News From Japan

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First Magnetic Resonance Imaging Under Three Tesla Magnetic Field

Superconducting (SC) technologies have long histories of research, and some of them are already in a stage for practical applications. For example, a SC current lead, a SC transformer, a SC cable in grid operation, and a SC flywheel for magnetic energy storage have been introduced in this “News From Japan” column as recent Japanese typical examples of SC applications [1]–[4].

Magnetic resonance imaging (MRI) is now widely used in medical diagnosis, since it can provide detailed images without using radioactive beams, as with X-rays. Especially for diagnosing brain diseases, SC-MRI capable of providing precise images by realizing high magnetic field is contributing to early diagnosis and early cure. As of March 2016, a total of about 6,000 SC-MRIs are working in Japan [5].

The mechanism of MRI is schematically shown in Figure 1. By obtaining computer-aided spatial distributions of nuclear magnetic resonance (NMR) signals from protons (^{1}H) in a human body induced by applying a powerful magnetic field and radio waves, we can have detailed pictures of the inside of the body. At present, since most SC-MRIs use NbTi SC coils, we need expensive liquefied helium.

Helium is not produced in Japan, which means that Japan needs to import all helium from foreign countries. Therefore, several research funding agencies affiliated with the Japanese government, such as the Japan Agency for Medical Research and Development (AMED) and New Energy and Industrial Technology Development Organization (NEDO), have been encouraging Japanese universities, research organizations, and industries to throw the development of MRIs capable of using high-temperature SC coils into high gear. Furthermore, the quality of SC coils using REBCO (rare-earth barium copper oxide) has become stable, and a long coil has become available [6].

With this background, Mitsubishi Electric, Tokyo, has been developing a SC-MRI using REBCO coils, together with Kyoto University and Tohoku University, under the supervision by NEDO. The project was started in 2013, and it aims at the completion of the design of SC magnets for imaging the whole body and putting the developed MRI on the market in 2026.

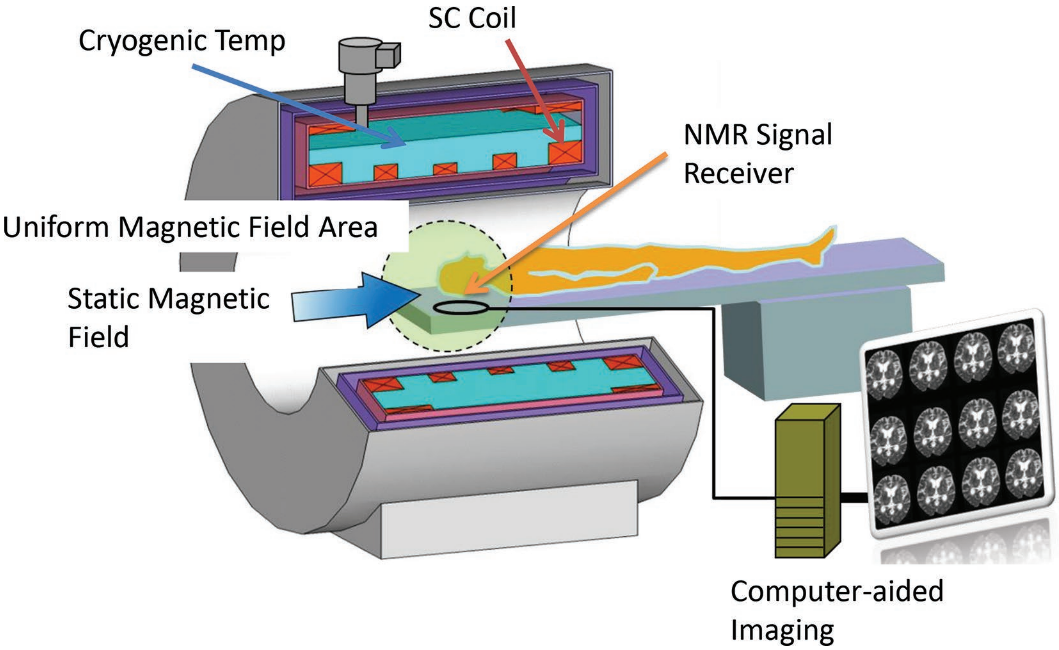

Figure 1. Schematic diagram to explain the mechanism of magnetic resonance imaging (MRI). SC = superconducting; NMR = nuclear magnetic resonance.

Table 1. Specifications of REBCO wire used for coils

Item	Specification
Dimension	
Conductor width	5 mm
Conductor thickness	0.16 mm
Structure	
Insulated film (dual)	Inner layer: fluorine-coated polyimide tape Outer layer: polyimide tape
Stabilized layer	Oxygen free copper (C1020)
Metal circuit board	Nickel based alloy (equivalent to Hastelloy RC-276)
Electrical characteristics	
Critical current	250 A or higher at 77 K, self-field

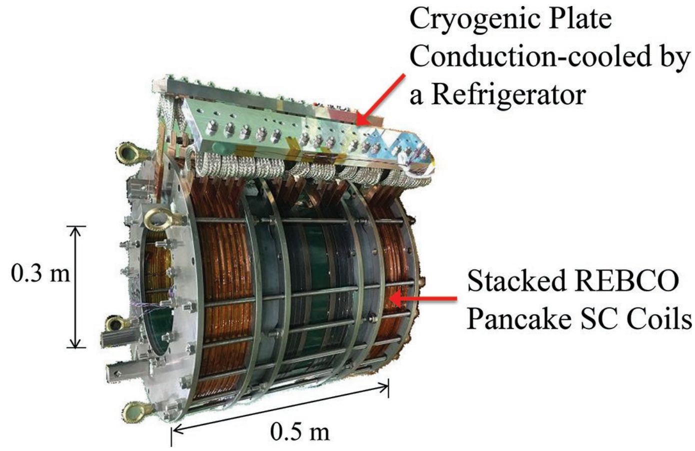

Figure 3. External view of the 3-T model coil for magnetic resonance imaging. SC = superconducting.

First, Mitsubishi developed a one-third model coil for a three tesla (3 T) MRI. One important development task is to realize a coil structure that would not cause deterioration of the SC characteristics. The REBCO SC wire for the coil consists of a 1.0-µm REBCO thin film fabricated on a Hastelloy (a highly corrosion-resistant metal alloy) substrate via a buffer layer and a copper film plated on the surface of the REBCO film or its periphery with Sn solder. If a mechanical stress is applied in the direction perpendicular to the thin copper-plated SC tapes, the plated copper would be delaminated and the SC properties would be easily degraded. As a countermeasure against this, fluorine coating was added to one side of a double-layer polyimide film to release the mechanical stress at the interface between the insulated film and the SC tape. Furthermore, the SC coil was vacuum impregnated with anti-cryogenic-cracking epoxy resin in order to increase the mechanical toughness. Table 1 shows the specifications of the REBCO wire used for the coil [7].

The second development task is the production of a high-precision coil. Usually, SC wire is wound to form a coil in the shape of a pancake as shown in Figure 2, and then, several pancakes are stacked [5], [6]. Although it is believed that each wire has to be wound with an error less than 0.1 mm, Mitsubishi wound the entire coil in one pancake within the error of 0.05 mm. In addition, each pancake was stacked up within a position error of 0.5 mm, although the designed allowable error was 1.0 mm.

An external view and the specification of the assembled SC model coil are shown in Figure 3 and Table 2, respectively [7]. The magnetic field at the center of the bore or area where a patient is scanned is 2.9 T, the same as general 3-T MRI systems. The maximum experienced magnetic field was 4.5 T when the central magnetic field was 3.0 T and the coil passed the current with a density of 113 A/mm^2. Note that the physical parameter with the unit of T (Tesla) is magnetic flux density and not magnetic field. However, in MRI-related industries or

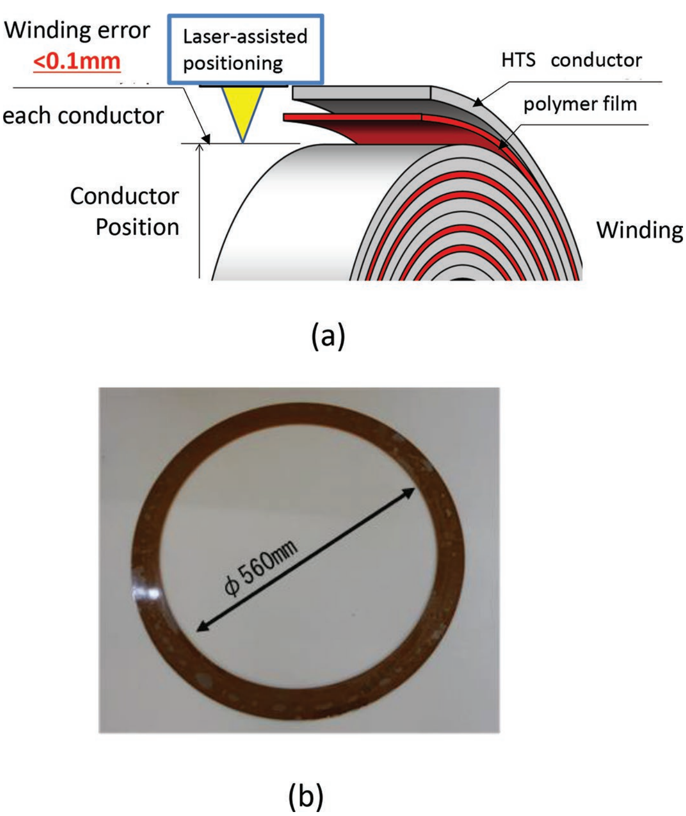

Figure 2. Superconducting coil for magnetic resonance imaging. (a) Method for adjusting winding positions in the coil. (b) External view of the pancake coil.

Table 2. Design specifications of HTS 3-T model coil

Item	Value
Inner diameter	320 mm
Maximum outer diameter	471 mm
Axial length	440 mm
Central field	3.0 T
Maximum field	4.5 T
Critical current of wire at field	351 A (20 K)/4.1 T (36°)
Current density of coil	113 A/mm^2
Inductance	32 H
Field uniformity	1.67 ppm/100 mm V_{rms}

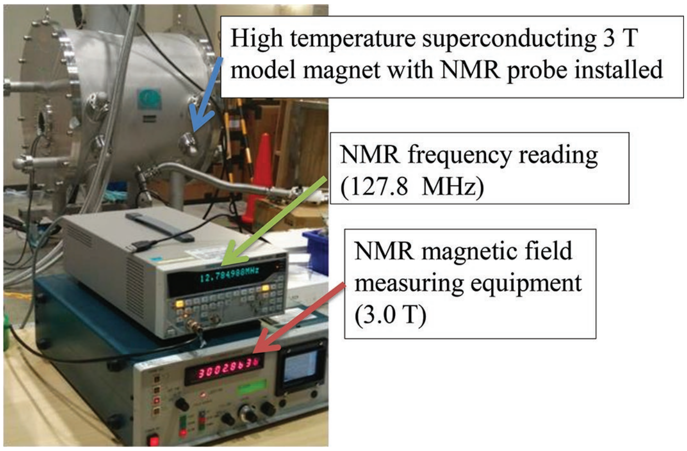

Figure 4. Photograph showing one scene in the experiment at the time of generating a central magnetic field of 3 T. Nuclear magnetic resonance (NMR) frequency of 127.8 MHz and the magnetic field of 3.0 T are clearly displayed.

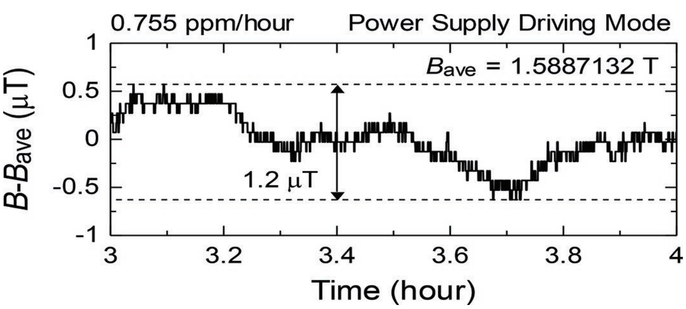

Figure 5. Stability of the output magnetic field for the nuclear magnetic resonance probe supplied from a highly stable excitation power source, achieving the variation of 1 ppm/hour or less.

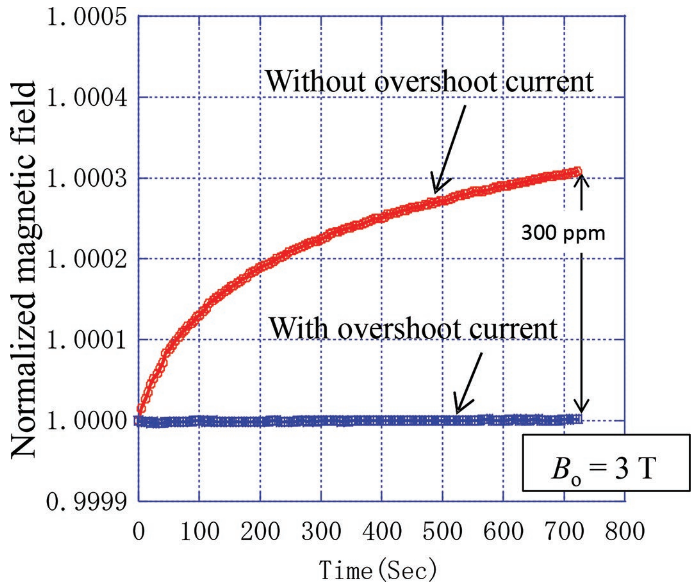

Figure 6. Time-dependent change in magnetic field of the developed HTSC 3-T test magnet right after the excitation and the effect of passing an "overshoot current."

among researchers in this field, the term "magnetic field" is far more often used. The maximum current density to pass through the coil is 120 A/mm^2. The inductance of the coil is 32 H, and the total length of high temperature SC wire is 16 km.

For cooling the model coil, a Gifford-McMahon refrigerator was used, by which the coil was cooled to a temperature below 10 K in one week and then its temperature was stabilized at around 7 K. Then, the coil was excited by passing the electric current, and the central magnetic field of 3 T was successfully attained. Figure 4 is a photograph that shows one scene during

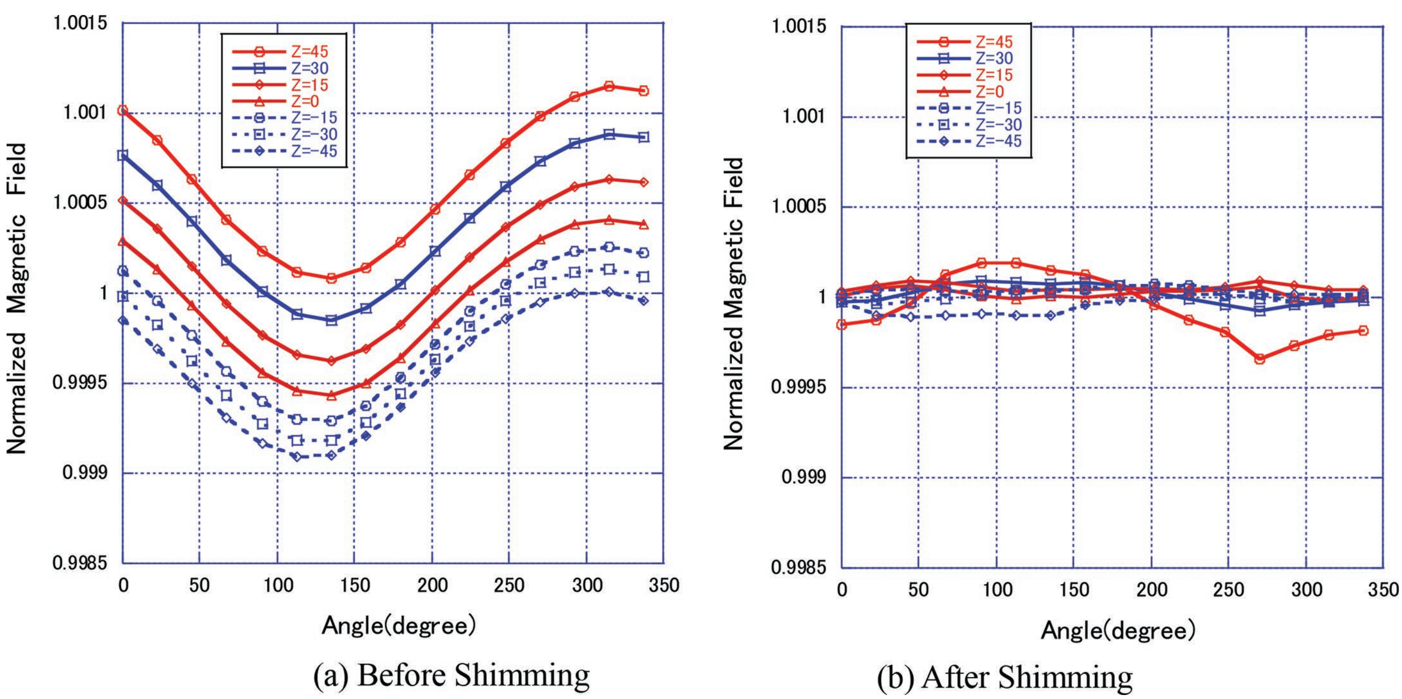

Figure 7. Effect of shimming on the angle-resolved distribution of magnetic field in the developed HTSC 3-T model coil.

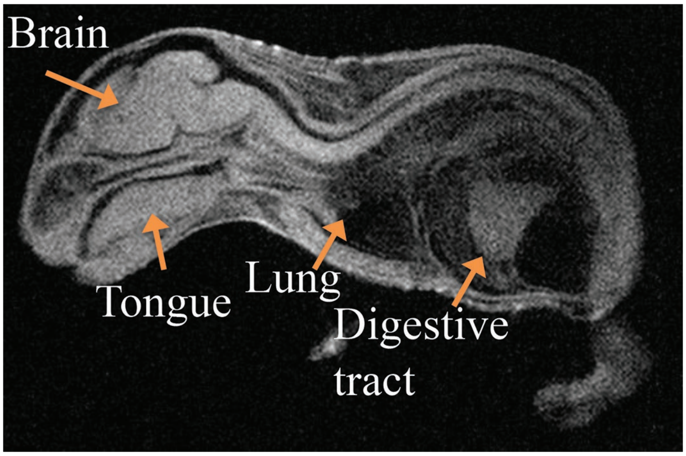

Figure 8. First magnetic resonance image of a mouse fetus taken under a magnetic field of 3.0 T using the developed HTSC 3-T model coil.

the experiment to measure the magnetic field. In the photograph, the NMR frequency of 127.8 MHz and the magnetic field of 3.0 T are clearly displayed on the measuring instruments. Note that the equipment at the back of the measuring instruments is a cryostat, in which the high temperature SC coil was installed. The cylindrical bore of the cryostat, where MR imaging would be done, is 230 mm in diameter and was kept at room temperature [7].

If the magnetic field varies with time, an electric current is induced by electromagnetic induction so that the current would suppress the variation in the magnetic field. This induced current would never decay in a SC substance. Therefore, the magnetic field for MRI must be very stable. This is a specific issue of SC-MRI. Regarding this, the project used a highly stable excitation power source, with a temporal fluctuation of less than 1.0 ppm/hour, as shown in Figure 5. Furthermore, by passing an "overshoot current" [8], [9], the magnetic field becomes even more stable as shown in Figure 6. In addition, the uniformity in spatial distribution of the magnetic field is also very important. As for this spatial uniformity, the 3-T model coil has attained a high uniformity, as shown in Figure 7, by adopting a proper "iron magnetic shim" method [7].

To demonstrate the MRI ability of the developed coil, an image was obtained for a mouse fetus 25 mm long. Figure 8 shows this example. It is clearly demonstrated that an image with a high spatial resolution less than 0.2 mm, which is sufficient for practical use, can be obtained. As of February 2016, the image shown in Figure 8 is the first MRI image of a mouse fetus taken under a magnetic field of 3 T using a high temperature SC coil [7].

This article was completed with the help of S. Yokoyama of Mitsubishi Electric Corporation.

References

[1] Y. Ohki, "News From Japan—Development of a RE-Ba-Cu-O Superconducting Current Lead," *IEEE Electr. Insul. Mag.*, vol. 30, no. 5, pp. 43–45, 2014.
[2] Y. Ohki, "News From Japan—Development of a Three-Phase Superconducting Power Transformer," *IEEE Electr. Insul. Mag.*, vol. 31, no. 3, pp. 46–48, 2015.
[3] Y. Ohki, "News From Japan—Japan's First In-Grid Operation of a 200-MVA Superconducting Cable System," *IEEE Electr. Insul. Mag.*, vol. 31, no. 4, pp. 61–63, 2015.
[4] Y. Ohki, "News From Japan—World's Largest-Class Flywheel Energy Storage System Using Superconducting Magnetic Bearings," *IEEE Electr. Insul. Mag.*, vol. 33, no. 2, pp. 62–64, 2017.
[5] S. Yokoyama, "Edge Cutting Applications of Superconducting Technologies, 9. SC Magnet for MRI," *Ohm*, March 2018, pp. 45–49. (in Japanese)
[6] Y. Ohki, "News from Japan—Development of a RE-Ba-Cu-O Superconducting Strong Magnet," *IEEE Electr. Insul. Mag.*, vol. 28, no. 2, pp. 60–61, 2012.
[7] S. Yokoyama, J. Lee, T. Imura, T. Matsuda, R. Eguchi, T. Inoue, T. Nagahiro, H. Tanabe, S. Sato, A. Daikoku, T. Nakamura, Y. Shirai, D. Miyagi, and M. Tsuda, "Research and development of the high stable magnetic field ReBCO coil system fundamental technology for MRI," *IEEE Trans. Appl. Superconduct.*, vol. 27, no. 4, 4400604, 2017.
[8] H. Miura, J. Miyazaki, M. Ide, D. Miyagi, M. Tsuda, and S. Yokoyama, "Estimation method of optimal amount of overshooting current for temporal uniform magnetic field in conduction-cooled HTS coils" *IEEE Trans. Appl. Superconduct.*, vol. 28, no. 3, 4401705, 2018.
[9] T. Yachida, M. Yoshikawa, Y. Shirai, T. Matsuda, and S. Yokoyama, "Magnetic field stability control of HTS-MRI magnet by use of highly stabilized power supply," *IEEE Trans. Appl. Superconduct.*, vol. 27, no. 4, 4702905, 2017.