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Advances in Whole-Body MRI Magnets

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***Abstract*— Magnetic Resonance Imaging (MRI) is the largest commercial application of superconductivity. MRI is a powerful diagnostic tool that the medical community considers as a procedure of choice for visualization of soft tissue. The recent decade has marked substantial progress in MRI magnets and systems. The 3.0 tesla horizontal field and 1.0 tesla vertical field open whole-body MRI systems have become commercially available. The superconducting magnet is the largest and most expensive component of an MRI system. The magnet configuration is determined by numerous competing requirements including optimized functional performance, patient comfort, ease of siting in a hospital environment, minimum acquisition and lifecycle cost including service. The factors that drive the magnet requirements are increased center field, maximized uniformity volume, minimized field decay and stray field, magnet compactness, long helium refill period, and more. Advances in the cryogenic technology and magnet design practice provide means for improvements in magnet performance while meeting the market requirement for continuous system cost reduction.**

***Index Terms*— Magnetic Resonance Imaging, MRI, superconducting magnets**

I. Introduction

The first practical superconducting magnets were built in the 1960's following the discovery of NbTi alloy. It was, however, the invention of Magnetic Resonance Imaging (MRI) that took superconductivity from the scientific laboratory to everyday use. MRI has transformed superconductivity from a scientific curiosity to a phenomenon that improves people's lives. It is also true that superconductivity benefited MRI by making it commercially feasible.

Since publication of the first human body images in 1977 [1], MRI has become one of the primary tools in medical diagnostics. MRI is the only chemically sensitive in-vivo imaging technique with high-resolution soft-tissue contrast. It allows physicians to peer deep inside the human body, producing clinically relevant images of soft tissue lesions and functional parameters of body organs, without the use of invasive procedures or ionizing radiation such as X-rays.

The low-field whole-body MRI magnets (<0.35 tesla) are a mix of resistive magnets with iron yoke and permanent magnets. Resistive magnets have the lowest installation cost among all types of MRI systems but require a large power consumption. The permanent-magnet MRI systems are heavy. Their installation cost is rather high but maintenance cost is low. The low-field magnets typically have relatively poor uniformity and stability. Poor uniformity results in poor image quality, although it might be adequate for some applications.

With few exceptions, MRI systems with a central field strength greater than 0.35 tesla use superconducting coils. MRI with superconducting magnets account for more than 75% of the installed MRI base. Advantages of superconducting MRI systems include, but are not limited to, better performance, higher signal-to-noise ratio as a result of higher field, higher resolution and lower lifecycle cost [2]-[4].

MRI uses the majority of superconducting materials produced worldwide. Averaged over the last decade, MRI magnets use about 60% of all superconducting wire (including copper), and about 40% of the NbTi alloy [5]. The higher fraction of conductor is due to the fact that MRI magnets use conductors with a high content of copper; a typical MRI conductor contains 80 to 90 volume percent of copper, and only 10% to 20% NbTi.

II. Superconducting whole-body MRI systems

A. Installed MRI base

After 30 years of commercial production, the industry of superconducting MRI has reached a state of maturity. The demands of the healthcare industry for high efficiency, low cost, reliable systems resulted in technically-challenging, well-integrated magnet designs reproducible in volume production.

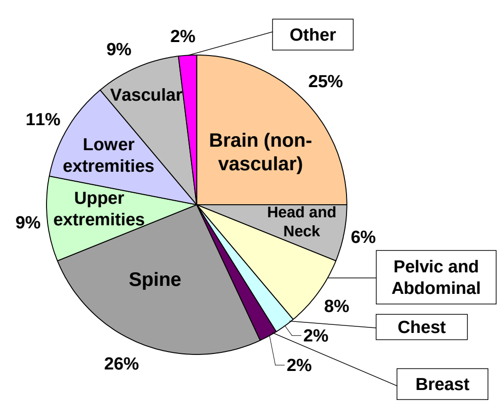

Figure 1. MRI procedures in developed countries, per modality (2007)

In 2008, the total installed base of superconducting MRI systems was about 26,500 units vs 14,600 systems in 2002. More than 2,500 superconducting MRI scanners are produced

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worldwide annually. An estimated 80 million MRI exams were performed worldwide in 2008, showing about 5% annual growth [6]. The US alone represents about 40% of the world MRI market, with about one system per 30,000 capita. In recent years, there is a significant increase in MRI systems installed in developing countries such as China, Brazil and India, typically in new facilities. In developing countries as many as 60% of the new scanners are being installed in new sites while in the USA 80% are replacement units and only 20% of the systems are installed in new facilities.

Spine and brain exams account for about 50% of all MRI procedures. Cardio-vascular and brain imaging demonstrate the highest growth rate, in part due to increased availability of higher field systems. Figure 1 illustrates the advantages of a versatile whole-body system. No single ‗dedicated' system such as one for extremities [7] or brain scanning would be able to serve more than 25% of the patients.

B. *Field strength*

Depending on field strength and shape, there are several types of superconducting MRI magnets. Thousands of commercial whole-body systems of 1.5 tesla and 3 tesla are produced worldwide. The very-high field 4.0 T to 9.4 T units are in the process of being evaluated at research sites and are for investigational use only. A unique 11.7 tesla scanner is being built to be installed in Saclay, France [8].

Table I summarizes typical parameters of MRI magnets. Within each column, magnet characteristics may vary significantly depending on uniformity, stray field, system dimensions, type of refrigeration, and other technical and commercial factors. The 1.5 T to 7 T magnets are assumed to be actively shielded (passively-shielded 7-tesla magnets use roughly half the conductor at a penalty of higher stray field and the need for a several-hundred ton iron shield). The weight in Table I includes cryogens and does not include the weight of other system components such as iron shielding, gradient coils or electronics. The Amp-Length is a product of the operating current and conductor length that is equal to the product of coil volume and engineering current density.

TABLE I
TYPICAL PARAMETERS OF CYLINDRICAL MRI MAGNETS

	1.5 T	**3 T**	**7 T**	**11.7 T** [8]
Length, cm	125-170	160-180	~300	400
Outer diameter, cm	190-210	190-210	>250	460
Stored energy, MJ	2 - 4	10 - 15	50 - 90	340
Weight, ton	3 - 6	5 - 10	>25	150
5-gauss line (Z x R), m	4 x 2.5	5 x 3	>7 x 5	9.6 x 7.5
Amp-Length, kA-km	15 - 25	35 - 60	120-180	~300

There is a drive towards making higher-field MRI systems. The high field systems potentially benefit from yet higher signal-to-noise ratio, contrast-to-noise ratio and higher scanning speed. On many occasions only high-field systems may provide sufficient image quality to identify abnormalities, especially in cases of brain and heart exams. There are, however, limitations that may restrict full realization of the high-field benefits [9]. Technological limitations for the magnet include such factors as an increase of the stray magnetic field, the need for stronger, higher linearity gradient coils, and, in some cases, reduction of the uniformity area in higher-field magnets. Table I illustrates that in 3 T systems the >5 gauss area with restricted access is about 50% larger than for 1.5 tesla scanners. High field changes relaxation kinetics in tissues, and may require changes in the scan protocols. Also, safety risk and patient discomfort factors may increase with magnetic field, although these can generally be managed.

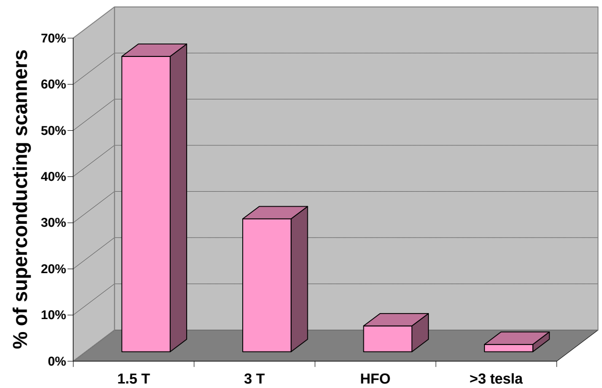

Figure 2. Delivered superconducting MRI systems by field strength (2008)

During the last decade, the lower-field superconducting 0.5 T and 1.0 T cylindrical scanners practically went out of production. Their marginally-lower cost was insufficient to outweigh the advantages of the commercial 1.5 T systems in faster patient throughput and better image quality. Until the late 1980's, the lower-field systems dominated the MRI market. In 2000, approximately 600 low-field systems were produced representing >20% market segment. By 2005, their production was practically stopped.

Figure 3. Philips' 3 tesla scanner in mobile configuration.

The 1.5 tesla units represent the majority of scanners produced in recent years (Figure 2). The 1.5 T systems are a good compromise between performance, patient comfort, ease of siting in a hospital environment, optimized installation, and life-cycle cost. Insurance reimbursement may also favor the 1.5 T systems: an average cost to purchase a 1.5 T whole-body

MRI system is $1.25M vs $2.0M for 3 T units while in many countries reimbursement per exam is about the same [10].

After several years of research use, the 3 tesla whole-body scanners from General Electric, Philips and Siemens entered the marketplace in the early 2000s. Initially, the commercial 3 T scanners were rather large and heavy, weighing 10,000 kg or more. The latest 3 tesla scanners have a significantly lower weight. Their dimensions and uniformity volume are now similar to 1.5 tesla scanners. The Philips‘ Achieva 3 T weighs only 5,600 kg (including cryogens) and may be delivered in either stationary or mobile configuration (Figure 3). Today, the 3 T systems are the fastest-growing segment in MRI industry.

About thirty higher-field whole-body 7 tesla to 9.4 tesla systems are installed at luminary sites around the world, usually university hospitals. Initially, these scanners were used solely for brain imaging that requires a relatively small image area with high uniformity. Now, a few research centers are extending imaging to cardiac, prostate, breast, extremity and other areas, with the ultimate goal being to expand the range of applications beyond high resolution anatomic and functional brain imaging. The high-field magnets are highly customized depending on the image volume, bore size, type of shielding, etc. The length of 7 tesla magnets varies from 2.6 m to 3.5 m. The whole-body 7 T system provided by Philips to several research sites has a magnet that weighs 32 tons and has steel shielding options of 218 tons and 406 tons (the magnet is built by Agilent Technologies). A similar Siemens 7 T scanner weighs 32 tons and requires 250 tons of wall shielding [11].

The last decade shows how definitions may change. In the 1980s, the 1.0 T MRI units were called high-field. In 1990s, the 1.5 T systems were called ‗high field‘ while 3 T MRI were ultra-high field. Today, 1.0 T cylindrical magnet is a low field unit, 1.5 T is the standard field, although 1.0 T Open magnet is considered a high-field unit for that geometry. The 3 T MRI are now high field units, and 7 T and higher-field MRI are called ‗ultra-high field‘. In the future, advances in the magnet technology could rename the 3 T MRI to the standard field.

C. *Magnet shape and orientation*

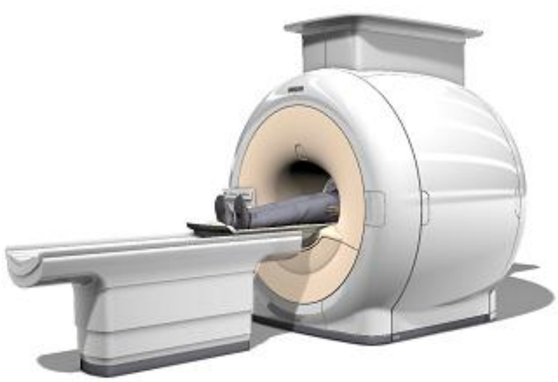

Figure 4. Cylindrical MRI scanner with patient bed

More than 95% of superconducting MRI magnets have a standard cylindrical shape (Figure 4). This mature magnet design provides a high-quality field for a large variety of applications. It allows minimization of the stray field, compact dimensions and low cost. This configuration permits mobile configuration that may move from location to location while at nominal field thus minimizing the setup time.

Figure 5. The 1.0 T high-field open (HFO) scanner Panorama from Philips

Cylindrical MRI systems have a known but accepted limitation: the narrow patient bore of typically 60-cm diameter and >100 cm length. This tunnel creates several issues: (a) obese patients may not fit into the tunnel, (b) a claustrophobic effect causes certain patients to reject the procedure creating financial loss for the image center and diagnostic loss for the patient, and (c) it restricts interventional medical procedures. More recently, the patient bore has become a limiting factor for certain image-guided medical procedures. The image-guided medical ablation is an example of such a procedure. It requires patient access of an open MRI with center field of at least 0.5 T. This procedure uses 3D MRI images to facilitate biopsy or treatment of tumors, e.g. liver cancer [12].

Figure 6. Resistive passively-shielded 0.6 T open magnet (Fonar Corp.)

There is a trade-off between more patient-friendly open MRI and the higher-field cylindrical systems that are the best for diagnostic information. Open MRI systems (Figures 5 and 6) are very challenging technically. Their installation cost may be higher than that of even 3 T systems. Open MRI are limited to a field of about 1 T. The open magnet uniformity volume is often smaller than in 1.5 T and 3 T cylindrical scanners. This might result in lower image quality and longer scanning time. The stray field may be higher than in 1.5 tesla scanners.

Fonar Corporation manufactures another type of open MRI. This resistive 0.6 T system (Figure 6) offers a unique variety of patient positions during imaging including images of the spine and knee obtained while bearing weight. This system requires 225 kVA to power the magnet [13]. The field of >0.5 T and wide patient opening are feasible for passively-shielded system only. The same-shape superconducting unit would require about 20 tons of iron to shield the magnet.

III. Aspects of the Magnet Design

A designer of superconducting MRI magnets must address multiple trade-offs. Table II lists some of these trade-offs.

TABLE II
REQUIREMENTS TO MRI MAGNETS

Image quality • Field strength • Field homogeneity and stability in large volume
Costs to the customer • Low initial cost of the magnet and system • Low operational costs: low helium loss, long refill interval, low power consumption • Small stray magnetic field: outside MRI suite and at the location of MRI components • Light weight
Customer needs • Safety: the 5 gauss line limited to MRI suite, Emergency Run-down function, standards • Reliability: maximum uptime, long service time, no quenches in hospital • Short scanning time, high throughput • Compact/Accessible • Patient friendly: wide and short bore, open
Installation & service • Light weight, compact size • Fast installation/adjustment • Service at field

A. *Compactness and Accessability*

Patient comfort, the ability to perform medical procedures during scanning and ease of installation require magnets to be compact. Compactness includes short and wide patient bore, reduced cryostat outer diameter to minimize ceiling requirements, and low system weight.

The early MRI systems were large and heavy [3]: their length was about 250 cm and weight >10 tons. Since 1988, Philips introduced a family of compact 0.5 T to 3 T scanners with the magnet length of only 157 cm and outer diameter of 188 cm. Even the heaviest 3 tesla magnet in the family weighs less than 6 tons. This family sets the industry standard for MRI compactness and assured wide proliferation of MRI systems to ordinary hospitals and imaging centers around the world.

Recently, Siemens introduced the Magnetom family of even more compact magnets. Siemens 1.5 T Espree has an increased patient bore diameter of 70 cm and 125 cm length although the uniformity volume is smaller than in a typical smaller-bore magnet. It is too early to judge, however, the size of the future market segment for the wide-bore magnets. While these systems provide better access to the patient and a higher comfort level, in large market segments the question remains whether these benefits outweigh the higher system costs.

B. *Uniformity and persistence*

In order to provide high-quality images, MRI magnets must generate a magnetic field with very high temporal and spatial uniformity on the order of several parts-per-million (ppm) over the whole imaging volume. The typical guaranteed field decay in MRI magnets is less than 0.1 ppm/hour. The typical requirement for the commercial 1.5 T and 3 T magnets is that the field uniformity is on the order of 10 ppm peak-to-peak in about 50 cm diameter volume. MRI system designers may trade off a reduced image volume and system compactness either at a penalty of longer scanning time that assumes, for example, multiple scans to achieve extended coverage, or limit system application to dedicated examinations such as brain scanning.

The high spatial uniformity in MRI magnets is achieved by precise multi-coil design that typically consists of six to ten coils (Figure 7). The design uniformity in such magnets is on the order of 10 ppm over the imaging volume. Increase in the number of coils allows an improvement of the magnet uniformity with the penalty of the magnet complexity and cost. Increase of the peak field in coils is another disadvantage of the multi-coil configuration. In a long solenoid the peak field in conductor is about the same as the center field, while in multi-coil configuration the peak field in conductor is significantly higher. In an actively-shielded 1.5 T magnet, the peak field in coils may be 5 T or higher.

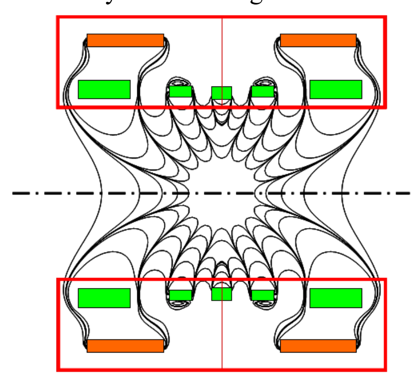

Figure 7 Typical coil configuration of a cylindrical actively shielded MRI magnet. The curved lines correspond to 10, 100, 1,000, 10,000 and 100,000 ppm uniformity.

In order to achieve the required uniformity, the coils must be precisely positioned with tolerances of fractions of a millimeter. Even using the best manufacturing practices, the standard commercial magnets have bare uniformity of several hundred ppm. Magnetic shimming is necessary to compensate for manufacturing variability and magnetic site environment. Shimming improves the magnet uniformity to the design value of 10 ppm over the image volume. Two shimming methods are used in MRI systems: active shimming using superconducting coils located in the cryostat, and/or passive shimming that uses small pieces of iron installed in the magnet bore. Either shimming method is magnet- and site-dependent: the magnet should be re-shimmed when moved to a new location. Passive shimming is the most cost-effective and reliable solution for commercial scanners [14].

The very high temporal stability required by MRI systems is an order of magnitude better than modern high-current power supplies can provide. Therefore, commercial MRI magnets operate in persistent mode: all coils are connected in a closed superconducting loop with a persistent current switch. For 1.5 T magnets, the total resistance of the loop shall be less than a fraction of one nano-ohm. Even a small interruption of only 0.01 mm in a superconducting circuit is unacceptable. Broken NbTi filaments is an example of such an interruption. Nearly-perfect superconducting joints with guaranteed resistance of $<10^{-11}$ ohm are required. Non-linear current-voltage conductor characteristics should be taken into account to assure the magnet persistence.

C. Stray magnetic field and shielding

MRI systems must be designed and installed such that the magnetic field outside of the scanning suite does not exceed the industry-standard safe magnetic field of 5 gauss. Five gauss is the maximum field at which a reliable operation of devices such as heart pacemakers can be guaranteed.

Magnetic field outside of a dipole may be estimated as

$$B = B_c\,(R/r)^3\,(1 + 3\cos\theta)^{1/2} / 2, \qquad (1)$$

where B_c is the magnetic field at the center of the dipole of radius R, and θ is the angle between the magnet axis and the direction to the field point r, $r >> R$. From Eq. (1) the 5-gauss field would be about eight diameters of a 1.5 tesla unshielded magnet in axial direction (θ = 0) and about six magnet diameters in radial direction (θ = 90º). This large area of restricted access is unacceptable.

Three types of shielding or their combination are used in the whole-body superconducting MRI systems:

1. Active shielding [15]: superconducting coils located at a larger diameter suppress magnetic field outside of the cryostat to values of Table I;
2. Passive (iron) shielding with iron attached to outer surfaces of the cryostat;
3. Iron shielding on the walls of the image suite.

Today, all commercial superconducting magnets and even some 7-tesla research MRI magnets are actively shielded. The compact, low-weight actively-shielded scanners dramatically reduce site setup cost associated with passive shielding. These magnets provide more stable magnetic field as they do not depend on ambient temperature variations. There is an increase of the conductor cost: typical actively-shielded MRI magnets require twice the amount of conductor than non-shielded magnets would use.

The increase in superconductor cost is smaller than the cost of passive shielding. A typical non-shielded 1.5 T magnet requires about 20 ton of iron regardless of whether iron is on the walls, or iron is applied to the outer surface of the cryostat [16]. In the US, room shielding may cost as much as $100,000.

D. Refrigeration

Development of commercial MRI systems has resulted in dramatic improvements in cryogenic performance. Early MRI magnets used liquid nitrogen thermal shields. These magnets had a helium boil off rate of about 0.4 liters per hour, requiring liquid helium (LHe) refill every 4.5 months, and nitrogen refill every one or two weeks. By the late 1980s, MRI systems adopted two-stage Gifford McMahon refrigerators that eliminated the need for the liquid nitrogen thermal shield, reducing helium consumption initially to less than 0.1 liters/hour. By 2000, LHe consumption was further reduced to less than 0.03 liters/hour resulting in a typical four-year interval between LHe refills [3].

In the last decade, zero boil off (ZBO) refrigeration became standard in commercial MRI systems, especially in 3 tesla and HFO units. ZBO refrigeration uses the same two-stage Gifford McMahon refrigerator but with advanced heat exchangers creating low enough temperatures to re-condense helium gas within the cryostat. ZBO magnets allow practically unlimited system operation without helium refill. In addition, ZBO units enable more compact magnet design, as only one thermal shield in the cryostat is required instead of two. Disadvantages of the ZBO configuration include higher refrigeration costs and higher power consumption.

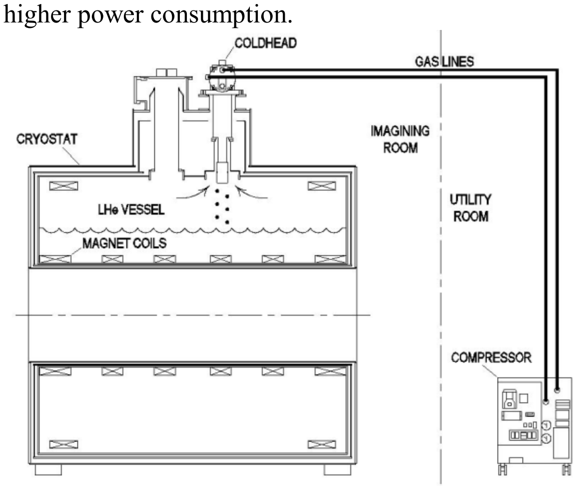

Figure 8. Principal schematic of ZBO refrigeration

The ability to create a ZBO unit depends on both excellent insulation and advanced refrigeration. Insulation techniques minimize all heat transport mechanisms (conduction, convection and radiation). A compact Cold Head directly integrated into the cryostat provides cooling (Figure 8). The Cold Head is connected to a remotely located compressor. Helium gas is circulated in a closed-loop fashion. The compressor compresses the helium gas and the Cold Head expands the helium gas to create low temperature cooling.

The refrigeration components are proven, reliable devices. However, because the components operate on a 24/7 basis, routine maintenance is required. Service intervals are extended, however, and are only required once every several years. Maintenance involves changing a filter medium in the compressor and the replacement of consumable items within the cold head. Both activities can be accomplished in a timely, scheduled manner requiring only a few hours of system downtime and a minimum of inconvenience.

IV. Future Opportunities

We have already discussed future magnet opportunities and trade-offs including very high-field, wide bore and open magnets. In this section, we will evaluate new superconducting materials that may add customer-oriented features and reduce system cost, especially the lifecycle cost.

Present commercial MRI magnets utilize NbTi conductor. NbTi conductor is a mature, mechanically-strong, manufacturing-friendly material. It is routinely available in long lengths of 10 km or more. The high critical current density J_c of >3,000 Amp/mm^2 at 4 T, 4.2 K and high value of index *N* above 40 allow production of high engineering current density, compact, cost-effective magnets. The NbTi wire is well optimized for MRI production. It is the lowest cost superconducting material, priced at about \$1/kAmp-m at 4 T, 4.2 K. Disadvantages of NbTi conductor include low critical temperature of 9.3 K and relatively low critical field of about 10.5 tesla at 4.2 K. These parameters require operation of NbTi magnets at liquid helium temperature resulting in high cryostat and refrigeration cost.

High-temperature superconductors (HTS) were considered for MRI application almost immediately after their discovery [17]. Unfortunately, HTS conductor is still rather expensive. An HTS conductor cost of \$10/kAmp-m at 77 K, self field marks the threshold point where HTS conductor may be considered for commercial dedicated MRI. Dedicated MRI systems use significantly less conductor then the whole-body units but benefit from operation at increased temperature and reduced cost of refrigeration.

Recently discovered magnesium diboride MgB_2 with critical temperature T_c = 39K offers the potential to become the MRI material of the future. MgB_2 promises quench-free, cryogen-free MRI systems. Several research MgB_2 MRI magnet projects are underway [7], [18], [19]. The MgB_2 conductor cost is driven by processing rather than by the material cost. Today, the MgB_2 price is about \$5/kAmp-m at 4 tesla, 4.2 K, i.e. it is significantly higher than that of NbTi. The conductor price is expected to be reduced in volume production.

There are still multiple material-related and magnet-related issues to be addressed. The highest critical current density achieved in MgB_2 samples is 30% below *Jc* in NbTi at 4 tesla, 4.2 K [20]. The *N*-value at 4 tesla is no better than 30 even in short samples. If not guaranteed to be more than 35 over 100% of the wire length, the low *N*-value will require either magnet operation at a relatively low fraction of critical current, or a driven-mode operation. Unless improved, these will result in higher material demand per magnet, and a less compact, more expensive magnet. Conductor should not require any additional treatment after coil is wound. Mechanical properties should be improved. Long conductor lengths of several km should be produced with guaranteed properties over 100% of the length. Magnet designers should develop efficient technologies for building MgB_2 magnets including, but not limited to, efficient winding technologies, quench protection, and superconducting joints. Refrigeration should be optimized for MgB_2 conductor: lower operating temperature results in lower conductor cost, while increasing the refrigeration cost.

V. Conclusion

The MRI industry is driven by the demand to provide high-quality service to patients in a cost-competitive environment. In the 30 years since introduction, superconducting MRI magnets for mainstream systems have reached a certain level of maturity. Volume production of MRI magnets has led to efficient, well integrated magnet designs. Still, there are opportunities for improvement to enable this excellent diagnostic tool to be available to patients worldwide.

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