The Clinical Utility of Magnetic Resonance Imaging According to Field Strength, Specifically Addressing the Breadth of Current State-of-the-Art Systems, Which Include 0.55 T, 1.5 T, 3 T, and 7 T

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Abstract: This review provides a balanced perspective regarding the clinical utility of magnetic resonance systems across the range of field strengths for which current state-of-the-art units exist (0.55 T, 1.5 T, 3 T, and 7 T). Guidance regarding this issue is critical to appropriate purchasing, usage, and further dissemination of this important imaging modality, both in the industrial world and in developing nations. The review serves to provide an important update, although to a large extent this information has never previously been openly presented. In that sense, it serves also as a position paper, with statements and recommendations as appropriate.

Key Words: magnetic resonance, field strength, image quality

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Information regarding clinical image quality and utilization across the breadth of available field strengths in magnetic resonance imaging (MRI) is generally lacking in the published literature, with this review article attempting to present a balanced perspective. For greater detail, reference can be made to the November 2021 special issue of *Investigative Radiology*, “Field Strength in MR, Clinical Perspectives.”[1] The review begins with a presentation of scans from 3 major anatomic areas, the brain, knee, and liver, providing a realistic comparison of the field strengths both as used clinically and as contrasted on the basis of signal-to-noise ratio (SNR). The review then continues with an in-depth discussion of each field strength, the pros and cons, as well as clinical utilization today and as projected in the future.

There are few examples in the literature comparing directly scans of the different anatomical regions at the different available field strengths under controlled circumstances and in the same individual, using similarly equipped scanners (much less with current state of the art).[2] Thus, with the cooperation and support of Siemens Healthcare GmbH, a normal volunteer was scanned on state-of-the-art systems at the headquarters in Erlangen, Germany, in 2021. Permission to perform such scans was provided by the internal regulatory medical system in place in the MR division. Two approaches were used to illustrate the similarities and differences between imaging at the 4 different field strengths. The first was that in which scan parameters were adjusted at each different field strength to obtain good image quality, as might be expected in clinical practice. The second was to compare images obtained with similar parameters, including specifically voxel size, number of averages, and scan time, to emphasize the inherent differences between the different field strengths. The commercial units used included the Magnetom Free.Max (0.55 T), the Magnetom Sola (1.5 T), the Magnetom Vida (3 T), and the Magnetom Terra (7 T). Of the major MR manufacturers, only one has current units that cover this field strength range, and thus the choice of systems. This also eliminates any difference that might be due to manufacturer.

The results provide a direct scan comparison of the 4 field strengths—0.55, 1.5, 3, and 7 T, not ever previously available, and offer important insight into current clinical MRI. The images provide insight into realistic differences and similarities in imaging at the different field strengths.

For the brain, sagittal T1-weighted fast spin echo (FSE) images are first presented (Fig. 1), comparing 0.55, 1.5, and 3 T. With scan techniques adjusted to provide a good T1-weighted image at each field, the improvement in SNR with field strength is still evident, although the scans are of diagnostic quality at all fields. The scan time as would be expected is longer at low field. Note should be made that T1-weighted FSE at 3 T is not the standard technique at that field strength typically used clinically to achieve a T1-weighted image of the brain. 7 T is not illustrated, once again, as the scan technique used to obtain a sagittal T1-weighted image at that field strength markedly differs from that at the lower field strengths. A comparison of diffusion-weighted imaging (DWI) using single-shot echo planar (ss-EPI) technique at the 3 fields (Fig. 2) provides similar results. All 3 field strengths provide clinically diagnostic images, with a progression in image quality with field strength despite the restriction of using ss-EPI, with multishot readout segmented EPI preferred in many institutions for brain DWI at 3 T. Note that bulk susceptibility artifacts increase with field strength (in this instance due to the adjacent petrous apex and sphenoid sinus at the lower anatomic level illustrated), although not to the degree one might expect, due to further optimization of the scan technique at the 2 higher fields.

Evaluating susceptibility-weighted imaging (SWI) in the brain, despite optimization of imaging technique at the 3 field strengths illustrated (1.5, 3, and 7 T), it should come as no surprise that the quality of the images improves substantially with field strength (Fig. 3). For best visualization of the veins and iron in the brain, 7 T stands clear as the field strength of choice.

For thin-section imaging of critical structures in the brain, 3 T and 7 T offer substantial advantages over 1.5 T. Illustrated are coronal, thin-section, T2-weighted images of the hippocampus (Fig. 4), with the images acquired using the same voxel size and number of averages, in an attempt to illustrate realistically the difference in intrinsic SNR between the 3 field strengths.

Three-dimensional (3D) time-of-flight MR angiography (TOF MRA) images are presented for the 4 field strengths (Figs. 5, 6). Diagnostic images are presented in each case, although image quality increases with each higher step in main magnetic field strength (Fig. 5). This occurs due to the combination of increase in SNR with B0 together with the prolongation of T1, the latter improving background tissue suppression. To illustrate fully the capability of 7 T, an example is shown from the published literature (Fig. 6), with an extremely small voxel size providing visualization of small critical vessels in the brain not possible at lower fields.

To showcase the similarities and differences between the 4 field strengths, for the musculoskeletal (MSK) system, 3 figures are presented

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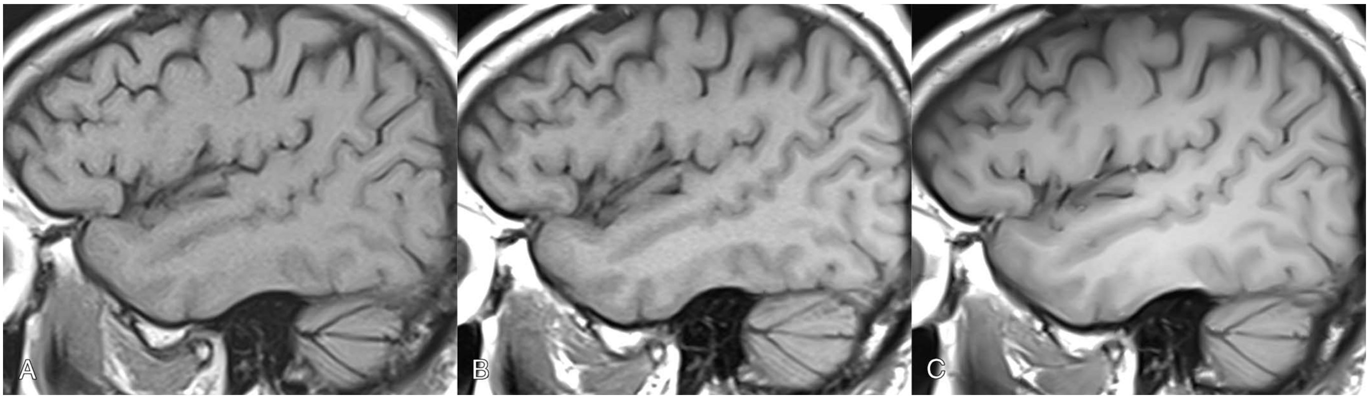

FIGURE 1. Sagittal off-midline T1-weighted spin echo images of the brain obtained at (A) 0.55 T, (B) 1.5 T, and (C) 3 T in a normal volunteer. The slice thickness was 5 mm at 0.55 T and 4 mm at both 1.5 and 3 T. Scan parameters were adjusted to obtain an acceptable image for a clinical screening examination at each field strength. Scan times varied from 2 to 3 minutes, with the longest being for low field. There is a substantial improvement in both delineation of cortical gray matter and signal-to-noise ratio (SNR) with increasing field strength.

with scans obtained to evaluate the knee and its internal components. When standardizing for technique, the marked increase in SNR with field strength is readily evident (Fig. 7). Optimizing the scan for each field strength, however, provides clinically diagnostic images across the entire range of main magnetic fields, from 0.55 to 7 T (Fig. 8). An additional example is provided to compare the extremes of field strength (Fig. 9), illustrating how even the lowest field strength (0.55 T) can obtain good quality diagnostic images of the knee, although at lower resolution. The figure compares 0.55 T to the ultimate in image quality, at 7 T. As a side note, despite its high cost, 7 T is likely to prove to be of substantial value in the future, in instances where anatomic detail is critical, in sports medicine, and in some cases before surgery.

The remaining 2 figures are presented to provide insight into body imaging at the different field strengths. With scan parameters held relatively constant, the increase in SNR with field strength, from 0.55 to 1.5 to 3 T, is readily evident, as illustrated on coronal breath-hold HASTE images of the liver (Fig. 10). However, when scan sequence parameters are optimized for each field strength, all 3 MR units provide diagnostic quality images (Fig. 11). The lower SNR at 0.55 T is regardless still apparent, despite the larger pixel size used. The exquisite quality of liver imaging at 3 T is also evident, principally due to the smaller voxel that can be acquired with adequate SNR.

In the main section of this review, which follows, the pros and cons of each field strength in terms of clinical imaging are discussed in detail. With the majority of clinical systems in place today being either 1.5 or 3 T, the discussion of these 2 field strengths is combined, enabling an improved comparison of the 2. The review then continues with 4 important specialty topics. These include the presence of metal (which favors today imaging at 1.5 T), MSK imaging in the hands of experts (favoring 3 T), the ongoing controversy regarding the use of 1.5 versus 3 T for cardiac imaging, and a section focusing on the evaluation of multiple sclerosis (MS) in the brain, an area where 7 T excels. Although the figures for the review have already been detailed, these are also referred to throughout the main section of the review as appropriate.

LOW-FIELD MAGNETIC RESONANCE

Magnetic resonance systems operating at 1.5 and 3 T today dominate clinical imaging. However, recent technological advances have made possible high-quality patient examinations at low field (such as 0.55 T). High-performance low-field MR systems are currently being introduced clinically, offering high flexibility, excellent clinical examinations, and reduced cost.

In the early 1980s, with the introduction of clinical MRI, low-field systems operating in the range of 0.04 to 0.35 T were showcased.[4,5] However, these systems failed to gain widespread acceptance in part due to lower image quality. Technological advances made possible the introduction of 1.5 T systems, and subsequently 3 T, which provided a marked improvement in SNR.[6] Today, these make up the majority of clinical MR systems. This evolution occurred despite the lack of clinical trials demonstrating the efficacy of higher field strength, and indeed the existence of several excellent studies showing equivalency for imaging of the brain[7] and the MSK system[8] between systems operating at 0.5 T and 1.5 T.

The choice of high field for clinical imaging came with a price, and not simply one of cost. These systems are large and complex, limiting where they can be installed, and requiring highly trained personnel both for operation and maintenance. Direct costs are high, with these including the purchase of the machine, its installation, and its maintenance. These conditions limit the expansion of MR in both industrialized and emerging nations. As a result, there are far fewer MR systems worldwide than computed tomography (CT) scanners, and far less when compared with ultrasound systems, due to lower cost, size, and complexity of operation. Fortunately, the past 40 years has brought a marked increase in knowledge in the field of MR, with radical innovation possible leading to high-performance, markedly reduced cost, low-field systems.[9]

Cost Drivers

Installation, setup, operation, and maintenance are not the only limiting costs of a 1.5 or 3 T MR system. Siting demands are also high. Both need to be addressed for wider dissemination of this clinically very important modality.

Major costs involving the MR device itself, which can be addressed and have been with the next generation low-field units, include the magnet, the gradients, and the cryogen exhaust system. The choice of low field also allows design of a compact and light weight system, simplifying and markedly reducing the costs associated with scanner location requirements, transportation, and installation. Together with reduced maintenance costs and simplified operation, these lead to a markedly lower breakeven point in terms of the number of examinations that are necessary to be performed per month.

Worldwide Access

Diagnostic imaging technology is a major cost driver of patient care across the world. Fifty percent of the world's population has in essence no access to MRI, mainly that in the low-income regions of the world.[10] For improved worldwide access, markedly reduced cost as well as simplified installation and operation and maintenance are essential. Innovations with teleradiography also allow, both in developing

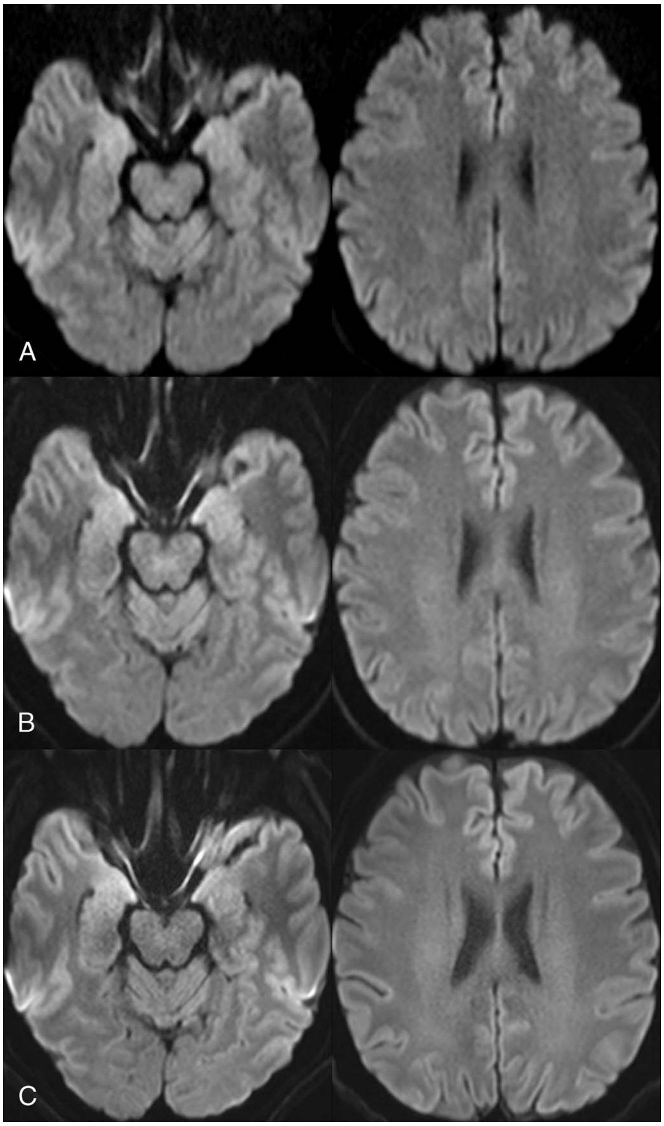

FIGURE 2. Single shot-echo planar diffusion-weighted (b = 1000) images of the brain obtained at (A) 0.55 T, (B) 1.5 T, and (C) 3 T in a normal volunteer. Images at the level of the cerebral peduncles (to assess susceptibility artifacts) and at the top of the lateral ventricles (to assess overall image quality) are illustrated in each instance. Scan parameters were adjusted to achieve good image quality at each field strength. Scan times were between 2 and 3 minutes for (A) and (B), and 1 minute, 29 seconds for (C).

countries and in more rural areas of the industrialized nations, assistance in terms of support and expertise both for technologists and radiologists.

Siting Logistics

Siting logistics are a major challenge both for hospitals in industrialized countries and for installations in developing countries. The smaller required footprint for a low-field scanner is an important attribute. Room costs are less, as well as the need for magnetic shielding, as might be required by location near large iron-containing equipment, for example elevators. The lack of need for a quench pipe, with new high-performance low-field systems, markedly simplifies the complexity and challenges in terms of siting. Siting in emergency rooms, operating suites, and intensive care units could thus become common, allowing point-of-care diagnostics, alleviating time delays (and thus improving quality of care) in terms of access to the examination and transport-related risks (and associated personnel costs).

Clinical Aspects

Low-field imaging can provide very high diagnostic image quality (Fig. 1), excluding those examinations in which submillimeter spatial resolution is required.[11,12] An extremely important measure of image quality is SNR, which is directly related to spatial resolution and scan time. Signal averaging can only be partially effective in achieving higher SNR, since SNR increases in direct proportion to the square root of the number of averages. Thus, images at low field are typically acquired with lower spatial resolution, both in plane and relative to slice thickness. The marked technological improvements over the last 40 years fortunately have led to a substantial increase in SNR at low field, making possible diagnostic quality images in a reasonable scan time with only minor adjustments in pixel and voxel size (Figs. 8, 9). In addition to providing diagnostic quality images over the broad range of clinical applications, there are areas where low-field carries advantages. Magnetic susceptibility is directly proportional to field strength; thus, susceptibility artifacts are markedly less at low field. Areas where this could be very important, clinically, include the visualization of tissue near metal implants and that at air-tissue interfaces. The latter include the lungs in particular, as well as the skull base and the bowel. Initial results at 0.55 T are extremely promising in the lung, with simultaneous imaging of pulmonary structure and regional function demonstrated, as well as assessment of local functional deficits.[13] Low-field MR also offers the possibility of serial lung imaging, in both acute and chronic lung diseases, without the radiation associated with repeated CT examinations.[14]

At 0.55 T compared with higher fields, T2* is longer and T1 shorter. The former allows the readout duration to be increased and the echo train to be lengthened, both methods to increase SNR. The latter allows repetition time (TR) to be decreased for T1-weighted scans. Low field should offer major advantages as well for MR-guided catheterization and interventional studies. Low field offers potentially substantially larger bore sizes (eg, the 80 cm diameter of the Siemens 0.55 T system) with the accompanying improvements in patient access. The tighter 5 gauss line and lower field strength within the area of intervention are additional major benefits.

Not to be downplayed are the marked advantages that low field offers in terms of patient acceptance. Further increasing the bore size from that available today with high-field systems offers a significant incremental benefit for claustrophobic patients and the obese. Acoustic noise from the gradients is markedly reduced as well.[15]

In terms of safety, low field offers further advantages. The translational force and torque associated with ferromagnetic objects are less. The specific absorption rate and thus heating is much less of a concern. A consequence of the lower magnetic field is a reduced risk profile for implants. These considerations extend as well to MR interventions, and the associated needles, guidewires, and endovascular devices.[16] This may open the field of MR to cardiovascular interventional procedures, with initial encouraging results.[17]

Current Perspectives

The advantages of a state-of-the-art low-field MR system are innumerable. Technical considerations include reduced siting logistics, simplicity involving the cryogen system (elimination of the quench pipe and the need for helium replenishment), the smaller footprint of the system, and the potential for point-of-care solutions. Socioeconomic considerations include reduced cost and complexity (for the system itself, siting, operation, and maintenance), making possible dissemination of this technology to emerging nations and further within industrialized nations. In clinical terms, these systems offer high-quality imaging of all organ systems and anatomical regions (Fig. 9), with true advantages

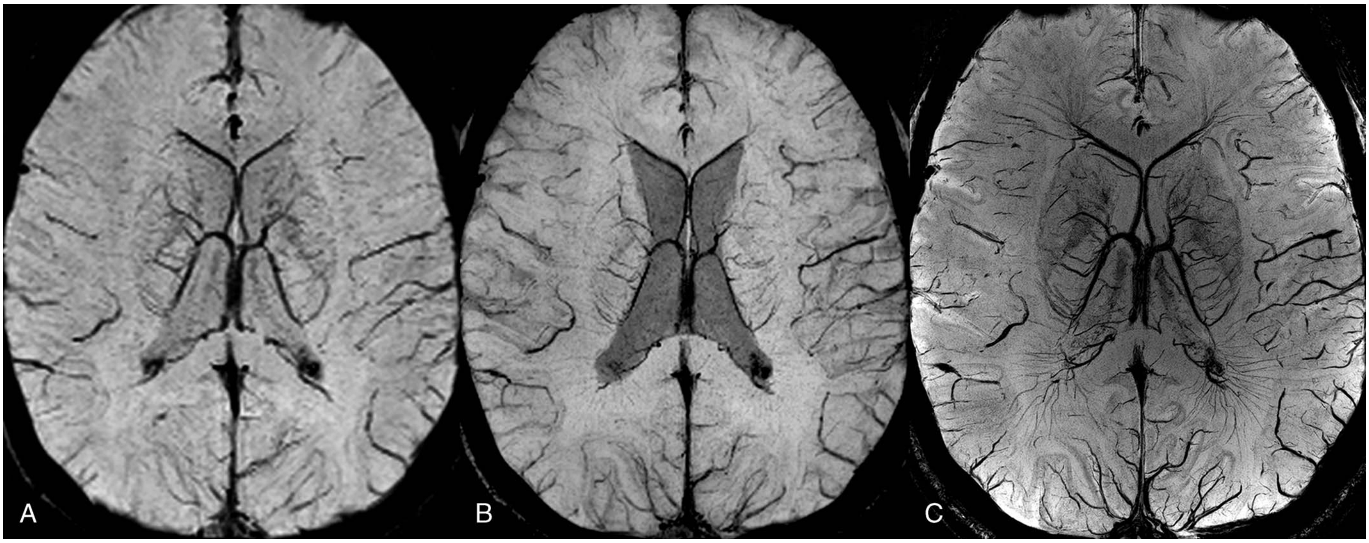

FIGURE 3. Susceptibility-weighted images, minimum intensity projection, of the brain at (A) 1.5 T, (B) 3 T, and (C) 7 T. The images presented are in the axial plane at the level of the lateral ventricles, with optimized parameters according to field strength. Note the improvement in delineation of the veins at 3 T as compared with 1.5 T (despite the scan time being half), with a further marked improvement at 7 T due to the higher susceptibility and available SNR, as well as the use of a smaller voxel size (28 times smaller than at 3 T, with the scan time being twice that at 1.5 T).

over high-field units in certain specific areas, including lung imaging and interventional examinations.

Technological advances have allowed radical redesign and marked improvements for low-field MR systems, providing high performance with low cost (Fig. 11) and thus expanding our diagnostic toolbox. Today these scanners complement those available at high fields and will play a major role in the future in expanding the use of MR. As emphasized throughout this review, as it continues, there is not just one optimal field strength for all diagnostic imaging applications. In many circumstances, a low-field MR system is sufficient, and in certain specialty areas, advantageous.

HIGH-FIELD MAGNETIC RESONANCE—1.5 VS 3 T

There are many clinical applications that benefit from the increase in SNR at 3 T, such as the brain and MSK system, but there are also applications where disadvantages at 3 T favor 1.5 T, including specifically the presence of metal implants and lung imaging.[18,19]

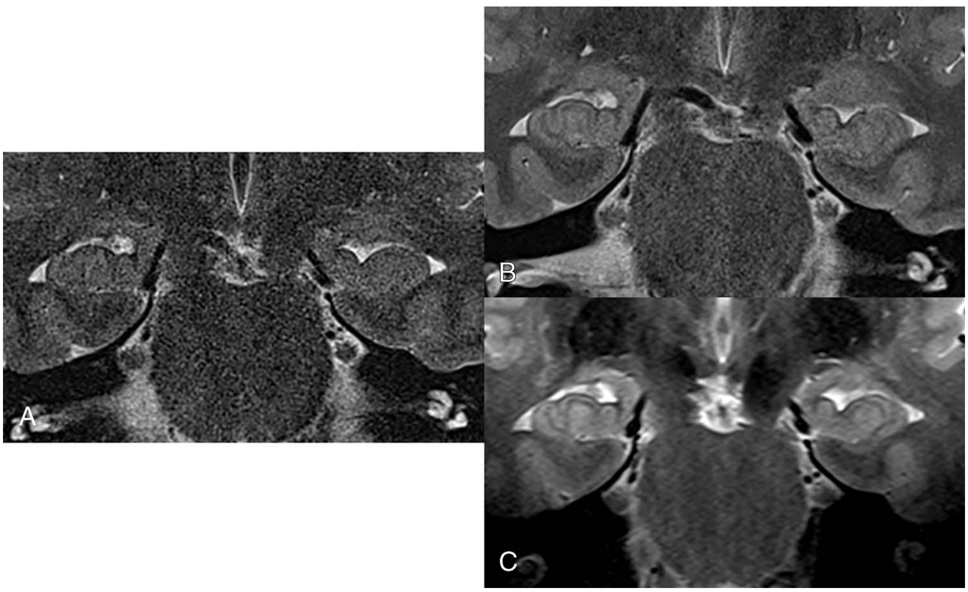

FIGURE 4. Coronal high-resolution, thin-section (2 mm), imaging of the hippocampus at (A) 1.5 T, (B) 3 T, and (C) 7 T in a normal volunteer. In this comparison, the number of averages, slice thickness, and pixel size were held constant, to emphasize the difference in SNR between field strengths. The general level of SNR can be readily assessed by the graininess of each image in the pons, scaling from low SNR at 1.5 T to high at 7 T. The scan time at 1.5 and 3 T was approximately 5 minutes, with that at 7 T being 2 minutes (using a shorter TR and an increase in acceleration factor). The internal structure of the hippocampus is best visualized at 7 T. In clinical practice, such scans however would be optimized for each field strength, improving substantially the image quality of the 1.5 T image. For such a specialty examination, likely the slice thickness would be increased at 1.5 T (to 3 mm), the pixel size slightly increased, and a greater number of scan averages used.

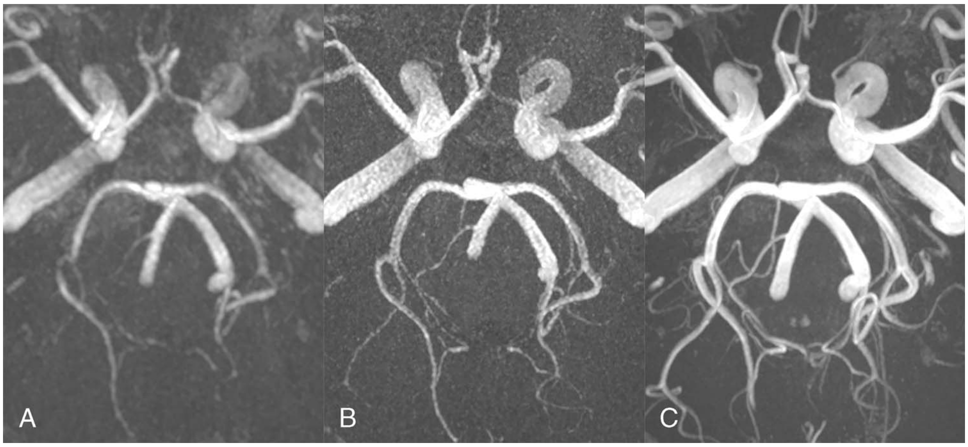

FIGURE 5. Thick axial maximum intensity projection (MIP) 3D time-of-flight magnetic resonance angiography images are presented, obtained at (A) 0.55 T, (B) 1.5 T, and (C) 3 T in a normal volunteer. Scan parameters were adjusted at each field strength to achieve diagnostic image quality, for a scan time of approximately 6 minutes. A substantial improvement in image quality can be noted with field strength, with the higher SNR due to both field strength and the prolongation of T1. This also allowed the pixel size to be decreased from (0.55 T) 0.44 mm^2 to (3 T) 0.26 mm^2.

When first introduced, 3 T was hampered by SAR limitations, inhomogeneous RF excitation, and the lack of well-designed array coils (for signal reception). Today, the landscape has changed, with these problems largely overcome. Despite the higher cost, 3 T systems may ultimately replace 1.5 T, as costs continue to fall. However, still today for new units, two 1.5 T systems are sold for every 3 T system.

The basis for SAR difficulties at 3 T comes because the energy of an RF pulse increases quadratically with the Larmor frequency. Thus, if the same excitation pulses are used, energy deposition is 4 times as high at 3 T compared with 1.5 T. The origin of B1 field inhomogeneities lies in the shorter RF wavelength at 3 T, leading to wave interference (dielectric effects) in larger parts of the body, such as the trunk, especially in patients with ascites. Two channel RF transmitters (dual-source parallel RF transmission) represent a major advance for improved B1 field homogeneity at 3 T.

For clinical field strengths up to 3 T, SNR increases linearly with magnetic field strength (B0), provided the receive bandwidth is the same. Strictly, this would translate to a reduction in scan time by a factor of 4 when moving from 1.5 to 3 T. However, other factors come into play that can reduce this advantage. T1 times are longer at 3 T (by 10%–30%), requiring in certain instances a longer TR. However, in TOF MRA, the longer T1 represents an advantage, with improved background suppression (Fig. 5). Relative to T2*, susceptibility effects increase linearly with field strength, representing a major advantage for techniques highlighting this parameter. Chemical shift increases linearly with field strength, an advantage for spectroscopy yet representing a major disadvantage in general for imaging. To achieve similar chemical shift at 3 T as at 1.5 T, assuming no other adjustments, the steepness of the frequency encoding gradients must be doubled, reducing SNR by 40%.

Diffusion-weighted imaging deserves special consideration (Fig. 2), in the comparison of 1.5 and 3 T, due to the more pronounced susceptibility artifacts with ss-EPI at higher field. Unless otherwise addressed, image quality can be degraded in the vicinity of air-filled structures or metal implants. State-of-the-art techniques at 3 T do much to improve DWI at 3 T, with higher-quality images (compared with 1.5 T) in most instances. Examples of such approaches include advanced shimming, parallel imaging, multishot readout-segmentation, and reduced field of view (zoom), which all reduce image distortion. Simultaneous multislice technique (SMS) is also commonly used to reduce scan time at 3 T, without SNR or image quality degradation.

Because of the increased available SNR, thinner slices can be achieved in breast examinations at 3 T allowing improved detection of small breast cancers. For prostate imaging, the increased SNR at 3 T allows high-quality prostate imaging without the need for an endorectal coil.

Brain

Neuroimaging generally benefits at 3 T, when compared with 1.5 T, from higher SNR.[20] This translates to images with higher spatial resolution and improved lesion contrast.

For brain tumors, 3 T offers higher lesion contrast-to-noise ratio (CNR) on contrast-enhanced scans as compared with 1.5 T. For metastases less than 5 mm, this enables detection of lesions that might otherwise be missed at 1.5 T, a very important point for determining appropriate therapy. Perfusion imaging (dynamic susceptibility contrast) at 3 T also offers improved differentiation between low- and high-grade tumors. Imaging of small pituitary tumors, as well as small lesions of the internal auditory canal, is also markedly improved at 3 T.

Using state-of-the-art techniques, 3 T has been shown to significantly increase the detection of small ischemic lesions, compared with 1.5 T. This includes specifically early, acute lesions detected with DWI. Studies of 3 T versus 1.5 T for MS reveal superior lesion conspicuity and overall image quality, for the critical imaging sequences—which include FLAIR, T2-weighted FSE, and gadolinium-based contrast agent–enhanced scans. As might be expected, most lesions that were missed at 1.5 T are small, less than 5 mm. Innovative imaging sequences can further the sensitivity of 3 T to MS lesions.[21] In Alzheimer disease, hippocampal atrophy serves as a major structural imaging marker, with improved assessment at 3 T compared with lower field strengths (Fig. 4). For epilepsy, high-resolution imaging at 3 T has been shown to be superior to that at 1.5 T for detection of structural lesions, including specifically cortical malformations. Likewise, 3 T better demonstrates hippocampal sclerosis, a major cause of temporal lobe epilepsy.

Spine

Despite the advantage of higher SNR (for 3 T), similar diagnostic quality has been reported for imaging of the spine and cord for 1.5 versus 3 T. Regardless, for the bone marrow, there is increased contrast between normal and pathological conditions at 3 T. In the hands of experts, with optimized thin sections, 3 T does prove to be superior for detection and delineation of disk herniations and associated disease.

Musculoskeletal Imaging

3 T holds great potential for improved MSK imaging, due to greater SNR (Fig. 7) and thus higher achievable spatial resolution. This leads to improved visualization (and depiction of internal detail) of

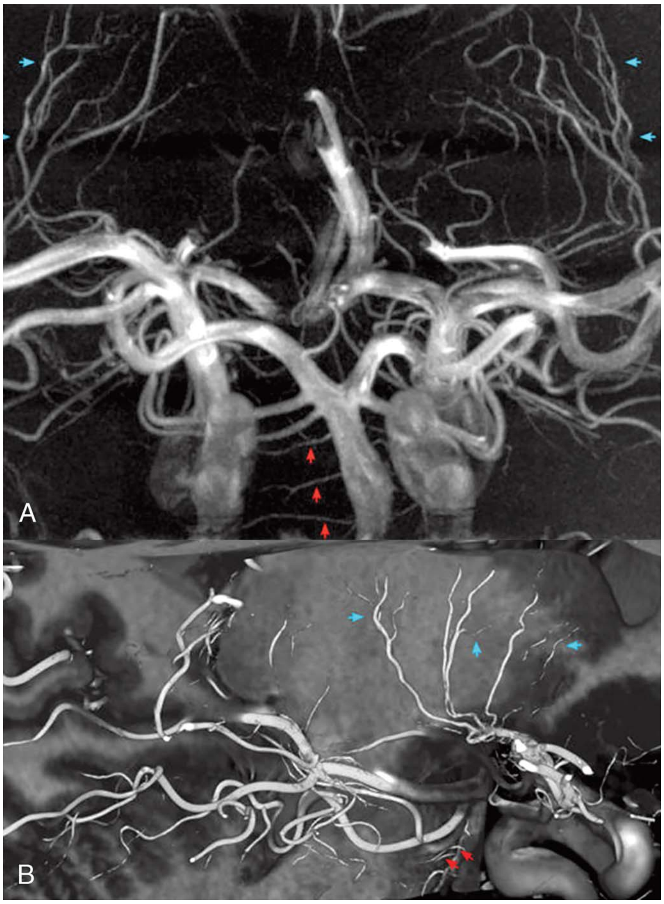

FIGURE 6. Three-dimensional time-of-flight magnetic resonance angiography at 7 T, 150 μm isotropic voxel, (A) maximum intensity projection and (B) 3D volume rendering with inclusion of data from the MP-RAGE scan (to provide visualization of the underlying brain anatomy). 7 T can provide clinically critical, fine vessel detail not achievable at lower field strengths. Note the excellent visualization of the lenticulostriate arteries within the deep gray matter (blue arrows) and very small pontine branches of the basal artery to the brainstem (red arrows). Published with permission from *Magn Reson Med* (2018;80:248–258).[3]

ligaments, tendons, and cartilage. As previously noted, images can be degraded by chemical shift, susceptibility artifacts, and image inhomogeneity—although with modern 3 T systems operated appropriately, these are much less of a concern.

In specific joints, such as the knee and ankle, 3 T has been shown to outperform 1.5 T in the assessment of cartilage and ligamentous lesions (Fig. 8). This is also true for the evaluation of smaller structures, such as the triangular fibrocartilage complex, pulleys, and ligaments in the wrist and hand. Care should be taken with the published literature, as many of the comparative studies are old and prior to the development of the high-quality, dedicated multichannel transmit/receive coils available today for MR MSK imaging. The reader is referred to the dedicated section later in this article concerning 3 T MSK MRI and its advantages. In the presence of metal, 1.5 T however holds an edge over 3 T, due to improved image quality (less degradation due to the presence of metal) and typically, but not always, less RF-induced heating.[22]

Abdominal Imaging

With time after its introduction, 3 T began to be accepted as the MR field strength of choice for abdominal imaging. Its strengths are SNR (Fig. 10) and CNR, providing higher spatial resolution and shorter scan times (Fig. 11). As in other areas of the body, the negatives to 3 T include higher SAR, increased magnetic susceptibility, and both B0 and B1 inhomogeneities, although today these have little impact on clinical imaging of the abdomen. Recent years have, however, brought a continual evolution in sequence, coil, and gradient design, lessening the difference in the 2 fields for routine abdominal imaging.

There are certain examinations, however, that favor abdominal imaging at 1.5 T. In the presence of ascites, despite the advent of parallel RF transmission at 3 T, image inhomogeneity due to the dielectric effect can still be problematic. Despite the potential benefit of 3 T in spectroscopy due to an increase in chemical shift, 1.5 T and 3 T perform similarly in the generation of proton density fat fraction maps. This recent scan technique uses a chemical shift-encoded approach to acquire maps of fat content in a single breath-hold, facilitating the diagnosis, grading, and monitoring of fatty liver disease. For quantification of liver iron concentration, 3 T at first glance might be thought to be superior, due to the greater effect on T2*. However, this is not the case. Low SNR at 3 T especially with high iron content makes 3 T unfavorable for quantification, with 1.5 T favored. Indeed, most of the published calibration curves are valid only for 1.5 T. For one last specialty application, MR elastography (today the noninvasive criterion standard for evaluation of liver fibrosis), 1.5 and 3 T perform similarly. However, technical failure rates are higher at 3 T. Overall, 1.5 and 3 T perform similarly in routine abdominal applications, with mild pros and cons detailed in the preceding paragraphs.

Prostate Imaging

Magnetic resonance imaging plays a critical role today in the screening for and evaluation of prostate cancer. Multiplanar T2-weighted, axial diffusion-weighted, and dynamic contrast-enhanced (DCE) sequences form the core of this assessment, together defining

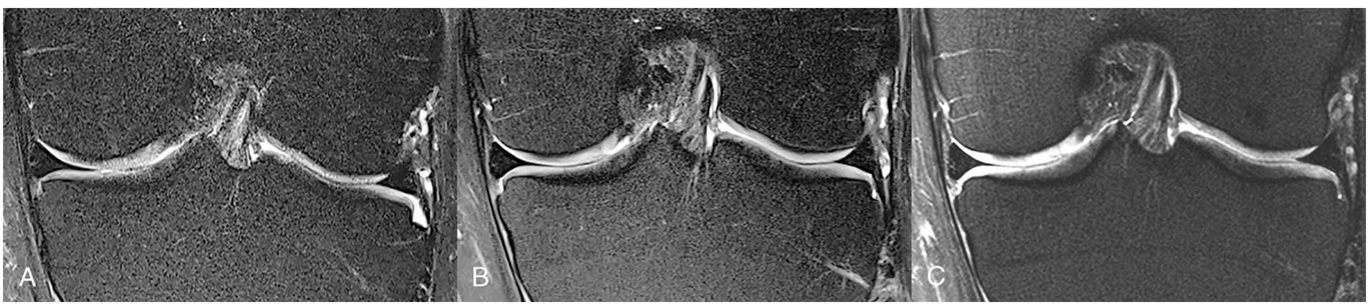

FIGURE 7. Coronal 3-mm proton density-weighted fast spin echo scans of the knee, visualizing the anterior cruciate ligament, articular cartilage, and menisci at (A) 1.5 T, (B) 3 T, and (C) 7 T. Imaging technique and sequence parameters were held generally constant to emphasize the difference in SNR between field strengths. As would be anticipated, SNR scales with field strength (best seen by evaluation of the marrow). Scan times were approximately 7 minutes for (A) and (B), and 5.5 minutes for (C).

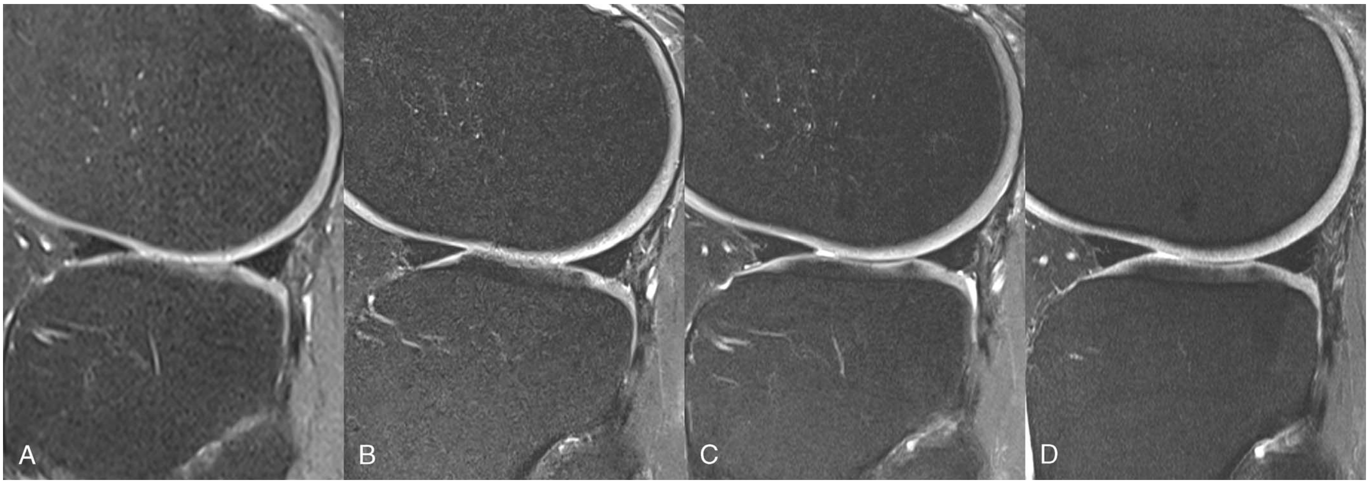

FIGURE 8. Sagittal proton density-weighted images of the medial meniscus and articular cartilage of the knee obtained at (A) 0.55 T, (B) 1.5 T, (C) 3 T, and (D) 7 T. Imaging technique and sequence parameters were changed for each field strength to obtain a representative clinical image from each, with scan times of 5.5 to 7 minutes. The slice thickness at 0.55 T was 3.5 mm, otherwise it was 3 mm. The pixel dimension decreased from 0.27 mm^2 at 0.55 T to 0.13 mm^2 at 7 T.

multiparametric MRI. Multicenter randomized trials have demonstrated the accuracy of prostate MRI, specifically reducing overdiagnosis of clinical insignificant prostate cancer and improving detection of clinically significant disease.

In routine clinical practice, both 1.5 T and 3 T systems are used for prostate cancer detection and staging. International guidelines support this practice.[23] 3 T does, however, provide higher SNR, which translates to a gain in spatial and temporal resolution (the latter for DCE), as well as decreasing scan time. For DWI at 3 T, it is important to use advanced techniques, such as zoomed imaging, to reduce image distortion, and susceptibility artifacts.

Oncologic Whole-Body Magnetic Resonance Imaging

Accurate tumor staging and surveillance are essential in oncologic patients. Whole-body (WB) MRI can be extremely useful in this role and is performed clinically at both 1.5 and 3 T. Although 3 T offers the potential benefit of a shorter examination, other factors favor 1.5 T. These specifically include its robust image quality, in particular with DWI (in WB examinations), and better suitability for metal implants (both in regard to artifacts and to safety/approval). Higher homogeneity of the main magnetic field over the complete field of view, with important consequences for DWI, fat suppression, and geometric distortion, is a distinct advantage for 1.5 T.

Pediatrics

Magnetic resonance imaging is an extremely important diagnostic imaging modality in pediatrics. Apart from the imaging of soft tissue, CNS and MSK pathology, MRI plays a major role in the imaging of the upper and lower abdomen in children. This is due to the poorer quality of CT scans in pediatrics in that area (because of the relative absence of fat) and the concern with CT due to radiation dose and risk of cancer later in life.

Adaptation of the MR examination to patient size is very important in pediatrics, and is fortunately facilitated by the abundance of differently sized receiver coils. The importance of obtaining sufficient SNR for a smaller voxel (necessitated by the smaller organ size) regardless favors 3 T in general over 1.5 T. Application of image acceleration techniques can also be important, due to limited patient compliance and the need to keep the examination short.

Because of the lack of ionizing radiation, MR urography represents the standard of care in pediatric patients (as opposed to CT). Magnetic resonance in this indication provides both morphological and functional information. Evaluation for congenital anomalies of the kidney and urinary tract is the most common examination. Given that most patients are evaluated as infants, and the small size of the critical structures to be evaluated, 3 T is favored over 1.5 T.

Of special note is cardiac MR, which is the accepted standard for noninvasive evaluation of congenital heart disease in children. Here, once again the benefits of higher field strength, specifically 3 T, apply, with the additional requirement for high clinical and technical expertise.

7 T MAGNETIC RESONANCE

The advantages of 7 T include substantially higher SNR and, in addition, higher CNR for certain specific types of scans.[24] Unlike at lower fields, where a linear increase in SNR occurs with increasing field strength, SNR increases at 7 T proportional to B0 raised to a power between 1.65 and 2.1, with noise no longer sample-dominated. These advantages are accompanied by significant challenges that include inhomogeneity of the RF excitation, high-energy deposition, and greater B0 field inhomogeneities.

Although 7 T was introduced near the turn of the millennium, regulatory approval did not come until 2017 (for the Siemens system), and is currently restricted to the head and extremities. Like when considering the move from 1.5 to 3 T, the SNR gain at 7 T can be used to achieve higher spatial resolution in a reasonable scan time, or shorter measurement times. Examples of the advantage of 7 T over lower fields in terms of higher CNR (other than that expected with the increase in SNR) include TOF MRA and arterial spin labeling, which both benefit from increased background suppression due to the longer T1 at higher fields. This advantage turns, however, to a disadvantage for techniques that require full T1 relaxation, which require longer measurement times. Another major advantage of 7 T is with techniques based upon susceptibility. Higher magnetic field provides increased susceptibility sensitivity, thus improving further for 7 T SWI (Fig. 3), quantitative susceptibility mapping, and functional MRI. The negative is increased susceptibility artifacts, which can lead to geometric distortion and signal dropout. Challenges in acquiring diffusion-weighted images (B0 and RF transmit inhomogeneity, shorter T2, higher SAR) have largely

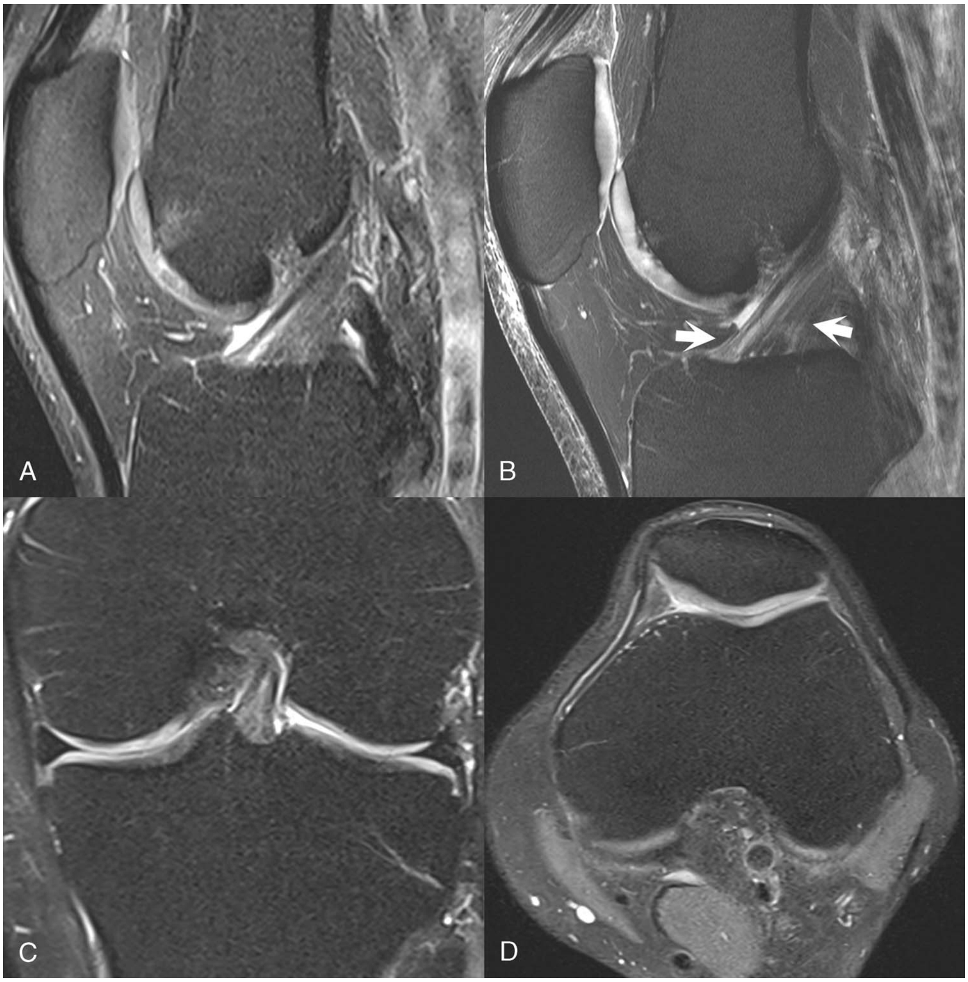

FIGURE 9. Proton density-weighted images of the knee (A, C, and D) at 0.55 T in the sagittal, coronal, and axial planes, and (B) at 7 T in the sagittal plane. The image quality is representative of that which can be achieved at 0.55 T, when scans are optimized for that field strength. The scan times at 0.55 T were 5.5 to 6 minutes, other than for (D) where the use of simultaneous multislice technique decreased the scan time to 3 minutes. The comparison to 7 T (with scan time held constant) is presented to show the higher spatial resolution (with the voxel size decreased 5-fold), SNR, and excellent image quality, which can be attained at the upper end of today's clinical field strengths. Note the exquisite visualization of the 2 major fiber bundles, the anteromedial and posterolateral (arrows), of the ACL at 7 T.

been met by technical advances, allowing brain DWI at 7 T. For spectroscopic applications, the greater splitting of resonance frequencies is advantageous for proton and phosphorus spectroscopy.

Field Inhomogeneity Issues

With increasing field strength, which equates to increasing resonance frequency, the wavelength for RF excitation approaches the size

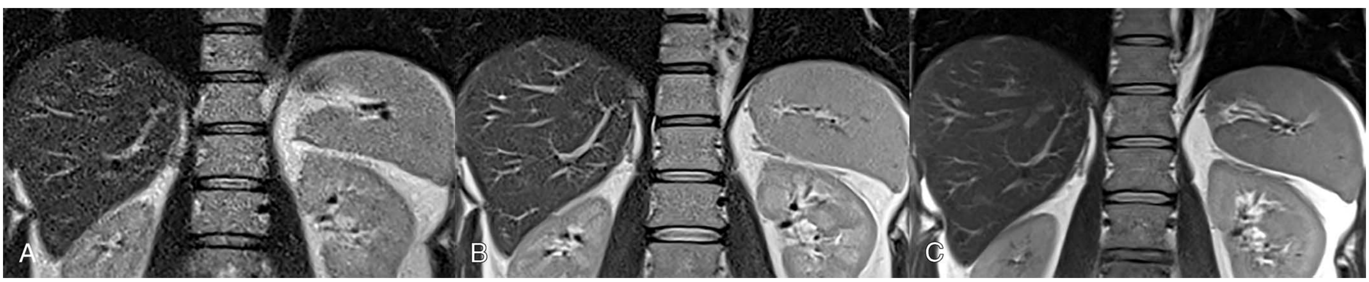

FIGURE 10. Coronal breath-hold HASTE images of the upper abdomen obtained at (A) 0.55 T, (B) 1.5 T, and (C) 3 T in a normal volunteer. Slice thickness, pixel size, and scan time (per slice) were held essentially constant to demonstrate the increase in SNR from low to high field. Specifically, the scans were not optimized for each MR system and field strength.

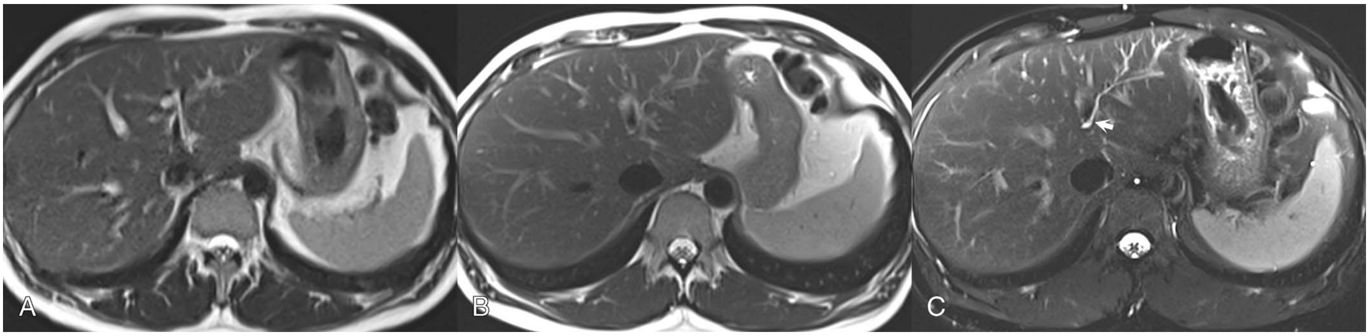

FIGURE 11. Axial breath-hold T2-weighted HASTE scans of the liver at (A) 0.55 T, (B) 1.5 T, and (C) 3 T, showcasing the image quality at each field. Higher field offers the possibility, due to the greater available SNR, to increase spatial resolution (decreasing both pixel size and slice thickness). This approach was combined with fat saturation, in the 3 T acquisition. Note also the visualization at 3 T of the hepatobiliary system (arrow) by its higher signal intensity. The voxel size at 0.55 and 1.5 T was $1.3 \times 1.3 \times 6$ mm^3, and that at 3 T $1.0 \times 1.0 \times 5$ mm^3. Additional scan parameters were TR = 421 milliseconds with 4 averages at 0.55 T and TR = 1660 milliseconds at 1.5 T and TR = 1240 milliseconds at 3 T, both with a single average. In a sense these scans, all in the same normal volunteer, offer 3 different perspectives of the liver (as viewed by field strength).

of the body parts to be imaged, with the result being standing wave effects. This leads at 7 T, in proton MR applications, to strong inhomogeneities in both transmission ($B1^+$) and reception ($B1^-$). The result is signal cancellation in the images, contrast variations, and regional SAR peaks. This has led to dedicated coil designs for each anatomical area to be imaged, for both RF excitation and reception, unlike those for lower fields. This is also the reason that the manufacturers do not offer an integrated WB coil, which for systems lower than 7 T in field strength is standard. The development, in the last decade, of RF coils with multiple transmit elements and channels for 7 T has provided an important advance. $B1^+$ shimming today can be applied dynamically, achieving homogeneous excitation for small regions of interest. Full parallel transmission techniques (pTx) exploit, as additional degrees of freedom, the amplitudes and phases of the multiple transmit channels. In addition to B1 inhomogeneities, hot spots can occur in local SAR, a safety issue that must be addressed. Likely the most pressing need for clinical 7 T, in terms of available technology, is the clinical approval/availability of pTx technology. This would greatly improve the available image quality throughout the body.

Brain Imaging

The major current clinical use of 7 T is for high-resolution, morphologic, proton imaging, both in the brain (described in this section) and in the MSK system (described subsequently). The increased spatial resolution achievable at 7 T can aid in diagnosis and clinical decision making. It should be noted, regardless, that for ultrahigh spatial resolution, subtle motion can cause substantial image degradation, with motion correction techniques being critical. Research continues in this area to develop improved approaches.

Methods based on magnetic susceptibility are particularly improved at 7 T (Fig. 3). Susceptibility-weighted imaging benefits, with improved MR venography, depiction of cavernous malformations, and characterization of white matter lesions.

Three-dimensional TOF MRA benefits at 7 T with improved visualization of small intracranial vessels (Fig. 6). There is enhanced depiction of small aneurysms and arteriovenous malformations, with detection of cerebral aneurysms ≤1 mm possible.

Depiction of small cortical infarcts is improved at 7 T as compared with 3 T. A growing body of evidence points to improved MS diagnosis at 7 T, with better visualization/detection of white and gray matter lesions, central veins within MS plaques, and iron deposition associated with plaques.

7 T offers as well an improvement in the assessment of neurodegenerative diseases. Evaluation of the hippocampus in Alzheimer disease, its internal structure, and volume reductions are improved. In patients with vascular dementia, there is improved sensitivity for white matter lesions and microbleeds, potentially allowing early diagnosis. The substantia nigra and its internal structure are better visualized, potentially aiding target identification for deep brain stimulation.

In patients with epilepsy, improved detection of epileptogenic areas, for example, focal cortical dysplasia, has been shown at 7 T as compared with scans at 1.5 and 3 T.[25] Better evaluation of hippocampal architecture is achieved (Fig. 4), as well, at 7 T in patients with mesial temporal sclerosis.

Musculoskeletal Imaging

Image quality, spatial resolution, and demonstration of critical anatomy far beyond that achievable at 1.5 and 3 T is readily accomplished today at 7 T for the MSK system, in particular for the knee, wrist, and ankle. Likely most important for clinical applications is the improved visualization of fine structures, and subtle lesions, in bone, menisci, cartilage, tendons, and ligaments. Assessment of deep and superficial cartilage zones is possible. It may also prove possible to assess cartilage matrix integrity and collagen damage, as demonstrated in recent clinical research. Attention to through plane artifacts is important for clinical imaging and interpretation.[26] Imaging of the spine however still presents major challenges, with novel coil technology likely required.

Body Imaging

Technical feasibility has been demonstrated for abdominal, pelvic (prostate), breast, and cardiac imaging at 7 T. These remain research areas, with potential routine clinical application in the future.

X-Nuclei, Phosphorus-31, and Sodium-23

In 2019, regulatory approval was granted for the Siemens 7 T system for sodium imaging of the head and phosphorous imaging and/or spectroscopy of the WB (excluding the head). These constitute "X-nuclei," nuclei other than protons that can be detected by MR. The strength of such imaging is the ability to provide information that cannot be provided by conventional proton MR. Low sensitivities and/or low in vivo concentrations however can lead to long scan times and low achievable spatial resolution.

Phosphorus-31 MRI/MRS enables detection of important energy substrates in the body, including specifically phosphocreatine, adenosine-triphosphate, adenosine-diphosphate, and inorganic phosphate. Absolute pH values can also be obtained. Most studies to date have involved the brain or calf muscle, with the largest number involving the former. Body applications, in the future, represent a promising clinical area.

Sodium-23 concentrations provide information about tissue state, reflecting energy metabolism and disruption of cell membrane integrity. Clinical studies in the brain have focused on neurodegeneration and neuroinflammation, brain tumors, and in particular MS. Sodium concentration is increased in brain tumor tissue, with potential applications for determining biopsy sites, for surgical/radiation therapy planning, and for monitoring of therapy response. There is a plethora of clinical studies at 7 T demonstrating the utility of sodium imaging in MS. Correlations have been demonstrated between brain sodium accumulation and disability, progression of disease, cognitive impairment, and lesion evolution, with potential clinical value for monitoring the disease course in MS patients.[27]

Proton Spectroscopy

Proton MR spectroscopy and MR spectroscopic imaging benefit at 7 T from larger frequency dispersion, increased SNR, and reduced J-coupling. Field inhomogeneities and SAR however present challenges. 7 T extends the list of metabolites that can be detected, and the sensitivity for such.[28] Promising investigations have been conducted in the brain in patients with glioma, MS, and Alzheimer disease.

Functional Magnetic Resonance Imaging

Neuronal activity in the human brain can be mapped by functional MRI. The BOLD (blood-oxygen-level-dependent) effect is used, exploiting the principle that oxygenated hemoglobin is diamagnetic, whereas deoxygenated hemoglobin is paramagnetic. Although the scientific explanation is somewhat complex, a higher BOLD signal is observed on T2* and T2-weighted images in activated brain regions. Submillimeter spatial resolution (<1 mm^3) can be obtained at 7 T (which has been demonstrated in visual and sensorimotor cortex), whereas spatial resolution is restricted to approximately 2 mm^3 at 3 T. Substantially improved mapping of neuronal activity is enabled in the cortex, cerebellum, and subcortical structures.

METAL—WHERE 1.5 T EXCELS

Metal artifacts are less severe at 1.5 T when compared with 3 T, due to the linear increase in susceptibility differences (tissue/metal) with field strength. When available, 1.5 T is thus preferred over 3 T.[29] The advent of high image quality 0.55 T (low-field) MR systems offers the potential to further decrease such detrimental scan artifacts, although this is limited by SNR, thus impacting the choice of spatial resolution and scan time.

Regardless of field strength, FSE pulse sequences should be used, as opposed to spin echo, due to reduced signal loss adjacent to the implant and overall improved image quality, whereas gradient echo sequences should be avoided. The most important, single, sequence parameter that effects, and can be used to reduce in-plane metal artifacts, is the receive bandwidth. Higher receive bandwidth requires a stronger gradient, improving gradient linearity (gradient-induced field differences are increased relative to those induced by susceptibility differences, ie, metal). Higher receive bandwidth will decrease SNR, but this is typically compensated by increasing the number of scan averages. Reducing the slice thickness can also have an important impact, as this requires a stronger slice selection gradient, with less through-plane distortion and intravoxel dephasing. Decreasing the pixel size also reduces dephasing and thus signal loss, but has little effect on in-plane distortion.

Increasing the RF excitation pulse bandwidth, such as decreasing slice thickness, leads to a higher strength slice selection gradient with similar benefits. SAR however is increased, and this parameter is not typically accessible to the user. Often this change will be incorporated in scan sequences supplied by the manufacturer specifically for decreasing metal artifacts.

Fat suppression should only be used when necessary, with selection of the specific technique critical. Both spectral and Dixon-based fat suppression techniques perform poorly, with the latter typically producing a high-quality image at a distance, but failing in close proximity to metal. STIR is the method of choice, due to its relative independence from resonance frequency. The major tradeoff however is lower SNR as compared with other fat suppression techniques. Another caveat is that enhancing tissue (on postcontrast scans) can have a T1 similar to fat and thus also be suppressed.

When large metal implants—specifically including most arthroplasties—are to be scanned, additional advanced methods for metal artifact reduction are typically necessary to obtain images for clinical diagnosis. View angle tilting (VAT), an early such technique, corrects for in-plane, but not through-plane, distortions. View angle tilting unfortunately also causes image blurring, which can be reduced by the use of thin slices. To achieve the in-plane distortion reduction (VAT), a second gradient is applied in the slice-select direction, in addition to the readout gradient. The result is a slight tilt of the readout direction, and likewise the voxels. Slice encoding metal artifact correction (SEMAC) and multiacquisition with variable-resonance image combination (MAVRIC) are more recently developed approaches. Both provide a substantial reduction in the metal artifact around implants when compared with FSE techniques alone. SEMAC is a 2-dimensional (2D) technique that reduces both in-plane misregistration and through-plane distortion. MAVRIC is a similarly effective 3D technique. SEMAC incorporates VAT, with each imaged slice additionally phase encoded in a third dimension (z). In-plane distortions are suppressed by VAT and through-plane by the z-phase encoding. The number of slice-encoding (z) steps is user defined, specifying the number of adjacent slices assessed for artifact reduction during readout of any specific slice. The larger the susceptibility artifact, the more slice-encoding steps needed for adequate artifact reduction. This increases scan time, a major drawback, with sufficient artifact reduction often requiring >10 slice-encoding steps (and thus greater than a 10-fold increase in scan time). The introduction of SEMAC regardless represented a major step forward for the evaluation of patients with metal implants at 1.5 T, providing markedly superior artifact reduction to the scan techniques previously available. SEMAC subsequently became a clinically viable technique (with acceptable scan times) by the addition of compressed sensing. With MAVRIC, phase encoding is performed along 2 dimensions (since it is a 3D technique), with thus less distortion as compared with gradient-based (2D) slice selection. Signal loss would however still occur since the nonselective RF excitation pulse cannot cover the entire range of off-resonance frequencies near a metal implant. This is solved by acquiring several 3D slabs with varied resonance-frequency offsets. Disadvantages include aliasing in the through-plane direction (due to the nonselective 3D volume excitation)—an important negative for hip and shoulder scans, and the lack of slice selectivity (with 3D scans being time-consuming). Compressed sensing has subsequently been implemented for MAVRIC, as with SEMAC, making possible reasonable scan times. Today there are many, many excellent scientific articles showing SEMAC accelerated to match the scan times of more conventional FSE sequences, with excellent overall image quality and a marked reduction of metal artifacts.[30]

MUSCULOSKELETAL IMAGING—WHERE 3 T EXCELS

3 T excels in MSK imaging due to the improved SNR provided by the increase in field strength. The available SNR defines the achievable spatial resolution, and likewise tissue contrast, with higher SNR leading to improved image quality and efficacy.[31] There is a 2-fold SNR gain when compared with 1.5 T, enabling theoretically a 4-fold faster data acquisition. If acquisition time is held constant, the gain in SNR at 3 T makes possible improved in-plane spatial resolution, and/or acquisition of thinner slices. The increase in field strength also leads

to an increase in tissue contrast, with T1 typically 20% longer, and T2 15% shorter. As with scanning at any MR field strength, repetition and echo times, echo train lengths, and interecho spacing need to be optimized. Longer TRs are required to recover sufficiently the longitudinal magnetization of fluid and articular cartilage. Because of the shorter T2, shorter echo times are needed for optimal tissue contrast and higher SNR. Sequence optimization commonly involves longer echo trains and shorter echo spacing, together with the use of parallel imaging, SMS, and compressed sensing.

The 3 T MR systems are commonly equipped with high-performance gradients and RF coils, further accounting for the image quality improvement when compared with lower field units. The 3 T MSK RF coils typically feature a higher number of elements and channels, leading to higher SNR. Faster excitation and refocusing radiofrequency pulses require less time within a pulse sequence and increase the speed of scan acquisition. Fast and strong gradients require less time for phase encoding, further speeding data sampling. Shorter echo spacing decreases susceptibility artifacts and blurring, resulting as well in faster acquisitions and sampling of earlier echoes with higher signal.

The high intrinsic SNR at 3 T, in combination with multielement, multichannel receive coils, provides an ideal platform for the routine use of parallel imaging, specifically to accelerate 2D and 3D scan sequences. Simultaneous multislice techniques are also routinely used, in particular for 2D DWI and 2D FSE pulse sequences, further accelerating acquisition. Combining SMS and parallel imaging can facilitate higher resolution and shorter scan times, and as such is commonly used in 3 T knee MRI. In certain instances, use of a combined transmit and receive coil can be important, as well as the use of dedicated RF pulse designs and lower flip angles, to stay within SAR limits.

Two phase encoding directions are required to acquire a 3D scan, as opposed to one with a 2D scan. Although this leads to higher SNR, scan times are also substantially longer. High-resolution 3 T MSK protocols using 3D scans are regardless possible and routinely used. Two-dimensional and 3D CAIPIRINHA, and compressed sensing, are all important techniques that enable the acquisition, in a reasonable scan time, of high-resolution MSK scans at 3 T, for example, that in the knee featuring isotropic $0.5 \times 0.5 \times 0.5$ mm^3 resolution.[32]

CARDIAC IMAGING—THE CHOICE BETWEEN 1.5 AND 3 T

1.5 T was for several decades considered the field strength of choice for cardiovascular MR (CMR) examinations. Today 3 T is also widely used, with some examinations better performed at 1.5 T and others at 3 T, with no single "best" field strength for CMR.[33]

Cine imaging is a key component of cardiac MRI, which is used to assess cardiac function, to provide ventricular volumes, and to depict wall motion abnormalities. The scan sequence of choice, whether at 1.5 or 3 T, is retrospectively gated balanced steady-state free precession. This scan technique provides high SNR and excellent blood/myocardium contrast. Studies have shown little difference in left ventricular indices, volume, and mass for examinations whether performed at 1.5 or 3 T. The pros for 3 T, due to the higher SNR, include improved temporal and/or spatial resolution, and/or a reduction in the time required for the patient to hold their breath. The cons for 3 T include higher RF power deposition, B0 and B1 inhomogeneity, and greater difficulty with "dark band" (off-resonance) artifacts. Solutions exist for each of these potential problems at 3 T, which otherwise might compromise image quality.

Edema imaging makes possible the differentiation between acute and chronic myocardial injury. Myocardial edema can be focal or diffuse, depending on etiology. T2-weighted scans are typically used, with suppression of both fat and the signal from flowing blood within the cardiac chambers. T2 mapping offers the possibility of a more objective measure, in particular for diffuse edema, which can be more difficult to visualize and evaluate. The pros and cons of 3 T in comparison to 1.5 T are similar to that seen with cine imaging, with the necessity of fat suppression (and potential artifactual myocardial signal suppression) as an additional complexity.

Late gadolinium enhancement provides visualization and quantification of permanent myocardial injury, whether due to fibrosis or acute infarction. This evaluation is an important component of nearly every CMR examination. There are many studies comparing 1.5 and 3 T directly for late gadolinium enhancement imaging. Regardless of the imaging technique used, 3 T provides higher CNR between normal and infarcted myocardium, whether the injury is acute or chronic. Once again, balanced steady state free precession (bSSFP) readout techniques lead to more image artifacts at 3 T.

Stress perfusion CMR is one of the methods of choice for detection of myocardial ischemia, along with SPECT, PET, and echocardiography. The examination uses DCE scans, acquired at peak vasodilatory stress and at rest during bolus contrast injection. Areas of reduced myocardial perfusion appear hypointense. At 1.5 T, balanced steady-state free precession, spoiled GRE, or echo planar scans are all used. At 3 T, the technique of choice is spoiled GRE. Although similar caveats exist as previously noted, stress perfusion has been consistently shown in the scientific literature to be of higher diagnostic value at 3 T as compared with 1.5 T.

Parametric myocardial mapping enables spatial visualization of quantitative changes in T1, T2, and T2*. For T1 and T2, both 1.5 T and 3 T are routinely used clinically today. However, myocardial T2* measurements are typically performed only at 1.5 T (due to susceptibility artifacts at 3 T, and the very short T2* at 3 T—allowing only 2–3 echoes, as well as other issues). Volume selective shimming is usually performed at both field strengths, and RF field volume shimming also at 3 T.

Complex flow patterns can be visualized and quantified with 4D flow technique, which provides 3D time-resolved, velocity-encoded data. This technique provides additional information to guide clinical therapy, including flow velocities, volumes, and wall shear stress. In terms of the choice of field strength, valvular disease and aortic hemodynamics are well evaluated at both 1.5 and 3 T, with published studies favoring 3 T.

No single best field strength (specifically when comparing 1.5 and 3 T) exists today for cardiac MR, with each field having its advantages. If a site can only scan on a single MR unit, 1.5 T is still preferred, due to ease of use, the ability to perform all major clinical cardiac examinations, and accessibility for patients with cardiac devices or implants.

MULTIPLE SCLEROSIS—WHERE 7 T EXCELS

7 T offers 2 main advantages over lower field strengths, and specifically 3 T, for the evaluation of MS. These are increased SNR and improved sensitivity to susceptibility changes.[34] The increase in SNR comes primarily from the increase in field strength. Improved SNR makes possible higher spatial resolution, as well as achieved tissue CNR. The higher field strength also emphasizes tissue susceptibility effects. The contrast achievable with SWI is thus markedly improved, with better visualization of central veins and paramagnetic rims that are seen with some MS plaques.

Cortical MS lesions are difficult to visualize at lower field strengths, even 3 T, something that has only been realized in recent years. In comparing postmortem MR with histological findings, it has been found that less than a quarter of cortical lesions can be visualized by imaging (when imaging at 7 T is not included). Likely reasons include the small size of some lesions, CSF partial volume effects, and the different intrinsic signal intensity of the cortex (due to lower myelination). Both in vivo clinical MR evaluations and postmortem MRI have shown the higher sensitivity of 7 T, in comparison to 3 T, to cortical lesions in MS.[34,35] Use of advanced/innovative imaging sequences can also be very important.[36] 7 T makes possible as well improved classification of lesions as cortical or subcortical in location, which is important for the specificity of diagnosis (given that subpial lesions are felt to be pathognomonic for MS). Detection of MS lesions in subcortical gray matter and the basal ganglia is also improved at 7 T.

A centrally located vein, occurring within a focal white matter lesion, can be seen on MR in some MS plaques. This is an additional, highly specific marker allowing improved differentiation of MS from other diseases that cause a similar pattern of white matter disease. Veins, in general, are readily detected on T2*-weighted MR scans due to the paramagnetic effect of venous blood. Thus, detection of such central veins is markedly improved at 7 T, as with cortical MS lesions (albeit on a different basis), when compared with lower field strengths and specifically 3 T.

Paramagnetic rims, which represent perilesional iron, are thought to mark chronic active white matter MS lesions. These are well visualized at 7 T and, from a theoretical basis, should be best seen at 7 T as opposed to lower fields. Paramagnetic rims also are of value in differential diagnosis, adding to specificity in the diagnosis of MS. Contrast-enhanced MR has further shown that these paramagnetic rims correspond to areas of blood-brain barrier disruption, thus representing active inflammation.

Some technical challenges remain for the routine clinical application of 7 T in brain imaging. These include RF power deposition, for example, with FSE T2-weighted sequences, and both transmit and receive B1 inhomogeneities. For example, 3D MP-RAGE, a pulse sequence widely used at 3 T to achieve high-quality T1-weighted scans, suffers at 7 T from such B1 inhomogeneities. This leads to artifactual signal heterogeneity in the final images. MP2RAGE represents one possible solution, replacing MP-RAGE in 7 T imaging.

CONCLUSIONS

State-of-the-art MR systems approved for clinical use operate across a broad range of field strengths, from 0.55 T to 7 T. Despite the popularity of 1.5 and 3 T in past decades, utilization is very likely to spread more generally across this spectrum. Each field strength, whether 0.55 or 1.5 or 3 or 7 T, offers benefits for clinical use not matched by the other magnetic fields. Features that distinguish state-of-the-art scanners across this breadth of field strength include ease of installation (0.55 T), cost (0.55 T), familiarity in regard to operation (0.55 and 1.5 T), broad acceptability for all clinical examinations (1.5 T), improved imaging across all anatomic areas together with higher spatial resolution and access to all advanced imaging techniques (3 T), and recognition as the new criterion standard for a few, but limited, anatomic areas (7 T).

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