Abstract: Today’s health care environment is shifting rapidly, driven by demographic change and high economic pressures on the system. Furthermore, modern precision medicine requires highly accurate and specific disease diagnostics in a short amount of time. Future imaging technology must adapt to these challenges.

Demographic change necessitates scanner technologies tailored to the needs of an aging and increasingly multimorbid patient population. Accordingly, examination times have to be short enough that diagnostic images can be generated even for patients who can only lie in the scanner for a short time because of pain or with low breath-hold capacity.

For economic reasons, the rate of nondiagnostic scans due to artifacts should be reduced as far as possible. As imaging plays an increasingly pivotal role in clinical-therapeutic decision making, magnetic resonance (MR) imaging facilities are confronted with an ever-growing number of patients, emphasizing the need for faster acquisitions while maintaining image quality.

Lastly, modern precision medicine requires high and standardized image quality as well as quantifiable data in order to develop image-based biomarkers on which subsequent treatment management can rely.

In recent decades, a variety of approaches have addressed the challenges of high throughput, demographic change, and precision medicine in MR imaging. These include field strength, gradient, coil and sequence development, as well as an increasing consideration of artificial intelligence. This article reviews state-of-the-art MR technology and discusses future implementation from the perspective of what we know today.

Key Words: magnetic resonance imaging, alternative field strengths, advanced gradient, capabilities, acceleration techniques, quantitative MRI

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Future magnetic resonance imaging (MRI) must adapt to demographic change and the high economic pressure on the health care system. This entails, in particular, a strong desire for speed that has driven a race of gradients becoming faster and more powerful. Technological and methodological advances have led to the development of ultra-high-field (UHF) MRI systems and high-performance gradient systems that significantly reduce acquisition time and improve resolution. In light of the economic pressure on the worldwide health care system, however, it seems worth rethinking alternative field strengths and new possibilities of making MRI accessible for a broader patient population. In the first part of this review, we want to discuss advances in hardware that address the challenges of high throughput and demographic change, including field strength, gradient, and coil development.

In the second part of this review, we will then further elaborate on acceleration techniques. Over the last 20 years, advancements in sequence technology, such as parallel imaging (PI), acceleration, and parallel acquisition techniques with either an image- or k-space based algorithm, volume-interpolated 3-dimensional (3D) gradient echo sequences, dynamic magnetic resonance (MR) angiography with k-space sharing, and radial imaging, have significantly accelerated MRI. These build the foundation for 2 fundamental advances in the present decade with major current and future impact on MRI on which we will focus in this article, namely, sparse imaging, that is, compressed sensing (CS) and simultaneous multislice (SMS).

Compressed sensing enables the acquisition of contrast-enhanced sequences (in the body) without requiring a breath-hold. In an aging and increasingly multimorbid patient population, breathing is a major cause for motion artifacts. Several publications have evaluated the benefits of CS for various clinical applications such as contrast-enhanced abdominal as well as cardiac imaging. Because demographic change is leading to an expanding number of orthopedic implants, such as total hip arthroplasty (THA), sequences for metal artifact reduction are increasingly important to prevent nondiagnostic scans. Compressed sensing acceleration has also been successfully applied to improve metal artifact reduction.

Simultaneous multislice imaging represents one of the most important acceleration techniques today that most clinical 2D acquisitions can benefit from. Studies have proven its benefit in particular for musculoskeletal (MSK) as well as diffusion-weighted imaging (DWI), which is essential for the characterization of oncological diseases.

In the last section of this article, we will address the impact of quantitative MRI. Modern precision medicine depends on quantifiable data as well as high and standardized image quality to develop image-based biomarkers. The emerging field of radiomics aims at analyzing quantitative features derived from medical images to characterize tissue, eg, tumor phenotypes, and support clinical decision making. The diagnostic potential of texture analysis applied to quantitative MR sequences, such as T1 and T2 mapping, is promising. However, the robustness and reproducibility of radiomic features vary in different MR sequences, in particular nonquantitative ones, emphasizing the need for extensive reproducibility and validation studies as well as standardization.

Among quantitative MRI techniques, DWI is a particularly promising noninvasive technique for a more sensitive and specific probe of local tissue architecture and structure. Over the past decade, several advanced DWI methods have been developed to enable quantitative, detailed investigation of intrinsic features of tissue microstructure in vivo. Clinically, DWI is therefore of crucial importance, especially in oncologic imaging.

The recent invention of MR fingerprinting (MRF) enables the estimation of tissue-specific parameters such as T1 and T2 relaxation times. Magnetic resonance fingerprinting is based on the use of transient signal evolutions to estimate several quantitative tissue property parameters simultaneously. This makes estimation of tissue-specific parameters such as T1 and T2 relaxation times feasible. Following the outline above, this review attempts to give a brief overview of state-of-the-art MR technologies and discuss future implementations from the perspective of what we know today.
ADVANCED HARDWARE FOR THE FUTURE OF MRI

Field Strength

Over the past quarter of a century, technological and methodological advances have led to the development of UHF MRI systems by overcoming the inherent physical challenges of UHF radiofrequency (RF) signal homogenization in the human body, including nonuniform RF fields, enhanced susceptibility artifacts, and higher RF energy deposition in the tissue. Today, with over 80 MRI systems installed worldwide operating at a magnetic field strength of 7 tesla (T) or higher, UHF MRI is already widely applied both scientifically as well as clinically. The signal-to-noise ratio (SNR) is substantially increased in UHF MRI. Techniques that benefit especially from high magnetic fields are susceptibility-weighted imaging and phase-contrast techniques, imaging with X-nuclei, MR spectroscopy, chemical exchange saturation transfer imaging, as well as functional MRI.

Compared with UHF MRI systems, mid- to low-field MRI systems operating at a magnetic field strength of 1 T or less are very attractive as they would be expected to have reduced production and operational costs while maintaining high image quality due to modern system engineering. Furthermore, the reduced specific absorption rate as well as local susceptibility/improved homogeneity in the static B0 and transmit B1 fields, respectively, show great potential for clinical applications and could be used, for example, for imaging of metallic devices to reduce artifacts.

Alternative Field Strengths—New Opportunities in Making MR Accessible for a Broader Patient Population

The strong desire for higher speed and new clinical applications in MRI has driven a race of gradients becoming faster and more powerful. The emerging technology of superconducting magnets in the 1980s enabled a continuous increase in field strength. Today, 1.5 T and 3 T are seen as clinical standards, the first clinical 7 T was cleared for clinical use in 2017, while imaging at low field (<0.4 T) and mid-field (0.4-1 T) has almost disappeared. The increasingly complex specifications and infrastructure requirements of today's MR systems have inherently made MR one of the most expensive modalities in radiology and have restricted patient access.

In light of the economic pressure on the worldwide health care systems, it seems worth revisiting the question: Is it possible to reduce the cost of the most expensive function in an MR system, namely, the creation of the B0 field, while maintaining the diagnostic value? Would this allow MRI to be accessible to a larger patient population? Is there a missed opportunity on the path that the MRI market and research community has taken over the last 2 decades?

Presently, low- and mid-field systems are burdened with several major disadvantages. As shown by Marques et al., most of the respective magnets are built on permanent magnet systems with large iron yokes. The enormous weight of these systems of typically 10 to 20 t is not only a hurdle for the installation environment. Combined with the inability to switch the magnetic field on and off, the complete logistics from manufacturing to installation and servicing is highly complicated and costly. The cost of rare earth materials needed to create these magnets often sees volatile fluctuations, thus rendering the technology uneconomic when compared with superconductors. The stray field originating from the gradient coils penetrates the iron yoke, with the hysteresis of the iron generating undesired dynamic field behavior. This, together with high eddy current fields, results in poor image quality for various applications (eg, DWI or even general robustness when it comes to spectral fat saturation).

Many of the achievements of research in MRI, especially in the field of software and digitization innovation, have not been widely applied to the way we develop the architecture of MRI systems, particularly in regard to field strength. Furthermore, it has not been extensively investigated whether the combination of low-field imaging with the latest technology can offer new clinical applications, eg, in the field of intervention and lung or cardiac imaging. Parallel imaging, iterative reconstruction, and spiral acquisition techniques, as well as many other new technologies, were not invented yet, when low- and mid-field systems were overtaken by superconducting 1.5 T MRI. Superconducting horizontal-field magnets with magnet coils enveloping the human body have proven to be the most cost-efficient way to generate the static B0 field while also overcoming the image quality deficiencies of iron-yoke magnets. Such low- and mid-field magnets, combined with PI techniques to speed up the measurement times, and novel sampling or postprocessing technologies such as image denoising have the potential to generate a new class of MRI systems.

Campbell-Washburn et al. have examined the potential of such a combination using a MAGNETOM Aera (Siemens Healthcare, Erlangen, Germany) ramped down to 0.55 T, RF hardware adapted to 23.6 MHz, and a gradient system with 200 T/m/s slew rate and up to 45 mT/m gradient field. Their study with 83 participants reports good image quality even for demanding applications such as DTI and cardiac MRI. Furthermore, the study demonstrates that systems with lower field strength using state-of-the-art technology can even open the door to new clinical applications, where MRI used to be a niche—partly because of disadvantages originating from high fields. Imaging results in areas with high susceptibility contrast such as the lungs are improved using lower B0. Cardiac applications can benefit from increased T2 and reduced T1 time constants. The lower RF frequency shows significantly reduced heating in interventional devices, potentially solving one of the road blockers for safety in MRI-based catheter tracking. Advanced read-out schemes such as spiral k-space trajectories can help by increasing the SNR for a given measurement time. Intrinsic problems of lower field strengths such as demanding situations for spectral fat-sat could be addressed in the future with advanced pulse design methods, where fat-sat pulses are automatically adapted to the individual examination.

The comparatively small amount of superconducting wire needed even for mid-field systems allows a cost-efficient design of ultra-wide-bore systems with patient bores in the 80 cm domain with increased patient comfort and improved access for interventions. In addition to the new options for affordable and accessible imaging, new paradigms in patient comfort and diagnostic value are possible. In addition, installation and infrastructure (weight, size) requirements can be substantially reduced. This could help enable MR to be put into new locations for patient care that will lead to better value-based care.

Advanced Gradient Capabilities

The improvement in resolution and speed in recent MRI scanner generations was only possible with the development of high-performance gradient systems, which are used for the spatial encoding of the received signal and provide substantial benefits for the diffusion encoding in DWI. These gradient systems consist of a gradient coil, which translates current and voltage into rapidly changing magnetic fields, and an amplifier, which can provide high currents and voltages. The maximum gradient strength in mT/m and the rise time, normally specified as “slew rate” in T/m s, are the main characteristics of a gradient system and determine its performance.

Present gradient technology allows gradient amplitudes of up to 80 mT/m and slew rates of up to 200 T/m/s simultaneously for clinically used whole-body systems. However, the achievable gradient strength depends not only on the used amplifier but also on the geometric boundary conditions, like patient bore. Therefore, even higher amplitudes and slew rates (up to 200 mT/m and up to 1000 T/m/s) are possible using dedicated coils like head gradient coils. A major advantage of these head gradient coils is that the high amplitudes can...
mostly be switched in short rise times without restrictions owing to peripheral nerve stimulation (PNS). Nevertheless, the clinical relevance of these systems is restricted.

With special diffusion gradient systems for the research segment, e.g., the Connectom gradients, even higher gradient amplitudes can be achieved for whole-body systems. This system for example consists of 2 gradient coils stacked in 1 structure and 4 gradient amplifiers, a control system that allows the parallel activity of these amplifiers, and a cooling system to handle the high power. Inside the gradient coil, an optimized cooling system deals with the high energy deposition, and the vacuum potting ensures the dielectric strength without any partial discharge. In addition, the vacuum potting should lead to the necessary mechanical stiffness of the coil to withstand the high Lorentz forces. The reduced patient-bore and an adapted linearity volume allow the usage of high amplitudes for the diffusion lobes without PNS.

The development of scanners with high gradient performance is only possible with a detailed understanding of all interacting components of the MR system, including specifically the gradient coil, magnet, RF body coil, and gradient power amplifier. One example is the interaction of the rapidly changing magnetic fields emanating from the gradient coil with the conducting layers inside the magnet. Simulations and tests lead to a system optimized with respect to image artifacts owing to eddy currents. Another aspect to be considered is the energy deposition inside the magnet due to the stray field of the gradient coil. High maximum gradient amplitudes are of particular interest for DWI. Given a certain \( b \)-value, stronger gradients allow for a shorter echo time (TE) and a higher SNR. Diffusion-weighted imaging is sensitive to pore/cell sizes, which makes it a promising tool for the examination of microstructure, e.g., in the brain—for further elaboration, we refer to the “Diffusion-Weighted MRI” section.

FIGURE 1. Diagrammatic representation of data acquisition and reconstruction using the concept of data sparsity, with undersampling of k-space and subsequent iterative thresholding and denoising, leading eventually to the final reconstructed image. From Runge et al.1

FIGURE 2. Comparison of respiratory motion artifacts between single arterial phase (AP) versus C-SENSE double AP phase technique in an 83-year-old woman with hepatocellular carcinoma and underlying liver cirrhosis. The scan time for each AP image acquisition is indicated in seconds (s). A, Single-phase AP image taken 3 months before the current examination shows substantial respiratory motion artifacts with mild to moderate diagnostic impairment, by which focal enhancing lesions are blurred. B, The first phase of double AP imaging (AP1) shows no respiratory motion artifact with clearly visible multiple enhancing hepatic lesions (arrow). Mild ringing artifacts are noted at the right hemiliver (arrowhead). C, The second phase of double AP imaging (AP2) also shows no respiratory motion artifact with good focal lesion detection. From Yoon et al.3
Functional MRI is another field where high-performance gradients are essential. Here, however, the slew rate rather than the maximum amplitude is decisive. Signal acquisition in blood oxygen level dependent functional MRI is based on an echo-planar imaging (EPI) readout. The achievable resolution is limited by echo-spacing, which is in turn limited by the gradient slew rate. With slew rates beyond the capabilities of clinical scanners (>200 T/m/s), it could be possible to resolve the functionality of cortical substructures, such as the underlying columns and layers.

Peripheral Nerve Stimulation

Image acquisition strategies such as echo-planar and spiral imaging techniques or SMS excitation techniques such as PINS require rapid switching of the gradient coils. With recent advances in gradient engineering, PNS has become a major limitation on fully using the available gradient coil performance for fast imaging. Despite its impact on operational limits, PNS metrics are only indirectly considered in the coil design phase, for example, by conducting stimulation experiments on healthy human subjects using coil prototypes or by decreasing the linear volume. Davids et al developed a framework based on a coupled electromagnetic-neurodynamic simulation to completely model and accurately predict gradient coil PNS thresholds and stimulation sites in the body, providing detailed information on the coil’s PNS capability without the construction of expensive coil prototypes. This PNS model can be incorporated into the standard approach for numeric gradient winding optimization as an additional constraint to enforce...
PNS-optimized gradient winding patterns. In addition, such a PNS tool can be valuable in assessing other PNS mitigation strategies. With their results, the authors hope to open up the gradient's operational parameter space that is currently inaccessible due to PNS to fully use the available performance of state-of-the-art MRI gradient coils.

60-Channel Receiver Coil

Multielement coils are a prerequisite for k-space undersampling techniques such as PI. Parallel imaging leads to a substantial loss in SNR by the square root of the acceleration factor “R,” which describes the number of undersampled k-space lines. In addition to the acceleration factor, the geometry factor (g-factor) also leads to reduced SNR. The g-factor depends on the geometry, number, and density of the receiver coils. Receiver coils with more than 32 channels are already used in brain imaging, allowing acceleration by a factor of up to 4. In abdominal imaging, R so far is limited to factors of usually 2 to 3. Higher acceleration factors in abdominal MRI require receiver coils with higher channel density. Weiss et al. evaluated a clinically applicable multichannel body coil with 60 receiver channels (30 dorsal and 30 ventral) for contrast-enhanced abdominal MRI. Their results showed significantly higher image quality at intermediate acceleration levels (R > 2) but showed no benefit at low acceleration levels (R < 2). The authors used a controlled aliasing in PI results in higher acceleration technique with various acceleration factors. The SNR gain was mainly found in the more central parts of the body. They concluded that the 60-channel coil setup facilitates improved SNR and image quality at high PI technique factors with diagnostic image quality, enabling an acceleration of contrast-enhanced MR imaging of the abdomen.

THE NEED FOR SPEED—ACCELERATION TECHNIQUES 4.0

Compressed Sensing

CS and Motion Artifacts

Motion artifacts are a major reason for reduced image quality. The causes are manifold: patient movement, eg, in children, in patients with dementia, or in severe pain, as well as respiratory motion due to a limited breath-hold capability. This can lead to greatly reduced image quality and to nondiagnostic scans. Various efforts have been made in recent years to address the problem of motion artifacts. A key solution is to reduce acquisition speed. Almost 20 years ago, acceleration of data acquisition became feasible with the invention of PI and its evolution, or view sharing. This laid ground for dynamic contrast-enhanced (DCE) imaging such as peripheral MR angiography as well as DCE-MRI of the liver and kidneys. However, imaging acceleration so far has been limited by a significant loss of SNR. The latest techniques that combine PI and CS allow for faster scans without reducing image quality. Compressed sensing semirandomly (incoherently) acquires just a small percentage of k-space. An image will be reconstructed from this undersampled k-space data using an algorithm with the following specific characteristics: the data must be random, sparse, and incoherently subsampled, meaning that the final image has a sparse representation in its transform domain. The randomly acquired subsampling artifacts are incoherent. A nonlinear iterative reconstruction is used to reduce the aliasing artifacts related to subsampling (Fig. 1).
Because of this acquisition technique, CS allows for high temporal resolution and a shortened scan time. The contrast agent bolus can therefore be fully resolved after intravenous injection. This is an important prerequisite for example for the characterization of liver lesions. Several publications have already proven the benefits of CS for contrast-enhanced dynamic liver imaging.

Yoon et al.\(^3\) were able to show that CS can be used for single-breath-hold, double-arterial-phase scans with optimal temporal resolution and reduced respiratory motion artifacts (Fig. 2).

Weiss et al.\(^50\) evaluated a continuous hepatic arterial multiphase protocol during free-breathing. They concluded that the use of CS makes robust multiphase arterial imaging during free-breathing feasible at a high spatial and temporal resolution, resulting in an improved lesion conspicuity vs. a conventional breath-hold approach.\(^50\)

Taron et al.\(^46\) demonstrated that CS can also be used for scan time reduction in MR cholangiopancreatography (MRCP). Standard MRCP is acquired either with breath-hold or with navigator-based approaches for the compensation of motion artifacts. However, patients often present with limited breath-hold capacity or irregular breathing cycles that interfere with the navigator. This makes MRCP acquisition during a single breath-hold attractive. Taron et al.\(^46\) were able to show that CS-accelerated MRCP is feasible in clinical routine at 1.5 T and 3 T—while recommending 3 T—and offers a significant reduction of acquisition time as well as an increase in spatial resolution.

Zhu et al.\(^51\) compared CS breath-hold, CS navigator-triggered (NT), and a conventional NT protocol. Acquisition times for the specific protocols were 17 seconds for CS breath-hold, 134.1 seconds for CS NT, and 364.7 seconds for the conventional NT protocol \((P < 0.01)\). The authors concluded from their data that CS NT MRCP had the highest diagnostic performance for detecting ductal anomalies, long-segment duct stenosis, abnormal branch ducts, and communication between cystic lesion and pancreatic duct (mean area under the curve value, 0.943-0.983).\(^51\)

Compressed sensing has also been proven to optimize cardiac imaging. Sudarski et al. evaluated the accuracy of left ventricular analysis with a 2D real-time cine true fast imaging with steady-state precession (true FISP) sequence featuring sparse data sampling with iterative reconstruction at 3.0 T. They were able to demonstrate its feasibility irrespective of breath-hold commands and show that the assessment of left ventricular function with sparse data sampling with iterative reconstruction is noninferior to measurements on the basis of conventional fully sampled input data.\(^4\)

Moreover, CS has been shown to substantially reduce acquisition times in neuroimaging without compromising diagnostic quality. Greve et al.\(^52\) demonstrated that a combination of spiral imaging with CS accelerates time-of-flight MR angiography and may improve the detection and diagnosis of cerebrovascular disease. Eichinger et al.\(^53\) applied CS to accelerate double inversion recovery sequences in multiple sclerosis (MS) and showed it to be significantly less prone to imaging artifacts.

**FIGURE 3. (Continued).**

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**CS for Metal Artifact Reduction in MSK Imaging**

The number of patients with implants is steadily increasing. The expected demographic growth will exacerbate this problem. Local magnetic field inhomogeneities affect the assessability of anatomical areas adjacent to the implant. Several approaches have been developed to address this issue: conventional high-bandwidth metal artifact reduction
MR sequences; using high bandwidth, small voxel size, short echo times, and others; optimized metal artifact reduction MR sequence techniques, including view-angle-tilting, multiacquisition, variable-resonance image combination, and slice-encoding metal artifact correction (SEMAC). Slice-encoding metal artifact correction corrects through-plane distortions using pseudo-3D acquisitions. Despite that the benefits of SEMAC artifact correction in terms of image quality have been proven by several studies, a major limitation of this approach remains the prolonged scan time of up to 15 minutes. To reduce scan time, CS techniques have been combined with SEMAC (CS-SEMAC). In combination with CS, scan time could be reduced to about 5 minutes. This is highly significant for further successful clinical implementation. Jungmann et al evaluated the optimal imaging protocol for metal artifact reduced MRI of THA using CS-SEMAC at 1.5 T. Their findings demonstrated that overall best soft-tissue contrast was achieved using STIR and T1-weighted sequences with a defined set of parameters (19 SESs, 10 iterations; normalization factor of 0.001). The authors concluded from their data that scan protocol parameter optimization is essential for optimized image quality in addition to the development of new sequences for metal artifact reduction (Fig. 3).

**Simultaneous Multislice**

Simultaneous multislice imaging represents one of the most important acceleration techniques today that most clinical 2D acquisitions can benefit from. Simultaneous multislice is an acquisition technique based on the parallel acquisition of several slices. As a result, fewer slice excitations are required for the same field of view coverage. Typically, the repetition time is reduced by the acceleration factor. The acceleration factor equals the number of slices acquired in parallel. In a similar way to PI, SMS uses the spatial sensitivity information of multi-channel array coils to separate the individual slices (Fig. 4).

**FIGURE 4.** Diagrammatic representation of slice acquisition and reconstruction using simultaneous multislice, specifically with excitation of multiple slices by using blipped controlled aliasing in parallel imaging results in higher acceleration followed by slice and then in-plane unaliasing both using generalized autocalibrating partially parallel acquisition (GRAPPA). From Runge et al.

**FIGURE 5.** Boxplot of the apparent diffusion coefficient values of malignant and benign lesions for simultaneous multislice–accelerated (“SMS”) and the reference sequence ("noSMS"). Box plots provide information on median (horizontal line in each box), quartiles with interquartile range, and minimum/maximum values. From Ohlemeyer et al.
Studies have proven its benefit in particular for DWI, which is essential for the characterization of oncological diseases. Ohlmeyer et al, for example, successfully used SMS in DWI based on EPI for the diagnosis of benign and malignant breast lesions. While EPI is a fast and widely used readout technique for DWI, the scan time rarely falls below 3 minutes. The implementation of PI in EPI is limited in terms of acquisition time (TA) could be more than halved by adjusting the field of view, from 4 hours 39 minutes to 1 hour 57 minutes with the use of SMS.

**FIGURE 6.** Simultaneous multislice (SMS) can be used to acquire proton density-weighted scans of the same resolution in a shorter time. In this case, the acquisition time (TA) could be more than halved by adjusting the field of view, from 4 hours 39 minutes to 1 hour 57 minutes with the use of SMS.

**FIGURE 7.** This figure shows representative volume fraction estimates obtained with linear multiscale modeling analysis for voxels within different white matter (WM) and gray matter (GM) structures. The WM voxel shows a high signal especially originating from the restricted diffusion water compartments (A), whereas the cortical GM voxel shows a high signal originating from the hindered diffusion water length scales (B). The voxel within the thalamus contains a relatively high signal originating from both the restricted and hindered diffusion water compartments, compared with WM and cortical GM (C).
of scan time reduction relative to other applications because a large proportion of scan time is needed for diffusion-encoding gradients. The aim of their study was to evaluate the potential of SMS for the acceleration of DWI in breast imaging. They found that SMS acceleration can be used for DWI in breast MRI because SMS achieved the same diagnostic accuracy in breast MRI compared with standard EPI but with substantially reduced scan times (Fig. 5). With an acceleration factor of 2, TE/repetition time set to the minimal values possible, and all other imaging parameters kept as constant as possible, they were able to reduce the scan time from 3 hours 27 minutes to 1 hour 53 minutes. Simultaneous multislice has also been used in EPI DWI for organ regions other than breast, such as brain, liver, and kidneys.

Simultaneous multislice can also be applied to fast (turbo) spin echo imaging, in particular T1- and proton density-weighted scans in MSK imaging. It can be used to either decrease scan time or to provide higher-resolution images within the same scan time (Fig. 6).

**IMPACT OF QUANTITATIVE MRI**

**T1 Mapping**

T1 mapping enables quantitative tissue characterization through an evaluation of the longitudinal relaxation time (T1) for each voxel in the image. Through a voxel-wise determination of T1 relaxation time, it provides a quantitative noninvasive method for tissue characterization. The signal intensity of each pixel in the color map reflects the absolute T1 value of the underlying pixel. T1 mapping has already proven its clear diagnostic value in cardiac imaging, especially for the characterization of myocardial tissue. In addition to cardiac imaging,
T1 mapping has also been successfully used in oncological imaging. Here, personalized medicine relies on the most specific characterization of diseases possible. Quantitative imaging is of crucial importance for the optimal characterization of tumor heterogeneity. As with DCE imaging and DWI, T1 mapping also allows the derivation of quantitative values from the image data. Adams et al successfully used a native T1-mapping protocol for the identification of higher-grade clear cell renal cell carcinoma. They showed that native T1 mapping represents a potential imaging-based biomarker to differentiate lower- and high-grade clear cell renal cell carcinomas.6

Magnetic resonance mapping techniques are also promising in terms of applying machine learning techniques. Baessler et al demonstrated greater stability in features derived from quantitative maps compared with standard sequences. Magnetic resonance mapping techniques therefore seem to be significant for the further development of imaging-based biomarkers using artificial intelligence.

**Diffusion-Weighted MRI and Microstructure Imaging**

Among quantitative MRI techniques, DWI is a particularly promising noninvasive technique for a more sensitive and specific probe of local tissue architecture and structure. Its unique sensitivity to the molecular self-diffusion of water at both a cellular and subcellular level makes it possible to quantify and map the histologic features of, eg, tumors in vivo at microscopic length scales.80,81 Over the past decade, several advanced DWI methods have been developed to enable quantitative, detailed investigation of intrinsic features of tissue microstructure in two10–11.

The development of scanners with high maximum gradient amplitudes, for example, made it possible to examine axon–diameter distributions,82,83 where axonal loss can be attributed to, eg, MS, especially in the vicinity of MS plaques, as well as in normal-appearing white matter.83 Achieving a high-enough resolution for the depiction of smaller axons is the main challenge for axon-diameter distribution. The achievable resolution of oscillating gradient spine echo84 and pulsed gradient spin echo85 sequences is strongly dependent on the achievable gradient amplitude and benefits from systems beyond current clinical limits. High-resolution DWI also enables more accurate tractography, allowing the disentanglement of smaller white matter connections, such as intralobal connections. State-of-the-art gradients enable shorter TE per b-value. Especially in combination with efficient non-Cartesian acquisition techniques, the accuracy of tractography can be maximized to reveal connections that would otherwise be missed.84

The development of stronger gradients also helps to extract microstructural features beyond fractional anisotropy and mean diffusivity. The accessibility of larger ranges of b-values and the increase in SNR make the underlying inversion problem more tractable and enable insights beyond single diffusion encoding.

Linear multiscale modeling is a recently developed analysis framework for diffusion MRI data that estimates tissue microstructure parameters, including volume fractions of restricted and hindered water for a range of length scales and orientation distribution information.86,87 Linear multiscale modeling is based on the restriction spectrum imaging (RSI) approach developed by Nathan White, which has been increasingly applied clinically.10,88,89 The RSI approach reconstructs the diffusion tissue orientation distribution over a spectrum of length scales by assuming a spectrum of Gaussian diffusion response functions. Linear multiscale modeling extends the RSI approach to represent restricted water compartments with non-Gaussian diffusion response functions. Thereby, linear multiscale modeling provides detailed characterization of tissue microstructure and orientation distribution with promise for identifying distinct diffusion microstructural signatures of pathology compared with healthy tissue (Fig. 7).

**Segmented Diffusion-Weighted Prostate Imaging**

Geometric distortions, low spatial resolution, and poor SNR can limit the quality of examinations using standard single-shot EPI for DWI.90 Akshit Ciris et al have developed an accelerated segmented diffusion imaging acquisition sequence that allows for an increase in geometric fidelity as well as spatial resolution. Using a reduced field of view in a configuration with a single endorectal coil, they showed that the integration of acceleration based on the reconstruction of sparsely sampled data allows segmentation to be acquired with a reasonable scan time and no substantial loss in SNR90 (Fig. 8).

**Magnetic Resonance Fingerprinting**

Magnetic resonance fingerprinting uses transient signal evolution to estimate several quantitative tissue property parameters simultaneously. Therefore, the estimation of tissue-specific parameters such as T1 and T2 relaxation times become feasible.12 Varying the acquisition parameters creates a spatial and temporal incoherence that leads to efficient signal encoding. As a result, quantitative parameters can be derived by matching the signal acquired to simulated signals, the MRF dictionary. Yu et al13 used MRF technology in a comprehensive prostate cancer imaging protocol. The authors concluded from their data that MRF (T1 and T2), together with standard apparent diffusion coefficient, allows the identification of prostate carcinoma foci within the peripheral gland.91 Moreover, high- and intermediate-grade cancers can be differentiated.91 In the future, MRF acquisition may play a key role, especially for the further implementation of radiomics into clinical routine because current studies have shown the limitations of nonquantitative MR data for radiomic analyses. Quantitative data maps derived from MRF help overcome these current constraints.

**CONCLUSION**

Over the last decades, numerous methods have been successfully developed and implemented to robustly address the challenges of high throughput, demographic change, and precision medicine in MRI. A few of these approaches were reviewed in this article. In the future, increasing implementation of artificial intelligence technologies and integrative analyses will most likely greatly impact patient care and diagnostics.

**REFERENCES**


