The Next Generation – Advanced Design Low-field MR Systems

Val M. Runge, M.D.; Johannes T. Heverhagen, M.D., Ph.D.

Department of Diagnostic, Interventional and Pediatric Radiology,
University Hospital of Bern, Inselspital, University of Bern, Switzerland

Low-field superconducting MR systems, operating between 0.35 and 0.6T, were only briefly evaluated for clinical use in the 1980s before they were superseded by higher field systems. An important question today is the potential of such units, operating at a known sweet spot – 0.55T, employing in design all the knowledge gained during the interval decades. Looking at cost, flexibility, image quality, and accessibility, there is a very bright future for advanced design low-field MR units, which should expand markedly, worldwide, the use and clinical value of MR.

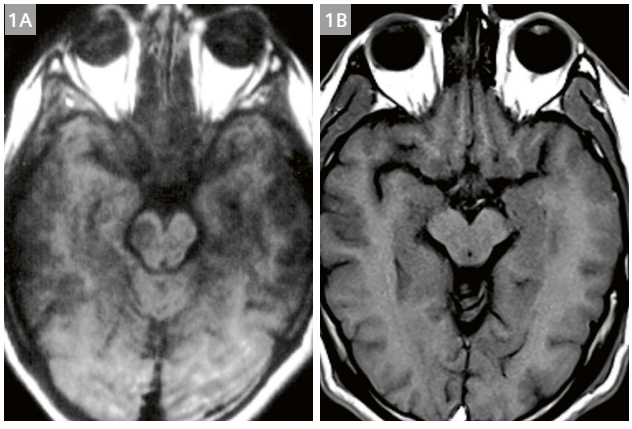
A brief history of the evolution of MR field strength for clinical systems

Paul Lauterbur and Peter Mansfield jointly shared the 2003 Nobel Prize in Physiology / Medicine for their fundamental work in the 1970s in the field now known as MRI. Soon thereafter, John Mallard introduced the first whole body MRI system, which operated at a field strength of 0.014T. The potential for improved SNR with higher field strength was quickly recognized, resulting in a second prototype operating at 0.028T, but still utilizing a resistive magnet.

The initial commercial development of clinical MR – in the early 1980s – was led by two companies no longer in existence, Diasonics and Technicare. Both used superconducting magnets, with the first company delivering 0.35T units, and the second company initially 0.5T units (Fig. 1) and subsequently 0.6T units. In the mid 1980s Siemens' first commercial units were delivered. These operated at a field strength of 1.0T, a theoretical optimum defended by many prominent scientists of the day.

In the late 1980s, a marketing blitz by one major manufacturer, who was yet to enter the field, led to all of the major X-ray manufacturers developing 1.5T systems. Standardizing on a field strength of 1.5T was a radical idea, with no clinical systems having been delivered at that time with such a high field strength. Much of the premise for development of this field strength was based on the possible clinical development and utility of techniques that would be thus enabled, such as phosphorus spectroscopy. This premise later proved largely false. Nevertheless, all the major vendors were forced to invest, largely due to marketing pressure, in the development of clinical 1.5T units. By the 1990s, delivery of 1.5T units dominated the industry.

Then in the 2000s, the debate began concerning 3T, primarily on the basis of brain imaging, which was indeed the only exam of sufficient quality for clinical diagnosis that these early 3T whole body systems could acquire. There were major challenges to make 3T an acceptable scanner, not only for the brain but also for the spine, musculoskeletal system, and body. The prolongation of T1, in plane and through plane chemical shift, and perhaps most prominently patient heating (SAR) all presented major challenges to overcome in making 3T clinically viable. Indeed, despite the very high quality of scans at 3T in many anatomic areas today, the debate continues regarding 1.5T versus 3T. Cost is a major impediment, being substantially greater for 3T in terms of the system itself as well as installation. Reflecting this debate and heavily these costs, today for new MRI units two 1.5T systems are still sold for every 3T.



1 A historical comparison of low field brain imaging in **(1A)** 1984 (on a Technicare 0.5T scanner) versus **(1B)** the current 2020 standard. The scan time was reduced from 10 to 4 minutes, accompanied by a marked improvement in SNR and spatial resolution due to interval technologic advances using the MAGNETOM Free.Max system.

MAGNETOM Free.Max is currently under development and is not for sale in the U.S. and in other countries. Its future availability cannot be ensured.

What data exists regarding low-field imaging from the 1980s and 1990s?

The development of 1.5T imaging in the late 1980s occurred despite the lack of substantial evidence at that time, supporting that field strength, for medical diagnosis and sensitivity to disease. More specifically, few large-scale clinical trials exist from that era comparing efficacy at low-field to that at 1.5T. It is to be granted that results today at 1.5T are indeed excellent, but let us turn to the little data that was available comparing field strengths during that era of rapid development.

From the scant scientific literature, two major publications / clinical trials stand out for their comparison of low and high field. These trials cover two important anatomic areas of clinical utility for MR, the brain and the musculoskeletal system. Both provide little evidence of an advantage to 1.5T in terms of either diagnosis or sensitivity to disease. In a large clinical trial involving patients with suspected multiple sclerosis, no difference in accuracy, sensitivity, or specificity was noted between 0.5 and 1.5T studies [1]. A similar design large-scale clinical trial was performed with patients referred for imaging of the knee [2]. Evaluation for anterior or posterior cruciate ligament and meniscal tears showed no advantage for the higher field strength in terms of accuracy and diagnosis.

In comparing 0.5 and 1.5T, the hypothesis still stands today as it did in 1996 "that applications that require very fast imaging, very high resolution imaging, or detection of very small image intensity changes *may* demonstrate diagnostic advantages for high magnetic field" [3]. It is important also to recall that some of the prominent arguments favoring 1.5T and above included techniques that are little used clinically today, such as spectroscopy, fiber tracking and functional MR. Also, making low-field much more viable are the many, significant technologic advances that occurred in the interval years. These are considered in the sections that follow. For a moment, however, let us consider the problem that we are confronted with, and that is the reality of physics in terms of signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR).

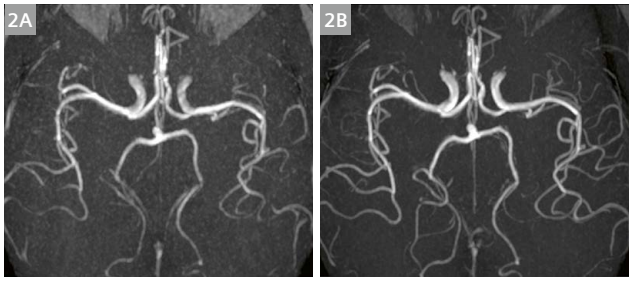
SNR increases linearly with field strength, with some assumptions. One is that receiver bandwidth is held constant. However, bandwidth for any particular scan sequence is usually increased at higher fields to account for chemical shift. When the pixel shift is held constant, then SNR scales with the square root of field strength (not linearly). CNR is a more complicated situation, in part due to the increase of T1 with field strength. Taking these factors into account, for T1-weighted scans the increase in CNR from 0.5 to 1.5T is in the range of 20%, while it can be more than 40% for scans with little T1 contribution.

One caveat to the consideration of the data from the 1980s and 90s is that Time-of-Flight (TOF) MR Angiography and contrast enhanced MRA were not evaluated. These techniques had not yet been invented. Thus, it remains a question – which is presently being answered – until next generation low-field units are further along in development how far interim software and hardware advances can close the gap in image quality for MRA between low and high field (Fig. 2).

What opportunities exist today for low-field MR?

The question is what opportunities exist for making MR accessible to a broader patient population, and/or more cost efficient [4]? The development of high field was pushed due to the promise of increased SNR and thus higher image quality. However, looking just at one area, and that is the spine, the march from 0.5 to 1.5 to 3T has not truly met one's expectations. Chemical shift and CSF motion created problems, some of which still exist today, and slice thickness for routine scans only moved from 5 mm to 3–4 mm. Regardless, a modern low-field system is expected to achieve comparable image quality, and thus have sufficient SNR. Alternatively, for the installed site, the number of applications demanding thin section imaging need to be very low, to justify purchase of a system without such capabilities. It should be kept in mind however that thin section imaging in certain instances can be achieved with longer scan times.

Along the way – in the development of higher field units – new problems were encountered, due to – for example – SAR, patient safety, tissue susceptibility, and not to be forgotten, cost. In regard to the latter, the specifications and infrastructure requirements of MR-systems have grown substantially over the years, keeping MR an extremely expensive imaging modality, limiting patient access and utilization.



2 Time-of-flight MRA was not developed until relatively late for MR, and thus the question remained – answered with this figure – about the diagnostic potential at 0.55T. TOF MRA performs well, comparable to 1.5T, with, as anticipated, a slight reduction in SNR. Thick axial MIP reformats are presented from scans at **(2A)** 0.55T and **(2B)** 1.5T with voxel dimensions of 0.5 x 0.5 x 0.5 mm^3 in each instance and approximately the same scan time.

Today there is tremendous economic pressure on healthcare systems worldwide. Thus, the following questions seem worth revisiting. Is it possible to reduce the cost of the most expensive part of an MR system, the magnet, as well as the next most expensive part, the gradients, and still achieve excellent image quality? Is it thus possible to add new diagnostic value, by making MR more accessible both in developed countries and in less developed areas, as well as for niche applications (such as interventional MR)? Is there a missed opportunity in the road that clinical MR and the research community have taken since the advent of this technology in the 1980s?

Intrinsic advantages for low-field systems include shorter tissue T1 and longer T2* (allowing more time efficient scan acquisitions), reduced susceptibility effects, and reduced specific absorption rate (tissue heating). Reduced SAR lessens scan parameter constraints (flip angle, TR, number of slices) and diminishes heating of metal devices and implants. Low-field technology was last explored in depth in the 1980s, long before the development of many current acquisition and post-processing strategies, including spiral acquisition, parallel imaging, iterative reconstruction and most recently deep learning reconstruction.

A new look at low-field technology today is highly recommended, holding the potential for development of advanced, next generation MR systems with markedly lower cost yet excellent image quality. Such a development could lead to a wide range of new scanners, from basic systems destined for small clinics or developing nations to higher end niche systems including dedicated emergency room, intraoperative, and interventional units.

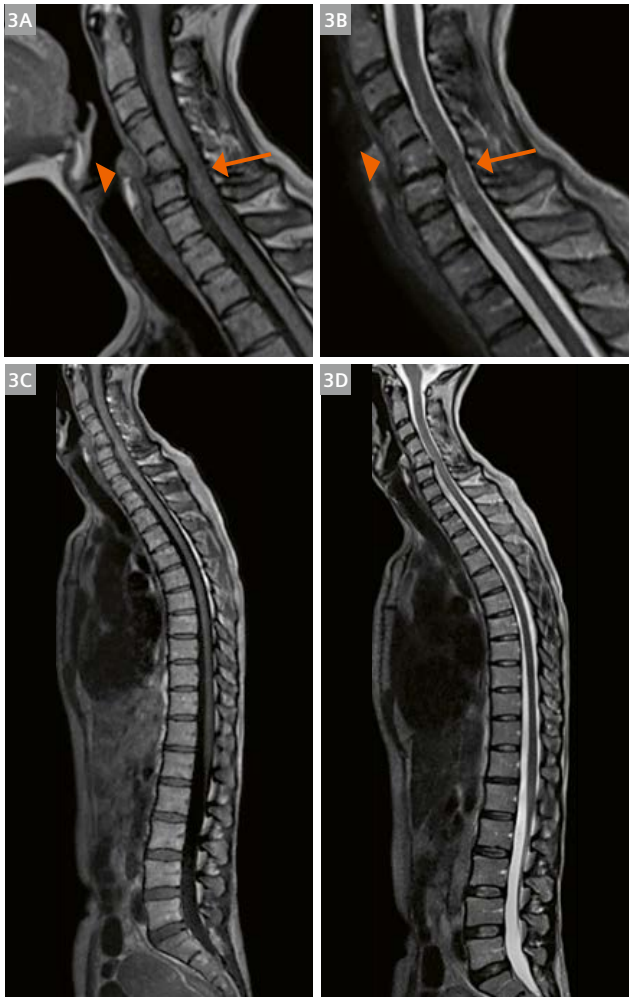
Magnet and receiver coil technology

Bore size is an important consideration in design of an advanced, next generation low-field system. The early, whole body, superconducting MR clinical units had a 60 cm width bore, although the bore was even slightly smaller in several designs that were generally not successful.

The first wide bore (70 cm) unit, the MAGNETOM Espree (which operated at a field strength of 1.5T), was launched in 2004. This design at the time was highly innovative, and the unit subsequently dominated the sales market, largely due to patient comfort and the increasing weight of patients worldwide. Since that time, wide bore high field (3T) units have also become available. In terms of the design of a next generation low-field system, the comparatively small amount of superconducting wire needed makes viable, cost wise, ultra-wide-bore systems, with bore dimensions in the range of 80–90 cm.

A huge barrier, both from a practical point of view and cost, is the siting of a unit, for example in operating rooms, remote clinical sites, and developing world clinics. A zero boiloff magnet, with elimination of the quench pipe and a markedly reduced foot print (including the 5 gauss line), are possible in the near future and offer great promise for dissemination of MR technology world-wide.

In terms of receiver coil technology, there have also been major advances since the 1980s that can be applied to next generation low-field systems. In the early years of MR, receiver coils were far from optimized, with for example head coils both larger in diameter and longer than needed. For body imaging, RF reception was

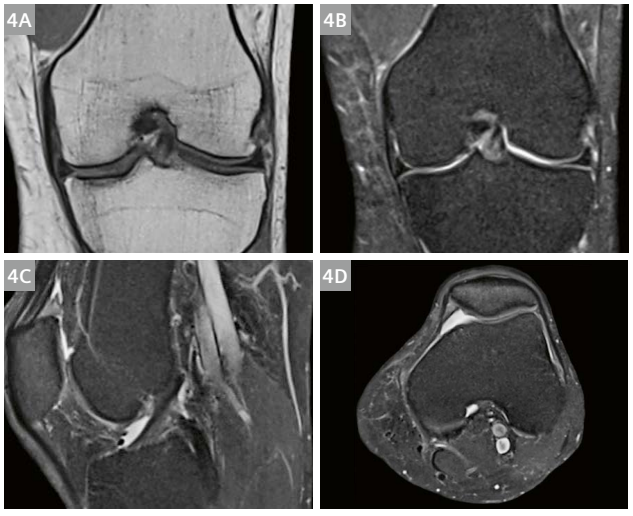


3 Sagittal **(3A)** T1- and **(3B)** T2-weighted 2D fast spin echo scans at 0.55T of the cervical spine. The slice thickness was 4 mm and 3 mm, respectively. Scan times were 3 minutes 10 seconds and 4 minutes 4 seconds. There is a mild retrolisthesis of C5 on C6. There is a disk-osteophyte complex at C5–6 with loss of disk space substance and mild endplate degenerative changes (arrow). The disk space at C4–5 is small (arrowhead), and given the appearance of the C4 and C5 vertebrae on the sagittal images, this likely represents a congenital block vertebral body (C4–5). The thoracic and lumbar spine that are displayed in the sagittal **(3C)** T1- and **(3D)** T2-weighted full spine images are essentially normal. The slice thickness in (3C) and (3D) was 4 mm. Scan times for the whole-spine scans were (3C) 9 minutes 30 seconds and (3D) 8 minutes 48 seconds.

performed using the body coil – placed far away from the patient and thus with relatively poor SNR. The advances we take for granted today – that have led to major improvements in SNR and paid into the capability for acquisition acceleration, such as multichannel, multielement coils, flexible coils, and specific contoured coils for body regions (for example the shoulder, knee, wrist, ankle, and neck) were in the 1980s and 1990s still decades away from development. Spine, musculoskeletal, and liver imaging at 0.55T will all benefit greatly from these technologic advances (Figs. 3–7).

Gradient performance

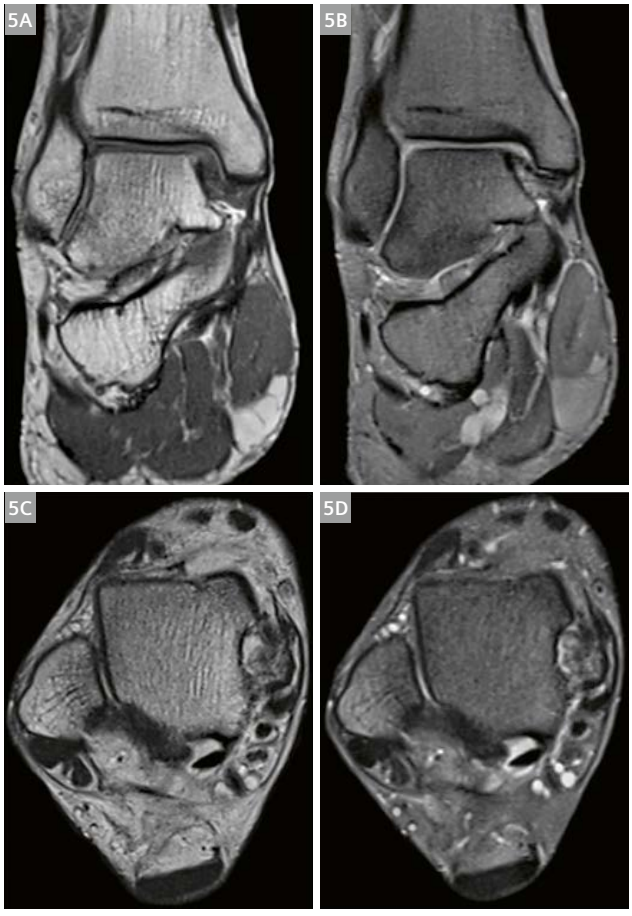
The gradient system is the second largest cost, following the magnet, for an MR unit. Over the years, there have been many advances regarding the magnetic field gradients, although today the slew-rate for whole body-systems is constrained not by technological limits but by physiology and specifically nerve stimulation. Research applications, for example high resolution DTI, however have primarily driven the quest for very high gradient amplitudes. There are cost issues here, due to manufacturing and design complexity, as well as increased power consumption and cooling requirements.



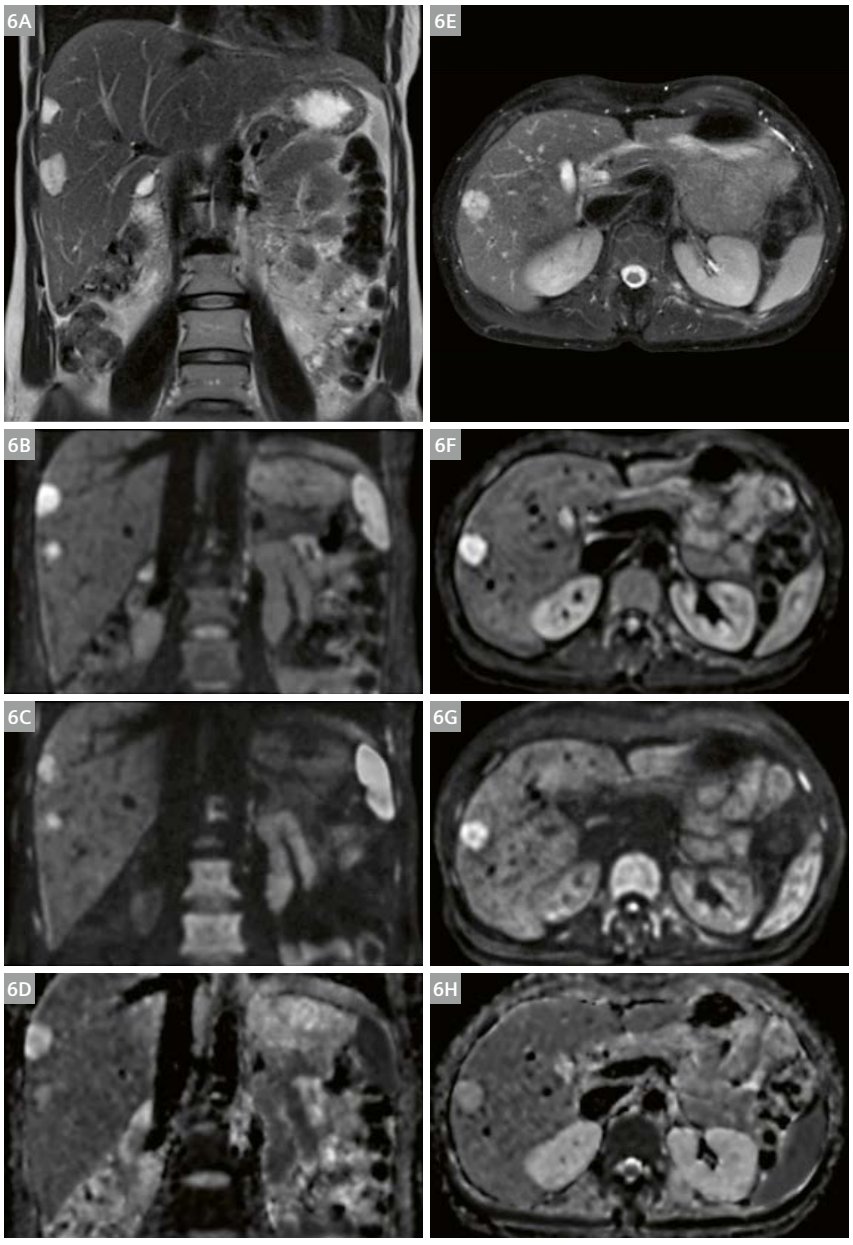
4 Coronal 2D fast spin echo **(4A)** T1-weighted images of the knee, and coronal, sagittal, and axial **(4B, C, D)** 2D proton density-weighted images with fat saturation, all images obtained at 0.55T in a normal volunteer. The slice thickness in each instance was 4 mm. Scan times were (4A) 4 minutes 4 seconds, (4B) 5 minutes 2 seconds, and (4C, D) 5 minutes 22 seconds, respectively.

However, much like cars, one does not always need a Ferrari, a BMW will do. One cannot always drive one's car at 200 kilometers an hour, and most Ferraris spend very little of their life doing such speeds. Thus, the question is, for MR, for daily, routine clinical use which techniques require peak amplitude of the gradients, how often are they used, and are there other ways to reach such requirements?

One of the techniques that drives the gradients the hardest is diffusion-weighted imaging (DWI). If one simply uses an older gradient system, with lower specifications, then the change required for high end DWI will be to



5 Coronal 2D fast spin echo proton density-weighted images of the upper ankle **(5A)** without and **(5B)** with fat saturation, obtained at 0.55T. Voxel dimensions were (5A) 0.5 x 0.4 x 3.0 mm^3 and (5B) 0.6 x 0.5 x 3.0 mm^3, with scan times of 3 minutes 47 seconds and 3 minutes 46 seconds.
Axial 2D fast spin echo T2-weighted images of the upper ankle **(5C)** without and **(5D)** with fat saturation, obtained at 0.55T. Slice thickness of the T2-weighted images was 3 mm, scan time was 3 minutes 37 seconds and 3 minutes 19 seconds.



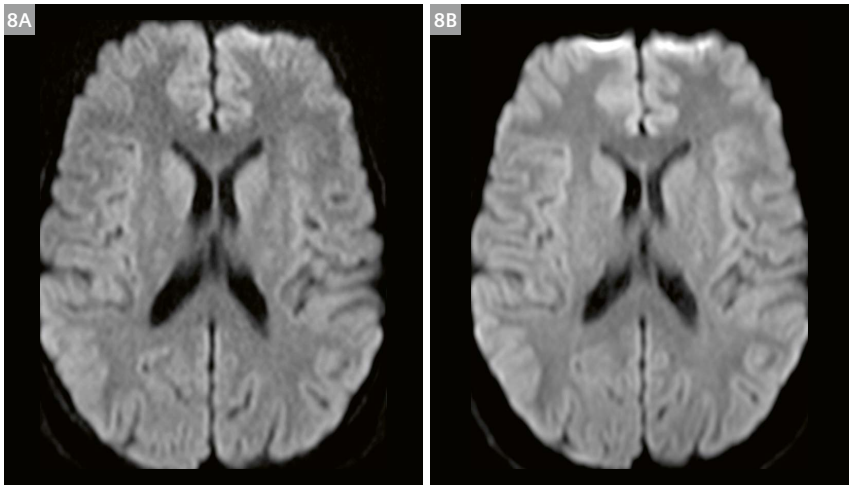
6 Coronal **(6A)** respiratory triggered 2D T2-weighted images of the abdomen, using a BLADE acquisition technique. Scan time was 2 minutes 26 seconds. Axial **(6E)** respiratory triggered 2D T2-weighted fat saturated images of the abdomen, using a BLADE fast spin echo acquisition. Scan time was 2 minutes 50 seconds. Images are presented from a normal volunteer at 0.55T and show two small liver hemangiomas with characteristic hyperintensity on T2w and hyperintensity on DWI and ADC. The b = 50 s/mm^2 **(6B, F)** and b = 800 s/mm^2 **(6C,G)** diffusion-weighted scans were obtained with single-shot echoplanar technique, the respective ADC-maps **(6D, H)** are also presented. Scan time for the coronal diffusion-weighted scans was 2 minutes 10 seconds, scan time for the axial diffusion-weighted scans was 3 minutes 26 seconds. The slice thickness was 6 mm in every instance.

7 Coronal breath-hold 3D T1 VIBE Dixon water images **(7A)** of the abdomen with a slice thickness of 3 mm. The scan time was 19 seconds.
MRCP **(7B)**: MIP reformat from a respiratory-triggered 3D T2 SPACE acquisition, displaying the right and left hepatic duct, the common hepatic duct, the common bile duct, and the pancreatic duct. Voxel dimensions were 1.1 x 1.0 x 1.0 mm^3, with a scan time of 4 minutes 19 seconds (CS factor 10).

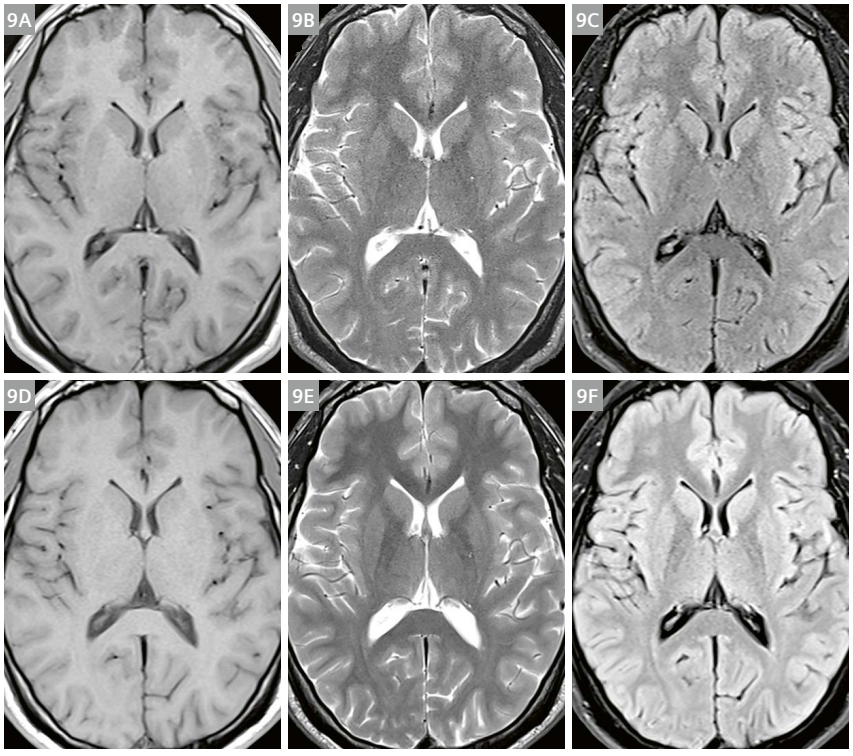
increase the TE on the order of 10–15 msec. At high field, this is actually a very substantial change. There, TE and echo spacing are kept as short as possible to minimize susceptibility and maximize SNR. However, at low-field, T2* decay and susceptibility are much less of an issue, with longer echo spacing acceptable and the SNR issue of longer TEs compensated with lower readout bandwidths (Fig. 8). This example illustrates the need to think outside the box for the design of low-field gradient systems. Balancing imaging parameters is an optimization problem with different boundary conditions at low-fields. By careful design, the potential disadvantages of a low cost gradient system can be mitigated with a non-traditional approach. High image quality can be achieved, with the lower cost of the magnet and gradients offering major advantages in terms of broadening access to MRI.

Image contrast

It is important to note that T1, T2, and T2* all change with magnetic field. Depending upon the specifics, this could be an advantage or a disadvantage for low-field (Fig. 9). T1 shortens by 1/3rd at low-field when compared to 1.5T,



8 2D single shot epi DWI (b = 1000 s/mm^2) at **(8A)** 0.55T and **(8B)** 1.5T, acquired with the same slice thickness (5 mm) and pixel dimensions. Known problems at higher field can be seen, for example the image distortion anteriorly (due to susceptibility effects from the frontal sinus), mild image blurring, and slight image foreshortening – which are not present on the 0.55T image. Due to doubling of the number of averages, the scan time at 0.55T was about twice that at 1.5T.



9 2D TSE **(9A, D)** T1-weighted, **(9B, E)** T2-weighted, and **(9C, F)** FLAIR images of the brain at (9A–C) 0.55T and (9D–F) 1.5T. To be noted is the mildly improved T1 contrast at 0.55T, due to the increase in T1 with field strength. The slice thickness was 5 mm. Scan times ranged from approximately equal for the two fields to about twice, depending upon technique.

which is advantageous for T1-weighted scans. Other advantages, specifically for echo-planar and spiral acquisitions, include that T2 is longer by a 4th, and T2* longer by almost half. Low-field may have as well applications for lung imaging, where the situation is unique, with T2* prolonged more than 3-fold [5].

Spiral imaging

Improved signal sampling efficiency can be achieved at low-field due to the prolongation of T2*. Using a spiral out acquisition (as opposed to Cartesian sampling), balanced steady-state free precession and spin echo techniques can achieve nearly double the SNR [5]. The rationale for high field imaging was the gain in SNR, in theory (although with many caveats) being linear with the increase in field strength. A spiral out acquisition at 0.5T can offer a gain of two times in SNR, largely negating the three times higher (in theory) SNR at 1.5T.

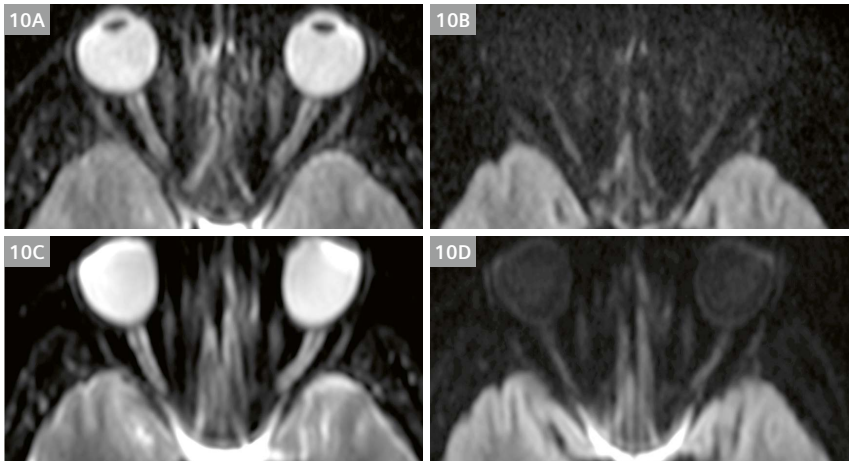
Simultaneous multi-slice technique

Simultaneous multi-slice (SMS) technique is not restricted to high field imaging, and is easily applied as well at low-field. As shown in many clinical applications, SMS can be used to reduce scan time as well as to increase the number of acquired slices within a given scan time [6]. Its primary application at low-field will likely be to improve SNR, while maintaining scan time. This technique is easily applicable to both single shot EPI (for diffusion-weighted scans) and turbo spin echo technique (for T1-, T2-, and proton density-weighted scans).

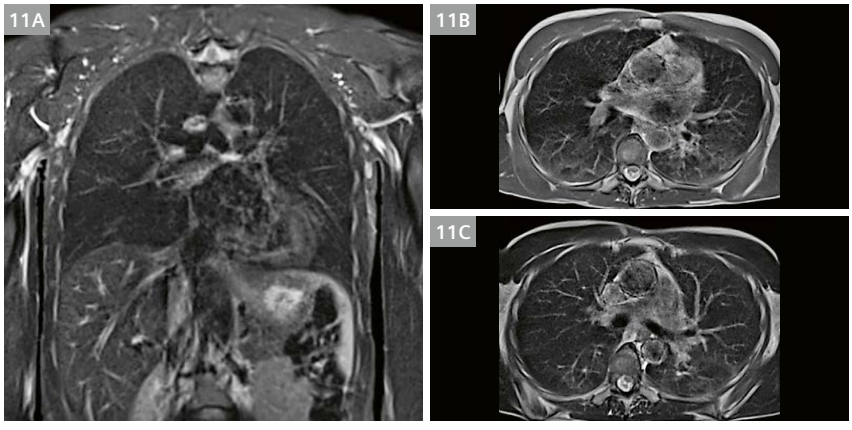
Iterative denoising

Iterative denoising is a relatively new technique that can be applied to improve image quality for low SNR scans [7]. The technique could thus be of particular value for low-field scans. Iterative denoising can be applied to almost all routine 2D and 3D MR acquisitions. It has the potential to increase SNR by 25%, or alternatively to reduce scan time by 30% (while maintaining SNR).

A short explanation of iterative denoising follows. Complex-valued image data is exported prior to interpolation and magnitude reconstruction, together with additional information regarding image normalization, *k*-space filtering, and noise calibration used. This data



10 **(10A, C)** b-value 0 and **(10B, D)** 1000 s/mm² single shot EPI DWI scans at (10A, B) 0.55T and (10C, D) 1.5T of the orbit. The increased magnetic susceptibility at 1.5T leads to marked distortion of the globes, poor depiction of the optic nerves, and prominent susceptibility artifact from the sphenoid sinus. Sequence specifics were similar for the two field strengths, with signal averages doubled for 0.55T.



11 Coronal **(11A)** T2 TIRM and axial **(11B)** 2D BLADE fast spin echo proton density weighted images and axial **(11C)** 2D T2-weighted BLADE images of the thorax.
All images were obtained with respiratory triggering in a healthy volunteer at 0.55T. Slice thickness was 6 mm.
Scan times were 6 minutes 8 seconds, 7 minutes 22 seconds and 5 minutes 44 seconds, respectively.

is then iteratively denoised by thresholding – spatially adapted to the local noise level – using orthogonal wavelet transforms. The data is re-imported to the reconstruction pipeline and magnitude images calculated. An important point is that the process is automatic, with the algorithm adapting to changes in acquisition and scan conditions.

Deep learning reconstruction
Another highly promising, yet very new approach is to utilize deep neural networks either in the direct transformation of raw data into images or to optimize the quality of otherwise non-diagnostic images. Consider for the moment the simple example of a fast, low resolution scan acquired at low field. If a neural network is trained with high-resolution images – from either the same field strength or higher, the network can establish "neural connections" to associate features in the lower quality image with those in the higher quality image. After training on a few thousand images, the network can then apply its "knowledge" to improve the resolution of the images. This approach is commonly termed superresolution processing. Beyond such image optimization strategies, deep learning might also be beneficial to limit the impact of artifact patterns, such as streaking in radial imaging.

Image degradation due to susceptibility
The differing susceptibility of tissues causes, at their interface, both geometric distortion and artifactual areas of high and low signal intensity in MR images. Susceptibility is substantially less at low-field, being proportional to magnetic field strength. Prominent susceptibility artifacts, interfering with clinical diagnosis, are well known in the orbits, internal auditory canal, skull base, lungs, bowel, and close to metal implants. Considering this issue by itself, image quality will thus be substantially improved in these areas at low-field (Fig. 10). Lung imaging in particular will benefit, with MR of course offering potential clinical value over CT due to its soft tissue contrast and in particular the ability for spatially resolved assessment of lung function (Fig. 11). Recent clinical images from a 0.55T prototype show great potential for the imaging of parenchymal lung disease [5]. A particular advantage for MR in this application would also be the elimination of the high radiation dose otherwise necessary over a patient's lifetime for the evaluation by CT of chronic diseases, in particular those that occur in the pediatric population, for example cystic fibrosis. The imaging of metal implants is another area expected to benefit greatly from low-field, due to less severe susceptibility artifacts.

Acoustic noise
The noise in MR, during scanning, comes from the gradient coils. If everything else is held constant, doubling the magnetic field increases the acoustic noise (which is measured on a logarithmic scale) by 6 dB(L) [8]. To put this in perspective, normal conversation is at 60 dB, a vacuum cleaner 75 dB, sounds above 85 dB harmful, and for a subway 90–95 dB. Many sound deadening designs for the gradients have been introduced over the history of MR, with all applicable regardless of field strength. In a comparison of a low-field and a 1.5T unit (performed in the early 2000s), acoustic noise ranged from 77 dB at low-field (with the lowest noise scan) to 98 dB at high field (with the highest noise sequence). With all else equal, and a scan that produces moderate noise, changing from a 1.5 to a 0.5T MR could reduce the noise of the gradients for example from that of a subway to that of a door bell.

Interventional MR
There are many special requirements for interventional MR. RF heating can be a concern, due to the use of biopsy needles and guidewires. Heating in MR generally scales with Larmor frequency, and thus the operating field strength. Low-field consequently offers major advantages over high field. This is particularly true for cardiac catheterization. A recent study at 0.55T demonstrated that heating, with a subset of currently available devices, did not represent a restriction, and specifically did not exceed 1°C during 2 minutes of continuous imaging [5]. For an interventional system, improved bore access (due to greater bore dimension) would also be a marked advantage.

The lower price of a low-field system (with the real cost including the system itself, installation, and service/cryogens) would make much more practical dedicated installations for interventional work. Patient monitoring should be simpler, due to the lower field strength, with fewer problems caused by the magnetic field (for example with monitoring equipment) and the 5 gauss line much closer to the unit. The lower cost and ease of installation could lead to dedicated systems not previously possible in many departments, like what evolved historically with CT as well.

A substantially lower main magnetic field also reduces susceptibility artifacts, specifically the artifacts from catheters and needles. TrueFISP, the pulse sequence of choice for interventional guidance, also performs better at lower fields. SAR limits are less of a constraint, and other image artifacts (such as bending) are also less. Overall, low-field offers a major advantage when compared to higher fields such as 1.5T for interventional work.

Summary

Next generation low-field MR units will greatly benefit by the knowledge gained in system development over the last 35 years. High image quality is dependent on magnet homogeneity, fast gradients with minimal eddy currents, multichannel receiver coils and advanced image reconstruction (including compressed sensing), all achievable on a low cost, low-field system today. Developing in addition an advanced design ultrawide-bore magnet would offer unrivaled patient comfort and ease of patient monitoring, sedation, and interventions. The reduced acoustic noise inherent to low-field offers a further improvement in patient comfort, as well as that for associated personnel.

Low-field MRI is inherently more cost-effective due to reduced magnet, gradient, RF transmitter, and siting costs. Installation and infrastructure (weight, size) requirements are substantially reduced. The need for helium refills, and even the quench pipe, could be eliminated with an advanced magnet design, further reducing costs. These all have important implications for technology dissemination – both in developed economies and in underdeveloped areas [4], and access to care.

Not to be neglected are the specific imaging advantages that come with low-field. Lower susceptibility leads to improved sequence performance, as well as improved image quality in many anatomic areas. Lower SAR adds scan sequence flexibility and diminishes the difficulties with metal implants and interventional techniques. Advanced readout strategies with increased SNR, such as spiral imaging, are possible. SNR-efficient long readout strategies can be employed, due to reduced T2*, providing the benefit as well of reduced image distortion and blurring.

Newly designed, advanced generation low-field MR imaging systems will radically increase access to disease diagnosis and surveillance both in developed countries and worldwide. MR systems operating in the range of 0.5T were briefly evaluated in the mid-1980s, in the early days of MR. Considering the subsequent hardware and software developments over the interval 35 years, those units were quite primitive and did not reflect the image quality that can be achieved today. The potential impact of new, low cost, advanced generation MR imaging systems is extremely high. These will lead to further dissemination of health care – both in the G20 nations and in developing countries.

The low system cost, low installation cost, ease of maintenance, and ability to operate even with electrical power issues, combined with high image quality, all predict a bright future for this development.

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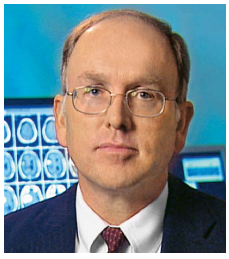

Contact

Val M. Runge, M.D.
Editor-in-Chief, Investigative Radiology
Department of Diagnostic,
Interventional and Pediatric Radiology
University Hospital of Bern, Inselspital
Bern
Switzerland
ValMurray.Runge@insel.ch