

BME 5030 Final Paper:

**Development of Radio-Frequency Coil
System in Magnetic Resonance Imaging**

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INTRODUCTION

Magnetic Resonance Imaging (MRI) system is a well-known non-invasive medical imaging instrumentation, which is commonly used to visualize high quality internal anatomical images and various bio-information and functions inside the human and the animal body. MRI is developed based on the knowledge of nuclear magnetic resonance (NMR), and various kinds of MR imaging techniques have been rapid developing to explore more sophisticated bio-information since the past decades.

Basically, a MRI system consists of three major hardware systems, which are:

1. Main Magnet
2. Gradient Coil
3. Radio Frequency Coil

The main magnet is mostly made from superconducting coils, which can provide a stable high magnetic field for imaging procedure. The gradient coil system is used to spatially encode the positions of protons

by varying the magnetic field linearly across the imaging volume, which is built within the main magnet. The RF system locates inside the main magnet and the gradient system, which can produce a B_1 magnetic field to rotate spins by different angles and transmit/receive an RF pulse from samples. [1] In the developments of MRI, these three systems have been remarkable advancing to improve imaging qualities of MRI since the first time the MRI was introduced to the world.

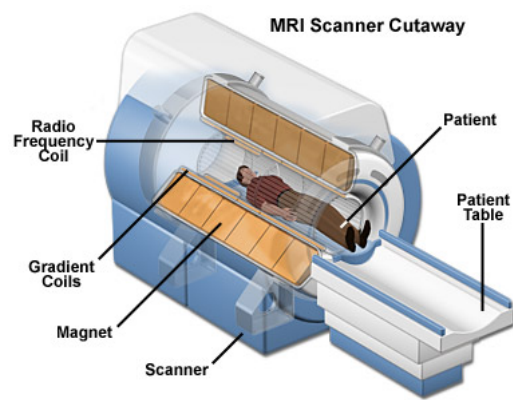


Figure 1 MRI scanner cutaway illustration From: <http://www.magnet.fsu.edu/education/tutorials/magnetacademy/mri/>

THEORY OF RADIO FREQUENCY COILS IN MRI SYSTEM

Basic transmitting and receiving concept of RF coils:

Radio frequency (RF) coil system is a critical component in a MRI system, which is an electrical device generally composed of multiple wire loops that can either generate a magnetic field or detect a changing magnetic field as an electric

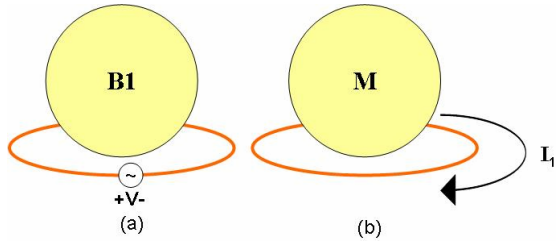


Figure 2 Illustration of transmitting and receiving concept. (a) a voltage source induces a B₁ field inside an object through an RF coil. (b) a magnetization within the subject induces a current in the RF coil.

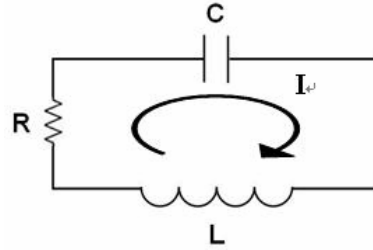


Figure 3. Equivalent circuit of a single loop RF coil

current induced in the wire. A RF coil system can be classified into three systems: 1) transmit and receive coil, 2) transmit only coil, and 3) receive only coil. A transmit and receive coil can send or transmit a B₁ field and receive RF energy from imaged samples. A transmit only coil is only used to create a B₁ field and a receive only coil is especially only used to detect signals from the imaged subject. In Figure 2, the illustrations show the brief transmitting and receiving concepts of a RF coil. After providing a transmitting excitation RF pulse, as shown in Figure 2(a), the current source within a RF coil induces a B₁ field inside the subject, which is called transmitting. In the Figure 2(b), the induced current in an RF coil is generated by the magnetization (M) in the imaged object, and this phenomenon is called receiving.

To understand the basic concepts of RF coils as resonators in MRI application, an equivalent circuit of a single loop RF surface coil is shown in Figure 3. Based on the Kirchhoff's law, the circuit can be expressed as [1]:

$$V(s) = I(s)\left(R + Ls + \frac{1}{Cs}\right) \tag{1}$$

Where $s = \sigma + i\omega$ is the angular frequency.

$V(s)$ is the voltage of the circuit.

$I(s)$ is the current in the circuit.

C is the equivalent capacitance.

L is the equivalent inductance.

R is the equivalent resistance.

The complex admittance $Y(s)$ can be derived as:

$$Y(s) = \frac{I(s)}{V(s)} = \frac{s}{L(s^2 + \frac{R}{L}s + \frac{1}{LC})} \quad (2)$$

Let $s = i\omega$, we can get

$$|Y(s)| = \frac{1}{\sqrt{R^2 + (\omega L - \frac{1}{\omega C})^2}} \quad (3)$$

Thus, a resonant frequency can be found as:

$$\omega_r = \frac{1}{\sqrt{LC}} \quad (4)$$

While $\omega = \omega_r$, this phenomenon is called resonance, and the current would be maximum at the resonant frequency. Thus, an RF coil can produce a desired magnetic field strength with a relatively low input voltage when it operates at the resonant frequency.

Q factor of RF coils:

Since the resistance of a RF coil is not zero, some energy will be dissipated in the circuit. Thus, a quality factor can be applied to determine the quantitative measurements of the coil quality, which is also known as a Q factor:

$$Q = 2\pi \frac{\text{maximum energy stored}}{\text{Total energy dissipated per period}} \quad (5)$$

Resonant systems respond to frequencies close to their natural frequency much more strongly than they respond to other frequencies. The Q factor indicates the amount of resistance to resonance in a system. Systems with a higher Q factor resonate with greater amplitude than systems with a low Q factor at resonant frequency. In a resonant system, the Q factor is defined as the resonant frequency ω_0 divided by the bandwidth (BW) $\Delta\omega$, which can be expressed as:

$$Q = \frac{\omega_0}{\Delta\omega} \quad (6)$$

In a tuned radio frequency receiver the Q factor is:

$$Q = \frac{1}{R} \sqrt{\frac{L}{C}} \quad (7)$$

From equation (4) and (7), the Q factor can be derived as:

$$Q = \frac{\omega L}{R} \quad (8)$$

The Q factor is a value that can measure the quality of RF coils under different circumstance.

Signal-to-noise ratio in RF coils:

The RF coil has always been considered as an importance role in MRI for achieving highest signal-to-noise ratio (SNR). Over the past decades, many different RF coils have been developed to enhance the quality of MRI images. In the evaluation of the quality of MRI images, Signal-to-noise ratio (SNR) is the best and most common way to describe. In MRI, the picked up voltage from sample voxel is a RMS value of the electromotive force (e.m.f), which is induced in the coil and can be expressed as [2][3]:

$$\xi_{rms} = - \int_{sample} \frac{\partial}{\partial t} \{ \vec{B}_1 \cdot \vec{M}_0 \} dV_s = \frac{1}{\sqrt{2}} \omega_0 B_1 M_0 V_s \quad (9)$$

Where B_1 represents a RF magnetic field, V_s represents the voxel volume, ω_0 represents the Larmor frequency, and M_0 represents the magnitude of the static equilibrium magnetization projected into the x-y or transverse plane. ξ_{rms} can be considered as the signal part in the MRI images.

In the noise terms for individual coil geometries, the noise power can be expressed as the well-known formula for Johnson thermal noise, which is [2][3]:

$$\sigma_n = \sqrt{4kTR\Delta f} \quad (10)$$

Where σ_n is the standard deviation of the noise power, k is the Boltzmann constant $k = 1.38 \times 10^{-23}$ (J/K), T is the operating temperature, Δf represents the effective noise bandwidth of the received signal, and R is the effective resistance. From equation (9) and (10), the SNR can be defined as [4]:

$$SNR = \frac{1}{\sqrt{2}} \frac{\omega_0 B_1 M_0 V_s}{\sqrt{4kTR\Delta f}} \quad (11)$$

Within the equation, the R is consisted of three different sources: 1) resistance of coils (R_{coil}), 2) losses from inductive interaction between magnetic field and sample (R_L), and 3) losses from dielectric interaction between electrical field and sample (R_D). Thus, the R in the equation (11) can be expressed as [4]:

$$R = R_{coil} + R_L + R_D \quad (12)$$

Based on the above equation, a simple case of the resistance of a simple single loop surface circuit is discussed first. In the above equation, the total resistance within the coil, R_{coil} , can be shown as:

$$R_{coil} = \frac{2\pi\rho r_0}{\delta w} \quad (13)$$

Where ρ is the resistivity of the coil, δ is the skin depth, w is the width of the strip line that forms the loop, and r_0 is the radius of the loop.

The sample resistance generated by magnetic field losses can be defined as:

$$R_L = \frac{\sigma\omega_0^2\mu_0^2r_0^3}{3} \quad (14)$$

The derivation assumes that the coil is a loop of radius r_0 positioned above a conducting half-space of conductivity σ .

The losses generated by stray electrical fields in MRI can be expressed as:

$$R_D \cong \frac{\omega_0 L^2 C_s^2}{R_c (C_s + C_c)} \quad (15)$$

Where L is the inductance of the coil, C_s is the lossless stray capacitance between the coil and sample, and C_c and R_c represent losses through the conductive pathways in the sample generated by the electric field.

Since R_D and R_L are all induced from the imaged sample, the equation (12) can be simplified as:

$$R = R_{coil} + R_{sample} \quad (16)$$

Therefore, the SNR of MRI images can be described as:

$$SNR = \frac{1}{\sqrt{2}} \frac{\omega_0 B_1 M_0 V_s}{\sqrt{4kT(R_{coil} + R_{sample})\Delta f}} \quad (17)$$

DEVELOPMENT OF RF COILS IN MRI SYSTEM

In the development of RF coils in MRI system, the goal always tends to enhance the SNR and reduce the imaging time. From the SNR equation in MRI (17), the relationship can be simplified to only consider the influence from a RF coil, which is:

$$SNR \propto \frac{B_1}{\sqrt{kT(R_{coil} + R_{sample})}} \quad (18)$$

In the equation, the B_1 field is related to the geometry of the RF coil and the distance between a sample and a RF coil. The T includes the temperature of sample and the temperature of the coil. Thus, the environment of a sample and a coil can also affect the SNR respectively. In addition, R_{coil} and R_{sample} that have been discussed in the above paragraph are also important factors that can influence SNR based on the geometry and the material of a RF coil. Therefore, to enhance the quality of MRI images under various conditions, current RF systems are developed in two directions: 1) structure, and 2) material. In the individual development, the purpose is to meet certain requirements in the imaging environment to achieve the best imaging quality.

Development in structure:

Basically, the development of RF coils in structure can be classified into three groups: 1) a surface coil, 2) a volume coil, and 3) an array coil.

1) Surface coil:

A surface coil is the most basic and fundamental RF coil system in MRI, which include single-loop and multiple-loop coils of various shapes. These coils are usually much smaller than the other types of coil system, thus it has higher SNR because they receive noises only from nearby regions. Several types of surface coils and the basic circuit for a surface RF coil are shown in Figure 4. In the circuit, the resonant

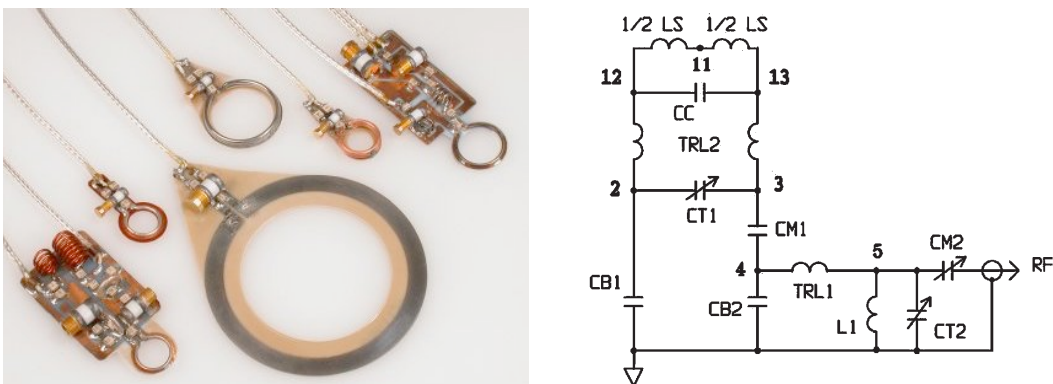


Figure 4. Left picture: various kinds of surface RF coils. [] Right picture: Basic circuit model of surface coil for the balanced T/R surface coil with remote tune/match. Transmission lines are designated by TRL_x , match capacitors by CM_x , balance capacitors by CB_x , and tune capacitors by CT_x . [5]

frequency and the quality of a surface coil can be controlled by the match capacitors, the balance capacitors, and the tune capacitors respectively. Since the RF coils would be affected under various circumstances, this circuit design can allow the coil to be adjusted to the best resonant condition.

2) Volume coil:

Volume coils basically include solenoid coils, saddle coils, and high pass and low pass birdcage coils. Within these coils, the birdcage coils are most popular because they can produce better homogeneous B_1 field over a large volume within the coil [6]. Volume coils are usually used for surrounding the whole body or a specific region, because they can cover larger imaging area and provide better magnetic field homogeneity in a large area than surface coils. A volume coil is usually used for brain (head) MRI, or MR imaging of joints, such as the wrist or knees. The basic schematics of the three basic volume coils: a birdcage coil, a solenoid coil and a saddle coil, are shown in the Figure 5, which are drawn in three dimensions. In the schematics, the capacitors are used to match the resonant frequency of MRI system [6].

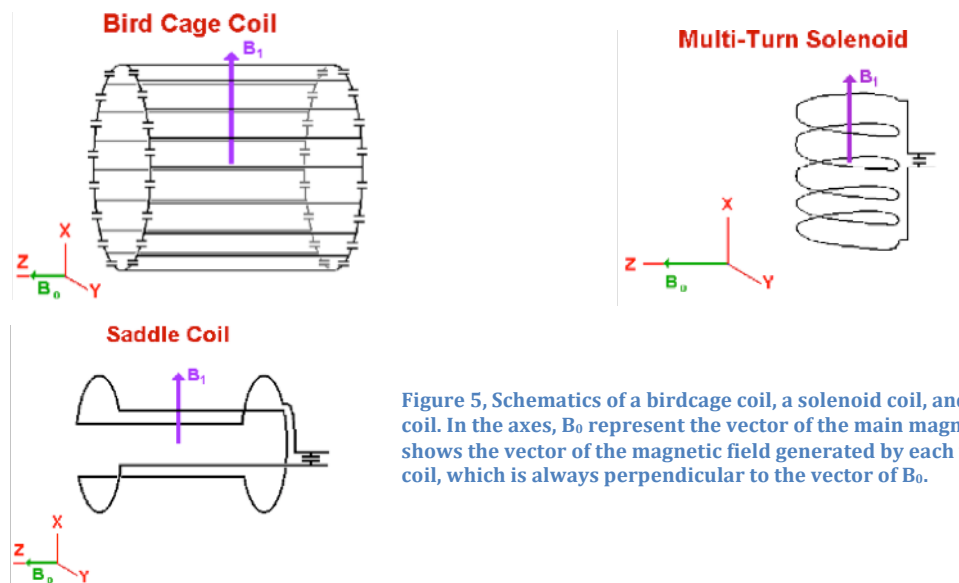


Figure 5, Schematics of a birdcage coil, a solenoid coil, and a saddle coil. In the axes, B_0 represent the vector of the main magnet. B_1 shows the vector of the magnetic field generated by each volume coil, which is always perpendicular to the vector of B_0 .

3) Array coil:

An array coil combines the advantages of a smaller surface coil and a larger coil, which can provide high SNR and cover a large imaging area. This type of RF coil consists of multiple smaller coils, which can be used individually or simultaneously. When an array coil is used individually, it can be considered as an individual signal

coil. When it is used simultaneously, there are three different types that can be utilized:

- Coupled array coils - electrically coupled to each other through common transmission lines or mutual inductance.
- Isolated array coils - electrically isolated from each other with separate transmission lines and receivers and minimum effective mutual inductance, and with the signals from each transmission line processed independently or at different frequencies
- Phased array coils - multiple small coils arranged to efficiently cover a specific anatomic region and obtain high-resolution, high-SNR images of a larger volume.

The data from the individual coils is integrated by software to produce the high-resolution images.

The schematics of commonly used array coils are shown in the Figure 6.[7]

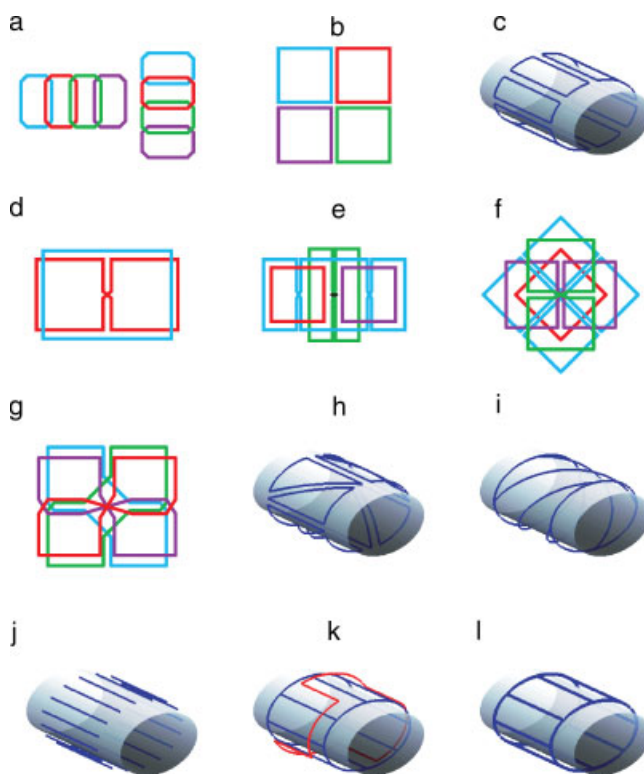


Figure 6. Schematics of array coils. The shown geometrical features of each coil array are the major design of array coils, and the element numbers and coil dimensions may be varied. (a) Linear arrays of loop coils; (b) 2x2 grid of loop coils. (c) 'Wrap-around' arrangement of eight loop coils. (d) Quadrature pair of butterfly and loop coils. (e) 'Saddle-train' coil; red and purple elements are simple loops, green element is a single butterfly, and blue element is a double-twisted saddle train coil. (f) 'Concentric' coil array; the red coil is a simple loop, green and purple elements are two-lobed butterfly coils in perpendicular orientations, the blue coil is a four-lobed cloverleaf element. (g) 'Diagonal' coil array; blue and green coils are top logically simple loops in diagonal orientations; red and purple elements are crossed saddle elements, also in diagonal orientations. (h) Triangular coil array, with eight right-triangular elements. (i) Spiral birdcage coil. (j) TEM-resonator arrays. (k) Degenerate-mode birdcage coil. Red conductor is used to simultaneously resonate the uniform and gradient modes. (l) Birdcage coil designed to produce spatial harmonic sensitivities. [7]

Development in material:

From the above section, we can understand the theory and the basic concept of SNR and Q factor in MRI RF system. In the equation (18), we can find that the material of RF coils is an important factor for SNR. Considering the equation in more specific condition, the relationship can be expressed as:

$$SNR \propto \frac{B_1}{\sqrt{(T_{coil}R_{coil} + T_{sample}R_{sample})}} \quad (19)$$

Where T_{coil} and T_{sample} are the temperature of a coil and an imaged sample respectively. From the above equation, we can expect a SNR improvement by applying material with lower resistance or lower the environment temperature of coils. The SNR gain can be expressed as:

$$SNR_{gain} \propto \frac{\sqrt{(T_{coil}^{copper}R_{coil}^{copper} + T_{sample}R_{sample})}}{\sqrt{(T_{coil}^1R_{coil}^1 + T_{sample}R_{sample})}} \quad (20)$$

Here we assume the comparison is between an improved RF coil and a conventional copper RF coil in the same configuration, thus we can ignore the effect of B_1 field. From equation (20), we can find that SNR gain will be more significant if the R_{sample} is lower. Therefore, in the small animal imaging, reducing the noise from a coil allows for a significant increase in SNR. In the following, we will discuss three commonly used means to improve SNR in the same RF coil configuration.

1) Silver RF coils:

Silver is usually used to replace conventional copper material to manufacture a RF coil because of its lower resistivity than copper. Comparing the resistivity of silver ($1.59 \times 10^{-8} (\Omega m)$) with the resistivity of copper ($1.72 \times 10^{-8} (\Omega m)$), we can find that the resistance of a silver RF coil is smaller than the resistance of a copper RF coil in the same configuration. Thus, the SNR acquired by using a silver coil would be higher than the SNR acquired by using a copper coil, and the SNR gain varies with different imaged samples.

2) Cryogenic RF coils:

From equation (20), we can find that the SNR would be enhanced by reducing the temperature of a cryogenic coil. Thus, various cryogenic RF coils have

been developed to improve the imaging quality in animal MRI or MRS. Liquid nitrogen is the most used medium to cool down the temperature of a RF coil to the temperature of 77K. The SNR gain in animal imaging can achieve around 1.5 – 2 folders comparing with the coils in normal condition (room temperature)[8].

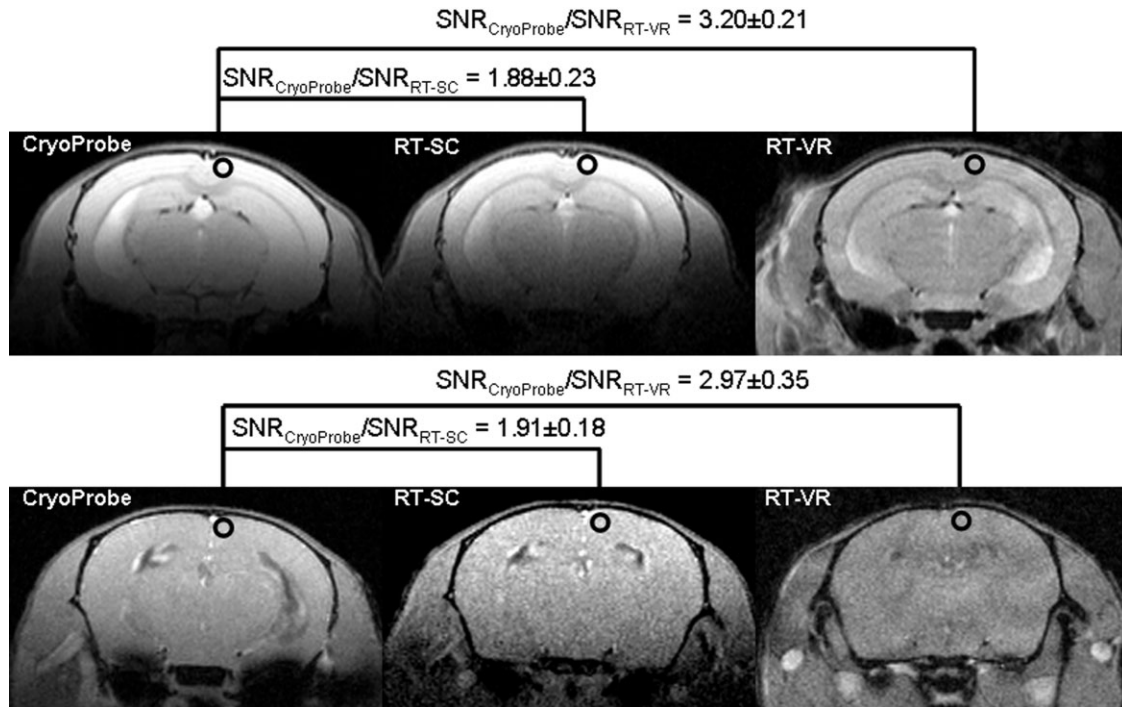


Figure 7. In vivo axial images of the murine brain. Images were acquired using the spin echo (upper row) and gradient echo sequences (lower row). Left: cryogenic probe (CryoProbe); Center: room temperature surface coil (RT-SC); Right: room temperature volume resonator (RT-VR). The circles indicate the analyzed ROIs and the individual SNR gains are specified. [8]

3) High temperature superconducting RF coils:

In the above two methods, we have already understood that SNR can be improved by reducing the noise of a coil and the temperature of a coil. To further enhance the imaging quality, high-temperature superconducting (HTS) material is applied to the RF coils to provide much lower coil resistance (almost zero) while the HTS RF coil achieves the superconducting status in the critical low temperature.

A HTS RF coil was first implemented by R.D. Black et al. in 1993 [9]. The HTS RF coil was manufactured from YBCO material, and the coil was patterned into a split ring shape (18-mm outer and 14-mm inner diameter) on each side of the substrate. SNR improvement of about 10 times was observed from phantom imaging results at 7-Tesla and at 4.2K compared with a room temperature copper

solenoid RF coil. As pioneers of HTS RF system, R.D. Black et al [9] provided an insightful analysis on the design principle of the signal coupling network and the scaling behavior of sample and coil noise. Thus, this is also an important reference for later researches. In the following developments, S.E. Hurlston et al [10] manufactured a HTS Helmholtz probe for microscopy at 9.4T MRI system. This HTS RF coil was also made from YBCO in a thin film type, but it was designed especially for a homogeneous resonator. The SNR gain could achieve around 7 in the comparison with a copper Helmholtz coil. In the above examples, the HTS RF coils were all manufactured from thin film coils. Hidehiko et al. first demonstrated the concept of using flexible HTS tapes to manufacture RF coils in 1994. Hidehiko et al. has developed a large size (31x34 cm) HTS spine coil using homemade Bi-2223 coated HTS tape. The SNR improvement was observed about 1.6 times in spine images by using HTS RF coil at 77K compared with a copper coil in the same configuration at room temperature. In 2005, Lee et al. developed a HTS receiving coil for MRI. Commercial Bi2223 tape coils were used to manufacture HTS RF coils [12]. The coils were designed in the type of surface coils, and the SNR gain was achieved around 2.56 in the kiwi imaging compared with a copper coil at 300K, and the SNR gain was 2 in the comparison between a HTS RF coil and a copper coil at 77K in murine brain images [11].

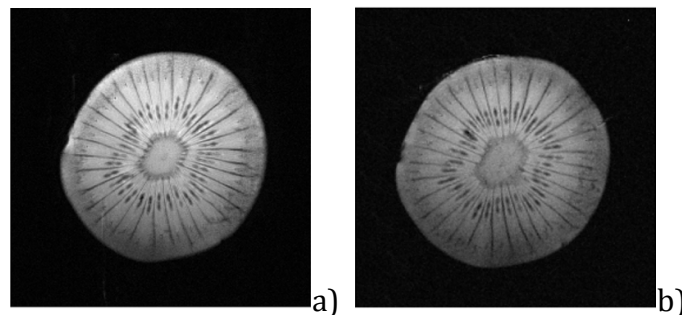


Figure 8. Images of Kiwi a) Using Bi2223 tape HTS RF coil at 77.4K, which SNR = 77.1, b)Using conventional copper coil at 300K, which SNR = 30. The SNR gain is 2.58 [11]

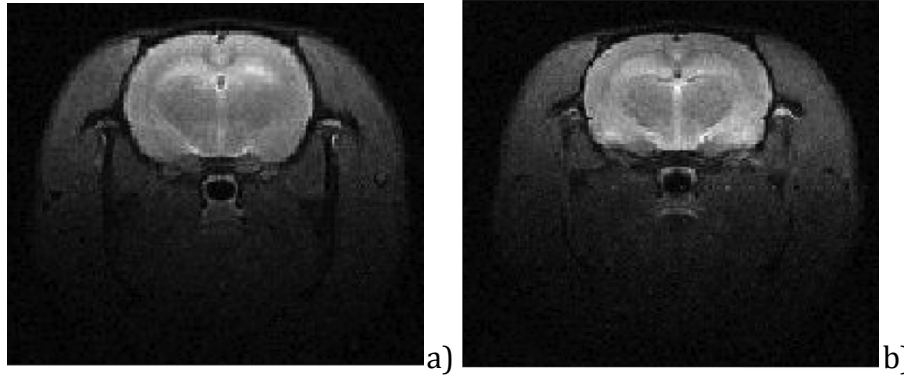


Figure 9. Images of murine brain. a) Using a HTS coil at 77.4K, SNR = 60. b) Using a copper coil at 77.4K, SNR = 30. The SNR gain is 2 by comparing these two images.[11]

FUTURE OF MRI RF SYSTEM:

In the past decades, the MRI RF system has been advancing in various means to improve the performance of MRI. Based on the previous development, the MRI RF system can be further improved to acquire with higher resolution and higher SNR. In the following, we are going to discuss 4 novel MRI RF systems that might lead to significant influence for the future MRI.

1) Multi channel RF array system:

The number of channels in phased array RF coil has been advancing from 2 to 256 in the past years. As mentioned above, the more channels for an array coil can provide larger imaging area but still maintain high SNR, which combines the advantages of both large coil and small coil. In the following development of array coils, the performance of array can be significantly enhanced by increasing the number of channels, improving the preamp and decoupling circuit, and changing the material. Based on the improvement of hardware, the array coils can achieve higher accelerating factors with high imaging quality by applying the parallel imaging techniques.

2) Micro-MRI RF coils:

A micro-MRI RF coil is the coil especially designed for the purpose of micro MR imaging. The modern RF coils basically are still mainly designed for imaging certain area of organs (brain, spine, knee, etc). With the development of RF coils, scientists try to design RF coils that can provide higher resolution for small regional images and provide more bio-information in detail. Thus, micro-MRI RF coils are rapidly developed to investigate the MRI images from micrometer to nanometer

scale. [12][13] In the future, a micro RF coil will play an important role in molecular imaging and high-resolution imaging, which can unveil the hidden information that has not been discovered until now.

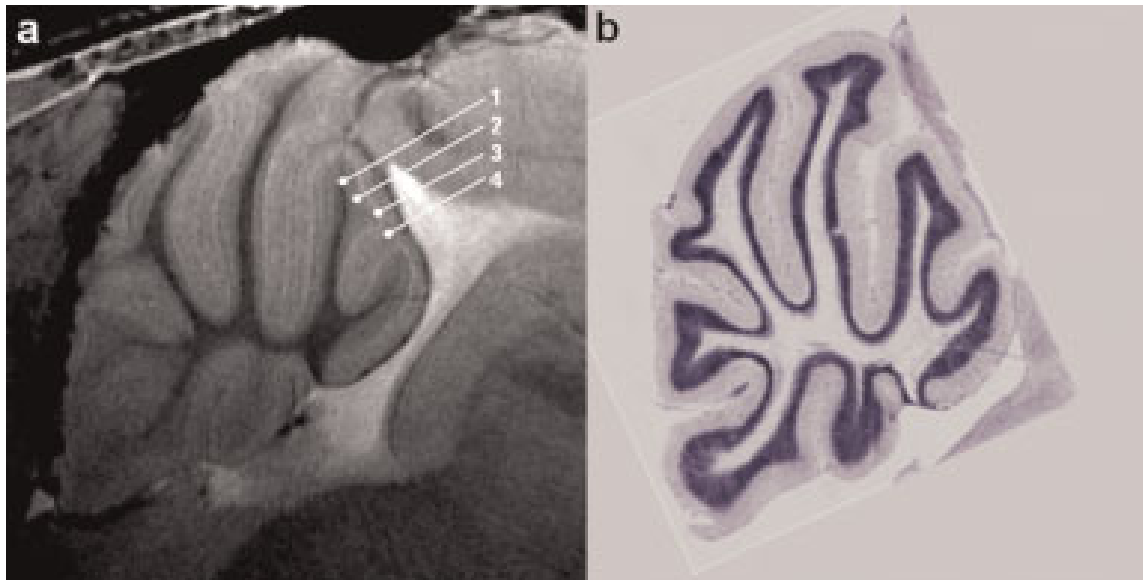


Figure 10. Structural in vivo micro MR image of the mouse cerebellum (a) and the corresponding histological section (Nissl staining) (b). An SNR value of 32 was found for the ROI indicated in Fig. 3a. The following micro-structures can be identified on the MR image: 1 1/4 white matter; 2 1/4 granular layer; 3 1/4 Purkinje cell layer; 4 1/4 molecular layer. The resolution of image (a) is 30 μm x 30 μm . [13]

3) Traveling-wave RF system:

In conventional MRI RF system, the signal detection is based on Faraday induction between a RF coil and an imaged sample, which requires one or multiple RF coils to be close to the sample. The traveling-wave RF system is a novel MRI RF system that can excite and receive signals by long-range interaction, which is developed based on traveling RF waves transmitted and received by an antenna. In 2009, Brunner et al. [15] successfully implement this method, and they also demonstrate a uniform in-vivo MRI image. The benefit of this RF system is that it can provide more uniform coverage of samples that are larger than the wavelength of the MRI signal, and the conventional RF coils that are close to a sample are no longer needed. However, this system is still not mature, and there are still some problems that need to be solved. Once the theory has been developed for more mature area, the full benefits can be realized more.

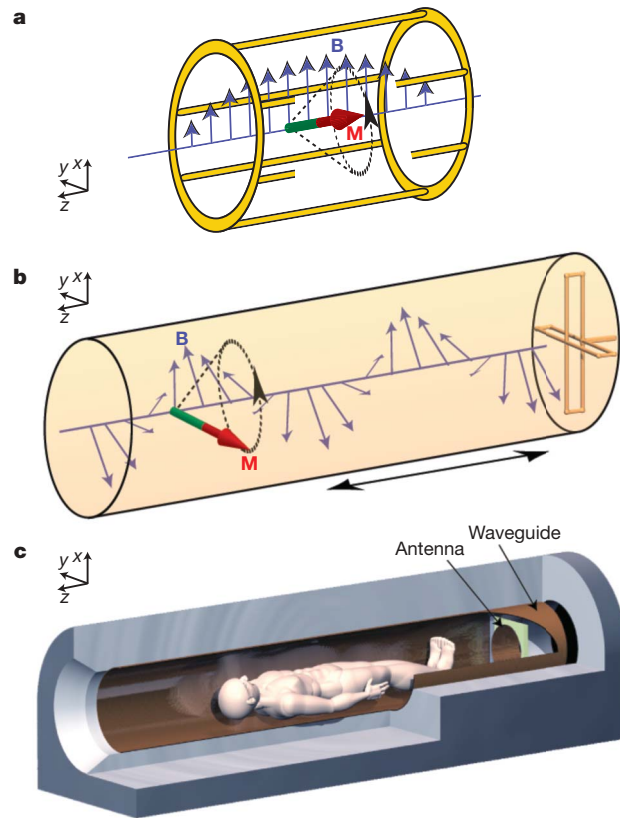


Figure 11. Concept of traveling-wave RF system. a, Traditional resonant probes transmit a RF wave within the sample. It generates B field that causes nutation of nuclear magnetization, M. b, In this approach, an antenna probe is utilized to receive the signals from sample through a travelling wave. c, In a wide-bore, high-field magnet, such waves can be guided by a conductive lining, permitting remote MRI excitation and detection with an antenna at the end of the magnet[15].

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