The principles and evolution of magnetic resonance imaging

M. Lakrimi, A.M. Thomas, G. Hutton, M. Kruip, R. Slade, P. Davis, A.J. Johnstone, M.J. Longfield, H. Blakes, S. Calvert, M. Smith, C.A. Marshall
Siemens Magnet Technology, Wharf Road, Eynsham, Witney, Oxon OX29 4BP, UK

E-mail: mhamed.lakrimi@siemens.com

Abstract. Magnetic Resonance Imaging (MRI) offers the best textbook example for the exploitation of both Heike Kamerlingh Onnes’s major discoveries, namely the liquefaction of liquid helium and superconductivity. This paper will briefly describe how MRI images are acquired, give a historical account about the early days, and then focus on the magnet. It provides a window into the world of physics and engineering of magnet technology and it will chart the magnet design evolution since inception. For example, over the past 20 years, magnet designers have taken 1.5T MRI magnets from a weight of 13 tonnes and a length of 2.40m in 1989 to a weight of 3.2 tonnes and a length of 1.37m in 2009. The 3T products have also undergone similar developments. Coils experience mechanical forces of order 380 tonnes. The cryogenics for MRI magnets have also undergone major improvements. Today’s magnets do not use liquid nitrogen and most importantly have zero helium loss, thus needing no refill. Combined with the weight, this makes the body scanner accessible to developing countries and easy to site in any room. Although the future may eliminate the need for liquid helium completely, MRI will always continue to depend on superconductivity.

1. Introduction and superconductivity work by Onnes

The achievement of Heike Kamerlingh Onnes to liquefy helium in 1908, which won him the Nobel Prize in 1913 [1], and his discovery of superconductivity in 1911 opened up a plethora of new physics and engineering. The units Ohm [2] and Volt [3] have been redefined in terms of physical observables at low temperatures as is much of the research on quantum encryption and QUBITs [4]. Several Nobel Prizes have also been awarded for discoveries made at low temperatures or in superconductivity.

The world waited 50 years before the first commercial research magnet was built, thanks to the breakthrough in manufacturing superconducting wire and the efforts of Sir Martin Wood and his wife Audrey who founded Oxford Instruments in 1959 with the blessing of Sir Nicholas Kurti [5]. Magnetic Resonance Imaging (MRI) uses the same underlying physics as Nuclear Magnetic Resonance (NMR) and exploits the spin relaxation of a small percentage of protons in response to a specially sequenced external excitation. Since its introduction in the early 1980’s, MRI quickly became accepted as one of the leading diagnostic imaging modality in healthcare. Being a safe and non ionising technique, almost all the tissues in the body could be non invasively and routinely scanned, enabling the diagnosis of tumours, tendinitis, stroke, multiple sclerosis, infections of the brain, spine and joints, visualisation of torn ligaments, or shoulder injuries, brain trauma (bleeding or swelling), and evaluation of bone tumours, cysts, bulging/herniated discs, structure of the heart or aorta or

1 To whom any correspondence should be addressed.
masses in soft tissues. Preventative care and the problem of an ageing population will continue to drive new applications for MRI.

Superconductivity, an electrical state discovered by Onnes at Leiden, manifests itself by zero (electrical) resistance when the temperature drops below a certain critical temperature. The cooling of the sample is achieved by means of liquid helium, which was first liquefied by Onnes.

After liquefying helium, Onnes dreamed that “in the future, I see all over the Leyden laboratory measurements being made in cryostats, to which liquid helium is transported just as the other liquid gases now are, and in which this gas also, one might say, will be as freely available as water”. Nowadays, helium is an expensive commodity and the advent of cryogen-free system and higher temperature superconductors are gnawing into this niche application.

Shortly after discovering superconductivity, Onnes thought of magnets. He told that “mercury at 4.2K has entered a new state, which can be called the state of superconductivity… Now that we are able to use these metals, all types of electrical experiments with resistance-free apparatus have become possible… A self-contained coil, cooled in the magnetic field, should, when the field is removed, be able to simulate for some time an Ampère molecular current… The absence of Joule heat makes feasible the production of strong magnetic fields using coils without iron, for a current of very great density can be sent through very fine, closely wound wire spirals. Thus we were successful in sending a current of 0.8 amperes, i.e. of 56A/mm², through a coil, which contained 1,000 turns… All that I have said so far about superconductors applies only to currents below a certain threshold value, which lies all the higher the lower one drops in temperature”. The far sighted Onnes was already using notions which nowadays are taken for granted, namely a current density, a critical current, a current-temperature relation. In April-June 1914, he achieved persistence for hours which he destroyed using a heater [1]. However, a practical superconducting electromagnet had to await the discovery of type-II superconductors that could withstand high magnetic fields. The first successful superconducting magnet was built by Yntema [6] in 1954 using niobium wire and achieved a field of 0.71T at 4.2K. On return from a conference in the USA, Martin Wood and his wife Audrey of Oxford Instruments built their first NbZr 4T in March 1962 [5]. Nb₃Sn, a higher current material and its tolerance of high background fields accelerated interest in building magnets [7]. Human MRI scanners became widely available in the late 1980s, producing images of the inside of the body. Current MRI scanners can produce highly 3-dimensional images and video of the human anatomy showing moving organs. A 3T scanner has an image resolution of better than 0.5mm. We shall first start with a brief review of the physics, the historical events, and then deal with the design and evolution of magnets over the past few decades and the role they now play in MRI.

2. The physics of MRI

2.1. Introduction
MRI has its roots in NMR. The word nuclear was in reference to the nucleus and has nothing to do with radioactivity. It is for this reason that the medical community dropped the word (nuclear) and it simply became MRI and not NMRI. Magnetism is a property of matter that is a result of orbiting electrons in atoms. Working independently, Block [8] and Purcell [9] made the first successful NMR experiments in 1946. The first patent in the field of MRI was granted in 1974 to Damadian. He performed a form of MRI prototype exam on a human being on July 3, 1977, and it took almost five hours to produce one image! Today a whole body can be scanned in about 20 minutes, thanks to faster computers and improved hardware for producing the necessary switched field gradients and improved radio frequency transmit/receive antennas.

The nucleus of a hydrogen atom is a single proton. The human body consists of 60% water which represents the biggest source of hydrogen protons in the body, followed by fat. The orientation of these tiny proton magnets is random in the absence of a magnetic field. In analogy to a pixel in 2D images, MRI uses a volumetric pixel called voxel. Since one mole of water fills a volume of 18ml and that there are two hydrogen atoms per mole, a voxel of 1×1×1mm³=0.001ml would contain 2×6.02×10²³
(molecules per mole) × 0.001/18 = 6.7 x 10^{19} total protons. The population between the two spin up and down levels obey Maxwell-Boltzman statistics. At 3 T and 37°C (the body temperature), the total number of excess protons = \(6.7 \times 10^{19} \times \Delta E/(2k_BT) = 6.02 \times 10^{15}\), where \(k_BT\) is the thermal energy, and \(\Delta E = \gamma B\) is the magnetic energy. \(B\) is the magnetic field flux density and \(\gamma\) is the gyromagnetic ratio associated with a Larmor precession frequency taking a value of 42.576 MHz/T for the hydrogen nucleus. One can see that the frequency is in the low Radio Frequency (RF) range and this is why MRI uses RF excitation. Figure 1 gives the population distribution for every 2 million protons from 0 T to 11.75 T. This means that for every two million protons available in the 1 mm^3 volume, there are, at 3 T, 18 more protons aligned with the field than against it. The vast majority of spins cancel each other and hence it is this small spin excess which is probed in MRI to yield the images. It is incredible that such a small number of spins contribute to the MRI images that we know of today, which is a testament to the remarkable sensitivity of the detection electronics developed for MRI.

2.2. RF pulse, T1, T2, and T2* relaxations

When a patient is loaded in the bore of an MRI magnet, the magnetic field, aligned along the z or longitudinal direction, settles the spin population between the low and higher energy states as given in figure 1. Under this steady state, not much happens except that there is a net magnetisation along the z-direction. The magnetisation in the transverse xy-plane is zero because of the randomness of the spins phases. When an external RF pulse excitation, matching or resonant with the Larmor frequency and oriented orthogonal to the static field is applied, a spin from the lower energy (up or \(\uparrow\)) state can be stimulated to jump into the higher energy (down or \(\downarrow\)) state. This is similar to intraband transitions in quantum wells. The net magnetisation experiences a torque which causes it to spiral away from the z-axis alignment; its vector will move a distance proportional to the duration of the RF field pulse, as shown in figure 2. The net magnetisation vector rotates an angle, \(\alpha\), called the flip or tip angle and maximum signal amplitude is achieved when this angle is set to 90°. After the RF pulse the spins precess around the static field like a spinning top.

When the RF pulse is switched off, the excited protons gradually return to their original lower energy state and MRI exploits this relaxation or decay back to the equilibrium state. The spins return to their equilibrium in two ways. They can relax to their lower energy state by loosing energy to the surrounding molecular lattice and hence the name spin-lattice relaxation, as known in solids. This relaxation is characterised by a time constant \(T1\) and a recovery in the longitudinal magnetisation, \(M_z(t)\), of \(M_z(t=0)\) the magnetisation at equilibrium, just before the pulse is applied. The recovery rate in \(M_z(t)\) towards equilibrium follows an exponential decay. \(T1\) is defined as the time when the magnetisation has returned to 63% of its original value. The relaxation is considered complete after 5\(T1\) duration.

Following the RF pulse, the number of spins in the upper state is increased and thus there is increased interaction between the spins spinning in the xy-plane, leading them to lose their phase coherence over time and resulting in a reduction in the net transverse field intensities.

**Figure 1.** Distribution of 2 million protons for spin up and spin down levels for a number of field intensities.

**Figure 2.** The magnetisation spirals away from the \(B_z\) alignment and decomposes into \(M_z\) and \(M_{xy}\).
magnetisation, \( M_{xy}(t)=M_{xy}(t=0)\times\sin(\alpha(1-e^{-t/T_2})) \). The precessing spins generate an alternating magnetic field which can be detected using a suitable pick up coil. The signal emitted during this relaxation is called a Free Induction Decay (FID). This timescale is defined by the T2 spin-spin or transverse relaxation time constant. In this dephasing process, there is no loss of energy to the surroundings. T2 is also characterised by the time taken to lose 63% of the transverse magnetisation and is always smaller or equal to T1. T1 and T2 are defined by properties of the materials under study and hence may be used to distinguish between types of tissue by differing image contrast. Table 1 gives some values for T1 and T2. In terms of the black and white MRI images, T1 weighted images depict fat and normal anatomic planes and fat suppressed T2 weighted images are used for abnormalities.

The magnet’s imaging field can never be perfectly homogeneous and is also influenced by differences in magnetic susceptibility between tissues and cavities, for example in the sinus region, or if a patient has metallic inclusions. This residual inhomogeneity causes a further dephasing or loss of transverse magnetisation with a time constant shorter than T2, called T2*. At first this appears to be a significant problem, but fortunately a sequence call Hahn-echo or spin echo can be used to rephase those spins which have undergone a T2* decay. To illustrate the spin-echo imagine protons precessing at different frequencies; let’s select for the sake of clarity just three spin s precessing at different rates due to the small variations in local magnetic field. These spins are analogous to a tortoise, a hare, and a snail racing.

At t=0sec, all three are standing on the starting line and, focused on the race, they are said to be in coherence. If left to race in a straight line, the hare would cover a much longer distance than the tortoise and the snail would finish last. As the gaps between them are increasing, they are said to lose coherence. However, imagine that after t=TE/2 they are summoned to return to the starting line. It follows that the hare will have a longer distance to travel back and the snail the shortest. In the end, all three would cross the starting line at the same time despite having covered different distances. As all three stand on the starting line again, they are said to have regained phase coherence causing the formation of an echo signal in the RF pick up coil. The event which causes the direction of the race to be reversed is the application of a further RF pulse which flips the magnetisation vectors around the third axis by 180°, rotating their precession.

A great variety of MRI pulse sequences have been developed, some basic ones being Spin Echo, Inversion Recovery, Gradient Echo, Turbo Spin Echo [10], etc. Important pulse sequence parameters are TR (Repetition Time between applications of the initial tipping pulse), TE (Time to Echo), TI (Time for magnetisation Inversion) and Flip Angle. The values of these parameters are chosen depending on the tissues and MRI examination. Finally, scientists also use contrast agents such as paramagnetic materials at the right dose to engineer a preferential T1 enhancement. For example gadolinium contrast agents cause preferential T1 relaxation enhancement, causing increase in signal on T1-weighted images and thus achieve better resolution. Contrast agents are used for bowel, liver, and lymph node imaging.

In section 3, we shall give a short description as to how the MR signals are used to give us the Cartesian two dimensional images. But how did the science of MRI start?

### Table 1. T1 and T2 values at 1T.

| Material            | T1(s) | T2(s) |
|---------------------|-------|-------|
| Cerebro-Spinal Fluid| 2.50  | 1.40  |
| Fat                 | 0.24  | 0.09  |
| Spleen              | 0.46  | 0.08  |
| White matter        | 0.68  | 0.09  |
| Gray matter         | 0.81  | 0.10  |
| Water               | 2.50  | 2.50  |
| Liver               | 0.49  | 0.04  |
| Blood               | 0.80  | 0.18  |
| Muscle              | 0.73  | 0.04  |
to demonstrate MRI for the first time on small test tube samples. In 1974, Mansfield’s group developed selective spin-echo excitation [12]. The University of Aberdeen followed Lauterbur’s article and built a small scale MR imager which produced an image of a mouse in 1974 [13]. In 1975, Ernst’s group proposed MRI using phase and frequency encoding and the Fourier Transform [14]. In July 1977, Damadian demonstrated a form of MRI called field-focusing nuclear magnetic resonance [15]. Aberdeen University (UK) conducted the first clinical whole-body MRI scan in the world on 28 August 1980 [13]. Edelstein et al developed the spin warp sequence [16] and Mansfield and co-workers the Echo-Planar Imaging (EPI) technique [17]. Nowadays images are produced at video rates of 30ms per image. Figure 3 shows a photograph of a modern MRI body scanner. The MRI images can be in any plane: axial (xy-plane), coronal (zx-plane), or sagittal (zy-plane) or any arbitrary angled plane as illustrated in the inset to figure 3. MRI creates cross-sectional images or slices of a body part.

In this next section, we will look at the history of the MRI superconducting magnet.

2.4. A short history of the MRI magnet
From the famous date of 1911, several superconducting materials have been discovered, the list and chronology of which is beyond the scope of this paper. Currently, NbTi, Nb3Sn, and MgB2 have emerged as the materials of choice for building superconducting magnets with NbTi being the true workhorse; its current cost is less than one fifth of that for Nb3Sn and MgB2.

The use of superconductors benefited from the fact that copper was expensive and consumed much greater electricity and the resulting heating was very challenging to address. Surprisingly, it was the high energy physicists in need of high magnetic fields who championed and propelled the use of superconductors. Rutherford Appleton Laboratory (UK) built the first superconducting accelerators using copper clad NbTi in 1960 and American scientists followed in 1967. Commercial NbTi was only developed in 1962 and in the same year General Electric (GE) built the world’s first 10T magnet in 1962. Although this was looked on as an achievement, the commercial aspect was a disaster as Bell Labs only paid GE the fixed price contract of $75k for a magnet that cost $200k to realise [18]. Intermagnetics General Corporation (IGC, a spin off from GE in 1971) and Oxford Instruments (OI, the second spin off from The University of Oxford) emerged as champions of the commercialisation of superconductors. In 1967, OI demonstrated a 10T magnet and a 30mK dilution refrigerator. However, it is MRI and to some extent NMR that expanded the use of superconductors.

OI built the first superconductor magnet with a bore diameter of 1m. It delivered, in 1980, the first MRI whole body magnets (0.15T) to Hammersmith hospital (UK) and the University of California in San Francisco and the third to GE. Orders for MRI magnets increased from £1million in 1981 to £25million in 1982 and OI commanded 90% of the market [18]. Following this achievement the company set up in 1982 Oxford Magnet Technology (OMT) to exploit its technology for MRI body scanners; the magnet volume was 50 systems a year. In 1984, a purpose built factory was established with a production capacity of 300 magnet systems per year. In May 1989, OMT became a joint venture between Siemens AG (51%) and OI (49%). In November 2003, Siemens acquired whole ownership of OMT and its name changed to Siemens Magnet Technology (SMT) in June 2004. By the end of the joint venture, OMT had supplied more than 7,000 MRI magnets which amounted to at least a third of all magnets installed in world hospitals. SMT continues to supply Siemens and other third party customers and following two further factory expansions, the manufacturing capacity is stretched.
to 1500 systems per year. In 2006, Philips Medical Systems acquired IGC. The top five MRI magnet suppliers are Siemens, GE, Philips, Toshiba, and Hitachi.

NMR and MRI remain the most significant profitable application of superconductivity to date. More than 75% of MRI magnets are made using superconducting wires. More than 2,500 superconducting MRI systems are now sold annually and more than 75,000 people are imaged daily. In 2008, the installed base of superconducting MRI systems was about 26,500 units compared to 14,600 in 2002 [19]. We shall now review the underlying physics and engineering behind the magnet design.

3. The design of an MRI magnet
Although, the design of the MRI magnet has undergone a major revolution, magnets only came of age recently. In 1980, a standard superconducting MRI magnet achieved only 0.5 Tesla compared with 1.5 Tesla today. Furthermore, in 1980, a 3 Tesla MRI magnet was almost unthinkable compared with today when it is gradually becoming the standard and systems with field strengths above 7T have been successfully built and 11.75T is in progress [20].

3.1. The magnet design
The magnet could be designed as a set of discrete coils or solenoids. However, the former geometry is preferred to make the magnet commercially viable, occupy the least volume and have a better control over the field homogeneity. Although a customer may only be interested in the specifications of the magnetic field intensity, homogeneity, drift, and temperature of operation, the magnet designer has to adhere to several other requirements for critical current, strain/stress, stray field, and other factors to balance the technical and economical risks. Designers continually refine their design limits through observing the performance of magnets and the measurements of certain parameters such as strain, temperatures and voltages during quench. Within one field system, there may be several products and the customer chooses the magnet to suit their purposes.

Magnets are designed by a team of people. Upon being issued with a specification for an imaging volume and the field strength, first the design and location of the coils to an accuracy of less than 0.1 mm and use wire of a particular geometry and insulation is carried out by the magnet designer. The mechanical engineer designs how the coils will be supported and how the magnet will be suspended within the cryostat. The cryogenic engineer will carry out the necessary calculation with regards to minimising conduction, radiation, and convection heating loads which affect the helium boil-off.

Magnets are designed using wire capable of carrying a few 1000 A/mm² at a certain field; domestic wiring carry typically 10A/mm². At low temperatures, say 4.2K, the wire specific heat drops by a factor of 4,000 compared to that at room temperature and this is why the minimum heating energy required to destroy completely superconductivity in the magnet can be less than 20 μJoules [21].

3.2. The magnet stresses and quench
Superconducting magnets operate at low temperatures and thus experience two types of stresses. Firstly, during thermal cooldown, the coils contract axially and radially, resulting in large negative strains and stresses. Secondly, during energisation, the coils experience intense Lorentz magnetic forces whereby the coils expand radially thus inducing a positive strain and simultaneously contract axially resulting in a negative strain. In addition, because of their proximity from one another the coils exert a force of a few hundred tonnes on each other and their supporting structure. It is also well known that if a local temperature exceeds that of superconducting wire critical temperature over a critical length known as the minimum propagation zone, the wire undergoes a transition to normal state and current will be dissipated through the resistive copper matrix, leading to excessive heating of the coil. This is termed a quench and the magnet suddenly releases its stored energy, typically tens of megaJoules, and a complex quench scenario ensues whereby a particular coil could be magnetically pumped up by its adjacent coils, leading it to experience higher stresses than intended [21]. To prevent any wire movement the coils are typically impregnated in wax or epoxy resin. Without adding other
coil refinements, it suffices to say that each individual coil is actually a complex composite that may not necessarily be easy to model *a priori*.

Onnes himself also pondered over the origin of quench as to whether [1] “*this first impulse proceeds from bad sections of the wire or is also produced in pure, tension-free, evenly crystallised metal. If the potential phenomena are to be attributed to bad spots, we shall learn to eliminate them and then it is perhaps possible that the conductivity could be increased still higher than to the value reached at present, which is thousand million times that at normal temperature*”. Fortunately, engineers rely on known rules to design magnets to avoid coils suffering mechanical damage and elasto-plastic deformation. If necessary, a quench protection circuit is designed to manage the stresses, the temperatures, and voltages arising during a quench [21].

### 3.3. The magnet stray field and homogeneity

The variation in homogeneity of the field within the bore of the magnet may be described by a series of Legendre polynomials. The designer adjusts the coils in order to minimise all impurities to an acceptable level while managing the other critical parameters. The very large inner diameter of 1m comes at a heavy price. The horizontal and vertical distances over which the stray magnetic field intensity for a 1.5T drops to less than 5G, deemed safe for patients with a heart pacemaker and electronics, can be very large indeed and need to be tightly controlled. In early magnets this was of the order of 10m axially, however this could be reduced by shielding consisting of a hundred or so tonnes of iron, making the requisite for reinforced floors and spacious rooms necessary. In 1986, OMT revolutionised MRI by introducing the Active Shield (AS) technology, allowing the stray field to drop to less than 5G within less than 4m axially and less than 2.5m radially, and thus enabling MRI systems to be installed at sites previously thought impossible. Active Shielding is achieved by engineering two outer coils in reverse polarity to the inner coils which generate the central magnetic field. This OMT innovation has become the industry standard. A typical MRI suite is sketched in figure 4. OMT/SMT have continued to focus on technological and manufacturing excellence. The business has won several Awards in recognition of its achievements in these areas and also for export success since 95% of the magnets produced at SMT in Eynsham are exported. The company has made significant contributions to the development of MRI technology and some of the notable world’s firsts to emerge from OMT/SMT are given in table 2.

For a high quality image, the magnetic field must not vary over the imaging volume by more than a few parts per million as shown in figure 5. This high homogeneity field is required to minimise the dephasing of the spins given by $T_2^*$.  

### 3.4. The magnet stability

The coils are then linked in series and the wires jointed together and there is also a stability specification that the field should not decay by $\geq$0.1ppm/hr. This means that the resistance of the joints must be much better than $10^{12}\Omega$ at the operating current (typically several hundred amperes) and in a background field. The MRI in a hospital is run in persistent mode, i.e there is no power supply attached. In order to achieve this, a superconducting switch, which in normal state has a resistance of

| Year | Innovative Technology                        |
|------|---------------------------------------------|
| 1980 | First MRI superconducting magnet            |
| 1986 | Actively shielded magnet                    |
| 1989 | 1.5T AS magnet                              |
| 1994 | Open magnet                                 |
| 1997 | New generation compact magnets              |
| 1997 | 3T AS magnet                                |
| 1999 | 1T Open magnet                              |
| 2000 | 4T AS magnet                                |
| 2000 | Shortest 1.5T AS magnet                     |
| 2004 | 1.5T AS open bore magnet                    |

*Figure 4. Sketch of an MRI suite using an Actively Shielded magnet.*
several ohms, is assembled in parallel to the magnet. When the MRI magnet is commissioned, the installation engineer first quenches the switch using a heater, thus causing its resistance to rise to a few Ω, and the current from the magnet power supply goes straight in the magnet which is in a superconducting state (0Ω). The rate at which the magnet is energised depends on the compliance voltage of the power supply and the inductance of the magnet. After reaching the desired operating current, the heater of the switch is turned off. After allowing sufficient time for the switch to revert to a superconducting state (0Ω), the current is ramped down to zero. Again given the finite internal resistance of the power supply, the current prefers to flow through the switch and it is only the current in the leads that drops to zero. The dissipation of the current with time is dictated by the quality of the resistance of the joints. A decay measurement is carried out to make sure that the drift stability is met.

The magnet engineer also includes in the design an external interference screening (EIS) circuit to protect the magnet against external disturbances such as lifts, vehicles driving by, electrical cables in walls, etc. because Faraday law of induction states that, in a magnetic field, a current or voltage will be induced in a fast moving conducting material. Every MRI magnet is equipped with an Emergency Run Down Unit (ERDU) to deliberately quench it in case of emergency.

3.5. The magnet cryogenics and mechanical stability

The magnet is then put in a cryostat as shown in figure 6. The cryogenic engineer is concerned with limiting the three types of heating: conduction, convection, and radiation. These are minimised by the use of suitable materials. Magnets have a vacuum chamber that is pumped to ~10⁻⁵ mbar at room temperature which then cryopumps to ≤10⁻⁷ mbar on cooldown. The cryostat is checked for leaks. Multi-layer insulation is used to reflect back any radiation heating that finds its way through the cryostat. The mechanical engineer has to deal with the mechanical stability and suspension issues. The MRI magnet could be transported by plane, ship, and road to its final site. Note that there are also mobile systems which are shared between different sites. These spend their lifetime on the road with the magnet always at field.

Early magnets made use of liquid nitrogen and needed a refill of nitrogen on a weekly basis and helium on a monthly basis. On modern magnets known as zero boil-off magnets, smart cryogenic engineering has eliminated the use of liquid nitrogen. These have a cryocooler inserted through the turret (shown in figure 6) which acts as a mini-liquefier whereby any helium gas boiling off is recondensed back into the magnet and so the systems only need topping up with liquid helium every few years! The cryostat holds approximately a thousand litres of helium.

3.6. The design evolution

Table 3 lists the design evolution for 1.5T over a period of 20 years. It can be seen that the length has been reduced to 1.37m, easing a patient’s claustrophobic feelings. The weight has been cut from 13 to 3.23tonnes, thus making the magnet easy to site on any floor of a building. Figure 7 shows a picture of
an old (left) and modern (right) MRI systems. In the short length magnet the patient’s head and feet can be seen and thus combined with the large 70cm diameter clear bore magnet reduces the claustrophobia felt by patients.

4. The gradient coils and cause of audible noise

So far we have only referred to the RF pulse and the magnet. As mentioned before, it was Lauterbur who invented the concept of imaging using a gradient coil [11]. The gradient coils, as the name implies, are used to produce magnetic field gradients in specific directions. The gradient coils are in pairs with equal and opposite polarity about the centre. One coil decreases the field whilst the other increases it, like an incline on a seesaw. These field gradients are \( \partial B/\partial z \), \( \partial B/\partial x \), and \( \partial B/\partial y \), where \( z \) is along the axis of the magnet bore, \( y \) is the vertical, and \( x \) is the horizontal. They are used to provide spatial localisation of the MR signal by using complex sequences of gradient coil and RF pulses. The spins in the higher field region will precess at a faster frequency than those located in a low field region. In figure 8, a transverse slice of spins is first selected to be in the xy-plane by applying the z-gradient simultaneously with an RF tipping pulse. The flip angle of spins within the slice is dependent on the resonant frequency and thus the thickness of the slice can be changed by using an RF pulse with a bandwidth and fixing the field gradient or by using a fixed frequency RF pulse which is held constant for a period of time whilst the field gradient is varied. Similarly applying the x-gradient will lead to a sagittal slice and the y-gradient will result in a coronal slice. The image resolution is dependent on several factors including the strength of the magnetic field. Increased image resolution usually comes at a penalty of increased acquisition time but this is mitigated by higher magnetic field strengths and modern innovations such as multiple parallel receivers. The image is typically a matrix of 256 x 256 pixels, i.e. made from 65,536 voxels.

The signal from different voxels within the slice can be split into 256 strips of say along the x-direction and one pixel height in the y-direction by application of suitable x and y gradient pulses during image acquisition. If a gradient is applied in the x-direction, the spins will precess at different rates dependent on their x-position; this is known as frequency encoding gradient. To differentiate the spins’ y-location, the y-gradient may be switched on for a very brief period of time to dial in a phase shift between spins in the y-direction, as shown in the top part of figure 8. In this example, the y-gradient is known as the phase encoding gradient. Table 4 gives the phase encoding gradient for any imaging plane. The signal acquired by the RF pickup or receiver coil is an integration of signals from all excited spins in the slice. Earlier descriptions referred to frequency, phase coherence, and relaxation times, it is better to switch from spatial locations in the (x,y,z) Cartesian to a reciprocal k-space where these are easily described. In k-space, the axes within a slice are frequency and phase. The computer generates an intensity histogram from the signal acquired by the receiver RF coil for all 256 voxels forming a strip. The location of spins in a particular row is determined by their different frequencies and in a

| Year | Product name | Length (m) | Weight (kg) |
|------|--------------|------------|-------------|
| 1989 | A            | 2.40       | 13,000      |
| 1994 | B            | 2.20       | 8,650       |
| 1997 | C            | 1.60       | 4,070       |
| 2003 | D            | 1.50       | 4,050       |
| 2004 | E            | 1.25       | 4,350       |
| 2009 | F            | 1.37       | 3,230       |

Figure 7. Long bore tube versus a modern MRI. The patient is not fully confined in the magnet.

Figure 8. Typical MRI sequence and images in k and real spaces.
particular column by their relative phases. Two-dimensional Fast Fourier Transform (2D-FFT) is used to convert the \( k \)-space signal to a Cartesian image, as shown in Figure 8.

People who have experienced examination or simply entered the imaging room during an examination procedure often remark on the noise. This is caused by the gradient coils being switched on for the sequences described above. The gradient coils sit inside the background field of the magnet and so when current is changed in the (gradient) coils to generate the field gradients they experience significant Lorentz forces which cause the coils to expand radially and contract axially as explained before in the context of the magnet design. As the switching is typically in the audible frequency range, the resulting vibration produces loud noises (clicking or beeping during scanning). The sound intensity is about 110dB but can reach 120dB (equivalent to the noise from a jet engine at take-off). Patients wear appropriate ear protection during the examination.

5. The other MRI sub-systems

Figure 6 shows a cross-section of an MRI magnet and a view of all its components; we have already discussed the magnet, the gradient coils, and the RF pulse. The installation engineer only uses a magnet power supply during commissioning of a magnet or servicing. There is electronics on the system to monitor the general state of the system. The gradient coils have their own power supplies which stay on site.

The bore contains the RF body coil which is used to generate the intense and uniform pulses of transverse RF magnetic field required to tip spins in the imaging volume. In figure 6, the RF pulse coil has the shape of a bird cage coil.

The RF receive coils are used to pickup and transmit the RF signal emitted from the hydrogen protons during relaxation back to equilibrium. The RF-receive coils are tuned to the magnet’s Larmor frequency and fashioned as local antennas which are body-part specific, as shown in figure 9. They exploit Faraday’s law of induction which states that a current is induced in a coil made from conducting material placed in a changing magnetic field. The signal is increased by close proximity to the region being imaged. Modern MRI systems use arrays of coils operating in parallel to reduce image acquisition or increase resolution.

The shim coils and iron are required to improve the uniformity of the magnetic field because realistic manufacturing tolerances make it impossible to achieve the required homogeneity without fine adjustments to the magnetic field, called shimming. Passive shimming hardware is simple and uses an array of small iron shim plates loaded in cartridges, typically located inside slots within the gradient coil. However, it has the drawback that the magnet must be ramped down to load the iron for obvious safety reasons and then re-energised again. The passive shims are not changed after set-up. There may also be active shimming utilising either superconducting or resistive coils.

The patient table or bed is used to load the patient into the central bore of the magnet. It can be fixed or demountable and often contains receiver coils (e.g. spine array).

The computer is for programming the gradient coils to select different imaging planes. The computer also processes the 2D-FFT images and creates the MRI images.

Finally the MR signal is extremely small as the earlier calculation showed and this is why it is important to minimise external RF interference. Thus, the MRI magnet is located in a Faraday cage which prevents external RF signals from reaching the patient under examination.

| Image Plane | Gradient | Encoding Gradient |
|-------------|----------|-------------------|
| XY          | X        | X or Y            |
| XZ          | Z        | Z or X            |
| YZ          | Y        | Y or Z            |

Table 4. Image plane and encoding gradient.
6. Conclusions
This paper reviewed a number of topics. It is in celebration of Onnes’s centenary for the discovery of superconductivity to illustrate the commercial and profitable business that is MRI. It provides the basics behind MRI and describes how images are acquired. It offers an insight into how commercial MR magnet production started and the history of SMT. It gives a description of how magnets are designed and how the design has evolved over a period of 20 years and finally an explanation of the various pieces of hardware that are used on a hospital MR whole body scanner.

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