**Conclusion**

The technological developments over the last decade have made CMR scanning feasible at 3 Tesla. Although the management of SAR limitations and susceptibility effects remains a primary concern, the improved CNR in contrast enhanced MRI, like MRA, perfusion MRI and late gadolinium enhancement, will drive further technical advances for a more widespread clinical use of CMR at 3 Tesla.

**References**


