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Ultrahigh field MR Neuroimaging

Gaurav Verma¹, Priti Balchandani¹

¹Translational and Molecular Imaging Institute, Icahn School of Medicine at Mount Sinai, New York, NY

1.0 Introduction

Four decades following the earliest acquisition of magnetic resonance (MR) derived images of the brain, MR technology has developed into an essential diagnostic tool across a diverse range of clinical applications. Central to this evolution has been the development of human MR scanners with increasingly greater magnetic fields, most recently those operating at field strengths of 7 tesla (T) or more (up to 11.7T) [1]. These ultrahigh field MR scanners have the potential to permit neuroimaging with unprecedented detail, leading to better characterization of normal tissue and pathologic lesions and improved planning of treatment and monitoring of response. As of early 2019, at least 76 whole-body MRI systems of 7T or greater field strength have been installed around the world according to a database developed by Renzo Huber of the layer fMRI blog. The 7T Terra (Siemens Healthineers, Erlangen, Germany) system has become the first ultrahigh field system to receive 510(k) clearance for clinical imaging.

Signal-to-noise (SNR) and tissue contrast, two considerations central to the utility of MRI, both scale proportionally with field strength (see Table 1). Previous comparative studies across high (3T) and ultrahigh field MRI (7T and above) have empirically demonstrated a potentially supralinear relationship between field strength and SNR [2, 3]. A consequence of this greater SNR compared to conventional scanners is that images at ultrahigh field can typically be acquired with greater resolution and better differentiation of fine anatomical structures [4]. This greater sensitivity of ultrahigh field imaging may help to establish imaging of biomarkers for diseases like brain tumors, Alzheimer's disease [5], neuropsychiatric disorders like depression, post-traumatic stress and schizophrenia and neurological disorders like epilepsy and multiple sclerosis. As a leading-edge technology ultrahigh field imaging is not without its technical challenges. Realizing the clinical potential of ultrahigh field neuroimaging requires addressing issues such as greater B_0 and B_1 inhomogeneity and specific absorption rates (SAR).

Corresponding Author: Priti Balchandani, priti.balchandani@mssm.edu, Phone: (212) 824-8457, Hess CSM Building, 1470 Madison Ave., New York, NY, 10029.

^{6.2}Conflicts of Interest and Disclosures

PB is a named inventor on patents relating to magnetic resonance imaging (MRI) and RF pulse design. The patents have been licensed to GE Healthcare, Siemens AG, and Philips International. PB receives royalty payments relating to these patents.

2.0 Parametric benefits of Ultrahigh field Imaging

Table 1 shows a list of the relationship between main field strength (B_0) and various imaging-related parameters which may ultimately enjoy a benefit from scanning at ultrahigh field. While parameters such as higher SNR or greater spectral separation are generally beneficial across a range of imaging applications, other parameters such as greater susceptibility and faster relaxation times may be exploited towards a benefit or act as a limitation depending on the particular imaging sequence.

2.1 T_1 and T_2 weighted structural Imaging

The currency of greater SNR provided by higher field strength can be for other performance improvements including higher spatial resolution and faster scan time. Ultrahigh field MRI benefitting from greater tissue contrast and higher resolution has facilitated the visualization of finer structures and improved differentiation of lesions and other regions of interest from the surrounding tissue. These improvements have advanced post-processing towards more precise segmentation and volumetric quantification of smaller structures such as the subfields of the hippocampus [17–19] and the nuclei of the amygdala [20]. Other small features of clinical interest with the potential for improved characterization with ultrahigh field MRI include tissue microbleeds [21], polymicrogyria [22] and subcortical features [23, 24].

The SNR benefit of ultrahigh field may be counterbalanced in the context of the T_2 -weighted turbo spin echo (TSE) sequence, which relies on trains of 180° radio frequency (RF) pulses. The greater SAR at ultrahigh field may limit the echo train lengths (ETL) or minimum repetition time (TR) of these RF-intensive techniques, resulting in longer scan times. Figure 1 demonstrates a coronal-oblique T_1 -weighted MPRAGE and T_2 -weighted TSE acquisition showing an in-plane resolution of 0.4×0.4 mm with a slice thickness of 2 mm. Figure 2 shows a fast gray matter acquisition T_1 inversion recovery (FGATIR) sequence employing an inversion pulse to null the signal contribution from white matter and improve anatomical contrast in deep gray matter structures. All images shown in this chapter were acquired using a Nova Medical 32-channel receive head coil on a Siemens 7T MAGNETOM MRI scanner.

2.2 Functional and Vascular Imaging

Contrast mechanisms based on magnetic susceptibility, including susceptibility weighted imaging (SWI), quantitative susceptibility mapping (QSM), blood oxygen level-dependent (BOLD) imaging and flow-dependent imaging such as time-of-flight (TOF) each benefit from the proportionally greater susceptibility effect at ultrahigh field. TOF imaging also benefits from longer T_1 relaxation times at ultrahigh field, yielding a near-doubling of observed contrast-to-noise ratio (CNR) [25]. Likewise, the arterial spin labeling (ASL) technique benefits from both the higher SNR at higher field and the longer T_1 , with signal decay described as:

$$\Delta S(t) \propto B_0 * e^{1/T_1}$$

The SNR advantages at ultrahigh field could lead to reliable detection of smaller arteries than is achievable at lower fields, with potential applications in detection of occlusions in acute ischemic stroke as well as in neurosurgical planning. Figure 3 shows a high-resolution axial SWI image acquired from a healthy volunteer at 7T with 0.2 mm² in-plane resolution and 1.5 mm slice thickness.

Resting-state functional magnetic resonance imaging (fMRI) studies demonstrate sensitivity benefits from greater SNR at ultra-high field, which may be leveraged towards higher spatial resolution. A recent comparison study between 3T and 7T by Morris et al. [26] demonstrates up to a 300% improvement in SNR and resting-state functional connectivity coefficients at the higher field strength. This greater sensitivity may allow improved characterization of subtle neuropsychiatric disorders including major depressive disorder (MDD).

2.3 Diffusion Weighted Imaging

Although the chief contrast mechanism of diffusion weighted imaging (DWI) and diffusion tensor imaging (DTI) is modulated by the magnitude of the applied directional gradient, diffusion-based sequences still benefit from the higher SNR at ultrahigh field. As with anatomical imaging, these SNR gains can be leveraged towards smaller voxel sizes, and therefore higher resolution while maintaining adequate signal quality. DWI/DTI studies typically rely on an echo-planar (EP) readout for fast spatial encoding, which is prone to susceptibility artifacts at tissue interfaces, particularly at ultrahigh field, and may furthermore be limited in duration by T₂* relaxation. Both limitations may be alleviated by a segmented EP readout method such as the one proposed by Heidemann et al. [27].

Higher resolution and SNR may also assist the accuracy of probabilistic tractography approaches popular in DTI post-processing to resolve crossing fibers and probe the connectomic properties of brain structures. Figure 4 demonstrates tractography from the whole brain generated from a 64-direction multi-band DTI sequence acquired at 7T with an isotropic voxel resolution of 1.05 mm³. Figure 5 shows tractography between the left and right hippocampus, amygdala and thalamus from a separate DTI study acquired with identical scan parameters at 7T. The left and right sides of the image show fiber tracts which pass through the hippocampus, amygdala and thalamus.

2.4 Spectroscopy & Spectroscopic Imaging

Magnetic Resonance Spectroscopy and Spectroscopic imaging (MRS/MRSI) [28, 29] and chemical exchange spectroscopy (CEST) [30] sequences derive a double benefit from scanning at ultrahigh field, with both proportionally greater SNR and chemical shift dispersion. These advantages result in the detection and quantification of metabolites with greater specificity and sensitivity than possible with equivalent sequences at lower fields [31–33]. Recent studies have attempted to leverage this improved performance to probe the role of low-concentration metabolites like gamma aminobutyric acid (GABA), glutamate (Glu) and glutathione (GSH) in neurological [34] and neuropsychiatric disorders [35, 36].

Gains in spectral separation at ultrahigh field may be partially mitigated by broader line-widths due to greater B₀ inhomogeneity, yet these limitations can be overcome by advancements in B₀ shimming algorithms and the implementation of higher order dedicated

shim coils. Greater potential for water contamination due to B_0 -inhomogeneity also prompted the implementation of variable power and optimized relaxation delays (VAPOR) water suppression by Tkáč et al [28, 37]. MRSI sequences at ultrahigh field may benefit from adiabatic radio frequency (RF) pulses such as the semi-localized by adiabatic selective refocusing (semi-LASER) [38, 39] pulse sequence to reduce signal loss from B_1 inhomogeneity. The typically-desirable greater spectral dispersion available at ultrahigh field may exacerbate chemical shift localization (CSL) errors endemic to most MRSI techniques, [40] yet may be alleviated by high spatial bandwidth technique such as the spatial-spectral adiabatic spectroscopic imaging (SASSI) sequence [40].

The two-dimensional (2D) correlation spectroscopy (COSY) technique benefits from the greater spectral dispersion of ultrahigh field along both its directly acquired (F_2) dimension and the indirectly sampled (F_1). This is not the case for the 2D J-resolved spectroscopy technique, as the J-coupling effects it measures are independent of the magnetic field.

2.5 Non-proton and multinuclear Imaging

While greater spectral dispersion may be beneficial to ultrahigh field imaging, it also contributes to greater B_0 and B_1 inhomogeneity for proton imaging applications. This is not so for non-proton applications whose gyromagnetic ratios (γ) tend to be smaller than the 42.58 MHz/T value for ^1H . These include sodium (^{23}Na) (11.26 MHz/T), carbon-13 (^{13}C) (10.71) and phosphorous-31 (^{31}P) (17.24) imaging. As an example, the resonance frequency of sodium at 7T is only 78.82 MHz, comparable to the dispersion of ^1H at 1.5T (63.87 Hz). Furthermore, sodium imaging also receives SNR benefits from the longer T_1 relaxation values at ultrahigh field. Sodium T_1 relaxation rates for brain, blood, cartilage and muscle have been observed to be in the range between 10–55 ms [41], necessitating short TEs for most imaging applications. The tendency for longer T_1 relaxation with increasing field strength, as shown in Table 1, makes the fast relaxation properties of sodium less of a limitation at ultrahigh field.

Because the MR-visible isotopes of those atomic nuclei occur at much lower natural concentrations than proton-containing molecules, their imaging applications tend to be relatively SNR-starved, and thus benefit greatly from the higher sensitivity of ultrahigh field. Non-proton imaging applications are limited by the availability of dedicated RF coils tuned to their particular resonance frequencies, particularly at ultrahigh field. Early development of non-proton ultrahigh field hardware was dominated by custom-built coils by dedicated RF hardware laboratories, but commercial availability of these coils has increased in recent years as manufacturers such as Nova Medical have entered the space.

3.0 Clinical Applications

The added resolution and signal quality of ultrahigh resolution imaging may be used to improve detection, characterization, treatment planning and monitoring for a range of clinical applications.

3.1 Brain Tumor

The greater sensitivity and spatial resolution of ultrahigh field imaging may help elucidate smaller or fainter lesions under contrast enhanced T_1 -weighted MRI, T_2 -weighted fluid-attenuated inversion recovery (FLAIR) imaging or proton density (PD) weighted imaging. The greater contrast and resolution offered by ultrahigh field may facilitate reliable detection of smaller or less well-differentiated lesions. Vessel imaging applications such as SWI or arterial imaging applications such as ASL or TOF may help to characterize tumor vasculature, detect microbleeds induced by radiation therapy [42], identify angiogenesis or identify major arteries near the tumor in the context of imaging-guided surgical planning.

Abnormal metabolism in brain tumors and the surrounding parenchyma may be non-invasively detected by MRSI. With greater SNR and spectral separation, MRSI at ultrahigh field may help in detection of the tumor-specific biomarkers of increasing interest in the era of personalized medicine. For example, specialized spectroscopy techniques including spectral editing and 2D MRS may enable reliable detection of the “oncometabolite” 2-hydroxyglutarate [43, 44], a highly-specific marker for mutation in isocitrate dehydrogenase (IDH1/IDH2, found in a majority of WHO Grade II/III gliomas). Other detected metabolic abnormalities may be markers for tumor activity including lactate, which is a product of anaerobic metabolism, and choline-containing metabolites, particularly phosphocholine, which are markers for membrane turnover. A sensitive and reliable MRSI technique may be used to non-invasively characterize tumor both before and after treatment, such as differentiation of recurrent tumor from pseudo-progression or treatment necrosis.

3.2 Epilepsy

In focal epilepsy, high resolution structural MRI obtained at ultrahigh field may facilitate the detection of small epileptic lesions that are difficult to detect at lower field strengths [45]. Even in non-lesional epilepsy these high-resolution scans may reveal abnormalities in the cortex such as polymicrogyria or abnormalities in white matter such as dilated periventricular spaces (PVSs), which may yield clues as to which cortical structures are pathologically involved [46, 47].

As deep-brain stimulation (DBS), particularly of the thalamus, has gained interest as a potential treatment strategy for non-lesional epilepsy, there is potential for ultrahigh field imaging to assist in the planning of electrode placement. High-resolution structural MRI and DTI may be able to elucidate small subcortical structures and probe their associated cortical connections to highlight potential targets for DBS.

3.3 Psychiatric Disorders

Higher SNR and spatial resolution of structural imaging may facilitate more reliable measurement of potential imaging biomarkers like cortical thickness and volume. Automatic segmentation packages like FreeSurfer and ASHS may leverage high resolution ultrahigh field imaging to more accurately parcellate fine structures such as the subfields of the hippocampus or the nuclei of the amygdala and thalamus. Higher resolution diffusion weighted imaging may also enable more reliable whole-brain tractography including

resolution of crossing fibers to establish connectomic imaging markers for MDD, which may correlate with increasing severity of symptoms [48–51].

MRSI at ultrahigh field may more reliably detect gamma-aminobutyric acid (GABA), the brain's chief parasympathetic neurotransmitter and glutathione (GSH), a marker for oxidative stress. Reductions in GSH and GABA concentration have been implicated in psychiatric disorders such as MDD and schizophrenia [35, 36, 52–54].

3.4 Alzheimer's Disease

A principal goal in imaging applications of Alzheimer's Disease (AD) is the direct or indirect detection of plaque-like pathology by "MR microscopy." [55] Even in applications where this direct characterization may not be possible, high-resolution ultrahigh field MRI may help to identify subtle volumetric loss in cortical or subcortical structures [56] to gain a better understanding of the nature of volume losses and cognitive deficits associated with the disease. Various imaging modalities at ultrahigh field, including T₂-weighted imaging, SWI and QSM, may be used to facilitate mapping of iron associated with amyloid disease [57]. MRS may also assist in the characterization of AD through the detection and quantification of brain metabolites, particularly N-acetyl aspartate and GABA [31].

3.5 Multiple Sclerosis (MS)

Ultrahigh field SWI and QSM applications have attempted to characterize multiple sclerosis (MS) lesions, by detecting the contributions of iron and myelin to their magnetic susceptibility [58]. Lesion subtype may have predictive power in characterizing physical and cognitive deficits in patients with MS [59]. The greater SNR and resolution of structural MRI at ultrahigh field may also help to reliably detect MS lesions too small or faint to detect at lower fields [60, 61]. Better identification may also improve the accuracy and reliability of automated segmentation techniques for detection and quantification of MS lesions [62]. The enhanced anatomical contrast at ultrahigh field may also help to detect MS lesions within the gray matter, where contrast is typically poorer [63].

3.6 Stroke

Signal quality of arterial imaging applications such as ASL and TOF is modulated by both the main magnetic field and the T₁ relaxation time, and therefore enjoys a double benefit at ultrahigh field. This greater signal quality may lead to the improved characterization of the microvasculature and the identification of small occlusions. Comparative studies at ultrahigh field have demonstrated better identification of microbleeds [21] and microinfarcts [64].

4.0 Technical Challenges and Engineering Solutions

There are several physical, technical and clinical challenges involved in scanning at ultrahigh field, with the potential to affect the performance, sensitivity, reliability and safety characteristics of the acquired sequences. Various software and hardware based engineering solutions are in development to meet these challenges including coils with multiple transmit elements to better shape the excitation field or adiabatic RF pulses [45, 65] with greater insensitivity to B₁ inhomogeneity.

4.1 Parametric limitations of Ultrahigh-field Imaging

4.2 B_0 Inhomogeneity

Inhomogeneity of the main magnetic field (B_0 inhomogeneity) scales proportionally with field strength, resulting in both spatial distortions and alterations in signal intensity and contrast. In MRS/MRSI applications, B_0 inhomogeneity may manifest both as a loss of signal and a mis-positioned spectral shift. This shifting may complicate the suppression of competing water and lipid signal and broaden the observed line-widths of metabolite peaks. The resulting overlap with other peaks or water/lipid background may complicate reliable quantification of metabolite signal through MRS. In large voxels, B_0 may contribute partial volume effects to the acquired signal and an overall loss in SNR.

4.3 B_1 Inhomogeneity

Scanning at ultrahigh field can be subject to greater inhomogeneity of the applied radio frequency field (B_1 inhomogeneity) resulting in varying signal intensity and tissue contrast in the acquired image. These artifacts are believed to be modulated by the dielectric properties of the scanned tissue, like conductivity and permittivity. For typical brain permittivity, half-wavelengths of the proton resonance are on the order of 10 cm, smaller than the cross-sectional size of the head. This can create standing wave patterns of constructive (bright) and destructive (dark) interference within the FOV. Furthermore, skin-depth effects and eddy currents opposing the applied B_1 field can cause reduction in the effective B_1 in the peripheral tissue.

Various types of adiabatic RF pulses, including hyperbolic secant pulses [70] and adiabatic Shinnar-Le Roux pulses [45, 71] have demonstrated reduction of B_1 -inhomogeneity effects. In the context of MRS/MRSI, the semi-LASER class of pulses is a commonly-used strategy for B_1 -insensitivity. Because adiabatic excitation and refocusing pulses tend to be higher bandwidth than their conventional counterparts, sequences employing these pulses may introduce additional SAR to the scans. This can be problematic as scanning at ultrahigh field is already subject to higher SAR independent of these pulses meaning their implementation may impose other limitations to the scan parameters, such as longer scan durations.

Parallel transmission (PTx) imaging with segmented readouts may be a hardware-based approach to the problem of B_1 inhomogeneity [72]. Along with a B_1 field mapping approach multiple independent coil transmit elements allow for more degrees of freedom in dealing with the complex interference patterns observed at ultrahigh field. PTx approaches have thus far been limited by safety concerns related to the complex SAR profiles from multiple transmit elements [73]. These interactions will need to be understood and modeled for the individual transmit arrays before ultrahigh field pulse sequences can fully take advantage of this emerging technology for clinical use.

4.4 Increased chemical shift localization artifacts

Because chemical shift dispersion is proportional to B_0 field strength, chemical shift localization artifacts also tend to become proportionally more severe at higher field. These artifacts occur when the effective excitation volume for a given metabolite of interest is displaced from the nominal excitation volume in proportion to its chemical shift and are a

major challenge in reliable MRSI. Acquisition techniques with high spatial bandwidths, such as the spatial-spectral pulses employed by the MR-SASSI technique [40], have demonstrated mitigation of chemical shift localization errors.

4.5 Specific Absorption Rate

RF deposition tends to scale with the square of the B_0 field strength, with RF power requirements for 7T shown to be roughly 1.8–2.0 times higher than needed for 4T. FDA-designated SAR guidelines (3.2 W/kg over 10 or minutes for head imaging) pose a significant limitation at ultrahigh field, limiting the minimum repetition times of RF-intensive techniques, reducing flip angles or spatial coverage, and potentially mandating the introduction of time delays between acquisitions. This is of particular concern in sequences requiring echo trains of multiple 180° RF refocusing pulses such as the turbo spin-echo (TSE) / fast spin-echo (FSE) sequence. Adiabatic pulses sequences tend to deposit more RF energy than their conventional counterparts - for example, in the implementation of 180° refocusing pulse pairs in semi-LASER. This further complicates their implementation as potential solutions to address the B_1 -inhomogeneity issue, and may ultimately lead to longer scan duration.

4.6 Lack of Body Transmit RF coil

At the current time of writing, no human 7T MR system ships with a body transmit coil, making it necessary to integrate RF transmit elements along with the receive elements in an RF head coil. Hardware manufacturers like Nova Medical have introduced 7T head coils with 32 receive elements and either a single transmit channel or eight parallel transmit channels.

4.7 Unconventional Contrasts and Relaxation Properties

Relaxation behavior differs between conventional and ultrahigh field strengths, including longer T_1 values, shortened T_2^* values for most brain tissues and shorter apparent T_2 values. This may complicate direct comparisons to tissue contrast at lower fields and introduce unreliability in automated contrast-based segmentation methods. This may introduce some T_1 -weighting to short TR spectroscopy sequences, also causing a slight loss in SNR.

5.0 Patient Experience

Though FDA guidelines state that main magnetic field strengths up to 8 tesla (8T) are not “significant risks”, there remain some basic considerations for patient safety and comfort at ultrahigh field. Magnet resonance safety studies have thus far not demonstrated adverse effects beyond transient phenomenon such as nausea, vertigo, balance difficulty and metallic taste in the mouth [74]. Among the most widely-reported of these is the sensation of dizziness experienced as the scanning bed moves into and out of the scanner core. As a consequence, the early generation of ultrahigh field scanners was deployed with manually-moving scanner tables and these tables tend to require slow movement to minimize patient discomfort.

Human studies involving 1-hour exposure to high magnetic field have not demonstrated significant changes to physiological parameters such as body temperature, heart rate or blood pressure [75]. Though RF deposition is greater at higher field strengths, scanning at ultrahigh fields conforms to the same safety guidelines on SAR as established for lower field strengths, and these levels may be achieved through adjustments in scan parameters including repetition time, flip angle or the introduction of specifically-designed lower-SAR RF pulses. Ultrahigh field scanners also adhere to the same peripheral nerve stimulation limits initially established by regulatory bodies like the FDA for lower field scanners. This stimulation is modulated by the speed of gradient switching rather than the strength of the main magnetic field, and therefore not observed to behave differently at higher field strengths.

Ultrahigh field MR scanners adhere to the same limits for peak unweighted sound pressure (140 dB) and weighted root mean square (rms) sound pressure with hearing protection in place (99 dB) as scanners at lower field. Noise due to rapidly switching field gradients is primarily modulated by the speed of that gradient switching, and therefore not expected to be higher at higher field strengths. Noise dampening, insulation, hardware and sequence design can all play a role in maintaining sound levels within guidelines for patient safety and comfort. Ultrahigh field scanners tend to have longer bores than their lower-field counterparts and currently lack the wider bores seen in newer generations of 3T MRI scanners. This may exacerbate feelings of claustrophobia when inside the bore. Overall, 7T scanners have been used extensively for human studies and are reported to be well tolerated by patients in research studies.

6.0 Conclusions and Future directions

Despite ongoing technical challenges, the promise of higher signal quality, greater resolution and more reliable quantification has spurred the accelerating adoption of ultrahigh field scanners. As the first generation 7T human scanners enter clinical use, the technology may enable the detection of more subtle structural, physiological and metabolic abnormalities than is detectable at lower field strengths. These sensitivity enhancements may lead to the reliable detection of smaller arteries in stroke, fainter lesions in MS, more peri-ventricular spaces in epilepsy or more subtle metabolic imbalances in depression. Technological advancements such as parallel transmit coils and adiabatic RF pulses may further improve the quality and reliability of ultrahigh field imaging and sophisticated post-processing algorithms such as automatic segmentation and probabilistic tractography may tease out still further detail from these high-quality images. The combination of these technologies will push ultrahigh field scanners into improving the detection, assessment, monitoring and treatment of a range of diseases and disorders.

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Figure 1:
T₁-weighted MPRAGE (right) and showing coronal-oblique section of a healthy volunteer.
Spatial resolution of the T₁-weighted MPRAGE is 0.7 mm³ isotropic.

Figure 2:

Top: T₂-weighted TSE (left) from the same volunteer as figure 1 with a spatial resolution of 0.4×0.4×2.0 mm³. **Bottom:** Zoomed-in inset of right hippocampus from T2-TSE.

Figure 3 (right):
Minimum intensity projection (mip) of an axial susceptibility weighted imaging scan from a healthy volunteer with 0.2 mm^2 in-plane resolution and 1.5 mm slice thickness.

Figure 4:
3D map showing whole brain tractography. Data was acquired using a 64-direction DTI sequence acquired at 7T with 1.05 mm^3 isotropic resolution and $b = 1200 \text{ s/mm}^2$. [47, 51].
Image courtesy of John W. Rutland.

Figure 5:

Tractography of left and right hippocampus-amygdala-thalamus from 64-direction DTI sequence acquired at 7T with 1.05 mm^3 isotropic resolution and $b = 1200 \text{ s/mm}^2$. Shown are fiber tracts which pass through the hippocampus, amygdala and thalamus. Image courtesy of Stephanie SG Brown.

Table 1:

The relationship between various MRI/MRS/MRSI related parameters and field strength (B_0)

Imaging Parameter	Relationship to Field Strength	Reference
Signal-to-Noise Ratio (SNR)	$SNR \propto B_0$	[6–8]
Spatial Resolution	$Resolution \propto SNR^{1/3} \propto B_0^{1/3}$	[9]
T1	$T1 = 1/ (.35 + .64 * B_0^{-0.7})$	[6]
T2*	$T2^* \propto 1/(7 + 3.5 * B_0)$	[6, 10, 11]
Apparent T2	$T_2 \propto 1/B_0$	[10–13]
Spectral Separation	$\delta \propto B_0$	[14, 15]
Susceptibility	$\psi \propto TE * B_0$	[16]

Table 2:

The relationship between various MRI/MRS-related limiting parameters and field strength (B_0).

Imaging Parameter	Relationship to Field Strength	Reference
Specific Absorption Rate	$SAR \propto B_0^2$	[66, 67]
Chemical Shift Localization Error	$CSL \propto B_0 * \text{Slice Thickness/Bandwidth}$	[68]
Susceptibility	$\varphi \propto TE * B_0$	[69]