Superconducting magnet designs and MRI accessibility: A review

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Abstract
Presently, magnetic resonance imaging (MRI) magnets must deliver excellent magnetic field ($B_0$) uniformity to achieve optimum image quality. Long magnets can satisfy the homogeneity requirements but require considerable superconducting material. These designs result in large, heavy, and costly systems that aggravate as field strength increases. Furthermore, the tight temperature tolerance of niobium titanium magnets adds instability to the system and requires operation at liquid helium temperature. These issues are crucial factors in the disparity of MR density and field strength use across the globe. Low-income settings show reduced access to MRI, especially to high field strengths. This article summarizes the proposed modifications to MRI superconducting magnet design and their impact on accessibility, including compact, reduced liquid helium, and specialty systems. Reducing the amount of superconductor inevitably entails shrinking the magnet size, resulting in higher field inhomogeneity. This work also reviews the state-of-the-art imaging and reconstruction methods to overcome this issue. Finally, we summarize the current and future challenges and opportunities in the design of accessible MRI.

KEYWORDS
accessible MRI, $B_0$ homogeneity, imaging in inhomogeneous fields, superconducting magnet design

1 | INTRODUCTION

Magnetic resonance imaging (MRI) is a noninvasive imaging modality that offers good soft-tissue contrast, leveraged for clinical benefits. The goals of improving its spatial resolution and signal sensitivity have driven it towards higher main magnetic field ($B_0$) strengths since its early days in the mid-1970s. The first whole-body superconducting scanners were operated at a field strength of 0.1 T and 0.15 T. By contrast, conventional clinical field strengths today are 1.5 and 3 T. In addition to these widely used magnetic field strengths, scanners at 7 T, 10.5 T, and 11.7 T are available, while even higher strength scanners are being developed. The quality of MR images depends on multiple factors and acquisition parameters, but most notably on the spatial uniformity of the main magnetic field. However, challenges related to field inhomogeneity are more significant as $B_0$ increases. Accurate spatial encoding mandates tight requirements on field homogeneity to prevent

Abbreviations: AI, artificial intelligence; BSCCO, bismuth strontium calcium copper oxide; C3T, compact 3 T; DL, deep learning; DSV, diameter of spherical volume; DWI, diffusion-weighted imaging; EPI, echo-planar imaging; FEM, finite element method; fMRI, functional magnetic resonance imaging; FOV, field of view; GA, genetic algorithm; GRE, gradient-recalled echo; HTS, high-temperature superconductor; LP, linear programming; MAVRIC, multiacquisition variable-resonance image combination; MAVRIC-SL, multiacquisition variable-resonance image combination selective; MgB2, magnesium diboride; MP-SSFP, missing pulse steady-state free precession; MQE, minimum quench energy; MRI, magnetic resonance imaging; MSiL, multispectral imaging; NbTi, niobium titanium; PNS, peripheral nerve stimulation; p–p, peak-to-peak; ppm, parts per million; REBCO, rare earth barium copper oxide; REMODEL, RE-construction of MR images acquired in highly inhomogeneous fields using DEep Learning; RF, radio frequency; SAR, specific absorption rate; SEMAC, slice encoding for metal artifact correction; SNR, signal-to-noise ratio; SWI, susceptibility-weighted imaging; TE, echo time; Vrms, volume root mean square; WHO, World Health Organization; YBCO, yttrium barium copper oxide; ZBO, zero boil-off.
artifacts in the images. It is reported that the field uniformity of a whole-body magnet after installation and placement of passive and superconducting shims must be of the order of 10 parts per million (ppm) peak-to-peak (p–p) over a 45–50 cm diameter of spherical volume (DSV). Some imaging techniques even require $B_0$ homogenization active shimming to further reduce this value below 2 ppm p–p over the field of view (FOV) during image acquisition.  

Minimizing field deviations and conductor wire volume in the magnet coils are the primary objectives of magnet design algorithms. Coil volume is a measure of the length of superconducting wire required to build the magnet, given by:

$$V = \frac{2\pi}{J} \sum_{k=1}^{N} \frac{r_k |i_k|}{J},$$  

where $r_k$ and $i_k$ are the radius and current in the $k$th coil, respectively, and $J$ is the current density in all $N$ coils. The current density depends on the superconducting wire critical current, which is determined by the type of superconducting material, its wire composition, the copper/superconductor ratio, conductor diameter, number of filaments, and filament diameter. The optimization process is iterative and integrates the solution of a linear programming (LP) or genetic algorithm (GA) with a finite element method (FEM) simulation software. The algorithm returns the optimum values for parameters such as the magnet’s number of coils, the number of turns per coil, their spatial coordinates, and dimensions. Subsequently, the simulation software inputs these parameters to guarantee that the resultant magnetic field distribution fulfills the homogeneity requirements. Theoretically, only an infinitely long solenoid can achieve a perfectly homogeneous field. Because of the unfeasibility of such configuration, algorithms set constraints on DSV size, final magnet dimensions, weight, and cost depending on the target application. However, the tradeoff between field uniformity and magnet length often results in exceedingly large and heavy MRI scanners. The amount of superconductor required to manufacture such magnets impacts the final product’s cost and hinders the systems’ transportation and siting options. In addition, other MRI components such as the gradient, shimving and radio frequency (RF) coils, and cryogenic system further increase the scanner’s complexity, power requirements, dimensions, and cost.

Figure 1 illustrates the whole-body superconducting magnet manufacturing trend over the past three decades. On the one hand, the push towards higher field strengths is perceptible. The first 1.5 T systems were introduced by GE around 1984, but this field strength did not dominate the market until the early 1990s, with 0.5 and 1 T scanners more prevalent in the clinic. The first 3 T magnet by Magnex Scientific was installed in 1991, and in 1999 Siemens launched the head-only 3 T MAGNETOM Allegra. However, it was not until the early 2000s that the first commercial whole-body 3 T by GE arose in conjunction with short and wide-bore 1.5 T systems, replacing lower field strength scanners as the clinical standard. For a fixed superconducting wire design and composition, the magnet’s required amount of superconductor increases with field strength. Consequently, the scanner complexity and previously mentioned cost and siting challenges also increase. For instance, the length and weight of a 60-cm bore 1.5 T clinical MRI are 1.71 m and 4.5 tons, respectively. It requires a minimum space of 28 m² and less than 1500 L of liquid helium for cooling purposes. By contrast, the first available clinical 7 T scanner is 2.97 m long, and weighs approximately 20 tons, demanding a 65 m² room and 4000 L of cryogen. The need for a compensation solenoid coil to maintain field strength and the large

**FIGURE 1** Evolution of whole-body superconducting cylindrical MR magnets across the last three decades (years). The magnets are classified by field strength from 0.5–3 T, indicated by the marker diameter, and each color represents one magnet design.
stored energy of 7 T magnets cause the increased weight, system footprint, and cryogen requirements. On the other hand, Figure 1 portrays how manufacturers have progressively reduced the length and weight of their magnets relative to early models thanks to innovations in optimization algorithms and superconducting wire design. Short-bore and lightweight designs achieve two fundamental goals: reducing the scanner’s cost and improving patient comfort. Additionally, they improve MR accessibility, which has become a critical objective for superconducting and permanent magnet design.

The list price estimate of an MRI system is roughly $1 M per Tesla, but the purchase price can be significantly higher because of additional installation, operation, and maintenance costs. This imaging technique provides remarkable image quality and continues to unfold new diagnostic possibilities. However, most scanners do not comply with the accessibility dimensions introduced by Geethanath and Vaughan. Their elevated cost and exigent infrastructure requisites have led to a heterogeneous distribution of MRI technology across the globe. While reports on the disparity of MRI density across world regions are available, it is crucial to further characterize this discrepancy according to field strength (Figure 2). Field strengths of 1.5 T and above represent 85% of the MRI market in the USA and Europe. Conversely, scanners below 1.5 T are still the most abundant in low-resource settings. This article considers high- and low-resource settings based on the World Health Organization (WHO) income level classification. The data illustrated in Figure 2, while limited in availability, represent the current state of scanner density. In this instance, low and lower-middle-income countries exhibit low MR densities and frequently coincide with higher proportions of low field strength systems. These units typically correspond to permanent magnet-based scanners that involve lower acquisition and maintenance costs.

MRI magnets must drastically shrink their size to impact MR access. Two proposed methods have been lowering the field strength and loosening the constraints on field homogeneity. While both are viable approaches, they involve reconsidering certain aspects of the imaging process to overcome challenges such as loss of signal-to-noise ratio (SNR) and image distortions. Novel ultralow, portable permanent Halbach array magnets have adopted both methods. They deliver substantial benefits in portability and cost but entail prolonged scanning times and reduced image quality. Moreover, advanced and critical imaging techniques such as diffusion-weighted imaging (DWI), susceptibility-weighted imaging (SWI), functional MRI, and angiography are not readily feasible at such low field strengths. For these reasons, this work only considers scanners using superconducting magnets of 0.5 T. Such systems present the most cost-effective configurations capable of delivering optimal image quality, signal sensitivity, and a wide range of contrasts and image applications.

This work reviews past and current efforts in superconducting magnet design to improve accessibility. For magnets with higher tolerances for field homogeneity, we outline the results of novel imaging techniques that avoid, correct, or mitigate the artifacts that nonuniformities induce in the images. Finally, the last section summarizes our findings and discusses the challenges and opportunities of accessible MRI magnet design.

**FIGURE 2** Proportion of MR units classified by field strength in various countries worldwide. The tones of blue in each wheel chart indicate the proportion of each field strength reported for each country. We considered three categories: below 1.5 T, 1.5 T, and 3 T. Country grayscale coding is based on the WHO’s country classification by income 2021–2022. The number in parentheses next to the name of each country is the MR density, that is, the number of MR units per million inhabitants. WHO, World Health Organization.
ACCESSIBLE MAGNET CONFIGURATIONS

The ubiquitous clinical whole-body MRI comprises a cylindrical multicoil, liquid helium bath-cooled niobium titanium (NbTi) superconducting magnet. This configuration is the most cost-effective while ensuring optimal $B_0$ homogeneity conditions. Nevertheless, it also raises challenges such as strong dependence on liquid helium, narrow mechanical tolerances, and limited siting flexibility. Jointly, the magnet and the cryogenic system account for approximately 38% of the scanner’s total cost. Hence, tackling these subsystems can vastly impact MR accessibility. While details about MRI magnets and cryogenic systems are proprietary information, this section summarizes the approaches of the field towards reducing MRI’s footprint and dependency on liquid helium using publicly available data.

2.1 Short, wide-bore MRI

Patient space and comfort have been the motivation for new magnet configurations. Such systems feature short and wide patient bore while keeping the system’s overall height and weight low. A wide bore leaves extra room within the scanner walls and the patient, while a short bore length permits the head to lie outside the cylinder in most procedures. However, these configurations require more superconductor material to achieve the same field strength and experience higher peak fields than the longer and narrower bore versions. Parizh et al. demonstrated that, in order to achieve 10 ppm homogeneity over the DSV and maintain the same stray field, the field strength has to rise by 0.1 T per cm cut off of magnet length. Because magnet cost hinges on the superconducting wire type and amount, the magnet designer must carefully assess tradeoffs on field strength, field homogeneity, and fringe field.

Bore length ranges from approximately 1.25 to 1.95 m for 1.5 T systems from 1.65 to 2.13 m for 3 T systems. Figure 3 exemplifies the effects of magnet length, bore diameter, and $B_0$ on field homogeneity. Shorter magnets entail increased field inhomogeneity, while long magnets are needed to attain satisfactory field uniformity for higher $B_0$. Ultrashort magnets may even require more than the standard eight coils (six main coils, two shield coils) to fulfill uniformity standards. As an example, the 1.5 m long MAGNETOM Avanto (2003) needed seven main coils to achieve 0.2 volume root mean square error (Vrms) ppm homogeneity at 40 cm DSV, resulting in a higher weight than similar systems (Figure 3). The magnet length also determines the dimensions of the homogeneity volume, which sometimes can be an ellipsoid instead of a sphere because of reduced uniformity along the axial direction. The shortest bore length observed in a commercial scanner was 1.25 m for the 1.5 T MAGNETOM Espree (2004). The advantages of this scanner are higher patient acceptance and success rates for claustrophobic patients and interventional procedures. Nevertheless, these advantages come at the expense of smaller maximum FOV (45 x 45 x 30 cm$^3$), increased probability of geometric distortion, and longer acquisition times when imaging large body parts compared with long bore systems of the same field strength.

**FIGURE 3** Magnetic field homogeneity (Vrms) at 40 cm DSV versus magnet length (left) and weight (right) for different commercially available magnets. The data represent publicly available specifications of the magnets used by the most common MRI vendors, classified according to patient bore diameter (60 and 70 cm) and field strength (1.5 and 3 T). DSV, diameter of spherical volume; Vrms, volume root mean square error.
The quest for openness moved the standard patient bore from a 60 to a 70 cm diameter. The advent of the so-called “wide-bore” MRI occurred in 2004, and most new MRI installations now present this configuration. The widest patient bore available corresponds to the MAGNETOM Free. Max (2021), with a diameter of 80 cm. The cost of a magnet is primarily dictated by the length of superconductor material it requires. For a fixed field strength, superconductor length increases with patient bore, as it directly impacts the magnet warm bore diameter. Similarly, for a fixed patient bore size, larger lengths of superconductor are required to generate higher field strengths. Hence, this wide-bore diameter constraint imposed a tradeoff in the system’s field strength and gradient specifications. To restrain the overall cost and gradient power consumption, the manufacturers had to reduce the field strength to 0.55 T and limit the gradient’s maximum slew rate to 45 T/m/s. Wide-bore magnets tend to also be shorter in length compared with narrow-bore systems of the same field strength to limit the amount of superconducting material, compromising B₀ homogeneity. Higher order of gradient nonlinearity correction might also be required when imaging large FOVs. Xu et al. determined that decreasing the patient bore diameter is the best method to reduce magnet cost and minimum magnet length. The MAGNETOM Free.Star has adopted this approach. It utilizes the same NbTi conduction-cooled magnet technology but has a 60 cm patient bore, which allows cost reductions and increases accessibility compared with the MAGNETOM Free. Max (510 k approval pending).

### 2.2 Reduced liquid helium MRI

NbTi is a mature, mechanically robust, manufacturing friendly superconductor material optimized for MRI production. However, its low critical temperature of 9.3 K requires operation at liquid helium temperature, which results in a higher refrigeration and installation cost. Liquid helium is a nonrenewable resource extensively used in industry and is paramount in the field of superconducting magnets. Its high demand and the limited number of suppliers have led to increased and fluctuating prices, global shortages, and uncertainty about its future availability. Furthermore, the lack of liquid helium sources is a crucial cause of the reduced access to high-field MRI in remote areas and developing countries. The regular NbTi liquid helium-bathed magnets operate at 4.2 K and contain approximately 1500–2000 L of liquid helium. Early systems required periodical cryogen refills, which progressively became more spaced as cryocooler innovations reduced their consumption per hour. Presently, MRI utilizes the “zero boil-off” (ZBO) cooling technology. The cryocooler recondenses the helium gas that evaporates from the cryogen bath, creating a closed-loop system and eliminating the need for any refill, except for after a quench event. Despite eliminating the cost and burden of the refill operation, the magnet still requires 500 or more liters of liquid helium, and the cold head has to be periodically monitored and replaced. Specifically, the average cold head lifespan is 4–5 years for a new system and 3–4 years for a refurbished system.

| Characteristic                  | (A) Liquid helium-bathed NbTi magnets | (B) Conduction-cooled NbTi magnet | (C) Conduction-cooled HTS magnet |
|---------------------------------|--------------------------------------|----------------------------------|----------------------------------|
| Refrigeration                   |                                       |                                  |                                  |
| Liquid helium capacity (L)      | ~1500                                 | ~7                               | ~1                               |
| Liquid helium refills           | In the event of quench               | NA                               | NA                               |
| Operating temperature Tᵢₒ (K)  | 4.2                                   | Higher than (A)                  | 4–20                             |
| Stability                       |                                       |                                  |                                  |
| Temperature margin (K)          | 1                                     | 1                                | Depends on Tᵢₒ but higher than NbTi |
| MQE (mJ)                       | 1–10                                  | 1–10                             | Up to several Joules             |
| Quench protection               |                                       |                                  |                                  |
| Protection system type          | Passive                               | Passive                          | Active (research stage)          |
| Quench pipe                     | Required                              | NA (sealed magnet)               | NA (sealed magnet)               |
| Persistence                     |                                       |                                  |                                  |
| Joint resistance (Ω)            | 10⁻¹²                                 | 10⁻¹²                            | Must allow persistent operation (research stage) |
| Iₒ/Iₘ                          | ~70%                                  | Lower than (A) due to reduced Iₒ | Depends on the material N-value and Iₘ but usually lower than (A) |
| Commercialization               |                                       |                                  |                                  |
| Conductor cost                  | $                                     | $                                | $$$ (expected to decrease if mass produced) |
| kAmp-km                         | 15–20                                 | Larger than (A) due to reduced Iₒ | Larger than (A) due to reduced Jₘ |

**TABLE 1** Characteristics of three different magnet configurations using LTS versus HTS and conventional liquid helium bath versus conduction cooling assuming a field strength of 1.5 T.

Abbreviations: LTS, low temperature superconductor; HTS, high temperature superconductor; MQE, minimum quench energy; NbTi, niobium titanium.
Recently, the advances of Gifford–McMahon two-stage cryocoolers have allowed conduction cooling of superconducting magnets. These so-called “dry” magnets eliminate the liquid helium bath. Instead, the cold head performs magnet refrigeration via thermal conduction of typically copper straps. Additionally, conduction-cooled magnets are fully sealed and do not require the construction of a venting pipe. This attribute allows for a considerably more flexible and affordable siting of these systems. Commercially available examples of this magnet configuration are Philips’s BlueSeal 1.5 T (2018) and Siemens’ DryCool 0.55 T (2021). Both NbTi-based systems reduced their liquid helium use to 7 and 0.7 L, respectively. However, the low minimum quench energy (MQE) and tight temperature margin of NbTi (Table 1) make these magnets less stable and require a reduction in operating current. Consequently, conduction-cooled NbTi magnets demand more superconducting material to maintain the same field strength.

Conduction cooling is more appropriate for cooling superconducting magnets with larger temperature margins, such as those based on high-temperature superconductors (HTSs). The use of HTSs for MRI application is a niche area of research in the pursuit of liquid helium-free accessible magnets. These superconducting materials include magnesium diboride (MgB₂), yttrium barium copper oxide (YBCO), rare earth barium copper oxide (ReBCO), and bismuth strontium calcium copper oxide (BSCCO). They all have a higher critical temperature than NbTi and allow higher operating temperatures, eliminating the need for liquid helium as a cryogen. Furthermore, HTSs’ high MQE of up to 1–2 Joules (J) renders very stable magnets and practically eradicates accidental quenches. However, their normal zone propagation velocity is several orders of magnitude slower than NbTi. This parameter measures how quickly the magnet can spread its stored energy during a quench event. If the magnet is not sufficiently fast, hot spots are more prone to occur before the traditional quench protection methods can detect them.

GE’s Signa SP 0.5 T open magnet for interventional MRI was a pioneer system in the field, operating at 10 K with a gaseous helium-based cooling system instead of liquid helium. However, its Nb₂Sn magnet is considered a low-temperature superconductor scanner. The MROpen EVO is currently the only HTS-based commercially available scanner. This MgB₂ 0.5 T magnet has an open upright configuration and operates in driven mode. Besides this scanner, in vitro imaging has been demonstrated for Neoscan Solutions’ 1.5 T neonatal MRI, a 1.5 T YBCO magnet for extremity imaging, a 1.5 T BSCCO magnet for head imaging, and a 3.0 T BSCCO small magnet. These prototype magnets were taped wound and operating at 20 K. There are still pending challenges before manufacturing a viable commercial HTS scanner. These include the conductor properties and price, availability of joints for persistent operation, and safe active quench protection systems.

Table 1 summarizes the most relevant characteristics of potentially accessible systems, comparing them among the standard liquid helium-bathed NbTi magnet, new conduction-cooled NbTi magnets, and prototype conduction-cooled HTS-based magnets. Most of the properties for the latter configuration correspond to MgB₂-based magnets, as Parizh et al. concluded that this HTS has the best success probabilities in their extensive review of superconductors beyond NbTi, and to date, it is the only HTS featured in a commercially available scanner.

### 2.3 | Specialty MRI

When compactness is a priority but field homogeneity must be preserved, system designers may reduce the imaging volume and limit the system’s application to dedicated examinations. These specialty MRI systems are anatomy-targeted to different body regions such as the head, extremities, breast, or imaging of neonates. Accordingly, the design of the system undergoes modifications tailored to its application.

The Compact 3 T (C3T) head-only system initially introduced by Foo et al. is another example of a sealed conduction-cooled NbTi magnet that requires only 12 L of liquid helium. Its predecessor is the head-only MAGNETOM Allegra 3 T scanner, optimized for fast brain imaging. The C3T system demonstrated safe brain imaging using an optimized gradient system with 80/m amplitude and 700 T/m/s slew rate. The increase in the slew rate allowed notably shorter echo times (TEs) in DWI and fast spin echo (FSE) sequences, and echo spacing reduction in echo-planar imaging (EPI). These reductions in acquisition achieved images with better SNR, sharpness, and geometric fidelity than images acquired in a whole-body 3 T scanner. Nonetheless, C3T images required higher-order gradient nonlinearity correction and experienced similar amounts of motion artifacts.

The Synaptive 0.5 T MRI is another compact conduction-cooled head-only alternative. Although at a lower B₀, its gradient system offers upgraded performance with maximum gradient amplitude and slew rate of 100 mT/m and 400 T/m/s, respectively. DWI and diffusion tractography imaging are feasible in this system with similar image quality, fractional anisotropy, and apparent diffusion coefficient mean values compared with whole-body 1.5 T scanners. Additionally, the lower field strength allows a 9- and 36-fold drop in specific absorption rate (SAR) compared with 1.5 and 3 T, respectively.

Various dedicated 1.5 T breast MRI scanners are commercially available (Aurora, Time Medical EMMA 1.5 T). These units include more powerful gradients and breast-tailored RF coils similar to head scanners. Research using Aurora reported better performance in breast cancer screening than whole-body 1.5 T scanners. However, breast-dedicated MRI entails large magnets as they must accommodate the entire torso inside the bore. Both models weigh approximately 3 tons and need a minimum room of 55 m², presenting similar siting challenges as 1.5 T whole-body scanners.

GE launched its 1.5 T Optima MR430s orthopedics-dedicated scanner in 2011. The magnet has a 21.8 patient bore, weighs approximately 400 kg, and requires only 50 L of liquid helium. The reduced gradient size provided low power deposition and reduced noise during acquisition.
also practically eradicates peripheral never stimulation (PNS) probability while delivering 70 mT/m amplitude and 300 T/m/s slew rate. 87 These characteristics made the system suitable for converting into a neonatal scanner in a clinical setting. 88

In most cases, the efficient use of the bore space renders more compact magnets with reduced costs and footprints. Additionally, the reduced size of the gradient coils allows the use of more powerful gradients at the PNS threshold. These size characteristics facilitate effortless siting, operation, and maintenance, while the gradient systems boost image quality. The reduced size also implies less liquid helium required for operation, making specialty MRI apt for conduction cooling and HTS magnets. 14 However, their implementation in traditional clinical settings is scarce because of their lack of clinical universality compared with whole-body scanners. 92

3  IMAGING IN HIGHLY INHOMOGENEOUS FIELDS

Differences in magnetic susceptibility at tissue interfaces or phase accumulation during long readouts can cause field deviations. The effects of these field nonuniformities on the images are well known; they are responsible for geometric distortions in EPI, blurring in spiral imaging, and signal loss or pile-up. These inhomogeneities are local and relatively small in magnitude, and postprocessing correction techniques can tackle the artifacts that they produce. These mitigation methods 89–93 are widely available in open-source packages 94–96 for fMRI and DWI data processing pipelines. 97–99 By contrast, the field inhomogeneity that an accessible, ultra-short magnet poses may involve smoothly varying bandwidths of tens of kHz over the imaging volume. The employment of inhomogeneous fields is a contrasting change in the conventional imaging methods, but is required to augment MRI accessibility. 25 Severe B₀ inhomogeneity poses issues to spatial selection and encoding. These include T₂⁎ local dephasing in gradient echo-based sequences and geometric distortions in the frequency-encoding direction and slice profile of 2D acquisitions. Reducing the TE as much as possible is the only mitigation option for the former artifact. By contrast, solutions to the latter require either lowering B₀ or increasing the amplitude of the readout gradients and using nonspatially selective pulses when possible. 10 However, nonsel ective pulses introduce spurious signal from outside the FOV. A good tradeoff is to leverage multiple slab-selective pulses to cover the whole imaged region together with dynamic shimming methods. 100–102 The challenges that such B₀ inhomogeneity poses to the imaging process and the available sequences to overcome them were recently reviewed by Mullen and Garwood. 10 Multispectral imaging (MSI) methods have shown potential among the reviewed sequences. These techniques assume large and rapidly varying field inhomogeneities as their purpose is imaging near metallic implants. Some of these sequences are multiacquisition variable-resonance image combination (MAVRIC), 103 slice encoding for metal artifact correction (SEMAC), 104 or the combination of both (MAVRIC-SL). 105

After manufacture, the field homogeneity of a whole-body superconducting magnet is several hundred ppm. 9,12,13 Manufacturers leverage passive and active shimming to reduce this value to clinical homogeneity standards. Passive shimming utilizes the induced magnetic field of diamagnetic and paramagnetic materials. These strategically placed materials correct hardware imperfections but are sensitive to temperature changes. 106 By contrast, active shimming uses superconducting coils and reduces the homogeneity to levels acceptable for imaging before data acquisition. Most B₀ imperfections within the magnet bore, including subject-specific susceptibility-based distortions, are mitigated by tailoring the coils’ current. 106–107 However, bore space for high-order shimming coils is limited within the bore of a compact magnet, 20 and they increase superconducting usage and the complexity of the cryostat design. Furthermore, active B₀ shimming relies on the field distribution measurement, and inaccuracies in this acquisition can degrade the shim performance. 107

Ideally, a distortion-free image could be reconstructed from a distorted one if the true field map distribution was known. 9,107 However, the signal loss during image acquisition would be nonrecoverable. A B₀ map is generally acquired using a double-echo gradient-recalled echo (GRE) sequence with short TE. This acquisition and subsequent phase image calculation are subject to errors caused by phase wrapping, eddy currents, low SNR, and coil-combination methods, even in a homogeneous field. It also involves a longer scanning time because of the extra sequence acquisition, during which motion can occur and cause registration errors. Furthermore, when acquired in an inhomogeneous magnetic field, geometric distortion along the frequency-encoding direction and spin dephasing typical of GRE sequences may exacerbate the problem, limiting the field map’s accuracy and the subsequent artifact correction of the images. Thus, techniques that estimate the field’s spatial distribution using alternative methods are essential.

Dual polarity encoding techniques first introduced by Chang and Fitzpatrick 93 estimate the field map from two sets of images acquired with opposite phase-encoding directions. Mullen and Garwood 106 used dual polarity in conjunction with a missing pulse steady-state free precession (MP-SSFP) sequence to obtain and correct images of a phantom with a metallic implant that mimics field inhomogeneities. They compared the results using as a benchmark the images of a MAVRIC sequence, which also provides a B₀ map estimate and is the clinical standard when imaging near metallic implants. 105,109 The dual polarity MP-SSFP sequence used low SAR, 20 kHz RF-pulse bandwidths for excitation and refocusing and reported an acquisition time that was 3.17-fold faster than single encoded polarity MAVRIC. The estimated field map showed high fidelity to theoretical simulations and leveraging it for correction resulted in near artifact-free images and avoided ripple artifacts characteristic of MSI. However, MAVRIC images showed recovered signal in closer proximity to the metal and required less processing time. The B₀ maps indicated off-resonance levels of ±47 ppm (3 kHz at 1.5 T). Nevertheless, the authors indicated that further enhancement of the sequence is necessary prior to its clinical feasibility assessment and implementation. MSI methods remain the clinical standard, implemented in all major vendors and widely used
when imaging near implants. A similar method utilized an unsupervised neural network to obtain the frequency field maps from 3D-MSI dual polarity images. The authors leveraged the predicted field maps to correct pile-up and ripple artifacts near metallic implants that caused off-resonance of up to ±156 ppm (10 kHz at 1.5 T).

A deep learning (DL)-based approach by Gowda et al. introduced RE-construction of MR images acquired in highly inhOmogeneous fields using DEep Learning (REMODEL). This convolutional neural network (CNN)-based tool inputs corrupted complex k-space data and outputs artifact-free image magnitude data. They tested the model on simulated data that included T1-weighted images corrupted with random generated field maps of up to ±390 ppm (50 kHz at 3.0 T). REMODEL demonstrated faithful reconstruction and root mean squared error smaller than 0.15 with respect to the ground truth images in all cases.

Figure 4 provides a visual summary of the timeline of methods for imaging in inhomogeneous fields since 1988. It comprises acquisition, $B_0$ map estimation, and reconstruction approaches.

### 4 | CHALLENGES AND OPPORTUNITIES

Accessible MRI requires making significant changes to the entire scanner and imaging process as we currently know it. As mentioned in Section 1, superconducting cylindrical magnets are the most cost-effective viable option for an accessible MR. In particular, mid-field (0.5–1.5 T) scanners can deliver sufficient image quality for clinical use while limiting accessibility challenges. Field strengths higher than 1.5 T offer SNR and image resolution advantages but entail increased cost, weight, size, and cooling requirements.

Past attempts to modify MRI design have primarily focused on enhancing image quality or patient comfort. Open systems accomplish the latter by better accommodating claustrophobic and obese patients at the cost of more challenges during siting and limitations on field strength. By contrast, anatomically targeted scanners can deliver high performance with a reduced footprint and cost when image quality is the priority. However, because of their lack of clinical universality, the purchase of several dedicated systems would be required to perform...
the most common imaging procedures, rendering a less favorable approach for sites with a limited budget. The democratization of MRI demands short-bore, lightweight, stable scanners that are easy to site and capable of delivering clinical quality performance with low power consumption. Reducing or eliminating the amount of liquid helium required for MRI operation can ease the siting and stability issues and MRI’s dependency on this cryogen. Because of their operating temperature and stability advantages (Table 1), innovations in conduction-cooled HTS magnets will be crucial. However, the feasibility of whole-body HTS still depends on solving some engineering challenges, such as the viability of permanent joints and quench protection systems.

In addition, significantly shrinking the magnet can achieve some accessibility requirements on 2.1. demonstrated that a wide bore increases the overall scanner’s cost and puts pressure on the gradients. Thus, an accessible MR will benefit from a bore narrower than 70 cm. Regarding magnet length, a shorter bore will share advantages with specialty magnets (Section 2.3). They would be able to house moderately powerful gradients, resulting in improved image quality with reduced PNS. However, it is evident from Section 2.1, that reducing the magnet length leads to an inevitable increase in field inhomogeneity. While Bₕ shimming methods have been the gold standard to correct these field deviations, these techniques introduce other penalties in the magnet design. Shimming hardware ultimately affects the final weight, bore dimensions, amount of superconductor, and cryogenic requirements (Section 3). Instead, opportunities will unfold if magnets relax their rigid restrictions on field homogeneity. Section 3 covered promising methods to acquire artifact-free or artifact-mitigated images in the presence of field inhomogeneities and to estimate the spatial field distribution. Field map synthesis and subsequent image correction using the resulting estimation have been demonstrated as feasible analytically and leveraging DL models.

Manufacturers have explored reducing the field strength of superconducting magnets below conventional values to reduce the cost while maintaining field homogeneity constraints. This approach aims to achieve sufficient image quality for clinical use while being conservative on B₀. The field strength choice depends on a tradeoff between cost, performance, and feasibility of advanced applications. A novel 0.55-T scanner has demonstrated the feasibility of this strategy by leveraging the past progress in MRI engineering and artificial intelligence (AI). Additionally, AI has demonstrated outstanding performance in essential tasks such as noise reduction, artifact detection, and mitigation, image quality improvement, and scan automation. These capabilities in an accessible MRI system will allow shorter scan times without compromising clinical quality and reduce the requirements for on-site skilled personnel.

In conclusion, an accessible MRI requires a breakthrough in magnet design and imaging techniques. Based on the accessibility demands and a review of the characteristics of existing magnets, we consider that the best candidate will be a whole-body superconducting, mid-field, conduction-cooled HTS-based, short magnet. Such a scanner will also leverage AI methods to optimize image quality, signal sensitivity, ease of operation, and patient throughput.

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ENDNOTE
* The 1.5 T scanner was chosen from the same manufacturer and with the same patient bore size as the 7 T system for a fair comparison.

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