A novel radio frequency coil for veterinary magnetic resonance imaging system*

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In this article, a novel designed radio frequency (RF) coil is designed and built for the imaging of puppies in a V-shape permanent magnetic resonance imaging (MRI) system. Two sets of Helmholtz coil pairs with a V-shape structure are used to improve the holding of an animal in the coil. The homogeneity and the sensitivity of the RF field in the coil are analysed by theoretical calculation. The size and the shape of the new coil are optimized and validated by simulation through using the finite element method (FEM). Good magnetic resonance (MR) images are achieved on a shepherd dog.

Keywords: veterinary magnetic resonance imaging (MRI), radio frequency (RF) coil, homogeneity, receiving sensitivity

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1. Introduction

The radio frequency (RF) coil is recognized as being of great importance for the image quality in magnetic resonance imaging (MRI). Unlike with a human, it is best to let the patient (animal) lie on its back with its legs up in veterinary MRI. In order to provide enough accessibility to the veterinary patient, an RF coil with a special structure is preferred. However, the majority of published articles on MRI RF coils focused on humans,[1−9] only a minority involved small size animals,[10−14] and few touched upon large size animals. And most of the coils were designed for superconducting magnet MRI systems with a cylinder-like structure or other shape structures that cannot provide enough accessibility to veterinary patient especially for the MRI imaging of big size puppy.

Recently, a V-shape wide gap open MRI system with a permanent magnet of 0.35Tesla has been developed for the imaging of veterinary patients exclusively, which is shown in Fig. 1. The new structure allows the access of animals of different sizes like big dog and small cat. The configuration of permanent magnet makes the new MRI system simple and cost effective. A V-shape patient table has been designed such that animal patients lie on their backs with their legs up within the magnet during scanning. This design optimizes the animal's orientation and position for spinal cord imaging. However, traditional RF coils mentioned above are not applicable to this novel design of the MRI system.



Fig. 1. V-shape MRI system with a 0.35 T permanent magnet.

Planar transmit coils close to magnet pole pieces,[1,2] which are required in most of open MRI systems, are eliminated to save the gap for patient. Surface coils with loop structure, though they have high signal-to-noise ratio (SNR) and are widely used in superconducting MRI system, are not suitable for the V-shape MRI system because of the poor B_1 field homogeneity.[3,4] Birdcage coils[5−7] have been a preferable option for many applications in horizontal bore MRI systems due to their high SNR and good RF field uniformity over a large field-of-view (FOV),[15] but they cannot be used in our new MRI system, for

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some of B_1 components are parallel to the B_0 direction in our new design of MRI system. Solenoid coils are known to have advantages for vertical field systems in sensitivity and uniformity.[16] Lee *et al.* have developed a solenoid receiving RF coil for small animal whole body imaging in a 3T superconducting MRI system.[14] In their design, another volume coil is an essential part for transmitting, which increases the requirement for patient gap that is very limited in permanent MRI magnet. But coils with a similar structure, such as Helmholtz pairs, are perfect for this veterinary MRI system.

In this study, we describe a transmit/receive RF coil with a modified structure of V-shape. In the new design, Helmholtz pairs serve as a basic unit to obtain high receiving sensitivity and adequate B_1 homogeneity over a large volume. The distance and the size of the Helmholtz pairs are optimized for the MR imaging of big dogs. Good images on a shepherd dog are obtained.

2. Structure analyses

Helmholtz pair as described by cylindrical coordinates of (ρ, θ, z) in Fig. 2(a), is found to serve as RF coils for applications in MRI because it can generate a uniform magnetic field in the vicinity of its axis and around the position $z = 0$. The magnetic flux density B in the space can be calculated by Biot–Savart law, and given by

$$B_z = \sum_{i=1}^{2} \frac{\mu_0 I k_i}{4\pi\sqrt{a\rho}} \left[K(k_i) + \frac{a^2 - \rho^2 - z_i^2}{(a-\rho)^2 + z_i^2} E(k_i) \right], \tag{1a}$$

$$B_\rho = \sum_{i=1}^{2} \frac{\mu_0 I k_i z_i}{4\pi\rho\sqrt{a\rho}} \left[-K(k_i) + \frac{a^2 + \rho^2 + z_i^2}{(a-\rho)^2 + z_i^2} E(k_i) \right], \tag{1b}$$

$$B_\theta = 0, \tag{1c}$$

where

$$z_i = z + (-1)^i \cdot \frac{a}{2}, \quad k_i = \sqrt{\frac{4a\rho}{(a+\rho)^2 + z_i^2}}, \tag{2}$$

$\mu_0 = 4\pi \times 10^{-7}$H per m (H/m) is the permeability of free space, I is the electric current carried in each ring, a is the radius of the ring, and $K(k)$ and $E(k)$ are Legendre's complete elliptic integrals of the first and second kinds and defined, respectively, as

$$K(k) = \int_0^{\pi/2} \frac{\mathrm{d}\theta}{\sqrt{1 - k^2 \sin^2\theta}}, \tag{3a}$$

$$E(k) = \int_0^{\pi/2} \sqrt{1 - k^2 \sin^2\theta}\,\mathrm{d}\theta. \tag{3b}$$

The inflection point of the field homogeneity is at $a/2$ from the coil centre along the axis. The radius of the coil is set to be 15 cm, ensuring a good field homogeneity over a region of approximately 15 cm, which is enough for the veterinary patient. Such a configuration matches the gap of the magnet. Larger coils offer a better field homogeneity but reduce the filling factor, thereby resulting in a lower SNR.

In order to increase the sensitivity of the coil, two Helmholtz coil pairs of radius a, separated by distance b, are combined together as shown in Fig. 2(b).

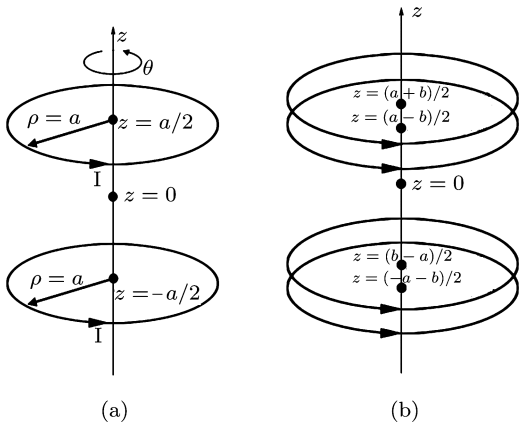


Fig. 2. Helmholtz coil pair (a) and Two Helmholtz pairs with coil radius a and separation b (b).

The magnetic field produced by these two pairs is the sum of the fields produced by each one. We obtain the magnetic field along the z-axis to be

$$B'_z = \sum_{i=1}^{4} \frac{\mu_0 I a^2}{2\left(d_i^2 + a^2\right)^{3/2}}, \tag{4}$$

where

$$d_1 = \frac{a+b}{2} + z,\ d_2 = \frac{a+b}{2} - z,\ d_3 = \frac{a-b}{2} + z,$$

and

$$d_4 = \frac{a-b}{2} - z. \tag{5}$$

Also μ_0 is the permeability of free space and I is the electric current carried in each ring. The field strength at the central point $z = 0$ is then

$$B_m = B'|_{z=0} = \frac{\mu_0 I a^2}{\left[(a+b)^2/4 + a^2\right]^{3/2}}$$

$$+\frac{\mu_0 I a^2}{\left[(a-b)^2/4+a^2\right]^{3/2}}. \quad (6)$$

We define a length l along z-axis such that the following condition is satisfied:

$$\left|B'|_{|z|<l} - B'|_{z=0}\right| < 0.05 \times B'|_{z=0}\,.$$

Therefore, the field homogeneity and the receiving sensitivity of the coil can be represented by l and B_m/I respectively. Both l and B_m/I are strongly related to the distance b between the two pairs. Their relationships are shown in Figs. 3(a) and 3(b).

It can be seen from Fig. 3 that the relationship between l and b is more complex than that between B_m and b. In Fig. 3(a), a peak point appears where b is about doubled a. When b is chosen as this peak point, a homogeneous RF magnetic field of a large volume is achieved. Meanwhile Fig. 3(b) shows that this configuration corresponds to a weak RF field and results in a poor receiving sensitivity. However, when b is rather small compared with a, the volume of the homogeneous RF field is almost the same as that produced by a single Helmholtz coil pair, and the receiving sensitivity will be significantly improved.

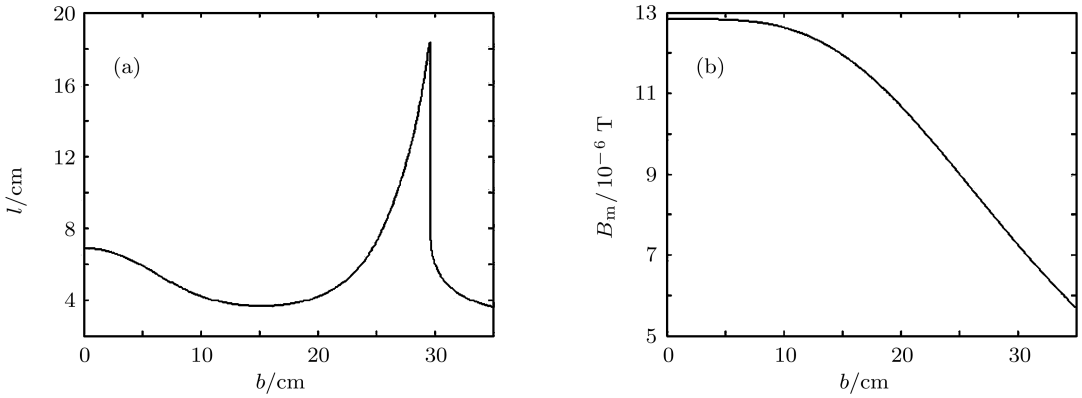


Fig. 3. Plots for (a) distance l versus b, and (b) field strength $B_{\rm m}$ versus b with $a = 14$ cm and $I = 1$ A in calculation.

3. Coil design and simulation

A novel RF coil for our new veterinary MRI system is designed. To fit the V-shape patient table in the new MRI system, the Helmholtz coil pairs are shaped into an English letter 'V', which are shown in Fig. 4(a). The variation of the coil geometry will change the RF field homogeneity. In order to make the shape of the new coil as similar to the standard Helmholtz coil pairs as possible, each 'V' shape coil is composed of several short straight lines (Fig. 4(a)).

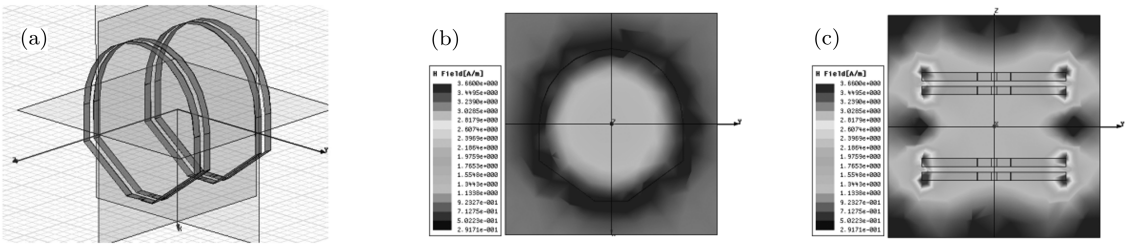


Fig. 4. Simulation results of (a) coil model, (b) B1 field map in x–y plane at $z = 0$, and (c) B1 field map in y–z plane at $x = 0$.

This new RF coil is made of copper tape of 15 mm in width and 0.2 mm in thickness, wrapped on a plastic frame as shown in Fig. 5. The distance between the two loops of a single Helmholtz coil pair is 155 mm and the two Helmholtz coil pairs are 25 mm apart. Capacitors (Temex Ceramics) are used for tuning and matching. The coil is tuned at 15.3 MHz and matched to 50 Ω.

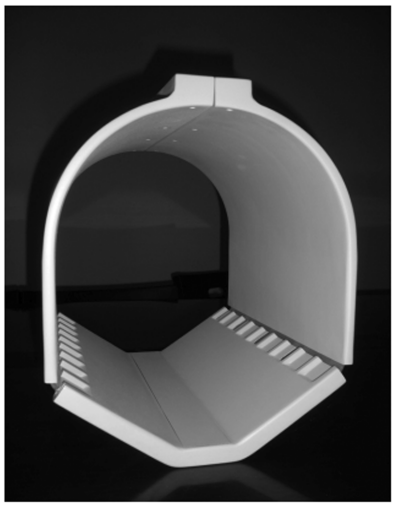


Fig. 5. Picture of our designed coil.

In order to achieve a high filling factor, the coil is almost the same size as the puppy's body. Unlike small animals, it is difficult to put the veterinary patient into such a closed coil. In order to achieve maximal patient comfort, plastic pivots are used to ensure that the whole coil can be open for the patient to enter into and exit from the closed coil. While scanning, copper plugs and plastic clips keep the coil closed and working regularly.

A simulation calculation of the distribution of RF field in the coils is performed by the use of a finite element method (FEM) program. The purpose of this simulation is to confirm the homogeneity of the RF magnetic field and optimize the configuration of the coil. The results are shown in Fig. 4. The z-axis is vertical to the planes of the coil loop (Fig. 4(a)), and the distributions of the magnetic field strengths in the x–y plane at $z = 0$ and the y–z plane at $x = 0$ are shown in Figs. 4(b) and 4(c) respectively. As shown in Fig. 4, a very homogeneous RF magnetic field is obtained in the central region within the new coil. In the y–z plane in Fig. 4(c), it can be seen that the homogeneity of RF field is deteriorated near the area outside the coil wire. However, inside the coil, a good homogeneity of RF field is achieved.

4. Experiments and results

MRI experiments were performed on a phantom and a shepherd dog. The phantom for SNR measurements consists of a 10 cm length cubic plastic tube, filled with water, doped with $CuSO_4$. The three-dimensional gradient echo (3D GRE) images of the phantom were acquired to have matrix size 256×133×7, time to repetition (TR) = 40 ms echo time (TE) = 7 ms, flip angle = 50°, slice thickness = 3 mm, FOV = 147×184 mm, and number of excitations (NEX) = 1. The GE-MRI image of the phantom (Fig. 6(a)) illustrates a very homogeneous distribution of image intensities across the phantom. It proves that the RF field produced by this new design of coil is very homogeneous over a large volume.

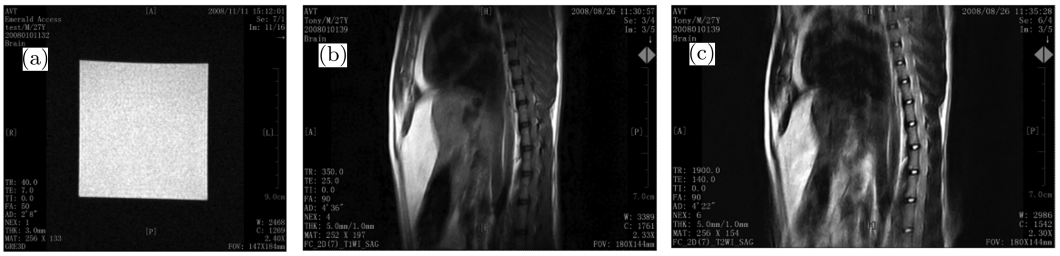


Fig. 6. Images of phantom (a), T1-weighted spine (b), and T2-weighted spine (c).

A shepherd dog with a length of about half a meter was anesthetized first, lay down on its back in the V-shape MRI system, and then it was scanned by MR in the V-shape coil. Lumbar images of T1 and T2 were obtained by spin echo (SE) and fast spin echo (FSE) sequence respectively. The MR parameters used to obtain the T1-weighted images had matrix size 252×197, TR = 350 ms, TE = 25 ms, slice thickness = 5 mm, FOV = 180×144 mm, and NEX = 4. The parameters for the T2-weighted images had matrix size 256×154, TR = 1900 ms, TE = 140 ms, slice thickness = 5 mm, FOV = 180×144 mm, and NEX = 6. The imaging

results are shown in Figs. 6(b) and 6(c). Figure 6(b) shows the coronal T1-weighted image and Fig. 6(c) exhibits the T2-weighted image of the spine of the dog. The spinal cord can be recognized clearly in both T1 and T2 MR images. As expected, there is a long and wide imaging area existing in the new design of coil, which is enough for the imaging of the lumbar of bigger dog. It proves that a homogeneous RF field of a large volume is produced by the new coil and the SNR is good enough for puppy imaging.

5. Discussion

The non-invasive nature of imaging makes MRI an ideal tool for diagnosing not only human patients, but also veterinary patients. The quality of the MR images can be simply assessed in terms of resolution and contrast, both of which depend directly on the receiving sensitivity that is dominated by the RF coils and the imaging sequence used to acquire the image. The novel V-shape coil used in permanent magnet possesses an imaging area that is big enough for the imaging of bigger puppy spine, and make MRI much more useful and cost-effective in the diagnosis of veterinary patients.

The RF field homogeneity and the receiving sensitivity are both important for RF coil. However, in most cases, the two characters cannot be optimized at the same time. One solution is to design transmitting coil and receiving coil separately,[13,17] but much attention must be paid to decoupling between the coils and the coil complexity increases. In this study, it is preferred to design an RF coil both for transmitting and for receiving due to the limitation of imaging gap of the magnet. However, the field homogeneity and receiving sensitivity must be balanced. In the current design, if the four rings are distributed uniformly, they can be considered as a solenoid coil and a high field homogeneity is achieved, but the receiving sensitivity is not the best. In this study, the four rings act as two pairs of Helmholtz coils close to each other, thereby achieving both high field homogeneity and high receiving sensitivity. Homogeneous RF excitation across a big area with about 15 cm in length is produced by the new coils during the transmitting period and high receiving sensitivity is achieved during the receiving period, which has been confirmed both by the simulation results and by the images acquired. Additionally, the new structure of the coil offer the highest possible filling factor for animal imaging.

The outline of the new coil is made look like an English letter 'V' to provide the better holding of animal. The RF field produced by such coils is similar to but not exactly the same as that produced by traditional Helmholtz coil pairs. However, from the results of FEM simulation and actual MRI images acquired, it is reasonable to use the standard Helmholtz coil pairs to replace the V-shape coils from the theoretic analysis and calculation.

Coils used for further larger puppies need more wires of ring for the field homogeneity. The distribution of all the rings becomes more complex. However, Helmholtz coil pairs still can be used as basic units. More Helmholtz coil pairs can be combined to improve the field homogeneity and the receiving sensitivity.

6. Conclusion

A novel RF coil is designed and built for the imaging of puppies in a V-shape permanent MRI system. It is proved that two sets of Helmholtz coil pairs with a V-shape structure can produce an imaging area which is big enough for the imaging of the lumbar of a bigger animal. The new coil will be useful and cost-effective in the diagnosis of veterinary patients.

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