Development of Superconducting Magnets for High-field MR Systems in China

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***Abstract* – In this paper we describe the development of superconducting magnets for high-field Magnetic Resonance Imaging (MRI) by various businesses and institutions in China. As the Chinese MR market rapidly expands, many foreign and domestic companies and research institutions are joining the race to meet this burgeoning demand by developing key MRI components for various magnetic field configurations. After providing a brief introduction to research on superconducting magnets dating back to the 1980s, the first large-bore 1.5T magnet with 50-cm DSV for whole-body MRI—successfully developed and manufactured by AllTech Medical Systems in Chengdu, China—is presented and its specifications are described.**

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I. INTRODUCTION

After Kamerlingh Onnes's pioneering demonstration in 1908 that the last so-called "permanent gas" Helium could indeed be liquefied, his follow-up discovery of superconductivity in 1911 introduced the world to zero electrical current resistivity. It was theorized that one could go beyond the resistive limit of a copper wire to develop a superconductor that could carry any amount of current but without ohmic loss. Although strides were made with this theory, the small critical field H_{c1} and the limited current density J_c of early superconducting materials stood in the way of practical applications in the early years. It wasn't until the 1960s—when practical superconducting materials that could support reasonably high magnetic fields and currents (such as NbTi and Nb_3Sn) were discovered and wire long enough for practical magnet winding were manufactured—that applied superconductivity was successfully implemented [1].

During the early era of cryogenic engineering in China, air separation, small-scale H_2 and helium liquefaction as well as their military and civil applications became the primary focus. In response to the expeditious development of applied superconductivity all over the world, Chinese scientists began in earnest to conduct research on the subject in the early 60s, which eventually produced a NbTi superconducting wire. The first LT

NbTi coil with a single-core superconductor in a copper matrix was built in 1965, and, lastly, the manufacturing of wire with multi-filaments began in the late 60s [2]. Even during China's Cultural Revolution period, Chinese scientists were able to make progress on the subject—provided the political environment allowed it and facilities were available—and the effort to engineering a superconducting magnetic energy storage (SMES) at the 100 kJ-level was successful [3].

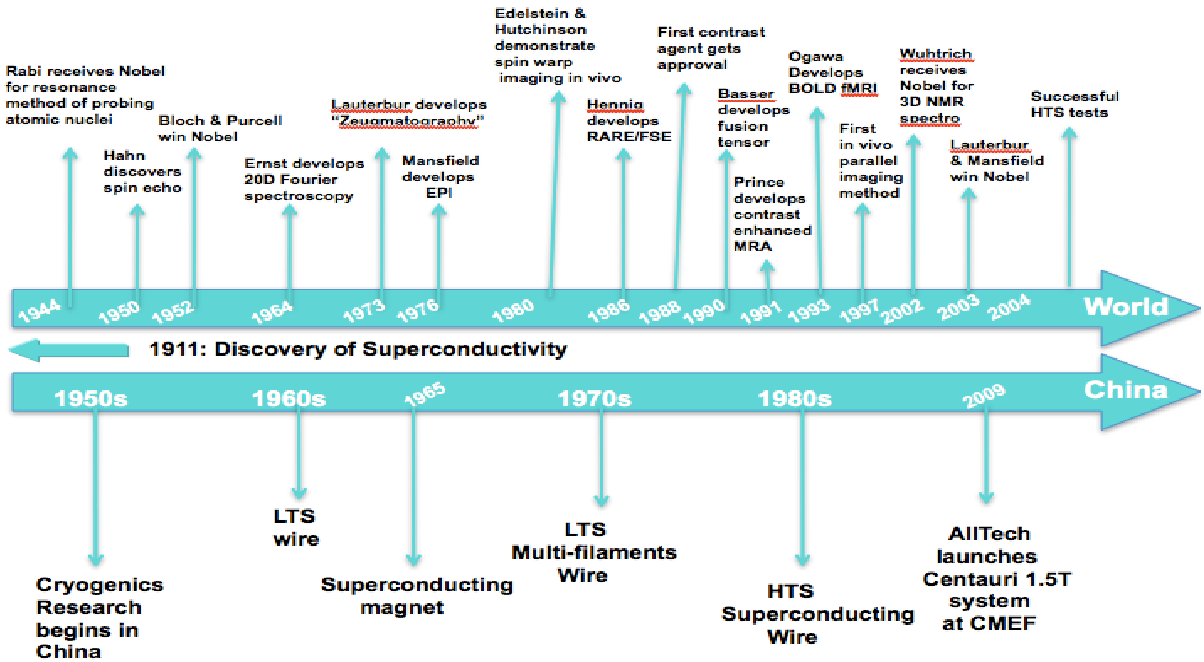

Fig. 1. MRI milestones in the global market and in China

Chinese scientists' efforts in high-temperature superconductivity took off in the tumultuous year of its discovery, 1987, and mainly focused on bulk materials, thin-film fabrication methods and electronics, physics of superconductivity and new materials. Projects such as superconducting magnets, accelerators, SMES units, superconducting generators, high-energy physics fusion research devices, magnetohydro-dynamic-propelled power generators, magnetic separators, and, finally, MRI were all researched by different institutions in China. The large-scale scientific project for fusion engineering, EAST (Experimental Advanced Superconducting Tokamak), began in 2000 and became fully operational in 2006 at the Hefei Institute of Plasma Physics. China also participated in international cooperative projects that included ITER (International Thermonuclear Experimental Reactor) and Alpha Magnetic Spectroscopy. Currently, a 40 T hybrid magnet consisting of resistive insert parts (providing 29T) and superconducting coils made of Nb_3Sn and NbTi (providing 11 T field on a 32-mm bore) is under construction at Academia Sinica's High-Magnet Field Laboratory and will be completed by 2013 [3,4].

II. SUPERCONDUCTING MAGNET FOR MRI

Magnetic Resonance Imaging (MRI) is established on the concept of nuclear magnetic resonance, where an atomic nucleus placed in a magnetic field absorbs the energy of radio waves at a specific frequency. Building on the combination of radio frequency coils and main magnets, MRI provides images of the distribution of hydrogen nuclei (protons) in the human body. Consequently, it has become a vital diagnostic tool for earlier, more accurate detection and analysis of diseases through non-invasive external magnetic fields. Continuing technological improvements have led to the development of MRI systems that have proven progressively more effective in screening for cancer, angiography and neurosurgical planning [5].

The imaging resolution is proportional to the field strength. The magnetic field strength of the MRI magnets currently and widely used in clinical settings is 1.5 tesla. When this strength is significantly increased, observation of finer body structures and analysis of the body compositions of significantly smaller forms becomes possible [6]. Since then, throughout the world MRI has become an indispensable medical tool to *in vivo* imaging, making it perhaps the greatest invention since the X-ray machine, which allowed the first diagnostic glimpse inside the human body. Unlike X-rays, which carry the risk of damaging human cells, MRI allows the physician to safely view the inside of the human body in 3D, with high-resolution, soft-tissue contrast and functional process and imaging.

Fig. 2. 1.5T MRI market in China

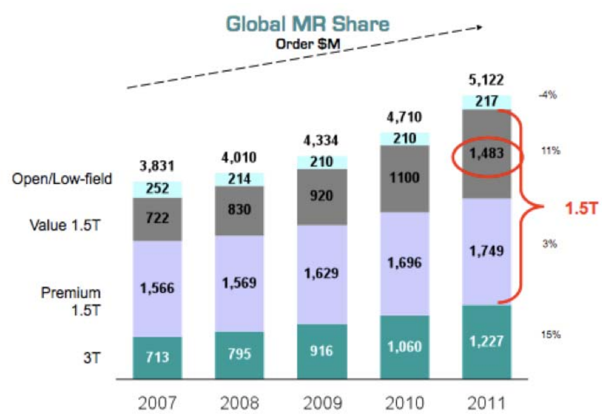

Fig. 3. 1.5T MRI global market

As a result of the MRI revolution in diagnostics, the global marketing picture is promising. In 2010, there were already more than 30,000 superconducting MRI systems installed worldwide, with over 2,600 systems built annually—a number which has increased more than 6 percent each year. It is estimated that a half-million scans are conducted every day around the world, up from roughly 300,000 in 2009 [7]. In consequence of this remarkable imaging activity, within the next five years the MRI equipment market is predicted to reach \$7.4 billion, thanks to a swell in clinical applications including neural, cardiovascular and breast imaging. Increasingly sophisticated high-field systems and innovations such as functional MRI and open MRI are fueling this major technological uptake both within China and foreign markets, as demonstrated in Figure 2 and Figure 3, respectively. One business report calculates that the Asia-Pacific market for closed MRI systems, specifically, will pass $687.6 million USD in value by 2014 [5]. The global MRI market is currently dominated by Siemens (SI), Philips (PHG), and General Electric Healthcare (GE), which collectively represent more than a three-quarter share of the market.

In response to financial backing by the Chinese government in the exploration of this new science's application, the strength and sophistication of the MRI main magnet has increased rapidly. In 1990, Kejian unveiled China's premier 0.6 T superconducting-magnet MRI with a uniformity of 14.93-ppm on a 50-cm DSV (diameter of the spherical volume of imaging) that consumed less than 0.5 L/h of liquid helium. In April of '92, the first system was installed. Later, scientists studied the development of a 1T model with both liquid nitrogen and liquid helium coolant-cooled magnet and liquid helium consumption of less than 0.2 liter per hour. This was the beginning attempt by a Chinese company to carve a place in MRI developmental history.

Several Chinese corporations and institutions went on to build superconducting magnets in the late 90s but were not successful for a variety of technical reasons. No further Chinese research activity on the MRI magnet front was carried out from 2002 to 2006. Then in 2007, a small bore 1.5 T MRI system boasting an imaging size of 30-cm in the Z direction—dubbed the Magnetom Essenza—was jointly built by Siemens and Mindt in Shenzhen, China. Following closely on its heels, in January of 2009 AllTech Medical Systems (AMS) developed the first large bore 50-cm DSV 1.5 T whole body superconducting magnet in Chengdu, China.

III. SUPERCONDUCTING MAGNETS FOR MRI: REQUIREMENTS

As aforementioned, the US, Europe and Japan collectively account for a prominent portion of the global MRI market. An increased need for dependable diagnosis prompts the demand for superior data images, while higher field strength magnets, improved image quality, and development of new techniques are driving the increased use of MRI images for evaluating the brain's complex structures. Similarly, an increase is expected in the use of interventional MRI for brain surgeries, as well as the use of MRI diffusion imaging in the diagnosis of strokes and brain injuries. The demand for better and better imaging has translated into the development of main magnets with higher field strengths. Among installed systems, more than 80 percent of the systems are 1.5 T machines, but there is now a significant increase in the rate of installation of 3 T systems. It is expected by 2015 that the 3 T model will become the mainstream product. In view of the development of newer and more powerful MRI systems, attention is now turned to desiderata for tomorrow's magnets. It is also worth noting that early MRI systems had only main coils to generate the primary field inside the coil bore, as the stray field was typically very large and required a correspondingly large installation room. Eventually, magnets with iron room shielding, iron yoke shielding, and passive shielding were added to the systems. Although these additions restricted the stray field to a certain area, they also added to the system room cost. Since 1990, most MRI systems use active shielding. Although they add to the magnet's complexity and cost, they greatly increase the flexibility of the installation of the MRI system.

The essential requirements for an MRI magnet are as follows:

- Field strength: The scanning speed and imaging resolution are proportional to this value because of the increase in signal-to-noise. This is the reason higher and higher magnetic field strengths are considered, despite their technological challenges.
- Imaging volume: The larger the imaging region, the better. This does require a trade-off owing to an increased cost.
- Field uniformity over the imaging volume: The greater the uniformity, the better. Non-uniformities should be due to the patient's image and the gradient coil used to reconstruct that image. The usual requirement is 5 ppm-15 ppm (parts per million) depending on the body region of interest being scanned.
- Field drift magnet persistence: The requirement here is less than 0.1 ppm per hour. This is a measure of the field stabilization.
- Loss of liquid helium: The "boil-off" of liquid helium must be less than 0.1L per hour. Currently, there is almost no loss throughout most of the main magnet systems, as the helium is recaptured through the 4K cryo-cooler re-condensing capability.
- External field shielding: The key here is the boundary outside of the magnet representing where the field has dropped off to 5 Gauss. This is usually described by a 5G line in its illustration.
- Diameter of the patient cylindrical access: This so-called bore size is usually 60 cm. However, 70 cm is increasingly the target to be used in North America, in view of the larger range of patient sizes.
- Length: The shorter, the better. The problem of claustrophobia has been a long-standing issue in the area of patient comfort.
- Weight: The limit on the weight of the magnet will depend upon the site conditions, the installation requirements, and the transportation capabilities.

Now higher and higher-field MRI systems are being put into application: 7 T, 9.4 T, and 11.7 T whole body MRI are all being developed and used for research. Clinical applications will not be far behind.

With economic developments as an added incentive, the Chinese government continues its concentration on initiatives aimed at improving healthcare infrastructure. In this regard, the MRI market holds enormous growth potential, with the preponderance of the demand expected from private hospitals and individual imaging facilities, as opposed to public hospitals. Increased instances of neurological, oncological, and cardiac diseases seen recently in China provide demand for developing the region's MRI equipment market, as well. Furthermore, the rise in specialized medical professionals—in addition to the constant upgrade of hospital services—additionally drives market growth.

IV. ALLTECH MEDICAL SYSTEMS MAGNETS

A. Magnet Field Profile: 1.5 T Magnet Test Line of AllTech Medical Systems

AMS's Centauri system features a highly homogeneous main magnet, the 16-Channel Orion RF coil system, high-performance gradient coils, and advanced MRI clinical applications that deliver high-resolution, real-time scanning. Parameters for the 1.5 T MRI magnet are listed in Table I. Figure 4 shows the magnet field profile of the 1.5 T magnet and Figure 5 illustrates the mechanical stress analysis for the coil former resulting from magnetic field forces.

Table I. Parameters of AllTech's 1.5 T MRI Magnet (note: ppm = parts per million)

Parameter	Value
Central Field	1.5 T
Frequency	63.75 MHz
Shield	Active+TeD
Shim Type	Passive
Drift	≤ 0.1 ppm/hour
50cm DSV (diameter of spherical volume)	≤ 10 peak-to-peak ppm
45cm DSV	≤ 1 VRMS
40cm DSV	≤ 0.2 VRMS
Weight	5400 kg including the gradient system
Length	172 cm
Stray Field 5 Gauss boundary	2.75 m radially and 3.80 m axially
Helium Consumption	< 75 cc/hour [Typical]

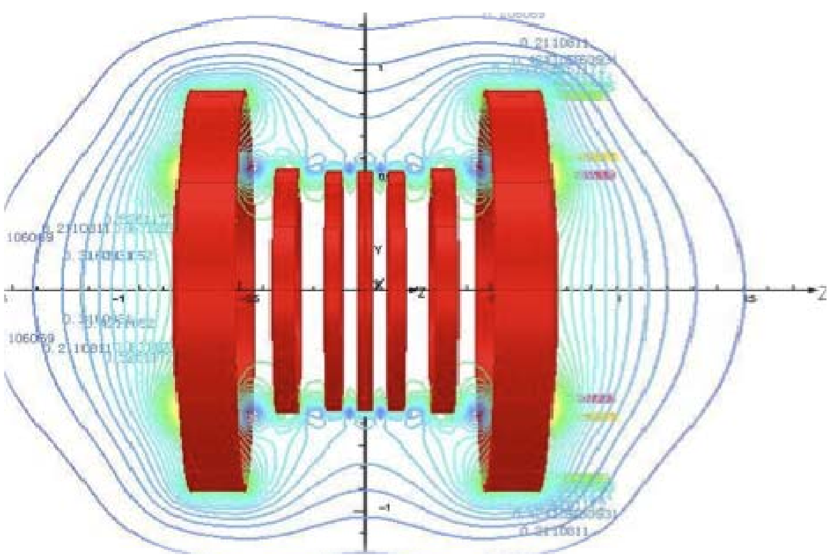

Fig. 4. 1.5 T Magnet field profile

B. Magnetic Field and Magnetic Force

In order to create a clear picture, the magnet must boast exceptionally high homogeneity over the imaging volume. In whole-body magnets, this "zone of homogeneity" is typically defined by a 50-cm diameter for the imaging volume. The degree of non-uniformity of the field over this volume is restricted to only a few parts per million. High-efficiency optimization methods are used for the 1.5 T magnetic coil's design with constraints on magnetic stress, wire turns, and coil dimension.

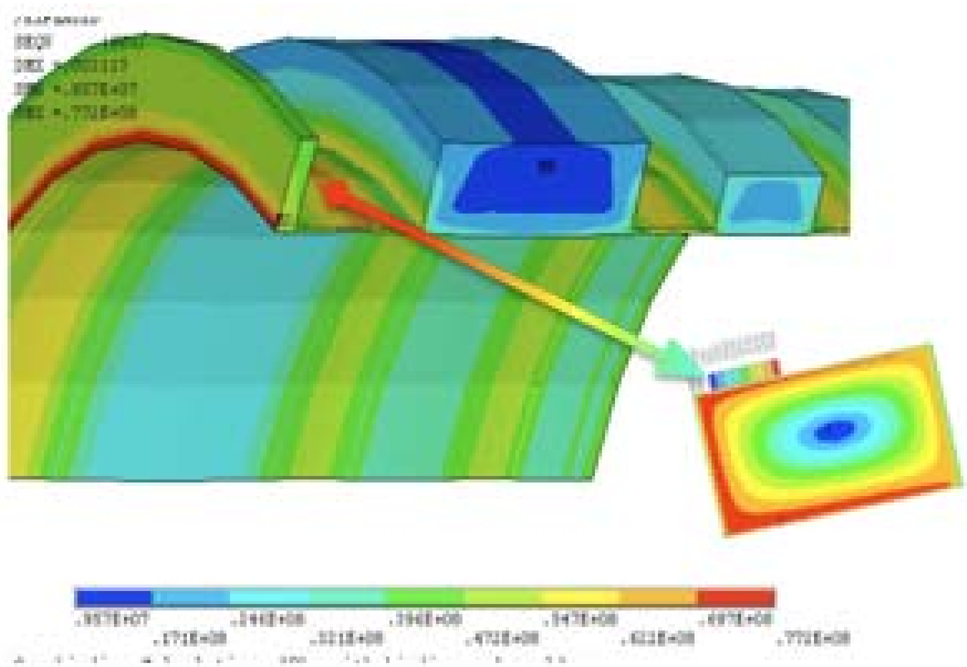

Fig. 5: Mechanical stress analysis and magnetic field

In view of the high-current density and the high magnetic field of the modern main-magnet coil, care must be taken that the magnetic forces and stresses are not too large. Finite element methods are introduced to analyze the limits on these forces, noting that magnetic forces on the end coil larger than 100 tons in the radial or axial direction will cause a deformation of the former that is greater than 1 mm.

Table II. Estimates of Magnetic Forces According to the Coil Position

Coil	Radial force in tons	Axial force in tons
Coil 1	14	-17
Coil 2	38	-6
Coil 3	54	-16
Coil 4	113	-93
Coil 5	219	26

C. Quenching and Its Prevention

Anomalous termination of magnet operation occurs in the abrupt change into the resistive state of the superconducting coil and is called "quenching." Common causes resulting in a sudden transition to the quenching include an excessively large field within the magnet, a rapidly increasing rate of change of this field, or both. In rare cases, a defect in the magnet triggers rapid Joule heating, increasing the temperature of the encompassing regions, which are then forced into the resistive state. That change triggers a chain reaction of additional heating, and so on, until the magnet becomes completely quenched. Such a rapid transition is marked by a big, characteristic "bang," as the energy converts to heat, and the cryogenic fluid boils off. An accompanying, sudden decrease in current may cause spikes in the voltage, so while permanent damage to the magnet itself may be rare, localized heating and large mechanical forces pose a risk to its components as a result of the quench.

Quenching remains very difficult to eliminate completely, especially during the ramping training period when there are several mega joules of energy stored in the magnet and thousands of volts distributed throughout the layers. Accordingly, it is all too easy to break the insulation and potentially burn the magnet. The quench protection circuit is designed to protect the coil of the magnet and, in accordance with the IEC requirement, the magnetic field must be decreased to smaller than 200 gauss in no more than 20 seconds. Diode and resistance bridge networks are used for controlling the protection circuit, while quench heaters are used to avoid heat concentration. As an example, a full-field quench test has been performed in an AllTech environment, where the field has been shown to be decreased to 200 gauss in less than 15 seconds. Figure 6 shows the Centauri 1.5 T magnet quench trace when EMMU acts by CE test qualification.

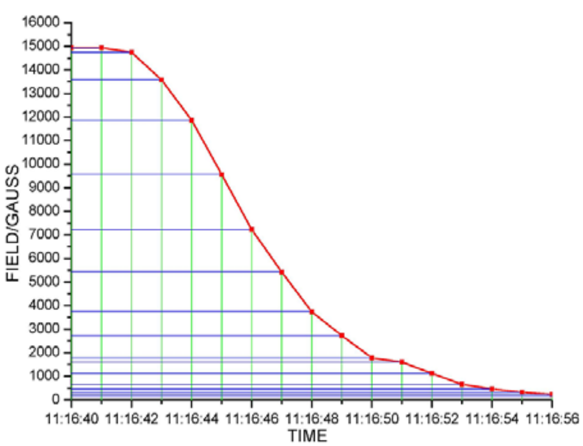

Fig. 6. Magnet emergency quench test

Fig. 7. Scanned imaging samples by Centauri 1.5T

D. Winding Technology

Because of the danger of quenching, quality coil winding technology is essential for the magnet. If there is uncontrolled movement of the superconducting wire, it will cause a premature quench, and may never reach the targeted performance after several successive quenches. Through the development of applied superconductivity, two methods have been introduced for winding superconducting wire: wet winding followed by impregnation and dry winding.

E. Wet winding and Vacuum Pressure Impregnation (VPI)

The coil is cured by epoxy or wax to constrain the movement of wire. However, this technique was not completely successful when it was first implemented, with premature quenches still persisting in these impregnated magnets. From many tests, it has been found that at the rigid corner of the coil there is a stress concentration that will cause the fracture of the epoxy or wax during cool-down and charging. This is a major factor for premature quenches. To solve this problem, a "floating coil" was introduced to avoid the stress concentration between the epoxy and the former. As a technical development, a high-efficiency technology for wet-winding was thereby invented.

F. Dry Winding

In dry winding, there is no epoxy or wax used during the wire-winding process. In operation, there can be movement between the coil and former, as well as between the layers. Comparing the wire movement in the coil produced by wet winding with the result from dry winding, an attempt to control the coil motion is easier said than done. Ten micrometers of movement by the wire may trigger the magnet to quench, which, as noted above, describes the warming of the superconducting magnets above a critical temperature. Operators must take care to control the floating point and allow the magnet to ramp up without quenching.

G. Shimming

The 1.5 T AllTech magnet has a relatively good uniformity, even prior to the shimming step, due to strict manufacturing tolerance control, with a bare non-uniformity that varies from 300 ppm to 1000 ppm over 50 DSV [8]. A novel shimming method is used for easier, safer, faster and more uniform shimming. On-site shimming for most systems can be finished with half-days reaching a final figure of 6 ppm.

H. Cryostat

Table III illustrates the gradual development of the cryostat for the MRI system's superconducting magnet. The development has resulted in a steady decrease in the consumption of liquid helium each year up until 2000, when zero liquid helium (LHe) consumption was achieved. Now and in the future no liquid helium loss is anticipated for the cryostat if no magnet quench occurs. Unfortunately, this possibility and the associated with it major refill expense cannot be fully excluded (see also Section V.D. below).

Table III. Evolution of Liquid Helium Consumption and Service by MRI Magnet Cryostats

Year	Cryogen	Cryo-cooler	Boil-off	Service Period	Notes
Pre-1990	LHe + LN_2	With or W/O	500 cc/hr	2-3 months	Difficult to build
1990	LHe	10 K GM 2 stage	40-100 cc/hr	6-12	LHe dependent
2000	LHe	4 K GM/PT	~0	24	Less LHe
2010	None	4 K GM/PT conduction	No LHe	>30	Like a home appliance

V. FURTHER DEVELOPMENTS

A. The 3 T Magnet

The signal-to-noise ratio is roughly proportional to the field's strength, which accounts for the development by competing manufacturers of 3.0 T whole body magnets. AllTech Medical Systems has a 3.0 T whole body magnet currently in development, the parameters for which are displayed in Table IV.

Table IV. Parameters for AllTech Medical Systems 3.0 T magnet

Parameter	Value
Central Field	3.0 T
Frequency	127.7 MHz
Shield	Active + TeD
Shim Type	Active + Passive
Drift	≤ 0.1 ppm per hour
45 cm DSV nonuniformity	< 10 ppm
42 cm DSV nonuniformity	≤ 1 ppm
40 cm DSV nonuniformity	≤ 0.1 ppm
Weight	< 7800 kg
Length	< 169 cm
Bore Size	90 cm
Stray Field Boundary (R radial × Z axial)	3.2 m × 4.8 m
Helium Consumption	ZBO (zero boil off)

B. The 7 T Test Magnet

In addition to 3 T whole-body imaging, there is great interest and progress in developing 7 T systems. AllTech Medical Systems has directed their initial 7 T development toward animal imaging. The parameters for their system are listed in Table V.

Table V. Parameters for AllTech Medical Systems' 7.0 T Test Magnet

Parameter	Value
Central Field	7.0 T
Frequency	298.032 MHz
Shield	Active + TeD
Shim Type	Active + Passive
Drift	≤ 0.1 ppm per hour
120mm DSV	≤ 10 ppm
100mm DSV	≤ 1 ppm
Weight	< 3000 kg
Length	< 100 cm
Bore Size	210 mm
Stray Field (R radial ×Z axial)	1.6 m × 2.2 m
Helium Consumption	ZBO (zero boil off)

Progressively improved advancements such as higher field strengths and new clinical applications have further stimulated the market growth of MRI systems. Experiments with higher-field systems such as 7T MRI for human research have brought within reach the ultra-high resolution of fMRI that was previously only available for animal studies. Soon this advanced technology will allow observers to study neuronal function at the sub-millimeter scale, with some possible clinical applications including the in-depth analysis of neurodegenerative diseases such as Alzheimer's.

Additional innovations include such features as an upright, multi-position version of the MRI system as well as explorations into superconducting material alternatives for background field coil windings.

C. Impact of Higher Temperature Superconductors

Currently, superconducting MRI systems use niobium titanium (NbTi) composite wire. The NbTi is a metal alloy superconductor that depends on liquid helium for its bath cooling, consequently increasing the size and cost of the total system [9]. To simplify designs and decrease the cost of production, engineers have been experimenting with substituting superconductor wires that can operate at higher temperatures (HTS such as BSCCO and YBCO) as well as the recently discovered material magnesium diboride (MgB_2) wire. The technical properties, present cost and manufacturing characteristics of superconducting wire materials are compared in Table VI.

Unfortunately, at this time all the superconductors (Nb_3Sn, BSCCO, and YBCO) cost more to operate at a given temperature and field than NbTi, which is presently used in full-body MRIs. For HTS the conductor cost increase is by two orders of magnitude. The first-generation BSCCO composite (Bi2223, bismuth strontium calcium copper oxide) relies on a batch process and considerable amounts of silver, creating additional cost issues. The second generation YBCO tape conductors present a slightly better solution by using cheaper raw materials and operating better at high magnetic field and current levels. However, the expense of vapor deposition equipment results in much higher composite conductor costs. It is hoped that in the longer term the cost of YBCO superconductors will decrease such that they can be operated more economically.

Table VI. Higher Temperature Superconductors

Wire Type	J_c/A/mm^2 at 77K	J_c/A/mm^2 at 4.2K-4T	T_c/K	B_c/T	Wire Length (m)	N factor	2011 US$/KA-m	Heat Treatment	Persistent *I* Joint
NbTi		>3000	9.3	10	>10000	80	1@4.2K-4T	No	Easy
MgB_2		>2000	39	8	>5000	40	6@4.2K-4T	Yes	Difficult
Nb_3Sn		>8000	18	24	>5000	30	5@4.2K-4T	Yes	Difficult
Bi2223	>1000	>2000	>77K	>20	>2000	20-30	>150@ 77K self field	Yes	Difficult
YBCO	>20000	>40000	>77K	>20	>1000	20-30	>200@77K self field	No	Difficult

In the short term, therefore, the best cost and performance compromise is the MgB_2 superconducting composite wire. It can operate at temperatures from 4 to 30K in reasonable magnetic fields while enabling the operation of conduction-cooled background MRI magnets in the 4 to 12K range. It is expected that in one to two years, the MgB_2 conductors could be manufactured at prices close to those of NbTi wires. Furthermore, the use of MgB_2 wires may partly circumvent the impending issue of helium shortage and increased life-cycle cost to the public, as discussed below.

D. Liquid Helium Shortage and Possible Remedy

The current major concern of MRI producers is the predicted shortage of liquid helium for the existing full-body MRIs using NbTi superconductors that require bath cooling. From 2006 to 2011, the cost of LHe has tripled worldwide and is expected to continue increasing. Moreover, global demand is expected to exceed supply by 2017. Customarily, most full-body MRIs are filled with liquid helium at the factory and then shipped while filled with liquid helium. The systems have re-condensing systems that result in negligible loss of helium during normal operation. The problem, however, arises when the magnet quenches, and all the helium is dumped. A typical MRI requires 2000 to 3000 liters of the liquid He to cool down the MRI and refill the reservoir. Presently, in most parts of the world liquid helium costs US$ 11 to 16 per liter. In some parts of the world the cost is already as high as US$ 25 to 30 per liter, resulting in costs between $50,000 and 90,000 merely to refill the machine. The refill may cost anywhere between $150,000 and $270,000 when liquid helium cost triples again in the next 5 years,. Consequently, the overall cost of existing system maintenance may increase at tremendous expense to the public, and the use of MRI might be curtailed for many regions.

To eliminate the need for liquid helium bath cooling and thus control the cost, we aim for a temperature margin of just a few extra degrees, in the 4 to 12 K range. At present, NbTi allows only a 2-K temperature margin (4 to 6K). However, by using a higher temperature superconductor such as MgB_2 that can operate in the 4 to 12K temperature range, conduction-cooled background magnets can be developed. These background field magnets may be cooled using the same types of refrigerators that are currently used for re-condensing the liquid helium.

By doing away with the liquid helium bath, we obtain additional advantages that ultimately should lower the cost. First, the cryostat no longer needs to be designed to meet the ASME pressure vessel code, a process that incurs extra expense. Second, the MRI magnets will no longer need to be air-shipped around the world while filled with liquid helium. Third, the hospital will no longer need to design and install a helium venting system. And finally, hospitals will no longer need to pay the price of liquid helium refills after a quench. In addition, the absence of LHe will make the magnets safer.

Currently, for new types of MRI systems, there are three driving demands: for smaller, less complicated systems, for a less confining structure, and for decreased diagnostic costs for patients (and, correspondingly, their insurance carriers). All three requirements can be met simultaneously by constructing more compact, less complex and potentially less expensive superconducting systems that are liquid-helium-bath-free. Therefore, MgB_2 conductively cooled magnets, which can be cooled with an efficiently-sized cryocooler (as opposed to large LHe compartment and connected re-condensing refrigerator) may enable the expansion of the MRI market and contribute to the reduction of LHe shortage.

VI. CONCLUSIONS

After 30 years of development and the efforts of several generations, remarkable progress has been attained in MRI superconducting magnets. Twenty years ago, China's Kejian was the first to attempt developing superconducting magnet at 0.6 T in China; two decades later, we (AllTech Medical Systems) are on the path to engineer the most advanced magnets in China while incorporating the latest innovations and improvements from all over the world. The development of superconducting magnets in China will support the ongoing evolution of diagnostic imaging, thereby promoting patient welfare worldwide.

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