MRI main field magnets

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Introduction

This paper presents an overview of the technologies used for generating the main magnetic field in whole-body MRI scanners.

Classification of MRI magnets

MRI scanners come in various shapes and they are based on several distinctly different technologies to create the static background field (B_0). Whatever the shape and the technology, all B_0 magnets for MRI have in common that they generate a strong field, which has, within narrow tolerances, the same value over a large volume and is very stable in time. The typical size of the imaging volume is a ellipsoid with 300-500 mm axis length, over which the field variations are less than a few ppm. The amplitude of time-dependent field variations is typically (far) less than 100 nT.

MRI magnets can be classified in various ways:

1. by their basic shape (open, with the field oriented perpendicular to head-feet direction of the patient or cylindrical/closed bore, with the field along the head-feet direction)
2. by the way the field is generated (by currents in superconducting or resistive coils or by permanent magnet material)
3. by the use of iron for flux return and/or magnetic field shaping

The table below shows the combinations found in commercial MRI scanners:

	Cylindrical/field direction // h-f	Open, field direction ⊥ h-f
Superconducting electromagnet	Nearly all high-field systems $B_0 \geq 1.0$ T (Fig. 1a) Iron shield if $B_0 \geq 7.0$ T	High-field open systems $B_0 \geq 0.6$ T (Fig. 1b) May use iron
Permanent magnet	Do not exist	Most low-field open systems $B_0 \leq 0.4$ T(Fig. 1c) Iron flux return yoke
Resistive electromagnet	No longer used	Some low-field open systems $B_0 \leq 0.4$ T (Fig. 1c) Iron flux return yoke

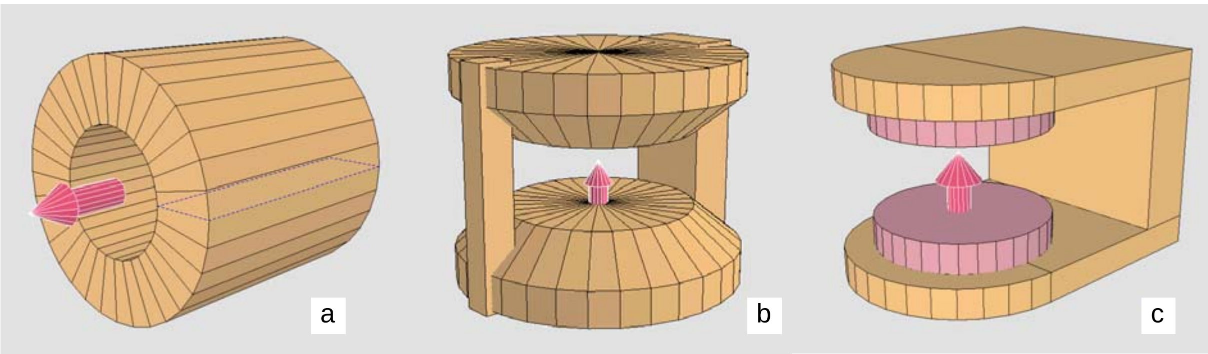


Figure 1: examples of MRI magnets: a) cylindrical, superconducting, no iron, b) high-field open, superconducting, c) low-field open, permanent magnet or resistive electromagnet, iron return yoke.

In the following sections, the types of magnets introduced here will be described in more detail. As cylindrical superconducting magnets account for more than 75% of the MRI installed base, this kind of magnet will receive most attention.

Cylindrical superconducting MRI magnets

In a cylindrical MRI magnet the windings generating the field are located on a cylindrical structure which surrounds the imaging volume. If the coil would be a simple solenoid with constant winding density along its length, the field would only be uniform if the magnet were infinitely long. If the length is truncated to the typical length of an MRI magnet, the field will behave as is shown in figure 2. At first sight, the field inside the coil looks pretty uniform, dropping to 50% at the ends of the coil (fig. 2b). The same field plot with the plotting range set to ± 1% of the central field reveals, however, that this field is not nearly uniform enough for use in an MRI scanner (fig. 2c).

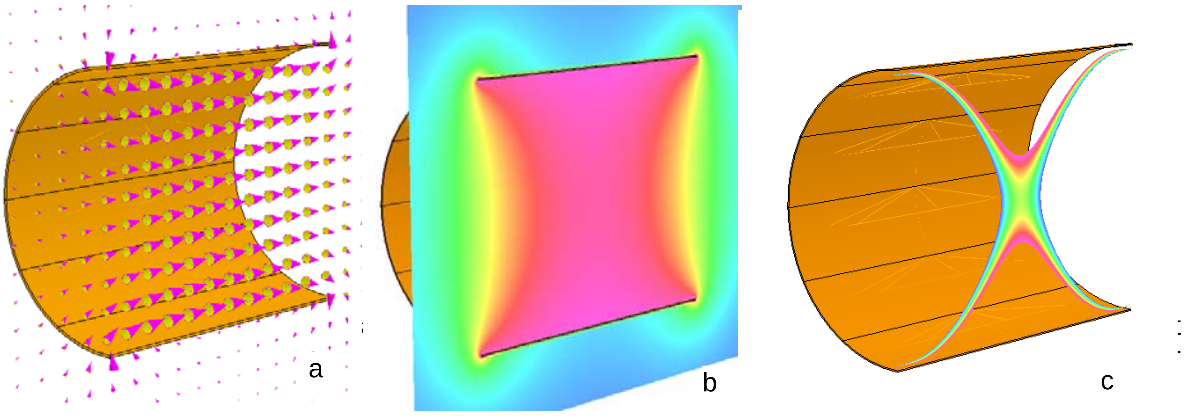


Figure 2. Field of a simple solenoidal magnet. a) field vectors, b) contour map, c) contour map of central field ± 1%.

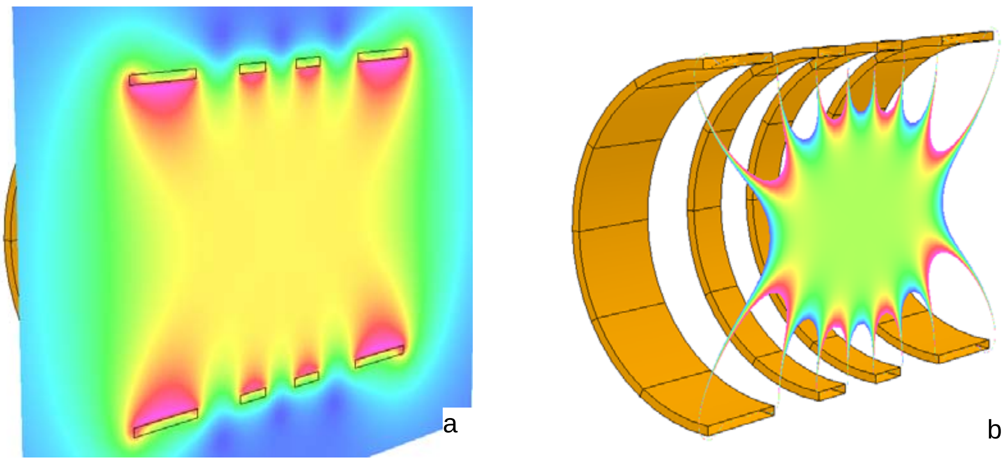


Figure 3: four-section magnet with better homogeneity. a) field range 0-max, b) central field ± 1%.

In order to make the field more uniform the current density must be profiled along the length of the magnet. In practice, this is generally done by cutting up the magnet into a number of discrete coil sections, with gaps in between. The position and number of ampere-turns in each section can then be varied to control the shape of the magnetic field [1]. A simple example of such segmentation is shown in figure 3. The region over which the field differs less than 1% from the central field has become much larger. Note that starting from the iso-centre, there are directions in which the field increases (red zones), separated by directions in which the field decreases (blue zones). Field maxima can be found in the vicinity of coil sections, minima are located towards the gaps between coil sections and along the axis of the magnet. Such patterns with a saw tooth-like boundary of the imaging volume are typical of all MRI magnets. They are a direct consequence of the fact that magnetic fields in free space have to obey Maxwell's equations. Note that the field pattern has rotational symmetry about the axis of the coil system (usually called the z-axis). Figure 4 shows a typical example of a high-field magnet such as they are now used in clinical systems.

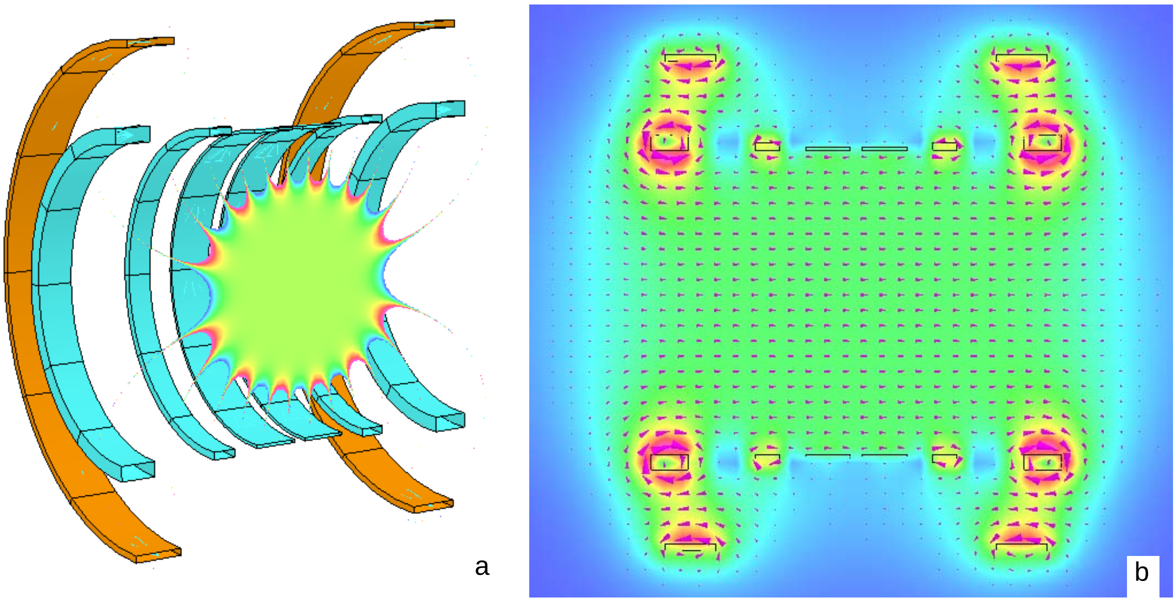


Figure 4. Typical actively-shielded high-field MRI magnet. a) coils and central field, b) active shielding

In addition to the 6-section main coil, this magnet features two additional coil sections on a larger diameter. The current in these coils runs in reverse direction and the main purpose of these coils is to reduce the magnetic field on the outside of the magnet (active shielding). Figure 4b shows how the field coming out of the bore is returned via the annular gap between the outer main sections and the shield windings. Without these shield coils, the magnetic field would have to be shielded by a heavy iron flux return structure (typical mass 15 tonnes for a 1.5 tesla magnet).

Figure 4 also shows that the highest field is located close to the coils (red zones). For typical compact MRI magnets, this peak field can be more than a factor of 3 higher than the central field. The peak field increases when the sections of the magnet are made short and thick (which is one of the things to do to make a very compact magnet). A large field on the conductor is problematic for two reasons. Firstly, the amount of current that can be carried by a piece of superconducting wire decreases with increasing field. Especially at conductor fields above 6 tesla, this reduction would lead to a dramatic increase in the amount of wire required to generate a certain central field. Secondly, the magnetic forces on the wire are directly proportional to the field, so high peak fields imply high mechanical stresses. At the ends of the outer sections of the main coil the field is mainly radially directed, leading to a large net force on this coil towards the midplane of the magnet. The windings on the inside of the shield sections experience a large axial fields; the resulting Lorentz force tends to increase the diameter of the coil, and it generally requires careful mechanical design to make sure that these coils

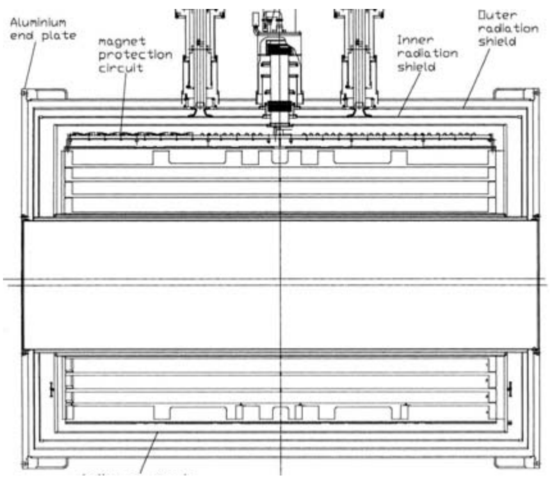


Figure 5. Very high field whole body MRI magnet

do not break. At 7 tesla and above, the peak fields and stresses of an actively shielded design like that of figure 4 would become unmanageable. In these magnets, the high field parts of the coil are long solenoids, extending over the entire length of the magnet. Figure 5 shows an example of such a very high field magnet [2]. There are no active stray field compensation coils; the external field is contained by an extremely heavy iron wall surrounding the system.

Mechanical and cryogenic structure

The essential parts of the mechanical structure of a cylindrical MRI magnet are shown schematically in figure 6. The superconducting coil is enclosed in a vacuum tight vessel, which is filled with liquid helium (grey coloured). The inner wall of this tank has a number of accurately machined circumferential grooves, which hold the coil sections in position. This coil support has to be very stiff since it has to react the large magnetic forces between the coil sections. In some magnet designs the coil support is a separate part, not integrated with the helium container. The helium tank is surrounded by a second air-tight vessel, called the outer vacuum container (red). The outer surface of this tank is the wall of the magnet as seen by the user. The space between the two walls is evacuated in order to inhibit heat transfer through gas conduction. The cold mass (helium tank and coils) is held in position by a number of thin support elements, usually made from glass reinforced plastic. The dominant heat transfer mechanism from the warm surface of the outer vacuum tank to the cold mass is radiation. In order to limit this heat load, there are one or two additional closed shells inside the vacuum space, made from a material with a good thermal conductivity (usually aluminium). When such a shield is connected to one of the heat stations of a small cryogenic refrigerator, nearly all radiation heat intercepted by the shield is extracted from the system and only a fraction of the radiation coming from the warm wall reaches the cold mass. Radiation heat transfer is further minimized by a large number of layers of aluminized plastic foil (superinsulation), loosely located in the remaining gaps. In a two-shield magnet, the net heat load on the helium tank, causing helium to evaporate, can be as low as 30 mW. This leads to a helium loss from the magnet of about a litre per day. In recent years, cryogenic refrigerators with a cooling power of about 1 Watt at 4 K have become available. MR manufacturers are gradually adopting this technology to reduce helium boil-off to zero. With such a refrigerator installed on the magnet, a single radiation shield is sufficient. This allows a smaller clearance between the walls of the vacuum space. This advantage can be used to reduce the manufacturing cost of the magnet or to make the magnet more compact.

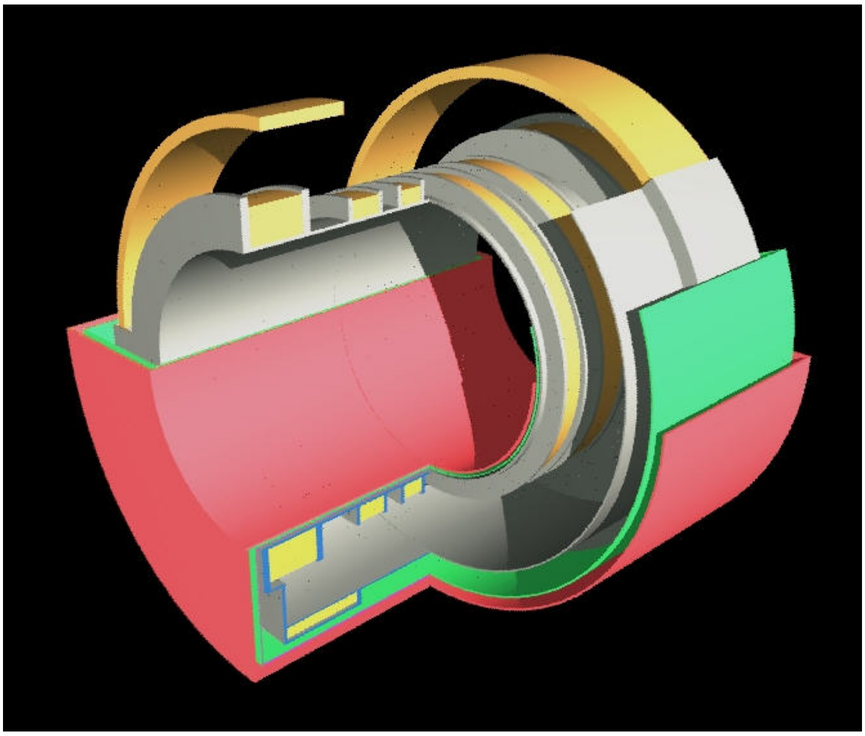


Figure 6: Cryostat structure of cylindrical MRI magnet

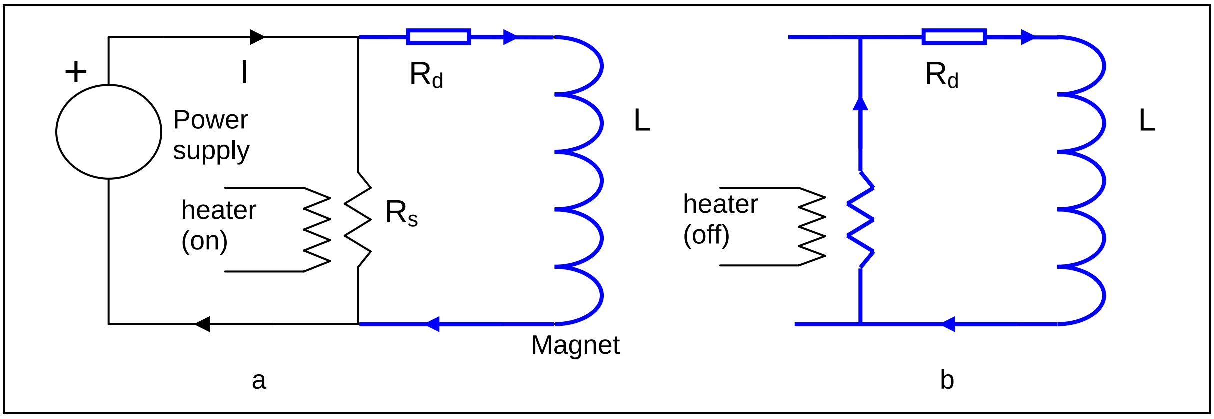


Figure 7. Magnet during charge (a) and in persistent mode operation (b).

Persistent mode operation

Superconducting MRI magnets are operated in persistent mode. The superconducting circuit in which the magnet's current is flowing is closed in itself and no external power supply is needed to sustain the current (fig 7b). In practice, there will always be some current decay due to the residual electrical resistance R_d of the magnet, which is of the order of 10^{-9} Ohm. For a magnet with an inductance of 40 Henry, the decay time-constant L/ R_d then becomes 4 10^{10} seconds, equivalent to a relative field loss of less than 0.1 ppm/hour. In order to be able to charge or discharge the magnet, a part of the superconducting circuit can be made non-superconducting by means of a heater. With the heater on, this so-called persistent mode switch has a resistance of the order of 100 Ohm. The magnet current then flows in an external circuit through a power supply. Depending on the polarity of this voltage source, the current increases or decreases. In order to limit the heat load into the helium tank during persistent mode operation, the current leads are usually retracted.

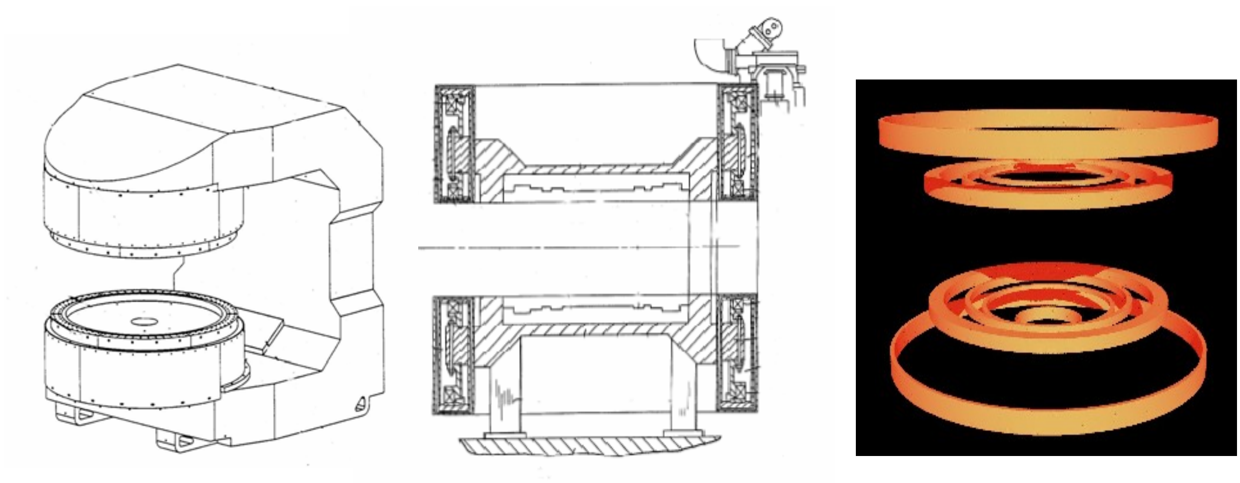
Stability and quenching

One of the major magnet engineering issues is the stability of the superconducting state of the conductor. If somewhere in the superconducting circuit of the magnet the conductor turns to the normal state, such a normal zone will, in most practical cases, grow rapidly. As the resistance of the coil becomes larger the current will rapidly decay to zero. At this point, all of the magnetic stored energy of the magnet will then have been converted into heat. This causes the rapid boil-off of a significant fraction of the helium inventory of the magnet, leading to a violent blow-out of cold gas. Such an event is called a quench. The design of the magnet must be such, that the likelihood of quenches is minimized and that, if a quench occurs, no damage is done to the magnet or to the environment. The amount of heat required to create a normal zone in the magnet is extremely small, of the order of microjoules. Such amounts of heat can be released if, under the load of the extreme magnetic forces in the magnet, a piece of conductor moves relative to adjacent conductors or relative to the coil support structure. Measures to prevent this from happening include rigidly bonding all turns of the coil and installing low-friction slip planes between windings and coil pockets. Quenches have also been attributed to events like RF interference with instrumentation wiring going into the magnet or pieces of frozen air falling onto the coil after vibration of the magnet. A quench will also be provoked when, in case of emergency, the fast magnet rundown switch is pressed. If a quench occurs, it must be avoided that all of the magnet's energy is dissipated at or near the spot where the quench started. In most large superconducting magnets, this is done by rapidly activating a system of heaters, with the objective to initiate many additional normal zones such that the energy is spread throughout the coil.

Open magnets

The first generations of open magnets were operated at very low field (< 0.1 tesla). At this field strength, it is advantageous to use a ferromagnetic yoke structure to carry the magnetic field around the system. The most common shape of this yoke is a large C (figure 8a). As long as the yoke is not magnetically saturated, it requires very little effort to magnetize it and the field generation system only has to create the field in the air gap, where the imaging volume is located. The field generation can be done either by means of a pair of electromagnets, located around the ends of the yoke, on either side of the air gap, or by layers of permanent magnet material close to the gap. The field is usually shaped by ferromagnetic disks with an accurately machined profile, facing the air gap (pole shoes).

The magnetic material commonly used for the permanent magnets of this kind is Neodymium-Boron-Iron. It is manufactured in the form of small blocks, which are stacked onto the pole structure of the magnet. The main advantage of permanent magnets is, that it does not require any energy to operate the magnet. This advantage is offset by a high cost of the material. The alternative solution, a resistive electromagnet, is cheap to build, but consumes a considerable amount of electrical power. Both types of open magnet have to deal with the problem that the strength and homogeneity of the field are affected by changes in temperature of the structure. The magnetic properties of steel and of the permanent magnet material are slightly temperature dependent and thermal expansion changes the shape of the structure. These magnets must either be used in a temperature controlled environment or the MR scanner has to be equipped with one or more field sensors to be able to compensate for temperature induced drift effects.



a b c

Figure 8. Examples of vertical field open MRI magnet configurations. a) low-field iron yoke [3], b) superconducting, ferromagnetic pole plates [4], c) superconducting, ironless, actively shielded

A practical upper limit for the strength of the field of the permanent magnet or resistive coil driven open magnets is approximately 0.4 tesla. Beyond this value, the amount of magnetic material or the power consumption of the drive coils would become prohibitively large (stronger fields can be made at the expense of patient comfort by narrowing the gap). For higher fields, open magnets have to use superconducting coil technology.

Several implementations of open magnets with superconducting coils exist in the MRI market. One approach is to retain the C-yoke flux return structure of fig. 8a and to replace the resistive drive coils by superconducting coils with an appropriate number of ampere turns. However, the amount of magnetic flux that has to be carried around via the yoke increases at least proportionately to the central field. The mass of the yoke iron needed to transport this flux exceeds 20 tonnes for fields above 0.5 tesla, which makes siting of these magnets somewhat problematic. A lighter and more compact alternative is a yokeless superconducting magnet with some degree of active stray-field compensation. The structure connecting the two halves of the magnet is not ferromagnetic; its main purpose is to keep the two parts of the magnet rigidly positioned relative to each other. The stray-field compensation coils are located at a greater distance from the midplane than the field-generating coils. Figure 8b shows an implementation of the yokeless open magnet which relies heavily on the use of iron to shape the field in the imaging volume. This approach allows a simple cryogenic coil, but keeping the ferromagnetic parts accurately positioned relative to the coils adds complexity to this system. The alternative is to shape the field in the imaging volume completely with coils, not using any iron (figure 8c).

Independent of the details of how the field is profiled and how the stray field is handled, the high-field superconducting open magnets have in common that they are larger and more complex than closed bore cylindrical magnets with the same performance.

References

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