MAGNETOM Free.Max: Access to MRI – How to Make it Big Inside and Small Outside

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MR systems have always been known to be large, heavy machines that require a complex infrastructure such as the supply of liquid helium and a highly reliable supply of electricity and cooling. While the new DryCool magnet technology has been presented in detail by Simon Calvert [1], this article will focus on how the MAGNETOM Free.Max[1] system is able to be big on the inside with the first-ever 80 cm patient bore on the market, making it at the same time one of the smallest whole-body MRI systems on the market. With a footprint of just 24 m^2, a transportation height of less than 2 meters, and a weight of only 3.2 tonnes we believe the system defines a new class of MRI systems. This paper will present a number of technical innovations, which in isolation could be seen as simple engineering tasks, but together they help to overcome long-established issues with the installation process and therefore can realize unmet customer needs. Furthermore, we would like to demonstrate how only a holistic system-perspective, which aligns all the engineering disciplines behind a common goal, is able to accomplish this.

How to make it big: Combining field strength and gradient power with new imaging techniques

MR systems with solenoid magnets have been available with 60 cm patient bores since the early 1990s. In 2004, Siemens Healthineers introduced MAGNETOM Espree, the first 1.5T system with a 70 cm bore and MAGNETOM Verio in 2007, the first 3T system with a 70 cm bore. This broadened access to MRI for growing patient groups by improving comfort, counteracting claustrophobia, and accommodating obese patients in the bore. Despite the larger bore on the 70 cm systems, the need remained for even more space in the bore for the same reasons that first triggered the development of 70 cm systems. But whereas 20 years ago, the market was able to deal with the associated increased costs of 70 cm systems versus 60 cm systems, the situation is different today. Radiology is under severe cost pressure, which calls for new ways of providing high-value imaging with improved patient access at an affordable cost.

The belief in MRI has long been that higher field strengths and gradient powers together with a high receive system channel count delivers better image quality and higher speeds. This belief still holds true but there are other ways to serve markets that require the diagnostic quality of a 1.5T system but not necessarily at exactly the same speed and contrast. A larger bore diameter is essentially what drives up the costs of MRI systems. This, in turn, reduces the accessibility of MRI to a large part of the worldwide population. The costs of the magnet (mainly the superconductive wire) increase rapidly as the size rises. Gradient coil power increases with $\sim R^5$, which would quadruple the power needed when going from a 60 to 80 cm patient bore. The only way out is to go against the grain and question existing assumptions on field strength and gradient power.

During an early prototyping phase back in 2016, a 1.5T MAGNETOM Aera system was ramped down to 0.55T and equipped with modified RF-electronics. With *in vivo* imaging, it was then possible to assess image quality and analyze the impact of different types of gradient engines.

Designed as our most compact whole-body MRI MAGNETOM Free.Max

Open	Small	Light	Low
80 cm	24 m^2	< 3.2 t	< 2 m

[1]*MAGNETOM Free.Max is pending 510(k) clearance, and is not yet commercially available in the United States.*

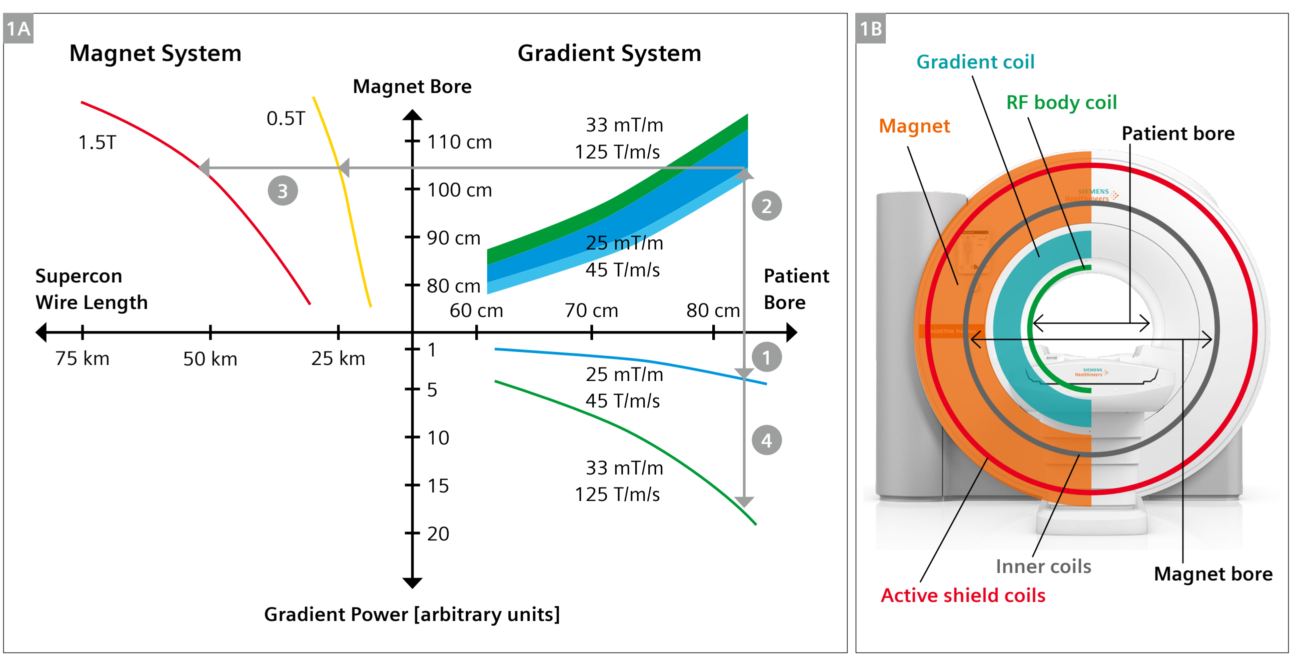
The in-house prototype and a replica of this system installed at the National Institutes of Health (NIH) helped to demonstrate that routine clinical questions in general radiology can be answered at a field strength of 0.55T [2]. Our internal analysis indicated that acceptable image quality and measurement times could be achieved with a gradient engine of 45 T/m/s and a gradient field of approximately 26 mT/m for this application field. In contrast to the 1990s, we were able to combine this MRI system with new imaging techniques that help overcome some of the drawbacks of mid-field imaging. These negative aspects had originally stimulated the design of 1.5T scanners, e.g., DeepResolve Gain and Sharp are image reconstruction methods enabling intelligent, iterative denoising using individual noise maps and an increase in image resolution using a deep neural network. These technologies can be used to reduce acquisition times and improve image quality simultaneously. At the same time, Deep Resolve can be combined with image acceleration techniques such as parallel imaging – which was not available in the 90s – and Simultaneous Multi-slice (SMS) on MAGNETOM Free.Max. In addition, Compressed Sensing has also proven to be a valuable tool for acquisition acceleration. For clinical examples, please refer to the Image Gallery [3].

The unique combination of a 0.55T scanner with various powerful new acquisition and reconstruction techniques laid the foundation for the innovation, MAGNETOM Free.Max. The reduced field strength on both the magnet and the gradient engine allowed the bore diameter to be scaled up from 60 to 80 cm, while still keeping the superconductive wire length and gradient power within a range that would make the system more affordable.

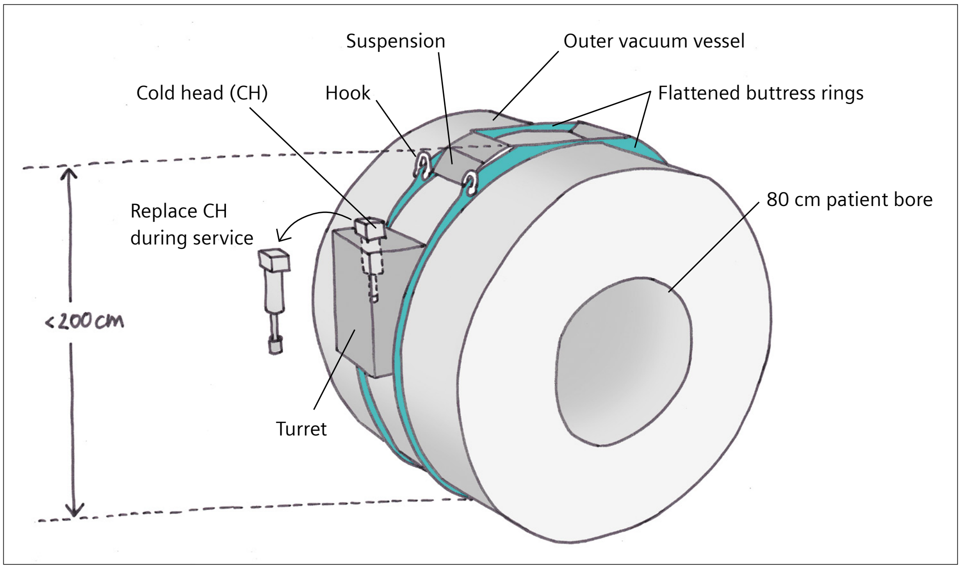
Figure 1A shows the optimization in the multiparameter space: superconductive wire length and costs increase with field strength and magnet bore diameter. When starting with an 80 cm patient bore (1), the thickness needed for the gradient coil and the body coil dictate the magnet bore diameter. For the sake of simplicity here, we assume a range of different thicknesses shown by the blue and green areas, with a slight tendency of gradient coils with higher G_{max} and SR to also require more radial space (2). With the magnet bore diameter derived from the outer diameter of the gradient coil, the impact on wire length for 0.5T and 1.5T are shown by the yellow and red curves (3), assuming similar boundary conditions on the stray field. These two curves show the huge scale of the nonlinear increase in the superconductive wire for the magnet when field strength and bore diameter are increased.

Starting from the 80 cm bore diameter, looking into the lower right quadrant (4), it becomes evident that the gradient power not only increases with higher SR and G_{max}, but it increases disproportionally with patient bore diameter.

Increased gradient power usually goes hand in hand with the additional power needed for the cooling system, which has to extract the heat from the gradient coil and the gradient power amplifier (GPA) and dissipate it in the



1 **(1A)** Scaling of superconductive wire length and gradient power with patient bore and field strength. Figures are merely illustrative to show the main correlations. Please note that the numbers in this article are also indicative to explain the physics and not related to a special design. **(1B)** When starting with an 80 cm patient bore, the thickness needed for the gradient coil and the body coil dictate the magnet bore diameter.



2 Figure 2 shows the flattened buttress rings, the outer vacuum vessel (OVC) with the shield coils, and the turret with the cold head that is moved further down on the side of the system.

air. Therefore, stronger gradients have a quadruple effect on the system design: They require more power to generate the fields, they require more cooling to extract the resulting heat and together this drives needs on local infrastructure up (connection power, space for chiller, etc.). They tend to also drive the gradient coil thickness and therefore the inner diameter (4) of magnet.

Scaling the Tx subsystem to provide sufficient B_1 amplitude on a 23 MHz system with an 80 cm patient bore is – compared with the magnet and gradient design – the easiest piece in this puzzle. Luckily, a lower Lamor frequency also requires less power to achieve the same B_1 field. Therefore, an existing 63 MHz amplifier can be tuned to 23 MHz and the additional power can be invested in overcoming the lower efficiency $\eta = B_1/P_{BC}$ of a body coil with a larger diameter. The lower conductivity of tissue at a lower frequency also makes SAR an almost negligible issue.

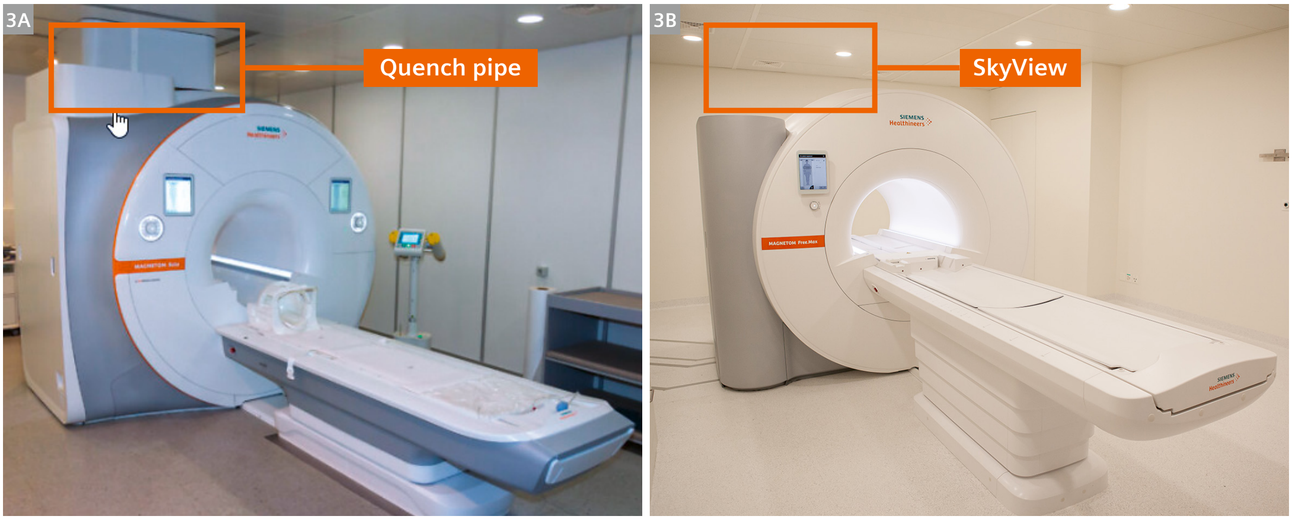
How to keep it small

While the inside of an MRI system should be as large as possible to provide space for the patient, it is more difficult to identify what the system should look like from the outside. When observing an installation process, it immediately becomes clear that the height of the system is a critical parameter: The system should be easily movable through doors. In the past, even removing doors often did not help and MRI installations frequently meant breaking up concrete structures, affecting the structural integrity of the building. Naturally, this was often associated with high costs and organizational efforts as well as other unwanted consequences. In many countries, two-meter high door openings are standard. Here is the simple but effective recipe with five major ingredients how to make sure the MRI system stays below 2 m height:

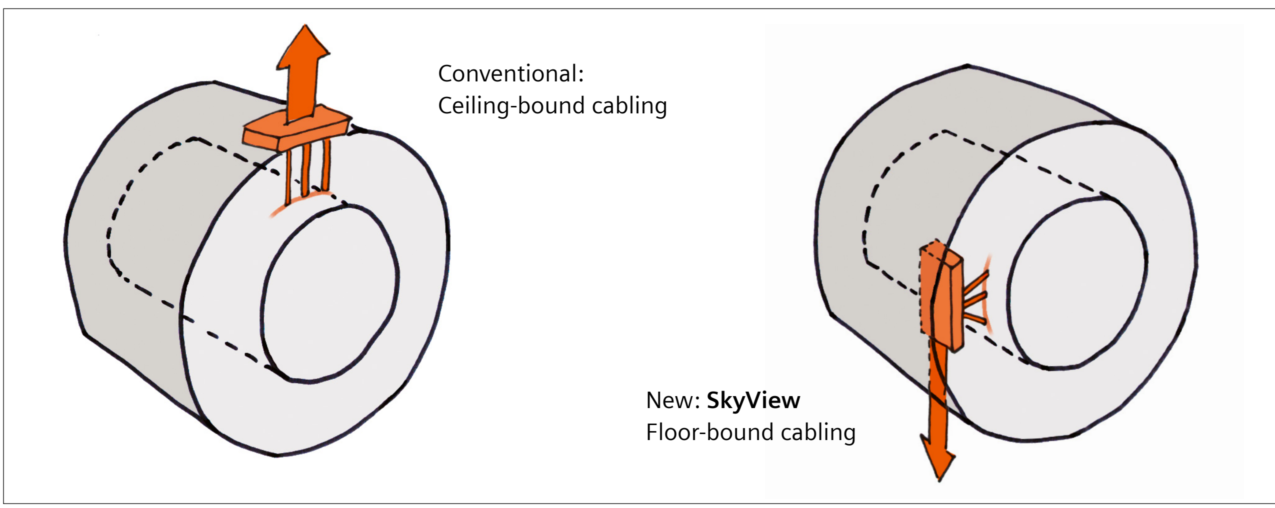
1. Outer vacuum vessel: Limit shield coil diameter
Underneath the plastic cover of an MRI system is the outer vacuum chamber (OVC), which contains the cryoshield and the superconductive magnet coils (dry magnets do not need a helium vessel). When the inner coils of the magnet are moved outward to accommodate the large patient bore, the shield coils also tend to move further out. Setting a boundary of ~1.95 m for the diameter of the OVC sets a clear design goal for the position of the shield coils. If the OVC is to be within the two-meter limit, it is essential that no other parts of the system design exceed this limit.

2. Buttress rings and magnet suspension
For the OVC to withstand the 1 bar atmospheric pressure from the outside, it is reinforced using circumferential buttress rings. These rings are flattened at the top of the magnet without any negative impact on their structural stiffness, which in turn allows the shield coils to be moved out as far as possible inside the OVC. Another structure that potentially affects the overall height of the system are the suspensions that hold the magnet coils in place. A tensile suspension connects the 4 K cold inside of the magnet with the warm outer vacuum chamber (OVC). The mechanical structure required to mount the tensile suspension to the outside of the OVC needs to be very slim so that this is not the highest point.

3. Quench pipe
The dry magnet does not need a quench pipe. It follows that there is no pipework on top of the magnet that could require extra height on top of the OVC and would require the connection of the magnet to the ceiling. This makes the new SkyView option possible (Fig. 3), which gives the system a unique visual impression by removing any connection between the MR scanner and the ceiling.



3 **(3A)** Conventional system with quench pipe. **(3B)** MAGNETOM Free.Max with DryCool magnet technology and easy siting.



4 Gradient connection and SkyView

4. Gradient connections

The gradient coil needs to be connected to the gradient cables that deliver the current from the gradient power amplifier (GPA). As in every electrical connection, the point where two cables are joined together is critical to maintaining good electrical contact. The gradient cables carry currents over 300 A and voltage up to 1200 V, so any loose connections could generate sparks that must be avoided. The connection of gradient cables is particularly critical due to the high Lorenz forces:

$$|\vec{F}_L| = |I|\, |\vec{\ell}|\, |\vec{B}|\, \sin\alpha$$

Locally, the actual field at the end of the magnet can be higher by up to a factor of 2–3 than the nominal field at the isocenter. For a 1.5T system with a strong gradient engine (I = 900 A), a 40 cm long gradient cable at the end of the magnet will experience a force of approx. 500–1500 N (equaling 50–150 kg) oscillating with the gradient pulses. This is why, historically, the connection of the gradient cables from the GC to the cables from the GPA on scanners from Siemens Healthineers was on top of the magnet. Here, the fields perpendicular to the wires and the resulting Lorenz forces are lower. This location for the gradient connection was never an issue on wet MRI magnets, because other parts (e.g., the cold head or pipework for the quench line) were located even higher.

I = current, ℓ = length of wire/cable, B = magnetic flux density aka magnetic field, α = angle between wire/cable and B

5 MAGNETOM Free.Max installation at University Hospital Basel in Switzerland. Even during one of the very first installations, the small size of the system and the eliminated quench pipe paid off to make the installation process much easier.

With the lower field strength and the lower gradient current, it was possible to reduce the forces by almost one order of magnitude. This facilitates a gradient connection on the rear side of the magnet rather than on the top. This is also required for the SkyView siting option. Here, gradient cables are routed together with all other cables through the floor rather than the ceiling (Fig. 3B).

5. Cold head

The cold head in conventional MRI magnets, where the superconductive coils are submerged in a liquid helium bath, needs to be located above the liquid helium level to allow recondensation of the gaseous helium. In dry magnets that have just a small helium reservoir as a liquid heat buffer rather than a large helium vessel, the cold head can sit in any vertical position. On MAGNETOM Free.Max, the cold head is located inside a turret, mounted on the side ~30 cm below the upper boundary of the OVC. This not only allows unhindered transportation through 2 m high doors, but also means that all later service activities (e.g., cold head replacement) can be performed within a ceiling height of just 2.2 m – even after the system has been installed in its final location. Since MAGNETOM Free.Max can be both installed and serviced in low-height premises, MR diagnostics can now be brought to new places such as small imaging centers. These are often located in residential buildings with limited available space.

This overview shows how a complete design overhaul of the magnet and gradient system together with the use of new imaging and reconstruction techniques results in an MRI system that achieves somewhat contradictory goals: A large 80 cm bore for the patient with a scanner that delivers diagnostic image quality AND easy installation with a small physical footprint and low connection power.

More background information on MAGNETOM Free.Max and DryCool magnet technology will be available soon on: www.siemens-healthineers.com/magnetom-world.

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