High Field MRI: Technology, Applications, Safety, and Limitations

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Introduction

The amount of available signal in conventional magnetic resonance imaging (MRI) is inextricably tied to the static magnetic field strength (B0) of the imaging system. Until recently, most clinical MRI scanners operated at field strengths at or below 1.5 Tesla, however, due in part to improvements in magnet design and shielding which ease siting requirements, 3 Tesla clinical scanners are now widely available and there is a push for even higher field whole body scanners (7-11 Tesla) throughout the industry.

The drive towards high-field MRI is fueled by the benefits of potentially higher signal-to-noise ratio, contrast-to-noise ratios and spectral resolution for certain applications. In many cases, these benefits will facilitate higher spatial and/or temporal resolution than previously possible with MRI. There are, however, technological, physical and safety limitations inhibiting the full realization of these benefits at high-field. Technology issues include homogeneity of the B0 and B1 magnetic fields, higher gradient coil performance and linearity, and the design of robust radiofrequency array coils for signal reception. At high-field, physics concerns include changes in relaxation kinetics, increased susceptibility effects and other changes in contrast mechanisms. Safety limitations include higher power radiofrequency pulses and the potential for tissue heating or coil burns, stimulation effects from stronger, faster switching gradients and moving within a higher magnetic field and, most prominently, the potential dangers associated with the main magnetic field, such as ferromagnetic projectiles in the scan room and effects on implanted medical devices, many of which have not been evaluated at fields above 1.5 Tesla.

Ultimately, the design of protocols and acquisition methods that account for these limitation will need to be pursued in order to reap the benefits of high-field MRI without compromising patient safety. Many MR imaging techniques have already seen demonstrable improvement at higher fields and have driven the development of high-field systems. Techniques in functional magnetic resonance imaging that rely on blood-oxygen level dependent contrast mechanisms, techniques in angiography and techniques in dynamic susceptibility contrast perfusion imaging all benefit from the effects of higher fields on local susceptibility contrast. Proton MR spectroscopy methods for brain and body imaging benefit from the higher spectral resolution of high field as do multi-nuclear techniques. Changes in relaxation kinetics provide enhanced contrast in angiography techniques and arterial spin labeling techniques.

Technical Challenges and safety concerns

Main field

After about 15 years of research use, 3.0 Tesla whole body scanners successfully entered the marketplace in 2002 after the FDA cleared whole-body clinical scanners from GE, SIEMENS and Philips. These systems accounted for approximately 8.5% of "high-field" revenue in 2003 after just one year on the market, and sales continue to climb. Whole body systems at 4.0 Tesla -8.0 Tesla are already in the process of being evaluated at luminary sites. The University of Minnesota,

Massachusetts General and NIH all have whole body 7.0 Tesla scanners in various stages of development for research. An 8.0 Tesla whole body scanner for research has been in use since 1999 at Ohio State University and a whole body 9.4 Tesla scanner is now nearly operational at the University of Illinois at Chicago. However, with current technology, the niobium titanium superconducting wire used to construct modern MRI scanners there is a limit to how high of a field can be produced because of limits on the critical field of these superconductors. This limit is reached in the range of 10-12 Tesla, with 12 Tesla being achievable only by further cooling of the NbTi conductors. Niobium tin can generate fields above this range, but the material is extremely brittle, making exclusive use of the material difficult. Likely high-field magnets beyond 10 Tesla will begin to interleave NbTi and Nb_3Sn windings to achieve higher fields.

With increased field strength comes increased weight, cryogen consumption and concern over fringe fields for siting. Weight, particularly in scanners using primarily passive shielding, can be an issue. Even with active shielding, commercially available 3 Tesla scanners are still roughly twice the weight as their 1.5 Tesla counterparts (currently about 10,000 kg versus 5,000 kg). A 7.0 Tesla magnet currently sites at about 30,000 kg (unshielded). Room shielding to contain the fringe fields can be > 100,000 kg. Current technology allows good active shielding for units < 4 Tesla, one should expect a fair amount of activity in developing active shielding technology for higher fields in the near future. While the 5 G line on today's 3 Tesla scanners extends out an additional 0.4-0.6 m axially from their 1.5 Tesla counterparts (which generally have a 2.5 m 5 G line), the radial 5 G line extends out 1-2 m on most 3 Tesla units (from about 4 m on most 1.5 T scanners). This means more room shielding is necessary to locate a 3 Tesla scanner in a room previously sited for a 1.5 Tesla system, adding to cost and room weight.

Additionally, vendors are becoming better adept at handling cryogen loss in an attempt to bring 3 Tesla scanners into the 0.03-0.075 L/hr range from the 0.15 L/hr range characteristic of older units. Further concern is the need for better shimming to maintain field homogeneity. In order to achieve 0.2 ppm over a 30 cm sphere or 0.1 ppm over 20 cm at higher fields, high-performance, automated shimming using a combination of passive and higher-order, room temperature shims will be needed to compensate the main field induced susceptibilities in the imaging volume. Careful consideration should be paid given to the system shimming ability in conjunction with the stated field homogeneity when considering a high-field system for applications such as spectroscopy or functional brain imaging.

Safety issues relating to increases in the main field include the possibility of ferromagnetic projectiles, increased torques on medical devices and implants and increased electromagnetic/magnetohydrodynamic effects. It is extremely important to note that medical devices declared "MR Safe" at 1.5 Tesla are not necessarily so at higher fields. The torque on a ferrous object increases linearly with the field strength while the translational force experienced by a ferrous object will increase as the product of $B_0(\nabla B_0)$. It is important to note that the increase in translational force for most heavily shielded units will be much larger due to the large spatial gradient of the magnetic field outside of the scanner.

As of 2003, the FDA specifies the criteria for significant risk in humans from the main field to be 8 Tesla for adults and infants older than one month and 4 Tesla for neonates. While there have been no reported health hazards reported in conjunction with physiological exposure to higher fields thus far, it should be noted that there are increased magnetohydrodynamic effects and human response to increased fields. Transient phenomena reported in association with patients moving in high fields include slight nausea and vertigo, headache, tingling/numbness, visual disturbances

(phosphenes) and pain associated with tooth fillings. All effects are transient, ceasing after leaving the magnet, and are usually reduced or avoided by making sure the patient moves *slowly* while in the main field.

Electrically conductive fluid flow in a magnetic field induces an electric current and a force opposing the fluid flow. The effects are greatest when flow is perpendicular to the field. The potential across such a vessel is proportional to B_0, while the force resisting flow is goes as B_0^2. This means that the well known phenomena of "T-wave swelling" distortion seen on an ECG during the period of highest flow through the aorta during an MRI exam, where induced potentials are on the order of 5 mV/Tesla, will be exacerbated at high-fields making it a challenge to obtain good ECG's in the MR environment. The increased blood pressure due to the additional work the needed to overcome the magnetohydrodynamic force has a negligible effect on blood pressure (< 0.2% at 10 Tesla). It is hypothesized that field strengths of 18 Tesla are needed before a significant risk is seen in humans.

Radiofrequency fields
At main fields of 3 Tesla and above, the B_1 field sensitivity increases approximately linearly, but becomes increasingly inhomogeneous due to permittivity, conductivity and the conformation of the patients. In tissue, the dielectric properties of tissue give rise to a region of hyperintensity at the center of the image referred to as "field focusing". While some effects can be compensated for improved coil design, such as transmission line (TEM), and post-processing methods based on simple models, signal intensity variations across the image are going to continue to be a significant challenge at high fields, driving and limiting development on both sides of the transmit-receive chain.

The specific absorption ration (SAR) is a measure of the power absorbed in tissue per unit mass. SAR goes approximately as B_0^2 for fields < 4 Tesla, but this power law is somewhat reduced for higher field strengths. The FDA limits SAR by anatomical site based on the potential effects of heating. Current limits beyond which an Investigational Device Exemption (IDE) is needed are a whole body dose of 4 W/kg averaged over 15 minutes, a head dose of 3 W/kg averaged over 10 minutes, a maximum dose of 8 W/kg delivered to any gram of tissue in the head or torso in a 5 minute period or a 12 W/kg dose in any gram of tissue in extremities over a 5 minute period.

At high-field, these SAR limitations create some of the most difficult challenges for high-field imaging by forcing tradeoffs between image acquisition rates, resolution and slice coverage. This is particularly true for traditionally high SAR sequences such as fast spin echo, chemical fat saturation or magnetization transfer sequences. A technical challenge for high-field imaging will be pulse sequence design, RF pulse design, new acquisition techniques and hardware designs that help overcome the limitations imposed by these tradeoffs. However, many of these techniques, such as the use of reduced flip angles or partially parallel imaging techniques such as SENSE, come at the price of altered contrast or concessions in signal to noise ratio (SNR). Nothing is "free" in MRI.

Gradient fields
Higher performing gradients for imaging and shimming are desirable at any field strength. However, at higher field strengths, high performance gradients are mandatory to take advantage of the potential for faster acquisitions or higher resolution scans available via the increased SNR from the main field. However, it is physiological constraints on dB/dt for safety that may ultimately limit gradient performance in high-field systems. One strategy for overcoming this is the design of high performance gradients linear over a smaller volume for specialized applications, such as

neurological or cardiac imaging, which facilitate faster slewing of the gradients for a particular gradient amplitude without inducing pain in the extremities. Other challenges related directly to higher field strengths include acoustic noise (force on the coils scales with the main field) and concomitant field effects, which also scale with the field.

Contrast changes at higher fields and applications

Image signal to noise ratio in the mid-field regime goes approximately as $B_0^{7/4}$, but transitions to a more linear dependence in the high-field range. The immediate implication of this increase in sensitivity is that higher resolution scans can be performed over a similar time span, faster imaging techniques can be used with similar resolution, or a combination of both. Applications limited by low SNR at 1.5 Tesla and could benefit from higher resolution, such as diffusion and diffusion tensor imaging, can benefit from these increases.

However, relaxation parameters also change with increased field strength, making a straightforward use of increase SNR difficult. The spin-lattice relaxation time, T1, increases and converges for tissues, decreasing both SNR and contrast for T1-weighted exams using the same TR used at 1.5 Tesla. The immediate implication here is that longer pulse repetition times, TR, must be used in T2 and T1 weighted imaging. The loss of contrast in T1 weighted images means that inversion recovery or magnetization prepared techniques may need to be implemented in order to achieve the desired contrast between tissues. However, the increase in T1 of contrast agents in tissue is much lower than the increase in T1 of the tissue, particularly blood. Therefore, normal tissue and blood are more easily saturated for applications such as contrast-enhanced angiography. Time of flight applications will also benefit from the increased T1 of tissue as it will be more saturated against the inflowing blood for similar acquisitions.

Spin-spin relaxation, T2, shows some shortening at higher fields > 3 Tesla. More importantly, however, the susceptibility gradients that lead to T2* shortening increase linearly with the main field (to first order) causing a shortening of T2* at higher fields. This leads to better T2* contrast in applications like dynamic susceptibility imaging for perfusion. Another important application is blood oxygen level dependent (BOLD) imaging. BOLD contrast, used to map function in the brain, gets a boost both from the increased SNR and the increased T2* contrast. However, a serious downside to all this is that shortened T2* values also lead to signal loss in for long TE gradient echo acquisitions and cause challenges for echo-train based acquisition techniques, such as echo-planar imaging (EPI). Once again, techniques such as partially parallel imaging, which reduce the echo-train length without sacrificing resolution, may aid in overcoming these limitations.

At higher fields, there is a larger spectral separation between different chemical species, leading to higher spectral resolution. Magnetic resonance spectroscopy applications will obviously benefit from this and the increase in signal to noise. MRS applications utilizing other, less sensitive, nuclei of biological interest, such as ^{31}P, ^{13}C, [23]Na, and ^{19}F, will also benefit from the increased sensitivity of higher fields. The chemical shift between fat and water goes from 220 Hz at 1.5 Tesla to 440 Hz at 3 Telsa, resulting in a faster accrual of phase-difference between water and fat for a given TE and exasperating the chemical shift artifact. Higher bandwidths will be needed to keep chemical shift artifacts under control.

Summary

Higher fields bring the promise of higher SNR and contrast changes that benefit some applications while making others more difficult. While commercial 3 Tesla scanners are now widely available, there are still a substantial number of technical hurdles that need to be overcome before the 3 Tesla systems can be used in lieu of the highly developed 1.5 Tesla scanners on the market. Serious thought should be given to the desired workflow and routine scan protocols desired before deciding on a high-field unit over a 1.5 Tesla system as many of the imaging protocols are still works in progress and a variety of coils and technology considered standard are not yet available on these higher field systems. However, as technology image quality advances with these higher field systems, expect to see more high field systems to become staples of the MR imaging market as well as a whole host of applications and techniques that exploit the particular advantages that come with higher field strength.