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Advances in Magnetic Resonance Neuroimaging

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MR NEUROIMAGING AT HIGH FIELD

Perhaps the most significant reason for the new advances in magnetic resonance (MR) has been the rapid development of magnet and field gradient designs at higher field strengths of 3 tesla (T) and more recently, 7 T.^{1–5} From its beginning in the 1980s, MR as an imaging method suffered from inherently low signal intensity (or signal-to-noise ratio [SNR]), in which approximately 5 hydrogen protons of every 1 million within a voxel contributes to the MR signal at 1.5 T. At higher magnetic fields (B_0) of 3 T for example, that number of protons contributing to the MR signal is 10 of every 1 million, which means that the all-crucial SNR in MRI is also approximately linear with the strength of the main magnetic field B_0 , such that the raw SNR from 1.5 T to 3 T is about a factor of 2, and a factor of 4.7 at 7 T. Why is this important? Two reasons: The only effective way to increase the SNR at a given field strength is to increase the number of signal averages, which at 1.5 T would be a fourfold increase in the number of averages, which typically translates into a fourfold increase in the scan time. A second important reason is that the extra SNR at higher fields can be used to trade off for other image enhancements, such as higher resolution, shorter scan times, and shorter echo times (TEs), to name only a few trade-offs. This second reason explains why the phenomenon of “parallel imaging” in MRI has accelerated the field so rapidly with the advent of higher magnetic field strengths; it facilitates these trade-offs.

The purpose of this article is to highlight recent advances in neuroimaging from two aspects: (1) those advances directly benefited by increases in field strength (increased T1, SNR, magnetic susceptibility-sensitivity, and chemical shift) and how the increased SNR can be used to trade off for other advantages and (2) those advances made in response to attempts to try to reduce the inherent artifacts encountered at higher field strengths (eg, reducing specific radiofrequency (RF) absorption in tissue and magnetic susceptibility).

Other game-changing inherent advantages of higher field strengths are the linear increase in the chemical shift, which, along with increased SNR at high field, has made MR spectroscopy more of a routine clinical use.^{6,7} In addition, longer proton relaxation times, T1s, are observed at higher fields, again typically a linear response. Because of the longer T1s, inflow methods such as MR angiography and postcontrast agent T1-shortening effects are better visualized at higher fields.^{2,8} The longer T1 values have also been important in the development of arterial

spin labeling or non-contrast-enhanced perfusion MRI for clinical use in such diseases as stroke when used at high field.^{9,10}

Looking back, perhaps the biggest reason why fields of 3 T and higher became important throughout the last decade was the improvement of the blood oxygenation level dependent (BOLD) approach used for functional MRI (fMRI). For fMRI, the sensitivity created by the nearly fourfold increase in T2* sensitivity at 3 T over 1.5 T (where T2* can be thought of as the relaxation time T2 measured in the presence of magnetic susceptibility effects such as blood, air, contrast agents, and so forth) has allowed for crucial fMRI advances, in single-event studies for example. Whether fMRI has made, or will make, a true clinical impact in neuroimaging is debatable; however, other blessings in disguise will likely overshadow the fMRI clinical advantages, to such an extent that 3 T is rapidly becoming, or has already become, the de facto standard field strength for clinical neuroimaging¹¹ for the reasons discussed below.

Not all T2* effects lead to discussions regarding fMRI. Magnetic susceptibility effects from increased T2* sensitivity at high fields,¹² although initially a curse of high-field MR (for causing image distortion artifacts, among other things) are today the means for cutting-edge MR applications such as susceptibility-weighted imaging (SWI) or T2*-sensitive imaging for detection of bleeds, calcium, oxygen, or hemosiderin. In other words, T2*-weighted MRI at 3 T is becoming a key sequence relevant for detection of microbleeds in vascular encephalopathies or for visualizing cerebral hemorrhage.^{13–16} Increased T2* sensitivity in high fields leads to more “dephasing” of the proton signal, depending on the proton microenvironment. The dephasing can be mapped using a phase-sensitive image reconstruction method. (The authors normally only map the magnitude of the proton signal).

Why the interest in high-field dephasing of the proton signal? It is simply that phase-sensitive methods can better visualize the microscopic effects of vascular blood oxygenation changes (eg, in stroke or in vascular disease) that might occur at resolutions below the threshold for MR angiography. SWI at high field has considerable potential^{13–18} in visualizing tissue oxygenation changes on a vascular level, ideal for venography. By mapping the proton phase changes due to the susceptibility T2* effects of iron, SWI becomes sensitive to small venous vessels. More broadly, phase-sensitive imaging will lead to an entirely new class of tissue contrast mechanisms based on the proton microenvironment and to the frequency shifts experienced during image acquisition, shifts that are sensitive to tissue oxygenation, structure, and heterogeneity. One may expect to see phase-sensitive MRI (and SWI) be used for amyloid deposits, calcifications, white matter heterogeneities, and other vascular abnormalities (Fig. 1).

THE RAPID RISE OF PARALLEL IMAGING: THE ONGOING REVOLUTION IN MRI

Since the early realization nearly 2 decades ago that multiple RF coils placed in an array could lead to increased coverage with improved resolution, the many new promising approaches to enhanced use of multiple coils have been termed “parallel imaging” (PI),^{19–22} all of which, in essence, permit the various trade-offs of SNR to speed or resolution per image scan time.

The underlying simplistic concept in PI lies in understanding that the reduction of phase-encoding steps necessary for image acquisition can be achieved by using multiple RF receiving coils, each coil being responsible for only a part of the image or the field of view, which has numerous advantages. PI can, for example, reduce the number of excitation pulses needed, thereby decreasing the specific absorption rate (SAR). With multiple coils, each can acquire the signal faster and in shorter TEs, reducing artifacts, distortions, and noise.

PI can, because of the multiple coil array, better detect and correct for motion because proton phase due to motion will affect each coil differently.²² In essence, each coil detects a different signal because of the motion, which can be monitored even on the fly and used to correct for rigid body motions (Fig. 2). This generalized approach opens the way for routine clinical motion correction and allows for motion navigation schemes between echo trains or “shots.” The use of motion navigation is essential for multi-shot high-resolution DWI (below).

The trade-off with PI is that the SNR will diminish. However, the SNR gain at 3 T easily justifies the cost. Sequences using PI at 3 T can, for example, produce image resolution and SNR similar to what one would expect from 1.5 T, but the entire image acquisition is four times faster or is “accelerated” by a factor of four.²² Likewise, one can acquire images with much higher resolution in similar scan times. As one would imagine, extensive (and exciting) trade-offs exist in PI regarding the number of coils, the coil arrangements, the desired resolution or scan time, and the number of echo acquisitions per echo train.

Not only is PI a powerful method of decreasing study times and thus reducing patient motion, it allows for significantly shorter TEs for sequences involving series (or “trains”) of spin echoes or gradient echoes (such as fast spin-echo [FSE] or echo-planar imaging [EPI]). The shorter TEs result in fewer distortions, and thus improved image quality, and also decrease blurring in FSE and EPI scans while reducing geometric distortions in EPI-used diffusion-weighted image (DWI).²³

Again, the key to understanding PI is the simple realization that the spatially inhomogeneous coil sensitivity profiles of individual surface coils can be stitched together by proper calibration routines to yield a larger two-dimensional Fourier Transform (2DFT) image or one requiring less scan time. For three-dimensional (3D) volumetric sequences, the speed-up acceleration can be applied to both phase-encode directions. Popular PI methods, such as SENSE²⁴ and GRAPPA²⁵ and many of its variations, will continue to be applied to yield numerous advantages. It is hard to imagine sequences that would not benefit from PI. Finally, although this discussion has been limited to methods of receiving the signal with multiple RF coils, one may expect similar ideas to apply to entirely new parallel RF coil excitation methods that will eliminate the need for the large and expensive homogeneous RF body or head volume coils now necessary for imaging in any field strength.^{22,26,27}

TAMING ARTIFACTS IN MRI ALWAYS LEADS TO NEW ADVANCES

Throughout the history of MR, 1 year’s imaging artifact becomes next year’s new and novel technology. One prime example has been the age-old issue of attempts at reducing the amount of RF power that a tissue absorbs (SAR), which is a major issue at high fields (the proton Larmor frequency at 7 T is about 300 MHz) and leads to local tissue heating effects, uneven tissue proton excitation (dielectric effects), and image intensity shading.^{28–35} Because RF energy deposition is proportional to the square of flip angle,³³ traditional sequences such as FSE or turbo spin-echo (TSE) sequences, which use long trains of 180° refocusing RF pulses, have been difficult to achieve even at 3 T^{28–30} because of excessive SAR. It took some time to realize that even small reductions of the flip angle lead to significant SAR decreases because at first glance, flip-angle reductions would result in a reduction in signal intensity, limiting the gain from higher field strengths.³⁰

HOW IT ALL STARTED: HYPERECHOES!

Higher field strengths in MR need more RF power for proton excitation at the inherently higher frequencies. More power of the RF pulses at higher frequencies will lead to intolerable SAR levels. Because of this, sequences such as FSE (a workhorse of neuroimaging) have to be modified or simply eliminated,³⁰ which is unfortunate because spin-echo sequences are

inherently superior in image quality because of the inherent correction for magnetic field inhomogeneities afforded by the refocusing 180° RF pulse. Excessive SAR with FSE also explains why higher-resolution 3D volumetric spin-echo sequences have not been available at 3 T or higher. Gradient-echo (GRE) 3D volume sequences are used instead because they minimize SAR (by eliminating the need for 180° refocusing pulses); however, the increased $T2^*$ sensitivity inherent in GRE sequences is not desirable in many neuroimaging protocols.²³

In 2001, Hennig and colleagues^{36,37} introduced a new methodology involving long trains of spin echoes using a new spin-refocusing strategy using “echoes” of echoes (termed “hyperechoes”). In a typical hyperecho long-train TSE acquisition (TSE is an equivalent of FSE), spin-echo train lengths of 16 and above can be acquired at 3 T at a preserved SNR but while reducing the SAR by 70% or more.

Although not immediately applicable for clinical use, hyperechoes nonetheless led to new advances in the creation of new and much more efficient high-field MR pulse sequence families. How do they work? By noting that the center of “k-space” (k-space is a plot of the frequency and phase changes that occur on a line-by-line basis in an image) contains most of the SNR and image contrast, these newer sequences reduce the excitation flip angles only for the outer or less important k-space lines while properly exciting the central k-space lines with higher flip angles, which reduces SAR dramatically while preserving much of the SNR and tissue contrast.³⁸ Because of this manipulation of the spin-echo train, entirely new 2D slice and 3D volume sequences can be built around the spin echo with all of its inherent advantages in neuroimaging while keeping the SAR within clinical limits.

MANIPULATION OF THE SPIN-ECHO TRAIN: XETA (CUBE), SPACE, VISTA

Efficient and safe 3D spin-echo sequences for neuroimaging at high field have always been in demand^{4,30–33} despite being limited by SAR and long scan times. New variations of 3D long echo train FSE or TSE sequences are now being introduced with not only a great deal of publicity but also a good deal of promise. These are called CUBE (formally FSE-XETA), T2-SPACE, and VISTA, to perhaps draw more attention to the 3D potential. These differ from conventional FSE sequences by allowing extremely long spin-echo trains of up to 200 frequency-encoded echoes obtained at minimum echo spacings.³⁸ Acquiring 200 phase-encoded lines within each repetition time (TR) is amazing; moreover, PI methods allow each echo to be frequency encoded with 512 points or more, resulting in 512×512 images to be acquired within a minute. This rapid acquisition speed can be thus traded off for a true 3D image acquisition scheme within a reasonable scan time of just a few minutes, which, in turn, results in a true volumetric 3D spin-echo sequence of sufficiently high resolution to allow for efficient and diagnostically useful reformatted images in a manner similar to that used today from 3D volumetric GRE sequences or from multidetector CT. Even curved planar reformats imaging can be conceptualized. Perhaps more importantly, the advent of 3D volumetric spin-echo images can provide diagnostically proper tissue contrast inherent from spin-echo images relative to the often mixed gray/white tissue differentiation seen from comparable GRE sequences (Fig. 3).

The secret behind these sequences is the alteration or modulation of the flip angle amounts during the FSE readout. The flip angles are modulated to preserve the echo train magnetization as long as possible while avoiding blurring and providing optimal signal intensity at an effective TE. Modulation may also be done to vary tissue contrast or to minimize TEs while also allowing for remarkable reductions in the SAR.

What's in a Name?

Within each new pulse sequence family comes the usual myriad of trademarked sequence names. Thus, for example, the 3D spin-echo sequence described above is called CUBE, T2-SPACE, or VISTA, depending on the MR scanner manufacturer. Complicated for physician, technologist, and physicist alike, sequence families and trademarked names have long since been frustrating for nearly all involved. How does one stay abreast of the thousands of common and trademarked sequence and method names? Although the published literature has always been a significant source and today more easily accessed because of archives such as PubMed (<http://www.pubmedcentral.nih.gov/>), several on-line resources exist that have truly made the effort easy and informative. These sites include (but not are exclusive to) <http://www.mr-tip.com>, <http://www.ismrm.org>, and <http://www.users.on.net/~vision/>. More to the point, these sites and those linked therein provide current relevant information on all things MR related, from safety information to manufacturer specifications.

PROPELLER (OR THE RE-EMERGENCE OF THE SPIN ECHO)

Spin-echo and FSE sequences have long been favored over GRE in imaging of numerous areas, such as the skull base and spine.³⁹ FSE is, in turn, preferred over conventional spin-echo sequences in that the image is acquired several “lines” at a time afforded by the train of spin echoes, or the FSE “blade.” One particular problem with T2-weighted FSE sequences, though, has been motion ghosting caused by cerebrospinal fluid (CSF) flow that can occur during the acquisition of the train or “blade.” Worse still, multiple-shot spin-echo trains or “blades” have never been suitable for higher resolution diffusion-weighted FSE because of the motion sensitivity of the diffusion-sensitizing gradients on the blades. For this reason, motion-sensitive sequences such as DWI have always required data acquisition in a single shot using the GRE EPI.

To overcome the FSE inherent motion sensitivity, Pipe and colleagues^{40–43} have used radial sampling of the echo train “blades” to produce a unique k-space acquisition scheme, “PROPELLER,”^{40–42} which is much less sensitive to these types of distortions and improves the diagnostic quality of such sequences. PROPELLER sequences are becoming extremely useful in reducing distortions, ghosting, and motion sensitivity in spin-echo sequences to such an extent that multishot diffusion-weighted sequences were first clinically viable using this radial acquisition FSE method.^{42,43} Because each spin-echo train or blade is acquired in a radial distribution about the center of k-space (by changing the direction of the frequency- and phase-encoding gradients), the exact frequency and phase differences between blades can be measured; any such deviations are due to motion and can be corrected for or “navigated,” resulting in a reduction in motion sensitivity so robust that intravoxel motion such as diffusion can be reliably measured and mapped (Fig. 4). One may expect to see many more PROPELLER sequences, applications, and variations.

THE ADVENT OF THE BALANCED GRADIENT-ECHO SEQUENCES: BEYOND STEADY-STATE FREE PROCESSION

The family of steady-state free precession (SSFP) GRE sequences uses a balanced gradient scheme.^{44,45} The use of balanced gradients can return or balance any proton signal on resonance between consecutive RF pulses to a constant given phase. A balanced GRE sequence is always initiated with an RF pulse of 90° or less, which, when applied rapidly enough, can bring the (predominantly long-T2) spins to a “steady state” of magnetization. Importantly, before the next TR is started, gradients along the slice-encoding, phase-encoding, and frequency-encoding directions are always balanced so their net positive and negative value is zero. By doing this, the proton magnetization is maintained from shot to shot and can actually

become part of the next TR magnetization. Because the gradient balancing can build up and maintain maximum transverse magnetization, the balanced GRE sequences can produce a near maximum of T2-weighted signal strengths such as that found in CSF and long T2 fluids.⁴⁶ In other words, the balanced GRE sequences excel in rendering long T2 fluids hyperintense. On the flipside, however, the tissue contrast result is often a mixture of T1 and T2 contrast, a persistent problem with many GRE sequences. Although the primary use of balanced GRE sequences has been in cardiac and body MR to detect long T2 fluids, these sequences are poised to provide new information in CSF and blood flow dynamics in neuroimaging.^{47–50}

What has been desirable, however, is a high-field sequence that could produce images with dramatically increased SNR from long T2 fluids while retaining much of the T1-weighted tissue contrast. These have now recently appeared⁵¹ because of newer developments in pulse sequences that allow for dynamic changing of RF pulses during the echo train by ramping the flip angles up and down during the acquisition. One may consider this as the GRE equivalent of what CUBE, T2-SPACE, and VISTA did for the spin echo (see earlier discussion). One of the new GRE sequences is termed “COSMIC” (coherent oscillatory state acquisition for the manipulation of image contrast) and represents improvements over traditional balanced GRE sequences such as SSFP or fast imaging with steady-state precession (FISP) (and true FISP) (Fig. 5).

In a similar fashion, multiple echoes acquired in the GRE trains can be combined into an image resulting in fewer artifacts while retaining a high SNR if performed in such a way that the earlier echoes provide superior SNR while the later acquired echoes increase tissue contrast. This sequence type of multiple gradient echoes is generally known as MERGE (multiple-echo recombined gradient echo) and is understandably also a focus of intense development for use at high field for neuroimaging and nonneuroimaging MRI applications (Fig. 6).³⁹

THE FUTURE OF FUNCTIONAL NEUROIMAGING SEQUENCES

Over the past 15 years, DWI has matured into an essential neuroimaging tool performed thousands of times daily as a routine screen for brain attacks and injury. It is fast, it is easy, and the hyperintense acute lesions from DWI are hard to miss. With a high diagnostic sensitivity and specificity (90% and above) to ischemic and infarction events following onset of stroke, DWI has significantly affected acute and long-term patient management strategies.^{52–54} In addition, because water diffusion in white matter and peripheral nerves is “anisotropic” (nonrandom or “ordered”), diffusion-tensor imaging (DTI) is now becoming an accepted surrogate measure for structural integrity and, in the minds of many cognitive researchers, functional connectivity of the underlying white matter tracts and pathways in the brain.^{5,55–57} DWI and DTI methods are experiencing profound growth and development, and as MR technologists and physicists in the field, we will see an accelerating number of new diffusion-sensitive sequences and methods added to our MR scanners.

IMPROVING DIFFUSION-WEIGHTED IMAGING AND DIFFUSION-TENSOR IMAGING THROUGH PARALLEL IMAGING

Why the excitement? Frustratingly, only single-shot EPI (ss-EPI) sequences have been able to acquire images rapidly enough to eliminate proton motion (because water diffusion occurs on a micron scale). Rapid EPI suffers from distortion artifacts and signal loss, and poor resolution (typically, 2 mm per pixel for 128×128 images). The introduction of multiple-coil RF PI method can “accelerate” the acquisition of the image by as much as the number of independent coils in the RF array. With the single-shot EPI sequence, however, speed is not the main gain; rather, it is the ability to shorten the TE and minimize distortion artifacts. With many scanners now using at least 8 parallel RF receive coils and some as many as 32 coils, one may expect

to see that number climb to 64 to 128 coils embedded in the patient table, the magnet bore, and the head and cardiac arrays. Where will it end? As long as multiring CT scanners suggest that “more is better,” PI in MR is likely to follow that trend.

Not only can the EPI TE be reduced, but the advent of “high-definition” DWI uses PI to detect and correct for motion by means of mapping motion-induced phase changes across the coil array. With this potential, navigation and motion correction of multiple shots of echo trains is possible.^{58–60} These sequences can achieve high-resolution DWI and DTI images up to 512×512 and beyond, all in conceivable scan times. With scan times on the order of 1 to 2 minutes for a typical stroke DWI study, the high-resolution DWI examinations offer breathtaking detail with minimal artifacts (Fig. 7).^{58,60} As these multiple-shot diffusion-weighted sequences mature, one may expect similarly high resolution DTI scans⁵⁹ to be routinely done in less than 15 to 20 minutes, with fewer artifacts and much higher resolutions, which was near impossible before (Fig. 8).

Tomorrow’s Diffusion-Tensor Imaging “Tractography”

This area is, to many, the most active area of diffusion imaging today, in which colorful computer modeling meets MR imaging to study the hot area of human cognition. Because the DWI always contain proton motion and directional information (because we image diffusion along at least three different gradient directions [X, Y, Z] with no upward limit), computers can “trace” the preferred path of proton diffusion along and through a stack of diffusion-weighted slices. This process is called “tractography”^{56,61,62} or white matter “fiber tracking” and is most often viewed as a 3D white matter structural series of “streamtubes” (which look like colorful spaghetti strands representing the major white matter tracts of the brain). From the orientation information contained in the fractional anisotropy (FA) maps (where brightness denotes the amount of ordering of the diffusion in that voxel), the physician can either “colorize” each fiber’s direction or “trace” the fibers within an image. These fibers can span many slices, of course, and are depicted as a 3D structure of tubes (Fig. 9). The emphasis today is on how to use these to better visualize white matter development in children, demyelization in diseases such as stroke, Alzheimer’s dementia, and schizophrenia, and in many different cognition applications. Although DTI has no clear “killer application” yet, it has tremendous potential. In the coming years, development in this area will more rapidly create these 3D white matter “roadmaps” of the brain, and be able to, for example, map white matter structural deviations of a specific individual relative to a standard template built from various patient populations.

Beyond DTI tractography, our standard model of how protons diffuse in vivo is built on the assumption that water will move faster along tube-like white matter fibers than across the same fibers. Questions soon arise, however, as to how protons move along or about fibers with complex structures. To answer this, our models need to be improved and the acquisition of “super” or “high-order” diffusion tensors are needed.^{63–66} The “super-tensor” is constructed from the image acquisition of hundreds of diffusion-weighted directions, often requiring 10 minutes or more (Fig. 10). The intravoxel proton motions contained in the super-tensor are fascinating, yet until scan times are reduced and a clear clinical need is discovered, one may expect that most DTI and tractography applications will continue to use the simple “rice grain” model of proton diffusion in white matter.

QUANTITATIVE MR PERFUSION: THE EMERGENCE OF ARTERIAL SPIN-LABELING

Accurate cerebral blood flow (CBF) measurements in clinical patients who have cerebrovascular disease remain a serious imaging challenge, regardless of magnetic field

strength or even imaging modality. Quantitative CBF values are frequently considered to weigh the risks and benefits of surgical versus endovascular versus medical management. Aside from MR, this frequently necessitates more invasive and costly imaging tests, such as stable xenon-enhanced CT, H₂O-15 positron emission tomography, or single-photon emission computed tomography (SPECT).

For CBF mapping in the brain using MR, two widely differing methods are used, both of which benefit from higher magnetic field strengths. Most often, MR-based contrast agent “bolus-tracking” brain perfusion measurements are performed. These contrast agent methods detect the dynamic T2* changes that occur during the first pass of the intravenously injected bolus of gadolinium-diethylenetriamine penta-acetic acid (Gd-DTPA), for example. These T2*-based methods are termed dynamic susceptibility contrast or perfusion-weighted imaging (PWI) and have proved difficult to use quantitatively, particularly in the setting of large vessel stenosis or occlusion, because of regional delay and dispersion of the injected intravascular Gd-DTPA contrast bolus that occurs before the tracer reaches the tissue of interest. Other errors that impact the quantitative PWI measurements include the need to know the agent concentrations in the arterial input function and the imaged brain parenchyma.^{67,68} Although some of these problems may be mitigated by the use of PI and multiecho approaches,⁶⁸ qualitative PWI yielding relative maps of blood volume, transit times, and flow acquired from single-shot GRE EPI sequences is by far the most common implementation in clinical practice.^{69–71}

In contrast to PWI, which uses a “nondiffusible tracer” bolus-tracking approach to map the bolus hemodynamics, other quantitative methods use an endogenous inflow proton T1-magnetization approach^{72–78} and are commonly termed “arterial spin labeling” (ASL), which is a “diffusible-tracer” measure of microvascular inflow (consider this as microscopic MR angiography). Several variations of this theme exist, depending on how the arterial protons are labeled and detected.^{77–79} Because the contrast-to-noise ratio of the ASL methods is much lower than that of PWI, sequencing is important. Most recently, rapid methods have become popular and offer good contrast-to-noise ratio with excellent image quality (Fig. 11).⁷⁹ One attractive feature of ASL is the observation that the CBF maps are less sensitive to large vessel effects and may offer improved quantitative CBF values. On the other hand, one major difficulty with ASL is the inaccuracy of CBF values caused by inflow transit delay times created, in turn, by stenotic vessels or anastomoses.

Regardless of the perfusion methodology used, selecting patients who have stroke for tomorrow’s neuroprotective and thrombolytic therapies will no doubt continue to require both DWI and a flow-sensitive method. Although DWI is sensitive to “dead brain” (DWI lesions are bright when blood flow falls below low critical levels), PWI using either contrast bolus injections or ASL methods shows surrounding brain that may die if not treated or reperfused (the origin of the diffusion-perfusion “mismatch” measure). Contrast agent “bolus-tracking” with single-shot GRE EPI imaging is commonly used, with all the image-quality distortions associated with rapidly-acquired GRE sequences. Here again, however, PI has dramatically improved image quality to the extent that PWI (or ASL) is becoming a “must do” for imaging of cerebrovascular disease. One may expect to see further refinements in multiple-echo and multiple-shot GRE or EPI sequences for PWI, more on-line and off-line PWI reconstruction refinements, and quantitation of real blood flow measures by taking into consideration the arterial inputs and corrections for blood volume and T2* sensitivity. Finally, many centers are now considering adding a complementary blood volume detection method^{13–18,80} such as SWI to the stroke protocol as an adjunct to the most commonly used sequences, such as DWI, GRE, fluid-attenuated inversion recovery, and MR angiography. The advantage of adding SWI is the excellent conspicuity and sensitivity to hemorrhage and bleeds over even conventional GRE,

afforded by the strong magnetic susceptibility effects of the local concentrated iron in bleeds on the proton frequency and phase.

FUNCTIONAL NEUROIMAGING IN NEUROLOGY

The ability to predict one's ability to read, process information, react to stop lights, and so forth, has long since been a hot topic in neuroimaging. The method of choice to map the minute dynamic changes in blood oxygenation/flow/volume that occur during, and following, a regional activation effect in the CNS has been the BOLD approach called fMRI.⁸¹ The key has been to make the fMRI sequence sensitive to T2*, which is easily done by using GRE sequences at high fields. As the blood oxygenation levels change (modulated by blood volume and flow dynamics during and after neuronal activation), these T2*-weighted effects can be as much as 20% of the fMRI image SNR at high fields. In fact, routine MR (using even spin echoes) at 7 T is often dominated by T2* effects! With the sensitivity of T2* at 3 T and 7 T (and beyond), "single-event" activations can be seen routinely.^{81–84} An exciting extension has been the explosion in research topics centered on "brain noise" or activation-mapping in the "resting" or "default-mode" brain.^{85–87} This research has the enormous potential of providing new tools to map the functional/nonfunctional brain regions that are so vital in understanding the extent of damage from stroke, trauma, coma, and demyelinating diseases, for example.

Assuming that the maps of FA acquired from DTI are a quantitative measure of white matter structure or "integrity,"^{88–90} good correlations have been seen in numerous studies between local white matter FA values and measures of cognitive or motor performance.⁹⁰ One may expect to see many new fMRI functional studies to be done together with DTI structural maps to provide novel depictions of brain pathology.

The number of publications and general interest in advanced neuroimaging such as diffusion imaging (DWI and DTI) coupled with other techniques such as fMRI continues to grow almost exponentially. This interest is certain to continue; new and faster sequences, better image quality, higher magnetic fields, and improved models of diffusion, perfusion, and functional connectivity are in constant development. Given the enormous usefulness of DWI and DTI in neuroimaging and the rapid acceptance of DWI to a near "gold-standard" status in stroke imaging, where is the next "killer application" for MR neuroimaging? Simply look for more integrated protocols based on structural, metabolic, and functional information to gain substantial attention in the basic neuroscience community for their ability to facilitate the understanding between functional connectivity and anatomic physiology.

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REFERENCES

1. Machann J, Schlemmer HP, Schick F. Technical challenges and opportunities of whole-body magnetic resonance imaging at 3T. *Phys Med* 2008;24(2):63–70. [PubMed: 18308606][Epub 2008 Mar 4]
2. DeLano MC, Fisher C. 3T MR imaging of the brain. *Magn Reson Imaging Clin N Am* 2006;14(1):77–88. [PubMed: 16530636][Review]
3. Schmitz BL, Aschoff AJ, Hoffmann MH, et al. Advantages and pitfalls in 3T MR brain imaging: a pictorial review. *AJNR Am J Neuroradiol* 2005;26(9):2229–2237. [PubMed: 16219827]
4. Tanenbaum LN. Clinical 3T MR imaging: mastering the challenges. *Magn Reson Imaging Clin N Am* 2006;14(1):1–15. [PubMed: 16530631][Review]
5. Nakada T. Clinical application of high and ultra high-field MRI. *Brain Dev* 2007;29(6):325–335. [PubMed: 17113259][Epub 2006 Nov 17. Review]

6. Henning A, Schär M, Kollias SS, et al. Quantitative magnetic resonance spectroscopy in the entire human cervical spinal cord and beyond at 3T. *Magn Reson Med* 2008;59(6):1250–1258. [PubMed: 18421679]
7. Mikulis DJ, Roberts TP. Neuro MR: protocols. *J Magn Reson Imaging* 2007;26(4):838–847. [PubMed: 17896388][Review]
8. Prince MR, Meaney JF. Expanding role of MR angiography in clinical practice. *Eur Radiol* 2006;16:B3–B8. [PubMed: 16802437][Review]
9. Jahng GH, Weiner MW, Schuff N. Improved arterial spin labeling method: applications for measurements of cerebral blood flow in human brain at high magnetic field MRI. *Med Phys* 2007;34(11):4519–4525. [PubMed: 18072518]
10. Golay X, Petersen ET. Arterial spin labeling: benefits and pitfalls of high magnetic field. *Neuroimaging Clin N Am* 2006;16(2):259–268. [PubMed: 16731365][Review]
11. Lin W, An H, Chen Y, et al. Practical consideration for 3T imaging. *Magn Reson Imaging Clin N Am* 2003;11(4):615–639. [PubMed: 15018114][Review]
12. Dietrich O, Reiser MF, Schoenberg SO. Artifacts in 3-T MRI: physical background and reduction strategies. *Eur J Radiol* 2008;65(1):29–35. [PubMed: 18162353][Epub 2007 Dec 26. Review]
13. Tong KA, Ashwal S, Obenaus A, et al. Susceptibility-weighted MR imaging: a review of clinical applications in children. *AJNR Am J Neuroradiol* 2008;29(1):9–17. [PubMed: 17925363][Epub 2007 Oct 9. Review]
14. Srinivasan A, Goyal M, Al Azri F, et al. State-of-the-art imaging of acute stroke. *Radiographics* 2006;26:S75–S95. [PubMed: 17050521][Review]
15. Smith EE, Rosand J, Greenberg SM. Hemorrhagic stroke. *Neuroimaging Clin N Am* 2005;15(2):259–272. ix. [PubMed: 16198939]Review
16. Thurnher MM, Castillo M. Imaging in acute stroke. *Eur Radiol* 2005;15(3):408–415. [PubMed: 15662493][Epub 2005 Jan 21. Review]
17. Raz N, Rodrigue KM, Haacke EM. Brain aging and its modifiers: insights from in vivo neuromorphometry and susceptibility weighted imaging. *Ann N Y Acad Sci* 2007;1097:84–93. [PubMed: 17413014][Review]
18. Sehgal V, Delproposto Z, Haacke EM, et al. Clinical applications of neuroimaging with susceptibility-weighted imaging. *J Magn Reson Imaging* 2005;22(4):439–450. [PubMed: 16163700][Review]
19. Larkman DJ, Nunes RG. Parallel magnetic resonance imaging. *Phys Med Biol* 2007;52(7):R15–R55. [PubMed: 17374908][Epub 2007 Mar 9. Review]
20. Larkman DJ, Atkinson D, Hajnal JV. Artifact reduction using parallel imaging methods. *Top Magn Reson Imaging* 2004;15(4):267–275. [PubMed: 15548957][Review]
21. Sodickson DK, Griswold MA, Jakob PM. SMASH imaging. *Magn Reson Imaging Clin N Am* 1999;7(2):237–254. [PubMed: 10382159][Review]
22. Bammer R, Aksoy M, Liu C. Augmented generalized SENSE reconstruction to correct for rigid body motion. *Magn Reson Med* 2007;57(1):90–102. [PubMed: 17191225]
23. Bammer R, Schoenberg SO. Current concepts and advances in clinical parallel magnetic resonance imaging. *Top Magn Reson Imaging* 2004;15(3):129–158. [PubMed: 15479997]
24. Pruessmann KP, Weiger M, Scheidegger MB, et al. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med* 1999;42(5):952–962. [PubMed: 10542355]
25. Griswold MA, Jakob PM, Heidemann RM, et al. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med* 2002;47(6):1202–1210. [PubMed: 12111967]
26. Adriany G, Van de Moortele PF, Wiesinger F, et al. Transmit and receive transmission line arrays for 7 tesla parallel imaging. *Magn Reson Med* 2005;53:434–445. [PubMed: 15678527]
27. Vaughan JT, Adriany G, Snyder CJ, et al. Efficient high-frequency body coil for high-field MRI. *Magn Reson Med* 2004;52:851–859. [PubMed: 15389967]
28. Frayne R, Goodyear BG, Dickhoff P, et al. Magnetic resonance imaging at 3.0 tesla: challenges and advantages in clinical neurological imaging. *Invest Radiol* 2003;38:385–402. [PubMed: 12821852] [Medline]
29. Baudendistel KT, Heverhagen JT, Knopp MV. [Clinical MR at 3 tesla: current status]. *Radiologe* 2004;44:11–18. [PubMed: 14740092][Medline]

30. Kim DS, Garwood M. High-field magnetic resonance techniques for brain research. *Curr Opin Neurobiol* 2003;13:612–619. [PubMed: 14630227][Medline]
31. Ross JS. The high-field-strength curmudgeon. *AJNR Am J Neuroradiol* 2004;25:168–169. [PubMed: 14970013][Free Full Text]
32. Shapiro MD, Magee T, Williams D, et al. The time for 3T clinical imaging is now. *AJNR Am J Neuroradiol* 2004;25:1628–1629. [PubMed: 15502155]author reply 1629 [Free Full Text]
33. Pattany PM. 3T MR imaging: the pros and cons. *AJNR Am J Neuroradiol* 2004;25:1455–1456. [PubMed: 15502120][Free Full Text]
34. Collins CM, Liu W, Schreiber W, et al. Central brightening due to constructive interference with, without, and despite dielectric resonance. *J Magn Reson Imaging* 2005;21:192–196. [PubMed: 15666397][Medline]
35. Hoult DI, Phil D. Sensitivity and power deposition in a high-field imaging experiment. *J Magn Reson Imaging* 2000;12:46–67. [PubMed: 10931564][Medline]
36. Hennig J, Scheffler K. Hyperechoes. *Magn Reson Med* 2001;46:6–12. [PubMed: 11443704]
37. Weigel M, Hennig J. Contrast behavior and relaxation effects of conventional and hyperecho-turbo spin echo sequences at 1.5 and 3 T. *Magn Reson Med* 2006;55(4):826–835. [PubMed: 16463351]
38. Gold GE, Busse RF, Beehler C, et al. Isotropic MRI of the knee with 3D fast spin-echo extended echo-train acquisition (XETA): initial experience. *AJR Am J Roentgenol* 2007;188(5):1287–1293. [PubMed: 17449772]
39. Vertinsky AT, Krasnokutsky MV, Augustin M, et al. Cutting-edge imaging of the spine. *Neuroimaging Clin N Am* 2007;17(1):117–136. [PubMed: 17493543]
40. Pipe JG. Motion correction with PROPELLER MRI: application to head motion and free-breathing cardiac imaging. *Magn Reson Med* 1999;42(5):963–969. [PubMed: 10542356]
41. Arfanakis K, Tamhane AA, Pipe JG, et al. K-space undersampling in PROPELLER imaging. *Magn Reson Med* 2005;53(3):675–683. [PubMed: 15723398]
42. Forbes KP, Pipe JG, Karis JP, et al. Improved image quality and detection of acute cerebral infarction with PROPELLER diffusion-weighted MR imaging. *Radiology* 2002;225(2):551–555. [PubMed: 12409594]
43. Pipe JG, Farthing VG, Forbes KP. Multishot diffusion-weighted FSE using PROPELLER MRI. *Magn Reson Med* 2002;47(1):42–52. [PubMed: 11754441]Erratum in: *Magn Reson Med* 2002 Mar;47(3): 621
44. Scheffler K, Lehnhardt S. Principles and applications of balanced SSFP techniques. *Eur Radiol* 2003;13(11):2409–2418. [PubMed: 12928954][Epub 2003 Aug 20. Review]
45. Fuchs F, Laub G, Othomo K. TrueFISP-technical considerations and cardiovascular applications. *Eur J Radiol* 2003;46(1):28–32. [PubMed: 12648799][Review]
46. Nitz WR. Fast and ultrafast non-echo-planar MR imaging techniques. *Eur Radiol* 2002;12(12):2866–2882. [PubMed: 12439564][Epub 2002 Jun 28. Review]
47. Deoni SC, Rutt BK, Jones DK. Investigating exchange and multicomponent relaxation in fully-balanced steady-state free precession imaging. *J Magn Reson Imaging* 2008;27(6):1421–1429. [PubMed: 18504765]
48. Lee JH, Dumoulin SO, Saritas EU, et al. Full-brain coverage and high-resolution imaging capabilities of passband b-SSFP fMRI at 3T. *Magn Reson Med* 2008;59(5):1099–1110. [PubMed: 18421687]
49. Paul D, Markl M, Fautz HP, et al. T2-weighted balanced SSFP imaging (T2-TIDE) using variable flip angles. *Magn Reson Med* 2006;56(1):82–93. [PubMed: 16755535]
50. Miller KL, Smith SM, Jezzard P, et al. High-resolution fMRI at 1.5T using balanced SSFP. *Magn Reson Med* 2006;55(1):161–170. [PubMed: 16345040]
51. Kijowski, R.; Klaers, J.; Takami, K., et al. Quantitative and qualitative assessment of multiple new 3T cartilage imaging MR pulse sequences. *Proceedings of the International Society of Magnetic Resonance in Medicine; Toronto, Ontario. 2008. Abstract*
52. Na DG, Thijs VN, Albers GW, et al. Diffusion-weighted MR imaging in acute ischemia: value of apparent diffusion coefficient and signal intensity thresholds in predicting tissue at risk and final infarct size. *AJNR Am J Neuroradiol* 2004;25(8):1331–1336. [PubMed: 15466327]

53. Saur D, Kucinski T, Grzyska U, et al. Sensitivity and interrater agreement of CT and diffusion-weighted MR imaging in hyperacute stroke. *AJNR Am J Neuroradiol* 2003;24(5):878–885. [PubMed: 12748088]
54. Schaefer PW, Hunter GJ, He J, et al. Predicting cerebral ischemic infarct volume with diffusion and perfusion MR imaging. *AJNR Am J Neuroradiol* 2002;23(10):1785–1794. [PubMed: 12427640]
55. Mori S, Zhang J. Principles of diffusion tensor imaging and its applications to basic neuroscience research. *Neuron* 2006;51(5):527–539. [PubMed: 16950152][Review]
56. Mukherjee P, McKinstry RC. Diffusion tensor imaging and tractography of human brain development. *Neuroimaging Clin N Am* 2006;16(1):19–43. vii. [PubMed: 16543084]Review
57. Ramnani N, Behrens TE, Penny W, et al. New approaches for exploring anatomical and functional connectivity in the human brain. *Biol Psychiatry* 2004;56(9):613–619. [PubMed: 15522243] [Review]
58. Skare S, Newbould RD, Clayton DB, et al. Clinical multishot DW-EPI through parallel imaging with considerations of susceptibility, motion, and noise. *Magn Reson Med* 2007;57(5):881–890. [PubMed: 17457876]
59. Liu C, Bammer R, Kim DH, et al. Self-navigated interleaved spiral (SNAILS): application to high-resolution diffusion tensor imaging. *Magn Reson Med* 2004;52(6):1388–1396. [PubMed: 15562493]
60. Holdsworth SJ, Skare S, Newbould RD, et al. Readout-segmented EPI for rapid high resolution diffusion imaging at 3 T. *Eur J Radiol* 2008;65(1):36–46. [PubMed: 17980534][Epub 2007 Nov 5]
61. Mukherjee P, Chung SW, Berman JI, et al. Diffusion tensor MR imaging and fiber tractography: technical considerations. *AJNR Am J Neuroradiol* 2008;29(5):843–852. [PubMed: 18339719][Epub 2008 Mar 13. Review]
62. Lee SK, Kim DI, Kim J, et al. Diffusion-tensor MR imaging and fiber tractography: a new method of describing aberrant fiber connections in developmental CNS anomalies. *Radiographics* 2005;25(1):53–65. [PubMed: 15653586][discussion: 66–8. Review]
63. Liu C, Bammer R, Acar B, et al. Characterizing non-Gaussian diffusion by using generalized diffusion tensors. *Magn Reson Med* 2004;51(5):924–937. [PubMed: 15122674]
64. Wedeen VJ, Wang RP, Schmahmann JD, et al. Diffusion spectrum magnetic resonance imaging (DSI) tractography of crossing fibers. *Neuroimage* 2008;41(4):1267–1277. [PubMed: 18495497][Epub 2008 Apr 8]
65. Hagmann P, Jonasson L, Maeder P, et al. Understanding diffusion MR imaging techniques: from scalar diffusion-weighted imaging to diffusion tensor imaging and beyond. *Radiographics* 2006;26:S205–S223. [PubMed: 17050517][Review]
66. Wedeen VJ, Hagmann P, Tseng WY, et al. Mapping complex tissue architecture with diffusion spectrum magnetic resonance imaging. *Magn Reson Med* 2005;54(6):1377–1386. [PubMed: 16247738]
67. Kakuda W, Lansberg MG, Thijs VN, et al. Optimal definition for PWI/DWI mismatch in acute ischemic stroke patients. *J Cereb Blood Flow Metab* 2008;28(6):1272. No abstract available
68. Newbould RD, Skare ST, Jochimsen TH, et al. Perfusion mapping with multiecho multishot parallel imaging EPI. *Magn Reson Med* 2007;58(1):70–81. [PubMed: 17659630]
69. Butcher K, Parsons M, Allport L, et al. EPITHET Investigators. Rapid assessment of perfusion-diffusion mismatch. *Stroke* 2008;39(1):75–81. [PubMed: 18063829][Epub 2007 Dec 6]
70. Luby M, Warach S. Reliability of MR perfusion-weighted and diffusion-weighted imaging mismatch measurement methods. *AJNR Am J Neuroradiol* 2007;28(9):1674–1678. [PubMed: 17885243][Epub 2007 Sep 20]
71. Moustafa RR, Baron JC. Clinical review: imaging in ischaemic stroke-implications for acute management. *Crit Care* 2007;11(5):227. [PubMed: 17875224][Review]
72. van Laar PJ, van der Grond J, Hendrikse J. Brain perfusion territory imaging: methods and clinical applications of selective arterial spin-labeling MR imaging. *Radiology* 2008;246(2):354–364. [PubMed: 18227536][Review]
73. Zaharchuk G. Theoretical basis of hemodynamic MR imaging techniques to measure cerebral blood volume, cerebral blood flow, and permeability. *AJNR Am J Neuroradiol* 2007;28(10):1850–1858. [PubMed: 17998415][Review]

74. Wolf RL, Detre JA. Clinical neuroimaging using arterial spin-labeled perfusion magnetic resonance imaging. *Neurotherapeutics* 2007;4(3):346–359. [PubMed: 17599701][Review]
75. Liu TT, Brown GG. Measurement of cerebral perfusion with arterial spin labeling: part 1. Methods. *J Int Neuropsychol Soc* 2007;13(3):517–525. [PubMed: 17445301][Review]
76. Wintermark M, Sesay M, Barbier E, et al. Comparative overview of brain perfusion imaging techniques. *J Neuroradiol* 2005;32(5):294–314. [PubMed: 16424829][Review]
77. Wong EC. Vessel-encoded arterial spin-labeling using pseudocontinuous tagging. *Magn Reson Med* 2007;58(6):1086–1091. [PubMed: 17969084]
78. Wong EC. Quantifying CBF with pulsed ASL: technical and pulse sequence factors. *J Magn Reson Imaging* 2005;22(6):727–737. 80. [PubMed: 16261572]
79. Wang J, Li L, Roc AC, et al. Reduced susceptibility effects in perfusion fMRI with single-shot spin-echo EPI acquisitions at 1.5 tesla. *Magn Reson Imaging* 2004;22(1):1–7. [PubMed: 14972387]
80. Hermier M, Nighoghossian N. Contribution of susceptibility-weighted imaging to acute stroke assessment. *Stroke* 2004;35(8):1989–1994. [PubMed: 15192245][Epub 2004 Jun 10]
81. Norris DG. Principles of magnetic resonance assessment of brain function. *J Magn Reson Imaging* 2006;23(6):794–807. [PubMed: 16649206][Review]
82. Toma K, Nakai T. Functional MRI in human motor control studies and clinical applications. *Magn Reson Med Sci* 2002;1(2):109–120. [PubMed: 16082132][Review]
83. Di Salle F, Esposito F, Elefante A, et al. High field functional MRI. *Eur J Radiol* 2003;48(2):138–145. [PubMed: 14680904][Review]
84. Rosen BR, Buckner RL, Dale AM. Event-related functional MRI: past, present, and future. *Proc Natl Acad Sci USA* 1998;95(3):773–780. [PubMed: 9448240][Review]
85. Raichle ME, Snyder AZ. A default mode of brain function: a brief history of an evolving idea. *Neuroimage* 2007;37(4):1083–1090. [PubMed: 17719799][discussion 1097-9. Epub 2007 Mar 6. Review]
86. Morcom AM, Fletcher PC. Does the brain have a baseline? Why we should be resisting a rest. *Neuroimage* 2007;37(4):1073–1082.[Epub 2007 Jul 3. Review]
87. Rogers BP, Morgan VL, Newton AT, et al. Assessing functional connectivity in the human brain by fMRI. *Magn Reson Imaging* 2007;25(10):1347–1357. [PubMed: 17499467][Epub 2007 May 11. Review]
88. Teipel SJ, Meindl T, Grinberg L, et al. Novel MRI techniques in the assessment of dementia. *Eur J Nucl Med Mol Imaging* 2008;35:S58–S69. [PubMed: 18205002][Review]
89. Durston S, Casey BJ. What have we learned about cognitive development from neuroimaging? *Neuropsychologia* 2006;44(11):2149–2157. [PubMed: 16303150][Epub 2005 Nov 21. Review]
90. Moseley M, Bammer R, Illes J. Diffusion-tensor imaging of cognitive performance. *Brain Cogn* 2002;50(3):396–413. [PubMed: 12480486][Review]

Fig. 1.

(A) Phase-sensitive MRI at high field. Images of a human volunteer acquired at 2048×2048 resolution ($88 \mu\text{m}$ in-plane resolution, 1-mm slice) in 8 minutes on a 7 T using a GRE sequence with a 16-channel ASSET (PI) head RF coil approach (acceleration factor = 2). Note the strong $T2^*$ effects, even in the magnitude image (*left*) from vascular structures. Figure on the right shows the raw phase-sensitive map. Repetition time 500 ms, TE 28 ms, number of excitations 2. (B) Phase-sensitive MRI at high field. The negative phase map on the left highlights even small vascular structures and is most similar to SWI. Image on the right is the positive phase map highlighting gray-white differences in phase. By mapping the relative intravoxel phase changes, novel tissue contrasts can be derived from the proton microenvironment and tissue heterogeneity.

Motion changes phase-Each coil sees differing phase.
Changes related to direction- Correction possible.

Fig. 2.

Motion correction using PI. Using a four-coil array, the changes in signal from each coil will depend on coil location, sensitivity, and motion, among other things. Rigid body rotation and translation can be readily detected and corrected given signal change. Single-shot EPI images are acquired with volunteer head motion before (*left*) and after correction (*right*).

Fig. 3.
High-resolution 512×512 CUBE image with fluid-attenuated inversion recovery acquired at 3 T. TR 4000 ms, TE 101 ms, 1.5-mm slices.

Fig. 4.

Comparison of single-shot EPI DWI acquired with PI (SENSE) with diffusion-weighted PROPELLER on a patient presenting with suspected acute hemorrhage. Comparable parameters when possible (1.5 T, 128×128 , 3-mm slice, $b = 1000 \text{ s/mm}^2$). Note the image degradation around the right-sided lesion in the EPI images due to susceptibility, whereas diffusion-weighted PROPELLER images show clear hyperintensity. PROP, PROPELLER.

Fig. 5.
High-resolution 512×512 GRE COSMIC image. TR 5 ms, TE 1.5 ms, 2-mm slices, acquired at 3 T.

Fig. 6.
High-resolution 512×512 GRE MERGE image acquired at same location as Fig. 4. TR 30 ms, TE 12 ms, 4-mm slices, acquired at 3 T.

Fig. 7. GRAPPA-DWI for high-resolution DWI. Three-shot DW-EPI with GRAPPA reconstruction to produce 192×192 DWI ($b = 1000 \text{ s/mm}^2$) acquired at 3 T. Scan time for 24 slices (TR 3000) was 2.2 minutes. (Courtesy of S. Skare, S. Holdsworth, and R. Bammer, Stanford, CA.)

Fig. 8. SNAILS-DTI for high-resolution DTI. Multishot spiral GRE diffusion-weighted sequence with motion navigation at 3 T. Images shown are diffusion weighted ($b = 800 \text{ s/mm}^2$) (*upper left*), the TraceADC or averaged apparent diffusion coefficient map (*upper right*), the fractional anisotropy (FA) map (*lower left*), and the color FA map (*lower right and magnified*). Colors correspond to the directional axis of preferred proton diffusion in white matter and offer novel means of white matter tract structures. AP, anterior posterior; GRE, gradient recalled echo; LR, left right; SI, superior inferior; SNAILS, self-navigating interleaved spiral.

Fig. 9. DTI and tractography. Multislice DTI FA maps (as in Fig. 8) are used to construct 3D streamtube structures corresponding to the preferred proton diffusion directions in white matter tracts. Colors represent the orientation of the tracts. Insert shows high-order tensor information on a pixel-by-pixel basis. Again, the colored shapes show the intravoxel preferred proton diffusion directions.

Fig. 10.

(A) High-order tensor and white matter structure. Diffusion images are acquired along many directions (a minimum of 6 and typically 32–256) from which the diffusion tensor is constructed. Consider the tensor as a map of the proton diffusion coefficient along a 3×3 direction grid. Given enough directions, however, appropriate models can describe higher-order tensors, which then graphically map intravoxel proton behavior in crossing or merging white matter tracts. The high-order tensor map here is useful in appreciating proton diffusion; diffusion directions (colors) and magnitude (shape of the object) are shown. Note the elongated diffusion within the splenium of the corpus callosum (hyperintense on the background FA map) along the tracts and the near-spherical appearance within the ventricles (indicating isotropic rapid proton diffusion). Integrated maps such as these will reveal an individual's complex white matter patterns. (B) Magnified region within the box in A. High-order tensor and white matter structure. Magnification of the voxels within the splenium shows the strong orientation of the tracts on the directional proton diffusion.

Fig. 11.

Arterial spin labeling. Quantitative CBF maps are shown from normal volunteer. Maps were created from 128×128 slices using a pseudocontinuous tagging scheme from which 10 2-mm slices were acquired in 5 minutes at 1.5 T. Note the excellent gray-white CBF differentiation. (Courtesy of G. Zaharchuk, Stanford, CA.)