Magnetic resonance imaging is a widely used noninvasive medical imaging technique to visualize the inner part of human body. It applied the basic principles of nuclear magnetic resonance (NMR) spectroscopy, which provides both chemical and physical information of molecules.

**Introduction**

Since a large fraction of composition of the human body is water and fat, the abundance of hydrogen atoms in fat and water makes the human body approximately 63% hydrogen atoms. MRI uses a powerful magnetic field to align the nuclear magnetization of hydrogen atoms in water in the body. When Radio frequency (RF) fields are added to systematically alter the alignment of this magnetization, the hydrogen nuclei produce a rotating magnetic field detectable by the scanner. Magnetic resonance imaging primarily images the NMR signals provided by the hydrogen nuclei inside the body. Those signals can be detected by gradient magnetic fields and build up enough information to construct an image of the body, especially important since it can display the different properties between normal tissues and tumor. The ability of depicting inner structure of substances also makes MRI of great use in physics, chemistry and many other areas.

Compared to PET and CT, MRI has a rather good resolution and is harmless to patient, since it doesn't require radioactive injections. This young and still growing technique should have a promising future. In 2003, there were approximately 10,000 MRI units worldwide, and approximately 75 million MRI scans per year performed. As the field of MRI continues to grow, so do the opportunities in MRI.

**History of NMR and MRI**

Back in the 1930s, Isidor Rabi conducted research on the nature of the force binding protons to atomic nuclei and eventually discovered that the spin orientation of nuclei in a magnetic field changes when radio frequency is added. This was the first human investigation on the interaction of atomic nuclei, magnetic field and radio frequency. He was thus awarded the Nobel Prize for Physics in the year of 1944 for this great work.

A decade later, in 1946, Purcel of Harvard University and F. Block of Stanford University independently discovered that magnetic nuclei in a magnetic field absorb and re-emit electromagnetic radiation. Their discovery of nuclear magnetic resonance and further expansion on this technique won them the Nobel Prize for Physics in 1952.

Immediate after the discovery of nuclear magnetic resonance phenomena, people started to apply this technique on investigation on nuclei structure and properties, such as the measurement of nuclear magnetic moment, as well as other practical researches. Chemists developed NMR spectroscopy according to the influences of molecular structure on the surround magnetic field of Hydrogen. NMR spectroscopy becomes a prevailing technique for molecular structure determination and it develops from one dimensional 1H spectroscopy to 13 carbon spectroscopy, two dimensional spectroscopy and other advanced methods. In the 1990s, scientist even applied this technique on protein, which made the accurate determination of molecular structure of liquid phase protein possible. Nowadays, nuclear magnetic resonance technique is widely used in various areas in chemistry, biology and medical research, from molecular structure and composition analysis to medical diagnose.
In 1971, R. Damadian found out that both the NMR longitudinal relaxation time (T1) and transverse relaxation time (T2) in tumor are longer than normal tissue, which means tumor and normal tissue can be distinguished by NMR in vivo. This work provided a magnificent chance of bringing NMR technique into medical research. Magnetic resonance imaging was first demonstrated on small test tube samples by Paul Lauterbur in 1973 and the first cross-sectional image of a living mouse was published in January 1974. In 1975 Richard Ernst proposed magnetic resonance imaging using phase and frequency encoding, and the Fourier Transform, which is the basis of current MRI techniques. While Richard Ernst was rewarded for his achievements in pulsed Fourier Transform NMR and MRI with the Nobel Prize in Chemistry in 1991, Paul C. Lauterbur of the University of Illinois and Sir Peter Mansfield of the University of Nottingham were awarded the Nobel Prize in Medicine for their discoveries concerning magnetic resonance imaging in 2003. To date, MRI is widely used for tumor detection and diagnose.

### Basic Principles

#### Spin physics

The hydrogen atom possesses a nuclear spin of 1/2. When placed in an external magnetic field, the spin vector of hydrogen atoms aligns itself with the magnetic field and the single energy state will split into two energy states. A particle in the lower energy state can absorb a photon, whose energy exactly matches the energy gap between the two states, and transit to the upper state. The energy of the photon can be described as:

\[
\text{E} = h \nu
\]

Where \( h \) stands for the Planck’s constant and \( \nu \) is called the resonance frequency and the Larmor frequency in MRI. It depends on the gyromagnetic ratio, \( \gamma \) of the particle as:

\[
\text{For hydrogen, } \gamma = 42.58 \text{ MHz/T.}
\]

Since the frequency of the photon lies in the radio frequency (RF) range in NMR experiment, \( \nu \) is between 60 and 800 MHz for hydrogen nuclei. In clinical MRI, \( \nu \) is typically between 15 and 80 MHz for hydrogen imaging. For example, at the most commonly used field strength of 1.5 Tesla, \( \nu \) is 63.87 MHz.

#### T1 processes

In the equilibrium state, the net magnetization vector lies along the direction of the external magnetic field \( B_0 \) and is called the equilibrium magnetization \( M_0 \). If we refer \( M_z \) as the longitudinal magnetization and \( M_x \) and \( M_y \) as the transverse magnetization, \( M_z \) equals \( M_0 \) while both \( M_x \) and \( M_y \) equal zero at the equilibrium state. However, if we apply the nuclear spin system to a radio frequency equal to the energy gap between the two spin states, the net magnetization can be changed and eventually saturate the spin system, so that \( M_z \) reaches 0. The so called spin-lattice relaxation time (\( T_1 \), also named as the longitudinal relaxation time) describes how \( M_z \) returns to its equilibrium value. The equation depicts this process is shown below:

\[
M_z = M_0 (1 - e^{-t/T_1})
\]

Obviously, \( T_1 \) is the time to reduce the difference between the longitudinal magnetization (\( M_z \)) and its equilibrium value by a factor of e.
If the net magnetization is saturated to the –Z axis (by an 180° pulse) and we record its return to the equilibrium state, the behavior can be described as:

\[ M_z = M_0 \left( 1 - 2e^{-t/T_1} \right) \]

**T₂ processes**

When the net magnetization was changed to the XY plane, after absorbing Energy from the radio frequency, it starts to dephase as different spin packets rotates at its own Larmor frequency due to experiencing slightly different magnetic field. The two main factors for this difference in magnetic field are molecular interactions and variations in B₀. The former is considered as a pure T₂ molecular effect while the latter is said to lead to an inhomogeneous T₂ effect. The combination of these two factors is the real result in the decay of transverse magnetization and the combined time constant is called T₂ star. The relationship is showed in the following equation:

\[ 1/T_2^* = 1/T_2 + 1/T_{2\text{inhomo}}. \]

To describes the return to equilibrium of the transverse magnetization, \( M_{XY} \), the time constant T₂ is defined as in the following equation.

\[ M_{XY} = M_{XY_0} e^{-t/T_2} \]

**Rotating frame, Bloch equations and pulse magnetic field**

If we define a rotating frame of reference which rotates about the Z axis at the Larmor frequency, it will be much easy to describe the behavior of the net magnetization vector in this frame than in the laboratory frame. In this case, the magnetization vector rotating at the Larmor frequency in the laboratory frame will be stationary in our rotating frame.

The Bloch equations are a set of coupled differential equations, used to describe the behavior of a magnetization vector under any conditions. When properly integrated, the Bloch equations will yield the \( X' \), \( Y' \), and Z components of magnetization as a function of time.

To create a magnetic field along the X axis, we can put a coil of wire around X axis and pass a current through it. An alternating current can create an oscillating magnetic field, but when we see it in the rotating frame, it just provides a constant magnetic field along the \( X' \) axis. This works just as we move the coil about the rotating frame coordinate system at the Larmor frequency. So the magnetic field generated by the coil passing an alternating current at the Larmor frequency is called the \( B_1 \) magnetic field. We can generate this pulsed magnetic field but turning the current on and off. 90° pulse and 180° are the mostly used pulses.
Magnetic field gradient

Magnetic field gradient makes it possible for different regions of spin to be exposed to a different magnetic field so that we are able to image their positions. In the following sections, we will use \( G_x \), \( G_y \), and \( G_z \) for a magnetic field gradient in the \( x \), \( y \) and \( z \) directions. The strength of the magnetic field increases along the axis.

A magnetic field gradient in the \( z \) direction is highly important for slice selection, which is the selection of spins in a plane through the object. When a 90° pulse is applied, together with our \( G_z \), only spins in the certain slice will be rotated to the \( XY \) plane. This is mainly because only spins in this slice matches the frequency provided by the RF, as showed in the following figure.

\[ \nu = \gamma ( B_0 + x G_x) = \nu_0 + \gamma x G_x \]

So when we turn on the phase encoding gradient, each transverse magnetization vector has its own unique Larmor frequency and the description of phase encoding is the same as frequency encoding.

In summary, the slice selection gradient is perpendicular to the slice plane while the phase encoding gradient is along one of the sides of the image plane and the frequency encoding gradient the other.

Pulse sequences

A pulse sequence is defined as a set of RF pulses applied to the object for a specific form of NMR signal. The 90-FID (free induction decay) sequence, spin-echo sequence and inversion recovery sequence are the most used ones.

In the 90-FID pulse sequence, net magnetization is rotated down into the \( X'Y' \) plane with a 90° pulse and the magnitude of the vector decays with time.

The spin-echo sequence also uses a 90° pulse at first so that the magnetization rotates down into the \( X'Y' \) plane and the transverse magnetization begins to dephase. However, an 180° pulse is then applied shortly after the 90° pulse and it rotates the magnetization by 180° about the \( X' \) axis. This creates a partially magnetization rephase and produces the called echosignal.

In the inversion recovery pulse sequence, an 180° pulse is first applied to rotate the net magnetization down to the -\( Z \) axis. The magnetization undergoes \( T_1 \) process and returns toward its equilibrium position along the +\( Z \) axis. A 90° pulse is applied before it reaches the equilibrium, so that the longitudinal magnetization is rotated into the \( XY \) plane.
Basic Imaging Techniques

Now we will talk about basic imaging techniques using various sequences. A good way to show how each imaging method works is to use timing diagrams. A timing diagram normally includes the radio frequency, magnetic field gradients, and signal as a function of time. For example, the simplest FT imaging sequence contains a 90° slice selective pulse, a slice selection gradient, a phase encoding gradient, a frequency encoding gradient, and a signal. The slice selection gradient and the slice selection RF pulse are the first things to be turned on. When they are complete, a phase encoding gradient is applied to the object. A frequency encoding gradient is turned on afterwards and a signal is recorded, which is in the form of a free induction decay. To collect all the data needed for an image, the sequence of pulses is usually repeated 128 or 256 times. We define the time between the repetitions of the sequence as the repetition time, TR. Each time the sequence is repeated, the magnitude of the phase encoding gradient is changed in equal steps between the maximum amplitude of the gradient and the minimum value. In this case, we can record 128 or 256 different free induction decays for resolving 128 or 256 locations in the phase encoding direction. Those recorded signals described above can be Fourier transformed to create an image of the location of spins.

A widely used parameter to quantify image quality is the signal-to-noise ratio (SNR). In an image, it is the ratio of the average signal for the tissue to the standard deviation of the noise in the background. The SNR of an MRI image has the following dependent factors: (1) the number of signal producing water molecules in the image voxel (2) the size of the signal each molecule produces, (3) the quality of the signal detection, and (4) the amount of spurious noise and the statistics of noise averaging.

Gradient-Echo Imaging

The gradient-echo sequence is intrinsically more sensitive to magnetic field inhomogeneities because of the use of the refocusing gradient or wind-up gradient. Together with a slice selection gradient, the slice selective RF is turned on to create a rotation angle of 90°. The phase encoding gradient $G_\phi$ is applied right after. Here, a negative dephasing frequency encoding gradient $-G_f$ is on at the same time to cause the spins to be in phase at the center of the acquisition period. When the frequency encoding gradient $G_f$ is applied, an echo is produced, because the gradient refocuses the dephasing which occurred from the dephasing gradient. This echo is named as a gradient echo, which is different from the echo created by an 180° pulse.

We define echo time (TE) as the time period from the start point of the RF pulse to the maximum of the signal. An equation to describe the signal intensity versus time and spin density can be written as:

$$S = k \rho (1 - \exp(-TR/T_1)) \exp(-TE/T_2^*)$$

Fig. 5 Gradient-Echo Imaging timing diagram

Spin-Echo Imaging

In spin-echo imaging, a spin-echo sequence is used and it displays the transverse relaxation time dependence to the
signal. This is especially useful for some tissues which have similar $T_1$ values but different $T_2$. So as showed in the timing diagram in figure 5, a slice selective 90\(^\circ\) pulse, together with the slice selection gradient, is applied to the object. After a time period of $TE/2$, an 180\(^\circ\) slice selective pulse is turned on in conjunction with the slice selection gradient. Between the two pulses, just like in the previous Gradient-echo imaging method, the phase encoding gradient $G_\phi$ and frequency encoding gradient $G_f$ are applied at the same time. This time, $G_f$ is necessary because it dephases the spins so that they will rephase by the center of the echo. Finally, the frequency encoding gradient applied after the 180\(^\circ\) pulse, during the time that echo is collected. The signal, which is the collected echo, is as follows:

$$S = k \rho \left(1 - \exp(-TR/T_1)\right) \exp(-TE/T_2)$$

Fig.6 Spin-Echo Imaging timing diagram

**Inversion Recovery Imaging**

An inversion recovery sequence is used in this imaging technique. It uses a spin-echo sequence to detect the signal so the RF pulses are 180-90-180. An inversion recovery sequence which uses a gradient-echo signal detection is similar, with the exception that a gradient-echo sequence is substituted for the spin-echo part of the sequence.

$$S = k \rho \left(1 - 2\exp(-TI/T_1) + \exp(-TR/T_1)\right)$$

Fig.7 Inversion Recovery Imaging timing diagram

**Multi-slice Imaging**

Multi-slice imaging makes a volume of anatomy to be imaged in the shortest time possible. The time to obtain an image is equal to the product of the $TR$ value and the number of phase encoding steps. While $TR$ takes time in seconds, the large steps of phase encoding last much longer. So when we take a careful look at the timing diagram of the gradient-echo sequence introduced above, we can easily find out that the most of the sequence time is unused. To make full use of the scanning time, other slices can be excited as show in figure 8. However, other RF frequencies of the 90\(^\circ\) should be used in order to prevent the interactions between those slices.

Fig.8 Multi-slice Imaging timing diagram

**$T_1$, $T_2$, and \(\rho\) Images**

Sometimes, properties of the spins in a tissue, such as the spin-lattice relaxation time ($T_1$), spin-spin relaxation time ($T_2$), and the spin density ($\rho$) are searched for. There are several methods of calculating $T_1$, $T_2$, and $\rho$ values. A $T_1$ image can be created from a series of images using the same pulse sequence with varying $TR$. The signal for a given pixel can be plotted for each $TR$ value and the best fit line from the spin-echo equation drawn through the data to find $T_1$, as show in figure 10.

Similarly, to produce a $T_2$ image, a series of images using a spin-echo pulse sequence with varying $TE$ is recorded. The signal for a given pixel can be plotted for each $TE$ value and the best fit line from the spin-echo equation drawn through
the data to find $T_2$, as shown in figure 12. The spin-echo equation is mentioned in the previous Pulse Sequences section. Once $T_1$ and $T_2$ are obtained, it’s easy to acquire spin density by using the spin echo signal equation and any spin echo signal.

**Fig.9** a series of imaging recorded with various TR under spin-echo sequence\[2\]

**Fig.10** signal – TR graph

**Fig. 11** a series of imaging recorded with various TE under spin-echo sequence\[2\]

**Fig. 12** signal – TE graph

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**Chemical contrast agent**

Although MRI has a good resolution and is very sensitive compared with other imaging modalities, such as positron emission tomography, chemical contrast agents (CA), substances which are introduced into the body to change the contrast between the tissues, play a vital role in the imaging procedure. Contrast agents may be injected intravenously to enhance the appearance of blood vessels, tumors or inflammation or directly injected into a joint in the case of arthograms. Normally, they can be divided into two main categories, namely $T_1$ contrast agents and $T_2$ contrast agents and they shorten the according relaxation time, respectively. The most commonly used compounds for contrast enhancement are gadolinium-based or iron-based. The former is paramagnetic and belongs to $T_1$ contrast agents, as it changes the contrast by creating time varying magnetic fields which promote spin-lattice and spin-spin relaxation of the water molecules. The latter is superparamagnetic and comes to the second group, as it changes $T_2$ of the water molecules around by distorting the $B_0$ magnetic field. Several well-known contrast agents are Magnevist (Gd-DPTA), Dotarem (Gd-DOTA) and superparamagnetic iron oxide (SPIO). Free Gadolinium ions in the human body may arose a certain kind of disease called nephrogenic systemic fibrosis (NSF). That’s why Gd needs to be connected with organic polymers with great care for safety reason.

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**Applications**

In clinical practice, MRI is used to distinguish pathologic tissue, such as a brain tumor, from normal tissue.

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**Hardware**

**Fig. 13** Hardware overview\[2\]

Figure 13 is an overview of the magnetic resonance imaging scanning system. Computer plays the most important role in this procedure as it controls all the components. The magnet produces the $B_0$ field which is critical for the imaging process, along with those gradient coils for gradients in $B_0$ in the X, Y, and Z directions. The RF coil generates the $B_1$ magnetic field for rotating the spins by $90^\circ$, $180^\circ$, or any other value selected by the pulse sequence and also functions as the signal detector from the spins within the body. The RF amplifier increases the pulses power from mW to kW while the gradient amplifier increases the power of the gradient pulses to a sufficient level to drive the gradient coils. Since
outer RF pulses may cause an additional impact on imaging, this scan room is enclosure with RF shields.

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Advantages and disadvantages

Despite of the rapid development and its improvement in the past several decades, MRI still has its shortcomings. The scan time of MRI for cardinal tissues lasts more than 30min, which is much longer than a CT scan and the expense of an MRI system is still high. Patients need to be stationary during the scan to prevent distortion of images. Besides, patients with pacemakers cannot have MRIs and claustrophobic patients cannot usually make it through a MRI. The machine makes a tremendous amount of noise during a scan due to the rising electrical current in the wires of the gradient.

Even with these disadvantages, we cannot ignore the tremendous contribution MRI has made in clinical practice in the past years. Because variations in $T_1$ and $T_2$ values are so much greater than variations in tissue density, MRI provides better soft-tissue contrast than plain radiography or computed tomography (CT).[3] In addition, it is easy to take a MRI slice in any direction, without changing the position of the patient. Compared to PET and CT, MRI is also harmless to the human body. The high resolution of imaging makes MRI an effective imaging modality and its opportunity in the future should be even more promising.

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References

2. The Basic of MRI, Dr. Joseph P. Hornak

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Contributors and Attributions

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