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Clinical Magnetic Resonance Imaging of Brain Tumors at Ultrahigh Field:

A State-of-the-Art Review

William T.C. Yuh, MD, MSEE, Greg A. Christoforidis, MD, Regina Maria Koch, MD, Steffen Sammet, MD, PhD, Petra Schmalbrock, PhD, Ming Yang, MD, and Michael V. Knopp, MD, PhD

Department of Radiology, The Ohio State University, Columbus, OH

Abstract

With the advancement of the magnetic resonance (MR) technology, the whole-body ultrahigh field MR system operated from 7 to 9.4 T becomes feasible for the routine patient imaging in clinical settings. The associated potentials and challenges from the perspectives of technology, physics, and biology as well as clinical application of the ultrahigh field MR systems are different from those systems operated at 3 T, 1.5 T, or lower field strength. In this article, we will present our initial experiences of brain tumor imaging using the 7 and 8 T whole-body MR systems at the Ohio State University Medical Center and provide a brief overview pertinent to the ultrahigh field clinical MR systems.

Keywords

ultrahigh field whole-body MR; 7 and 8 Tesla clinical MR; brain tumor

In this article, ultrahigh field (UHF) is defined as the magnetic field strength (B_0) of 7 T or higher. Imaging systems for humans are currently used from 7 to 9.4 T. We will focus on the clinical aspects of ultrahigh-field whole-body human magnetic resonance imaging (MRI) systems.

Since the introduction to clinical utility back in the 1980s, human whole-body MR scanners with various field strength have contributed significantly to patient care and became an indispensable tool for the management of most diseases of the central nerve system (CNS), including the focus of this issue, brain tumor imaging. For the past decades, the technical development for clinical MR scanners also has been making remarkable progress particularly in coil design, gradient, and radiofrequency (RF) performance as well as software development. The field strength, or B_0 , of the MR scanners increased constantly from the initial ultralow (<0.1 T) to low field (0.1–0.5 T) around the 1980s and from mid (0.5–1.5 T) to high field (3 T) during the early 1990s to 2000s. Meanwhile, the clinical experiences and applications for various CNS pathologies expand constantly as the field strength of the whole-body MR systems scanners continues to grow.

Until recently, clinical MR scanners operating at field strength (B_0) of 1.5 T have been considered as the optimal means to evaluate neurological diseases. Just few years ago, there

were many doubts about the worthiness and cost-effectiveness to develop a 3-T human system mostly because of technical challenges and cost perspectives. Fortunately, most of the technical problems have been resolved, and 3-T human scanners now become the state-of-the-art imaging modality for CNS diseases. With the improved signal-to-noise ratio (SNR) from the increased magnetic field strength, 3-T whole-body MR systems provide superior neuroimaging and functional capability as compared with those of the commonly available 1.5-T systems. Because the 3-T clinical systems have established tremendous credibility and clinical acceptance for the last few years, currently, the enthusiasm in building even higher field-strength clinical systems becomes the fastest growing segment of the MR marketplace. After the initial demonstration of feasibility at few academic institutions, whole-body MR systems with field strengths of 7 T or higher have become a new forefront of MR.^{1,2} Major MR vendors now embrace the potential of these systems and embark on the development of commercial ultrahigh-field whole-body systems. Therefore, there is a rush for even higher field whole-body human scanners throughout the industry, whereas the optimal field strength for the cost-effective whole-body system for clinical utility remains to be determined. From the neuroimaging perspective, the overall quality, capability, and potential have been consistently improving in a linear fashion with the increase of the field strength of MR scanners up to 3 T. The question remains how high the field strength should go and where the optimal field strength for CNS imaging may be. In this article, we will discuss the potentials, challenges and pitfalls of ultrahigh-field CNS tumor imaging based on our experiences in both 7- and 8-T whole-body MR systems.

ADVANTAGES AND DISADVANTAGES OF ULTRAHIGH-FIELD MR SYSTEMS

Potential Advantages

From the clinical radiologists' perspectives, there are 2 major factors to be considered with higher-field strength MR systems for clinical applications. These include much greater SNR and sensitivity to susceptibility contrast (eg, blood oxygen level-dependent [BOLD] functional MRI [fMRI] and susceptibility-weighted imaging) of the ultrahigh-field systems than those observed in the commercially available lower-field whole-body scanners ($B_0 < 3$ T). Generally, the gain of SNR from the higher field strength provides the most improvement, capability, and potentials that otherwise cannot be achieved with the current lower-field strength MR scanners. However, the gain of SNR is also offset by the major technical and clinical challenges associated with the increase susceptibility artifacts due to B_0 field inhomogeneity near air-tissue interfaces and RF field or B_1 inhomogeneity as the results of the ultrahigh field. Technical challenges and the associated cost have been the main limiting factors for developing an ultrahigh-field MR scanner. With the benefit of increased SNR, the ultrahigh-field strength whole-body MR scanners can afford much higher spatial and temporal resolutions to further improve the anatomical details (Figs. 1–4) and functional capability (Figs. 5 and 6) that otherwise cannot be achieved from conventional lower-field strength MR.

The quality and capability of fMRI have been greatly improved using ultrahigh-field MR systems. With gains in SNR, the quality and potentials of spectroscopy (Fig. 5), diffusion-weighted or diffusion tensor imaging (DWI and DTI, respectively) (Fig. 6), and cortical activation-based BOLD imaging are greatly improved. For spectroscopy, ultrahigh-field systems provide not only the needed SNR but also much better differentiation among different peaks (spectral resolution). Excellent spectra can readily be obtained in a clinical setting during a single acquisition with much shorter acquisition time, larger sampling volume, smaller voxel size, and more voxels (Fig. 5). With the increased SNR and high external field strength (B_0), ultrahigh-field MR provides the capability and opportunity to

study multinucleus spectroscopy including carbon 13, phosphor 31, and fluorine 19. The increased SNR of the ultrahigh-field MR also improves the quality of DWI and DTI (Fig. 6).

Magnetic susceptibility can be either advantageous or disadvantageous for ultrahigh-field MRI. By taking advantage of the susceptibility of deoxyhemoglobin, microvasculature as small as 100 μm can be depicted, and for cortical activation studies, increased magnetic susceptibility effects of ultrahigh-field MR also greatly enhance the BOLD effect.³

Conversely, magnetic susceptibility differences between air and tissue and resultant B_0 inhomogeneity causes image degradation in form of local signal loss, geometric distortions, and banding artifacts.

Potential Disadvantages

Potential disadvantages for the ultrahigh-field human clinical MR systems include technical and clinical challenges as well as higher costs.

Technical Challenges—Despite numerous and difficult technical challenges, significant progress has been made to have ultrahigh field in clinical setting. Technical challenges include, but not limited to, magnetic susceptibility, degrading the homogeneity of the static magnetic field (B_0), and the RF field (B_1) inhomogeneity requiring alternate the design of better coils for RF transmission and systems for signal reception. Several approaches have been proposed to resolve the susceptibility and B_0 inhomogeneity, including gradient compensation, tailored RF pulses, active and passive shimming, and postprocessing.^{4–9} The problem of field inhomogeneity in echo-planar imaging and spiral techniques was likewise tackled by the advancement in of parallel imaging techniques such as SENSitivity Encoding (SENSE),¹⁰ which permits shortening the readout duration.

Homogeneity of RF (B_1) distribution throughout the head is rather complicated and depends upon the shape, orientation, dielectric, and conductive properties of different anatomic regions and tissue types. RF inhomogeneity is more challenging with ultrahigh-field MR systems. It causes differences in flip angles and, therefore, worsens the image contrast, SNR, and overall image quality. Proposed solutions include the use of shaped RF pulses and multiport excitation of TEM coil or the use of multiple transmit coils analogous to multiple receive coils in SENSE imaging.¹⁰

Clinical Challenges—There are many clinical challenges associated with ultrahigh-field MRI including higher specific absorptions rates (SAR), changes in tissue relaxation times, increased susceptibility effects, resultant changes in contrast mechanisms, and potential safety issues.

Changes in Clinical Image Appearance—Tissue relaxation times and susceptibility change substantially with the field strength MR. Tissue T1 relaxation times increase with field strength, whereas T2 relaxation times as measured with multiecho spin-echo sequences remain approximately constant, resulting in different imaging appearance. Because of susceptibility effects, T2* relaxation times become significantly shorter at ultrahigh-field MR. Another important finding is that at 8 T, multiecho T2 are nearly twice as long as single echo T2 as shown in Table 1.¹¹ Similar observations regarding T2 were reported by Bartha et al.¹²

Specific Absorptions Rate Issues—Compared with lower-field strength systems, a much higher RF and SAR is associated with the operation of the ultrahigh-field MR systems. Pulse sequences requiring high RF excitations such as fast spin-echo, fat saturation,

or magnetization transfer sequences are limited by SAR at ultrahigh-field system. The advancement of parallel imaging techniques can help resolve the SAR issues associated with ultrahigh fields.

Safety Issue—Magnetic forces increase greatly with ultrahigh-field MR systems and increase danger from ferromagnetic projectiles as well as increased torque on implanted medical devices, thus leading to a greater potential to cause serious injuries. In addition, it is known that the incidence of the magnetohydrodynamic effects and human response increases with the magnetic field strength. Nausea, vertigo, headache, tingling, numbness, visual disturbances (phosphenes), and discomfort associated with tooth fillings have been reported in patients moving in high fields.^{17–19} These effects are transient and disappear after leaving the magnetic field and are usually reduced or avoided by making sure the patient moves slowly while in the main magnetic field. Although there have been no reports of health hazards in the literature with physiological exposure to ultrahigh-field MR systems, the long-term effects remain to be determined.²⁰

CLINICAL APPLICATION

Ultrahigh-field human MR systems have come a long way with the rapid advancement in MR technology to overcome many of the technical barriers and resolve these challenges. Even with the limited clinical experiences in ultrahigh-field MR, there are early evidences and potentials to gain further improvement in CNS imaging, including fMRI, spectroscopy,^{21,22} and DWI and DTI.

Pulse Sequences

Gradient- and spin-echo (T1, T2, and proton density) are the most commonly used pulse sequences for brain imaging, including tumors at ultrahigh-field MR scanners.

Spin-Echo Pulse Sequence

Spin-echo images are profoundly impacted by RF (ie, B_1) and B_0 inhomogeneity and changes in tissue relaxation times: T1 relaxation times increases with field strength; with the increase of T1 relaxation time, a longer TR must be used for the T1-weighted imaging for ultrahigh-field MR systems, and because of the flip-angle variability, true T1 contrast may not be achieved. T2 relaxation times for multiecho pulse sequences remain approximately constant, whereas singleecho T2s are approximately 50% shorter. This is because water molecules diffuse in the magnetic field created by magnetic entities such as deoxyhemoglobin in blood or parenchymal ferritin. As a consequence, different water molecules “see” different magnetic fields, resulting in irreversible dephasing. The longer the time between repeated RF pulses, the more the dephasing, and because the time between RF pulses is longer in a single-echo sequence (ie, $TE/2$) than in a multipulse sequence, such as a CPMG sequence for measuring T2 or the turbo spin-echo imaging sequence, different sequences have effective T2. Thus, T2 contrast depends on the type of spin-echo sequences.

Gradient-Echo Pulse Sequence

T2* is significantly shortened at ultrahigh fields, which can be advantageous for the microvasculature, microhemorrhage, or BOLD imaging. However, a shortened T2* value also contributes to loss of SNR for the longer TE gradient-echo acquisitions and causes challenges for echo-train-based acquisition techniques, such as echo-planar imaging. Studies in post mortem specimen have demonstrated that 8-T images with spatial resolutions of $0.25 \times 0.25 \times 2.0$ mm can depict vessels less than 100 μm .²³ This contrast mechanism is analogous to the BOLD effect used in fMRI, and thus susceptibility-weighted imaging has also been called BOLD venography. It was shown that phase rather than magnitude

reconstruction can further improve susceptibility-weighted images at 1.5 and 3 T, and even more at 7 and at 8 T.²⁴ Gradient-echo and turbo gradient-echo sequences with very short TE (<3 milliseconds) can minimize T2* effects and can be used to generate T1 or proton density-weighted images. Inversion recovery prepared turbo gradient-echo sequences have especially shown very promising high-quality T1-weighted images.

Magnetic Resonance Spectroscopy

Magnetic resonance spectroscopy has been extensively used to study of human cerebral metabolism, both in health and in disease at 1.5 and at 3 T.²⁵ Magnetic resonance spectroscopy provides in vivo noninvasive characterization and quantification of tissue metabolites. Clinically, its major impact has made it possible to investigate metabolic changes associated with brain tumors, stroke, epilepsy, and multiple sclerosis. Magnetic resonance spectroscopy, which includes single-voxel and multivoxel techniques, has a role in diagnostic imaging. With single-voxel spectroscopy, the concentration of some metabolites can be estimated in a volume of tissue. The potential of in vivo single-voxel magnetic resonance spectroscopy at ultrahigh fields was demonstrated in several studies.^{1,26} In multivoxel technique (chemicals shift imaging), the metabolite concentrations are mapped voxel by voxel within the volume of interest that is sampled. The multivoxel technique allows the metabolite concentration to be displayed as an image. Chemical shift imaging using high-field MR system can greatly increase the sensitivity and spectral resolution from the much-needed improvement in differentiation, characterization, and quantification of molecular markers for the management of a variety of neurological diseases.²⁷ High-order shimming is necessary at ultrahigh fields to reduce broadening of spectral peaks induced by stronger susceptibility effects at higher field strengths. Chemical shift imaging at ultrahigh magnetic fields is possible and increases the sensitivity and spectral resolution in magnetic resonance spectroscopy.²⁸ Spin-echo slice excitation and stimulated echo acquisition mode (STEAM) for chemical shift imaging at 7 T allows the acquisition of highly resolved MR spectra of the human brain in vivo. For STEAM SI excitation pulses with larger bandwidth reduced chemical shift displacement. PRESS localization is not applicable for signal intensity (SI) at ultrahigh fields because of a large chemical shift displacement.²⁸ Resonances of choline, creatine, *N*-acetylaspartate, and *myo*-inositol could be easily identified in different chemical shift imaging voxels. Chemical shift imaging with short echo time resolved additional resonances in the spectra such as glutamate and glutamine (Fig. 5). SI at ultrahigh field requires distortion correction of the data to improve the noninvasive characterization and quantification of molecular markers with clinical utility for improving detection and treatment of a variety of neurological diseases.²⁸ The tremendous gain in SNR at 7 T compared with the lower field strengths of today's clinical MR scanners will boost ultrahigh-field MR Spectroscopy into many clinical applications.

Diffusion Imaging

Diffusion tensor imaging allows the observation of molecular diffusion in tissues in vivo and therefore assessment of the structural organization. Although holding vast potential, DTI with single-excitation protocols still faces serious challenges: limited spatial resolution, susceptibility to magnetic field inhomogeneity, and low SNR are the most prominent limitations. These shortcomings can be effectively resolved by the transition to parallel imaging technology and high magnetic field strength.²⁹ DTI using the parallel imaging technique SENSE improves image quality at ultrahigh magnetic fields.²² Diffusion tensor imaging (Fig. 6) is an emerging and promising tool to provide information about the course of white matter fiber tracts in the human brain. Fiber tractography from ultrahigh fields provides a promising method for exploring the neuronal connectivity of the brain. Applications limited by the lower SNR at 1.5 T such as DTI have shown great improvement from the higher SNR and resolution of increased magnetic fields. DTI at ultrahigh fields can

be applied to study white matter diseases (multiple sclerosis, progressive multifocal leukoencephalopathy, leukodystrophies) and other diseases such as brain tumors that affect the integrity of white matter structures.^{22,30}

Ultrahigh-Field Brain Tumor Imaging

The goals of brain tumor imaging using ultrahigh-field clinical systems are to first establish a reliable diagnostic tool with high spatial resolution and functional capability to be readily implemented in clinical settings as well as to improve the detection, delineation, and assessment of intrinsic characteristics of tumors for treatment strategy. The ultrahigh-field clinical MR scanner, even with limited clinical experience, has shown several promising potentials aiming toward further improvement for all these goals as compared with the commonly available lower field MR systems.

Higher grade or more aggressive glioma is known to have higher incidence of microscopic evidence of neovascularity, disruptive internal texture, and hemorrhage. Similarly, early small brain metastasis also has high incidence of microscopic changes such as hemorrhage or hypoxia (BOLD effect). With the increase of susceptibility and spatial resolution, ultrahigh-field MRI provides an excellent opportunity to assess the microarchitecture, microvasculature, and microhemorrhage associated with glioma or metastases.

Microvasculature

Tumor vessels are frequently more tortuous, larger, and disorganized than normal blood vessels in the brain. Angiogenesis is considered a marker for tumor aggressiveness in astrocytic neoplasms, specifically glioblastoma multiforme (Figs. 1 and 2). Accordingly, histopathologic assessment of tumor vessels using endothelial cytology or staining techniques has been used to quantify angiogenesis and have been shown to correlate with disease-free survival.³¹ By taking advantage of the greater sensitivity to magnetic susceptibility effects induced by the ultrahigh field, gradient-echo sequences can depict vessels as small as 100 μm (Figs. 1 and 2) and have been used to evaluate for the neovasculatures of the brain tumors.^{32,33}

Tumor growth is believed to be dependent on angiogenesis. Folkman proposed that without angiogenesis, tumor growth is limited to several millimeters in diameter.³⁴ Furthermore, tumor cells are believed to induce angiogenesis by secreting a number of growth factors (angiogenesis factors), which stimulate endothelial cells.^{34–37} Glioblastoma has been shown to be one of the most vascularized tumors in humans and pretreatment assessment of the status of neovasculatures may provide insights in the aggressiveness of the tumor for prognostic assessment and treatment planning.³⁸ One therapeutic strategy for patients with brain tumors involves the development of agents that modulate angiogenesis.^{34,37,38} Validation of techniques that monitor angiogenesis is the key to the development of such strategies. It, therefore, follows that the imaging identification of angiogenic vessels may serve as a marker for response to antiangiogenesis treatment. Histopathologic methods grading angiogenesis have been based on the number of vessels, degree of glomeruloid vascular structure formation, and endothelial cytology.³⁸ More recently, microvascular density (MVD) counting using panendothelial staining techniques (ie, factor VIII, CD34) have been used to quantify angiogenesis and have been found to act as independent prognostic factors.^{31,39,40} Therefore, an imaging assessment of vascular density within a tumor bed may be an indicator for tumoral angiogenesis. Similarly, follow-up with the angiogenesis of the treated tumor may also provide early signs of treatment failure and thus provides a window of opportunity for alternative treatment strategy.

Microarchitecture

Brain tumor imaging obtained with ultrahigh field provides exquisite spatial resolution to resolve lesions with anatomical size as small as 200 μm (Figs. 1–3). Microscopic internal tissue structures that cannot be appreciated by the conventional lower field MR systems can now be appreciated with ultrahigh-field MR. These include microscopic septations, compartments, fascia planes, and orientations. Evidence of disruption or destruction of these microarchitectures is highly suggestive of a high degree of aggressiveness or malignancy, whereas those lesions with intact microarchitecture are likely representing a more benign nature of the tumors (Fig. 3).

Microhemorrhage

Microscopic bleeding frequently occurs in brain metastasis even in very early stage small seeding.⁴¹ With the increased susceptibility of the ultrahigh-field MR, very small metastases that are not apparent on the contrast MR study obtained from 1.5-T MR scanner (Fig. 4) can now readily be appreciated. The improved detection rate of the ultrahigh MRI is likely because of much higher SNR, increased sensitivity to magnetic susceptibility, and spatial resolution when compared with those obtained from the lower-field strength conventional MRI. It should also be noted that the size of lesions demonstrated with the ultrahighfield MRI is larger than that from a conventional 1.5-T postcontrast T1 MR study. The reasons are unknown and may be related to the exaggerated size from the susceptibility artifact induced by ultrahigh-field MRI and partial volume effects of these very small lesions on the conventional contrast study.

CONCLUSION

Significant technical advancement has been made to overcome many challenges related to ultrahigh-field whole-body systems with field strengths from 7 to 9.4 T. The implementation of ultrahigh-field whole-body systems in the clinical setting becomes feasible to provide many advantages and potentials for imaging of brain tumors, although there are many technical and clinical challenges yet to overcome. For the clinical applications with this rather new and advanced imaging tool, physicians such as radiologists, oncologists, surgeons, as well as scientists need to be familiar with the new and different appearance of those MR images, underlying tissue biology, and MR physics. By taking advantage of increased SNR, spatial resolution, and susceptibility effects, ultrahigh-field MR can further improve the efficacy of neuroimaging. It greatly improves the capability to detect particularly small metastases and to delineate the extent of tumor involvement as well as to characterize intratumoral microenvironment by depicting the neovasculature, integrity of the fine architecture, and microscopic bleeding suggestive of the aggressiveness of the tumors. Similarly, ultrahigh-field whole-body MRI greatly improves the quality and capability of DTI, BOLD imaging, and spectroscopy including multinuclear capability. The disadvantage of the much-prolonged tissue T1 relaxation time and susceptibility are partly resolved with fast imaging techniques including parallel imaging acquisition. Conversely, the longer T1, in combination with the increased SNR, may be the reason for the significant improvement in depiction of small arteries with 7-T time-of-flight MRA. The value of MR contrast agents is mainly used for increasing T2* effects, whereas the efficacy of T1-shortening contrast agent, such as gadolinium-labeled contrast agents, commonly used in the clinical practice, remains to be studied.

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FIGURE 1.

Normal vessels and neovasculation in a high-grade glioma. Comparison is made between T2-weighted FSE images of 8 T (A and C, 1024×1024 matrix size) and corresponding images from 1.5 T (B and D, 192×256 matrix size). Superior imaging quality of the 8-T images (A and C) over those of the 1.5-T becomes obvious as judged by the better SNR, spatial resolution, and visualization of the microvasculature (susceptibility). The ultrahigh-field MR images readily show the abnormal microvasculatures with irregular and distorted configuration (small white arrows, A and C) that cannot be appreciated by the 1.5 T-MR images (B and D). Similarly, the ultrahigh-field MR depicts the normal microvasculatures with relatively regular shape and straight course (black arrowheads, A and C), whereas the 1.5 T can only demonstrate the relatively larger sized branches of the middle cerebral arteries (black arrowheads). Images were cropped to demonstrate microvascular structures. 3V indicate third ventricle; CS, centrum semiovale; VOG, vein of Galen.

FIGURE 2.

Neovasculature in a high-grade pleomorphic xanthoastrocytoma. Superior imaging quality with high SNR and spatial resolution of the gradient-echo MR image (A) obtained by the 8-T MR scanner can be appreciated by comparing with the contrast MR studied with a 1.5 T-MR scanner (B). Microvasculature of the neovascularity of this high-grade tumor (circle, A) and the fine venous structures of the choroid plexus can be readily appreciated on the ultrahigh-field MR image. BVR indicates basal vein of Rosenthal; Chor, choroid plexus; MCA = middle cerebral artery.

FIGURE 3.

Fine microarchitecture of a low-grade tumor without neovascularity. A T1-weighted contrast MR image (A) obtained from a 1.5-T MR scanner shows a nonenhancing grade II oligodendroglioma. The 8-T MR images demonstrate fine septations within the low-grade glioma on the FSE T2-weighted image (B) and absence of abnormal focus of neovascularity on both FSE (B) and gradient-echo (C) images.

FIGURE 4.

A, Detection of early brain metastasis with ultrahigh-field MR. Multiple small metastases from breast cancer are demonstrated on the 8-T gradient-echo image (arrows and arrowheads) as focal hypointense lesions near the gray-white junction. B, Only 2 lesions are seen on the 3-dimensional gradient-echo volume contrast MR (arrow and arrowhead) obtained from a 1.5-T MR scanner. An incomplete “ring enhancement” (B, white arrowhead) of a “single” lesion demonstrated on the 1.5-T contrast study is not a typical finding for metastasis and is more consistent with the enhancement pattern of multiple sclerosis. This actually is caused by the partial volume effect of the 1.5-T MR images from a cluster of 4 very small lesions that are clearly shown on the ultrahigh-field MR image (A, white arrowheads). Overall, the lesion size on the 8-T MR images is much larger than that on the 1.5-T MR images, which was likely caused by the increased susceptibility artifact on the 8-T MR image and partial volume effect on the 1.5-T MR image, respectively.

FIGURE 5.

A, Chemical shift imaging of the brain of a healthy human volunteer at 7 T using spin-echo slice excitation with a long echo time (TE = 144 milliseconds). The resonances of NAA, Cr, and Cho (curve fits are shown in blue) are resolved and allowed to calculate metabolite images (color-coded overlay on anatomical images). B, Chemical shift imaging of the human brain at 7 T using STEAM with a short echo time (TE = 18 milliseconds) resolves additionally to the main resonances of NAA, Cr, and Cho also metabolites with smaller amplitudes in the spectra like lactate, glutamine and glutamate.

FIGURE 6.

Diffusion tensor imaging and fiber tracking of the human brain at 7 T. Diffusion tensor imaging and fiber tracking of the brain of a human healthy volunteer at 7 T. The images show directional color coded white matter nerve fiber tracts of the human brain overlaid on anatomical images.

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TABLE 1

Relaxation Times at 7 and 8 T

	PD*	T1†	T1	T2 Multiecho‡	T2 Single Echo§	T2*
Frontal WM	0.69	1017	1300	74.5 ± 3.4 (n = 3)	39.0 ± 1.6 (n = 6)	20.5 (n = 1)
Motor GM	0.80	1654	1800	74.3 ± 9.2 (n = 2)	37.5 ± 1.1 (n = 5)	20.3 (n = 1)
Frontal GM				71.4 (n = 1)	44.8 (n = 1)	

* Proton densities are averages of values listed by Tofis. 13

† The 7-T data superior midbrain regions are from Wright et al. 14

‡ CPMG multiecho sequence with 20-millisecond interecho time.

§ Single spin-echo GESSE sequence with TE = 50 milliseconds. 15

|| Blipped multi gradient-echo slice excitation profile imaging. 16

WM indicates white matter; GM, grey matter; CPMG, Carr-Purcell-Meiboom-Gill; GESSE, gradient echo sampling of the spin echo.