Revisiting the Physics behind MRI and the Opportunities that Lower Field Strengths Offer

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This article outlines why moving to higher magnetic fields may not always be advantageous in MRI. We focus on phenomena that have a substantial impact on daily routine, and aim to provide a solid insight into underlying physical mechanisms. Some discussions have been simplified in the interest of addressing a broad range of readers.

Introduction

When MRI first transitioned from research into clinical practice in the early 1980s, typical magnetic fields were between 0.2 and 0.5T, mainly due to technical limitations. The clinical MRI community quickly moved towards higher fields, converging on 1.5T, which became the new standard. Later, around the turn of the millennium, 3T magnets started to penetrate the market, fueled by the quest for higher signal to noise ratio (SNR) with higher field strength. Today, 1.5T and 3T are the clinical workhorses, present in almost every large hospital radiology department. Lately, Siemens Healthineers have pushed for even higher field, introducing the MAGNETOM Terra – the first clinical 7T scanner with FDA clearance.

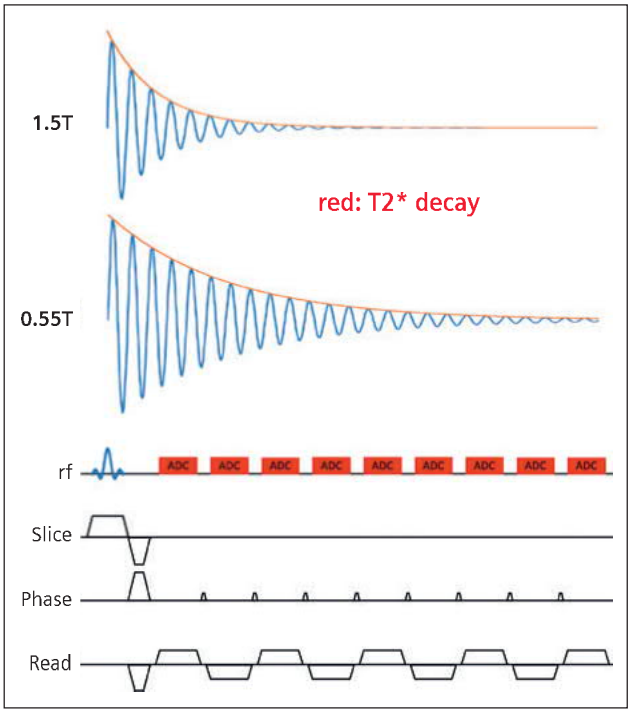
There was an immediate clinical advantage in moving to higher field. It increases the magnetization available for imaging, which improves SNR. The increased signal can be invested into either higher spatial resolution or reduced imaging time. For human MR imaging in general, SNR increases linearly with the field [1]. However, in order to maintain image quality, the chemical shift between different nuclei can be kept constant by adjusting the receiver bandwidth, in which case SNR is proportional to the square root of the field [1, 2]. Recent research in ultra-high-field human brain MRI suggests faster than linear increase of SNR with field [3].

Then again, there is no such thing as a free lunch. Increasing the field increases radiofrequency absorption in tissue, increases artifacts, and degrades imaging in various other ways, through the physical phenomena described below.

Radiofrequency absorption

MRI scanners use high-power radiofrequency pulses to manipulate nuclear spins. This RF energy would become a health hazard if it were intense enough to overheat tissue, so scans have to stick to safety limits.

Specific absorption rate (SAR) is the relevant measure of RF absorption, in watts per kilogram of body tissue. The International Electrotechnical Commission [4] sets maximum SAR of 2W/kg in normal mode, and of 4W/kg in first-level mode (both for whole-body SAR averaged over 6 minutes).



1 During a typical EPI gradient echo readout, the later echoes suffer from low signal intensity at 1.5T compared with 0.55T, and the resulting images are blurred (see Figure 2). The red envelope in the upper two sinusoids corresponds to the observed T2* decay (neglecting gradient-induced dephasing).

The absorbed energy depends on the square of the field, so it increases by a factor of 4 when doubling the field. Most clinicians and researchers have faced SAR limitations, e.g., in SAR "heavy" examinations such as Cardio or MSK, especially with the advent of 3T scanners in clinical routine. Operating in first-level mode is often mandatory at 3T, but even then the SAR threshold limits the use of clinically relevant sequences.

Using 0.55T instead of 1.5T or 3T reduces SAR by a factor of 7.5 or 30, respectively. Consequently, many SAR-intense sequences, e.g., TrueFISP or TSE, do not hit the SAR limit at 0.55T. This can help increase SNR by applying higher flip angles (e.g., 180° instead of lower flip angles such as 150° at 3T for TSE imaging; 90° flip angle for TrueFISP possible), or shortening radiofrequency (RF) pulses to minimize echo time (TE) and TR.

Relaxation times

The relaxation of spins is the main mechanism for the observed image contrast in MRI. In this section, we will review the types of relaxation and their impact on clinical routine.

T1 relaxation

The spin of a proton gives it a magnetic moment. Just as a compass needle aligns with the magnetic field of the Earth, the proton spin tends to align parallel to an external magnetic field B_0; this is the lower energy state of the proton spin, while the anti-parallel state is higher energy. In an MRI, a radiofrequency pulse can be used to convert parallel proton spins to their anti-parallel state; then spin-lattice relaxation (also known as T1 relaxation) gradually brings them back to their energetically preferred parallel state.

This is triggered by the random magnetic field fluctuations – sometimes referred to as "magnetic noise" – of the surrounding magnetic moments caused by, e.g., protons, electrons, and various molecules. The energy difference between the parallel and anti-parallel state corresponds to a resonance frequency, proportional to the field, and relaxation is triggered by "magnetic noise" close to that frequency. Generally speaking, this "noise" at lower frequencies is stronger than at higher frequencies – so at lower field, relaxation is more efficient.

The time constant of this relaxation, T1, is defined by [5]

Equation 1

$$T1 \propto \frac{B_0^b}{a}$$

with constants a and b that must be determined by experiment. Recent investigations [6] across a range of clinical fields (0.55T, 1.5T, 3T, 7T) revealed the following linear dependency between T1 and B_0:

Equation 2

$$T1 \propto \frac{B_0}{12.2}$$

with a = 12.2 and b = 1. This means that T1 gets shorter if the field decreases.

How does this help us from a clinical point of view? If T1 is shorter, the repetition time (TR) can be reduced, resulting in decreased scan time. For example, T1 for gray matter decreases from 1000–1300 ms at 1.5T down to 700–800 ms at 0.55T [7]. Due to this, TR can be reduced by the same proportion, at least 25%.

We can address the loss of SNR by averaging. SNR scales with the square root of the field [1, 2], so to have SNR at 0.55T comparable to that at 1.5T, three averages are necessary. Instead of tripling your imaging time, it will only increase your scan time by a factor of 2.25 because TR could be selected 25% shorter than at 1.5T. Physics comes in favorably from the T1 perspective at lower field strength.

T2 relaxation

Spin-spin relaxation, commonly referred to as transverse or T2 relaxation, does not show high dependency on field.

To observe T2 relaxation, a radiofrequency pulse first flips proton spins so that their net magnetization is perpendicular to the external field. Magnetization then precesses around the B_0 direction at the Larmor frequency

Equation 3

$$\omega = \gamma \cdot B_0$$

where γ is the gyromagnetic ratio of the proton.

The RF pulse also puts proton spins in phase. Gradually, magnetic noise knocks some spins out of phase, which decreases the net magnetization (the vector sum of all spins). After time T2 magnetization decreases by a factor of e, so after approximately 5T2 it has effectively vanished. By this argument the strength of the field does not affect T2 relaxation.

Nonetheless, empirical research shows that T2 increases slightly as the field falls. For gray matter, T2 values at 1.5T are 90–110 ms, while at 0.55T they are 110–120 ms

[5]. This would allow turbo spin echo (TSE) imaging to employ longer echo trains at 0.55T, reducing the number of required readouts, potentially reducing the required scan time compared with 1.5T – if the SNR penalty is neglected.

T2* relaxation

Imaging the diffusion of water molecules is one phenomenon that often relies on T2* contrast, and other applications such as BOLD imaging and DSC neuro perfusion suffer from T2* loss or blurring as well. All gradient-echo readouts, such as echo planar imaging (EPI), are modulated by T2* decay.

Several phenomena are affected by both regular T2 decay and another decay process known as T2'. T2' is caused by static inhomogeneities in the magnetic field, such as susceptibility-induced magnetic field gradients, which combine to form T2* relaxation.

The overall timescale is

Equation 4

$$T2^* = \left(\frac{1}{T2} + \frac{1}{T2'}\right)^{-1}$$

By employing TSE sequences, T2' can be recovered and true T2 decay can be observed.

Susceptibility-induced magnetic field gradients scale with B_0, so at high field T2* is shorter than at low field. In gray matter, T2* at 1.5T is typically 70–80 ms, while at 0.55T it is 80–90 ms [5]. This allows more SNR-efficient EPI sampling at 0.55T, since the available echo signal amplitude is relatively large for later echoes in the EPI train (see Figure 1). A side effect is reduced blurring (see Figure 2).

Susceptibility

Magnetic susceptibility measures how a material responds to an external magnetic field. The induced magnetization inside a material under external magnetic field strength *H* is:

Equation 5

$$\vec{M} = \chi\vec{H}$$

where χ is magnetic susceptibility[1].

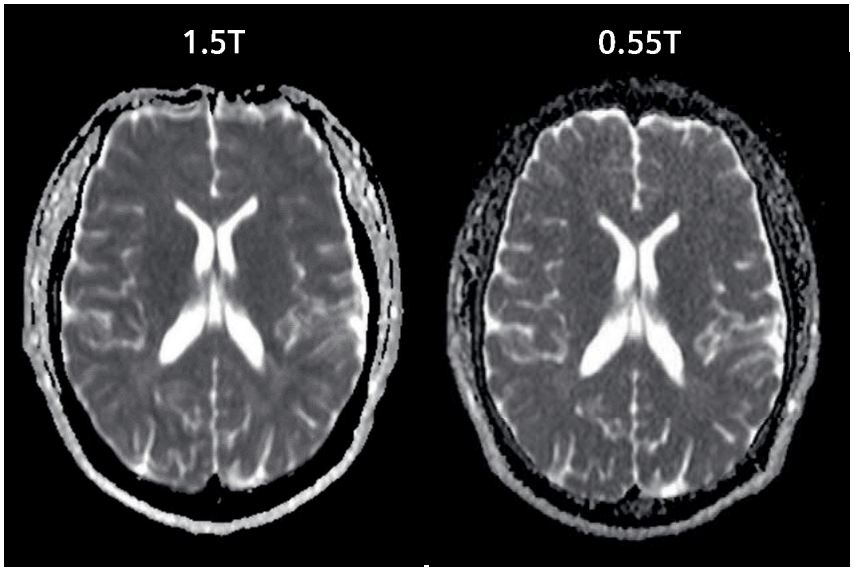
The induced magnetization *M* and the external magnetic field *H* result in the magnetic flux density *B*:

Equation 6

$$\vec{B} = \mu_0\vec{H} + \vec{M} = (\mu_0 + \chi)\vec{H}$$

where $\mu_0 = 4\pi \cdot 10^{-7}$ is the magnetic permeability of vacuum.

Eq. 6 tells us that as *H* increases, the change in *B* depends on the value of χ. If $\chi > 0$, *B* increases; if $\chi < 0$, *B* decreases; and if $\chi = 0$, *B* doesn't change.



2 Due to faster T2* decay at higher field (see Figure 1), the apparent diffusion coefficient map at 1.5T is slightly blurred along the anterior-posterior direction. This is particularly apparent in the hyperintense gyri. Both images were acquired on the same volunteer. Also, a slight geometric distortion in the anterior brain region is visible at 1.5T (see Susceptibility).

[1] *There are different physical phenomena behind susceptibility. Diamagnetism is a universal property of any material which decreases the magnetic field inside it. It is often "hidden" underneath the stronger phenomenon of paramagnetism, which increases the magnetic field. Both dia- and paramagnetism can only be observed when an external magnetic field is present. As a side note, ferromagnetism is a property that is maintained even when the external magnetic field has been turned off.*

Susceptibility changes can be a valuable contrast source. Susceptibility-weighted imaging (SWI) exploits slight differences in χ to visualize veins in the brain, enabling the diagnosis of cerebral hemorrhage, for example. Quantitative susceptibility mapping (QSM) produces a map of susceptibility that might constitute a new biomarker [8–10].

At the same time, susceptibility is a source of artifacts. Near the sinuses and auditory canal, for example, the changes in susceptibility moving from air to tissue cause geometric distortions in conventional EPI-based imaging, including diffusion-weighted imaging (DWI) and functional MRI (fMRI). Eq. 6 shows that reducing the external field also reduces the absolute differences in local magnetic fields, which makes susceptibility-induced artifacts much less prominent. This can be seen in Figure 3, showing the results of DWI scans on the same volunteer performed on a 3T, a 1.5T, and a 0.55T system. While SNR clearly decreases as field decreases, the geometric distortions are significantly reduced as well.

Diffusion imaging of the optic nerves is also substantially improved at lower field (see Figure 4). A similar behavior is to be expected for imaging of metallic implants such as hip implants. This has been shown in the literature [11–13] and might have a significant impact on clinical practice in the aging societies of industrialized nations.

B_0 and B_1 Homogeneity

B_0 homogeneity is of utmost importance for image quality, and less effort is required to guarantee this homogeneity at lower B_0 values.

B_1 homogeneity is also crucial, to ensure homogenous excitation of the desired slice or volume. B_1 – or more precisely B_1^+ – is the transmitted radiofrequency field that excites the protons at their Larmor frequency. The flip angle is directly related to B_1 through:

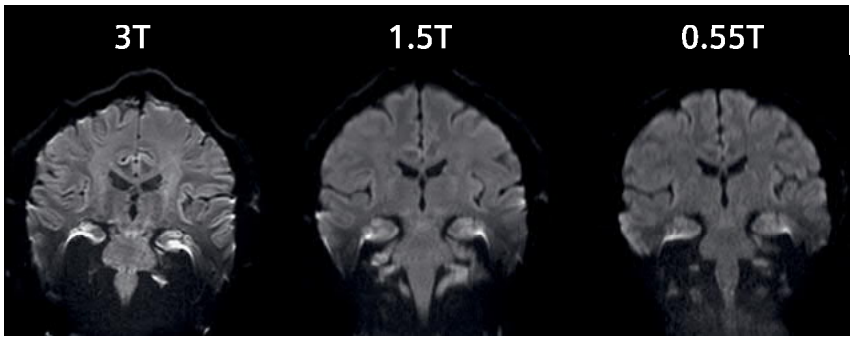
Equation 7

$$\alpha = \gamma \cdot B_1 \cdot t_p$$

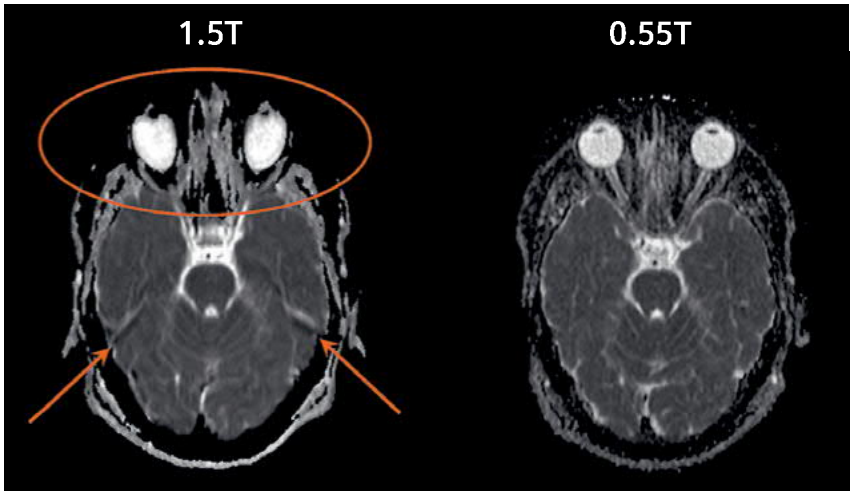
where α is the targeted flip angle of the applied RF pulse, γ is the gyromagnetic ratio of the proton, B_1 is the magnetic field of the applied RF pulse, and t_p is the pulse duration.

From this equation, it is clear that if B_1 is not homogenous, the excited slice or volume will show varying flip angle, leading to variations in contrast and SNR.

When the Larmor frequency ω increases, the associated wavelength λ decreases. At 3T, the wavelength within the human body is approximately 25–30 cm [14], close to the dimensions of a human torso. This results in construc-



3 Coronal EPI diffusion scans (b = 1000 s/mm²) acquired on the same volunteer with three different fields. Moving to lower field typically comes with a reduction in SNR; but at the same time susceptibility artifacts near the temporal lobe due to vicinity of the air-filled internal auditory canal are substantially reduced.



4 Geometric distortions caused by the air-filled sinuses and internal auditory canal are substantially reduced at lower field. In this example on the same volunteer at 1.5 and 0.55T, the eyeballs maintain their almost spherical shape, and the optic nerves are clearly visible (ellipse); the susceptibility artifacts near the temporal lobes (arrows) are almost invisible at 0.55T.

tive and destructive interference within the body, leading to inhomogeneous B_1 excitation. This phenomenon has triggered lots of research efforts to mitigate this problem and can now be considered understood; countermeasures such as B_1 shimming are available, but come with increased costs and hardware requirements.

Conversely, at lower field, the Larmor frequency is lower, so the associated RF wavelength is longer. This reduces or even eliminates the effect of interference. For example, at 0.55T, the wavelength is approximately 130–140cm, much larger than typical body diameter. So B_1 will be more homogeneous at lower field, giving more homogeneous contrast and SNR behavior.

Summary

For more than 20 years, the clinical standard for MR imaging has been either 1.5T or 3T. While many of the physical advantages of low field have been well known in the scientific community, the push for higher SNR favored higher fields. However, in the light of technical improvements made over the last 20 years, other metrics may be favored. Improvements in image reconstruction – from parallel imaging [16,17] and compressed sensing [18] to deep learning [19] – mean that clinicians can make optimal use of the available signal while exploiting the physical advantages of low-field MRI, such as reduced artifacts.

If SNR is sufficient at two different fields, then the focus may fall instead on diagnostic or financial value. Scientific literature even back in the mid 1990s showed no significant difference in diagnostic sensitivity or specificity between 1.5T and lower-field systems [20, 21]. Low-field machines are also less expensive, which is particularly relevant given the cost pressure in many healthcare systems.

All these factors could help bring MRI to places it has not been before – spreading into new geographical and clinical areas. The future of low-field MRI looks bright.

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