Shielding of Sensitive Electronic Devices in Magnetic Nanoparticle Hyperthermia Using Arrays of Coils

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Abstract. In Magnetic Nanoparticle Hyperthermia (MNH) an externally applied electromagnetic field transfers energy to the magnetic nanoparticles in the body, which in turn convert this energy into heat, thus locally heating the tissue they are located in. This external electromagnetic field is sufficiently strong so as to cause interference and affect sensitive electronic equipment. Standard shielding of magnetic fields involves Faraday cages or coating with high-permeability shielding alloys; however, these techniques cannot be used with optically sensitive devices, such as those employed in Optical Coherence Tomography or radionuclide imaging. In this work we present a method to achieve magnetic shielding using an array of coils. The magnetic field generated by a single coil was calculated using the COMSOL physics simulation toolkit. Software was written in C/C++ to import the single-coil data, and then calculate the positions, number of turns and currents in the shielding coils in order to minimize the magnetic field strength at the desired location. Simulations and calculations have shown that just two shielding coils can reduce the magnetic field by 2-3 orders of magnitude.

1. Introduction

Hyperthermia (HT) in Oncology refers to the artificial increase of local temperature for the treatment of cancer, typically in the range of 41° C to 45° C for 1 hour or more [1, 2]. Although hyperthermia alone does not seem to have a curative effect, there is significant evidence that a strong synergistic effect is achieved when HT is combined with other treatment modalities. With Radiation Therapy (RT), in particular, HT acts as a powerful radiosensitizer, especially addressing the weaknesses of RT, such as hypoxic regions and cells in radioresistant phases [3-5]. In addition, the abnormal vasculature of the tumor is unable to respond to the external heat the way normal blood vessels do, i.e. increasing blood flow in order to cool the tissues involved, so that the effect of HT remains localized in the tumor [2, 3].

HT has been traditionally administered using radiofrequency, microwave, ultrasound and infrared radiation techniques. In recent years, magnetic nanoparticles have emerged as a possible alternative to these techniques. In Magnetic Nanoparticle Hyperthermia iron-oxide nanoparticles in the form of an aqueous solution are directly injected in the tumor and heated by an externally applied time-varying magnetic field [6, 7]. The strength of the magnetic field required to achieve the desired thermal effect, typically $10^4$ kA/m, is sufficiently high so as to cause interference in nearby electronic devices. Standard shielding of magnetic fields involves Faraday cages or coating with high-permeability shielding alloys; however, these techniques cannot be used with optically sensitive devices, such as...
those employed in Optical Coherence Tomography or radionuclide imaging. In this work we present a method to achieve magnetic shielding using an array of coils or solenoids.

2. Methods

The basic setup is illustrated in figure 1. All the coils are coaxial. The HT coil is held fixed at \( x = 0 \) and the shielding coils are placed between the HT coil and the imaging device in order to reduce the magnetic field \( B \) at the desired location where the sensitive electronics lie.

![Figure 1. The experimental setup showing the HT and shielding coils, and the magnetically sensitive imaging device.](image)

Two scenarios were investigated: in the small-size scenario the dimensions are similar to those found in a typical small-animal experiment, whereas in the large-size scenario the dimensions are similar to those found in a typical large-animal or human experiment. In both scenarios both the HT and shielding coils consist of 10 turns each and are 5 cm in length and thickness. The current through the HT coil is held fixed at 100 mA. The remaining parameters depend on the scenario and are listed in table 1.

| Scenario | HT radius (cm) | Shielding coil 1 | Shielding coil 2 |
|----------|----------------|------------------|------------------|
| Small    | 10             | 20               | 20               |
| Large    | 40             | 80               | 100              |

The axial field of a thin, finite coil in the SI system is given by [8,9,10]:

\[
B = \frac{\mu_0 IN}{2L} \left[ \frac{x_2}{\sqrt{x_2^2 + r^2}} - \frac{x_1}{\sqrt{x_1^2 + r^2}} \right]
\]

where \( \mu_0 \) is the permeability constant (1.26 x 10^-6 H/m), \( I \) is the current in the coil, \( N \) is the total number of turns, \( L \) is the length of the coil, \( r \) is the radius of the coil, and \( x_1 \) and \( x_2 \) are the distances from the two ends of the coil.

For a solenoid of finite thickness (1) becomes [8,10,11]:

\[
B = \frac{\mu_0 IN}{2(r_2 - r_1)} \left[ x_2 \ln \left( \frac{\sqrt{r_2^2 + x_2^2 + r_2^2}}{\sqrt{r_1^2 + x_2^2 + r_1^2}} \right) - x_1 \ln \left( \frac{\sqrt{r_2^2 + x_1^2 + r_2^2}}{\sqrt{r_1^2 + x_1^2 + r_1^2}} \right) \right]
\]

where \( r_1 \) and \( r_2 \) are the inner and outer radii of the coil.

Alternatively, the field of each coil can be calculated using finite elements techniques, such as the COMSOL physics simulation toolkit (www.comsol.com). The advantage of this approach is that the
calculation produces the $B$ field in full 3-dimensional space, not just along the axis of the coils. The $B$ field can then be exported for further use.

If inductive currents and fields and other interplay effects between the coils are ignored, then the resulting field is a superposition of the field generated by each coil separately. Software was written in C++ to calculate the total axial field, using the data calculated by COMSOL.

The shielding effect was quantified using the shielding factor, which is defined as:

$$Shielding \ Factor = \log_{10} \left( \frac{B \ without \ shielding}{B \ with \ shielding} \right)$$

(3)

It should be noted that since $B$ is proportional to the product of the number of turns times the current through the coil, for the purposes of this investigation it suffices to keep the number of turns constant and modify only the current. In an actual implementation, one might choose to change the number of turns in order to increase or decrease the absolute magnitude of the current.

3. Results

The relative magnitude of $B$, normalized to the maximum value when only the HT coil is present, is shown in figures 2 and 3 for the small and large-size scenarios, respectively. In the large-size scenario the currents through the two shielding coils are $-3.5$ and $-0.0253$ mA (the minus sign indicates that the current flows in the opposite direction from the current in the HT coil). For $x >= 2.5$ m the field is uniform and the shielding factor is 2.6–2.8.

In the small-size scenario with one shielding coil, the current through the shielding coil is $-14$ mA. A uniform field is achieved for $x >= 46$ cm with a shielding factor of approximately 1.8–1.9. When two shielding coils are used, the current through the first one is $-14.45$ mA and the current through the second one is 0.11 mA. The resulting field is not uniform; however, the shielding effect is much higher and it begins closer to the HT coil. In the range $41 \leq x \leq 57$ cm the shielding factor is no less than 2.6. At $x = 42.5$ and 52.1 cm the shielding factor increases to 3.5 and 4.4, respectively.

4. Discussion and Conclusion

We have presented a method to reduce the magnetic field at a given distance from the hyperthermia coil, without affecting the optical properties of the imaging device. In the large-size scenario, a uniform field is achieved using two shielding coils, both of whose currents flow in the opposite
direction from the current in the HT coil. In the small-size scenario, a uniform field is achieved using a single shielding coil. The addition of a second coil may significantly increase the shielding factor, by more than an order of magnitude at specific points, at the expense of creating a non-uniform resulting field. Furthermore, it causes the shielding effect to begin closer to the HT coil. Both these characteristics are important since: (a) depending on the construction of the imaging device, the sensitive electronics may be only a few mm in size, and (b) in general, the imaging device, and thus the sensitive electronics, should be as close as possible to the organ or object to be imaged.

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