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A new method for detecting cerebral hemorrhage in rabbits by magnetic inductive phase shift

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ABSTRACT

Cerebral hemorrhage, which is an important clinical problem, is often monitored and studied using expensive devices, such as magnetic resonance imaging (MRI) and positron emission tomography (PET) that are unavailable in economically underdeveloped regions. Magnetic induction tomography (MIT) is a new type of non-contact, non-invasive, and low-cost detection technology, and exhibits prospects for wide application, especially for the detection of brain diseases. However, the previous studies on MIT have focused on laboratory models and rarely on *in vivo* applications because the induced signals produced by biological tissues are notably weak. Based on the symmetry between the two brain hemispheres and the fact that a local brain hemorrhage will not affect the contra-lateral hemisphere, a symmetric cancellation-type sensor detection system, which is characterized by one excitation coil and two receiving coils, was designed to improve the detection sensitivity of MIT. This method was subsequently used to detect the occurrence of cerebral hematomas in rabbits. The average phase drift induced by a 3-ml injection of autologous blood was 1.885°, which is a fivefold improvement compared with the traditional single excitation coil and single receiving coil method. The results indicate that this system has high sensitivity and anti-interference ability and high practical value.

1. Introduction

The characteristics of cerebral hemorrhagic stroke include high incidence, high recurrence rates, high morbidity, high mortality, and high medical costs. Trauma to the brain may result in the accumulation of blood in a certain region of the brain, which is a pathological condition known as a hematoma. This accumulation of blood occurs gradually, i.e., over a period of hours or even days, and has severe consequences unless detected in a timely manner (Ayata and Ropper, 2002; Rojas et al., 2008). The conventional noninvasive medical imaging modalities used to detect the occurrence of cerebral hematomas include magnetic resonance imaging (MRI) and positron emission tomography (PET). However, MRI and PET are expensive, involve the use of large equipment, and are unavailable to rural and medically underserved populations (Ayata and Ropper, 2002; Kidwell et al., 2004). Magnetic induction tomography (MIT) is a new and emerging type of noncontact, noninvasive, and low-cost detection technology and exhibits prospects for wide application, especially for the detection of cerebral hemorrhage and edema. However, MIT exhibits technical difficulties in the detection of biological tissues. The electrical conductivity of biological tissues is notably low ($\sigma < 3\ \mathrm{S\ m^{-1}}$); thus, the induced magnetic field is typically only 1% of the primary magnetic field at a frequency of 10 MHz (Griffiths, 2001; Watson et al., 2004). To improve the detection sensitivity, back-off coils (Griffiths et al., 1999), gradient coils (Tarjan and Mcfee, 1968; Netz et al., 1993, Riedel et al., 2002, Scharfetter et al., 2001; Xu et al., 2003), and normal sensors (Watson et al., 2004) are used to reduce the sensitivity of the receiving coils to the primary field (Igney and Watson, 2005). However, these methods are not clinically important and can only be used in laboratory models.

When the traditional single excitation coil and single receiving coil (single coil–coil) method is used to detect cerebral hemorrhage or edema, such as cerebral ischemia in rats (Gonzalez et al., 2009), the magnetic field sensed by the receiving coil can be divided into three parts: the primary field, the secondary field generated by the bleeding or ischemic infarction sites, and the secondary field generated by other brain tissues.

Moreover, the field of interest is the secondary field generated by the bleeding or ischemic infarction site, which only accounts for a small proportion of the entire secondary field (the entire secondary field is significantly smaller than the primary field), and is thus smaller than the secondary field generated by other brain tissues.

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Hence, to improve the measurement sensitivity, both the primary field and the secondary field generated by other brain tissues should be canceled. Therefore, a time-difference method is used to analyze the imaging differences between the MIT signals from the head before and after the simulated lesion occurs (Zolgharni et al., 2008). However, in practice, a before-stroke data-set is unlikely to be available for the imaging of hemorrhagic stroke (Zolgharni et al., 2009). Thus, frequency-differential methods, which depend on the ability to distinguish the 'frequency signature' of a lesion from that of the surrounding tissues, are being explored to overcome this limitation. However, the visualization of the hemorrhage using this method is significantly more difficult given that the conductivity of all of the tissues of the head change with a change in the frequency (Zolgharni et al., 2010).

In this study, we expanded on previous work and designed a new cancellation scheme. Previous animal studies have shown that, during the early and middle phases of cerebral hemorrhage, a majority of hematoma and all edema occur in the injured cerebral hemisphere, whereas the contra-lateral normal hemisphere is almost unaffected (Wagner et al., 1996; Wagner et al., 1999; Nagasawa and Kogure 1989). Accordingly, considering the natural symmetry of the left and right brain hemispheres, we used two receiving coils to receive the primary and secondary fields produced by the injured hemisphere and the normal hemisphere; note that the primary fields received by two coils is the same. The phase shifts of the two hemispheres are later subtracted to obtain the phase shift difference generated only by the bleeding sites. This method was named the contra-lateral hemisphere cancellation method (CHCM) and can theoretically greatly improve the detection sensitivity. Based on this concept, a novel coil detection system (with headgear) was designed to detect cerebral hemorrhage specifically in rabbits.

2. Materials and methods

2.1. MIT theory

In MIT, the measured sample is always placed between the excitation coil and the receiving coil, and a magnetic field is generated by the current flowing in the excitation coil. This primary magnetic field induces eddy currents in the sample which in turn produce a secondary magnetic field. Both the primary and the secondary fields are detected by the receiving coil. The primary and secondary signals (V, ΔV) can be represented by the phasor diagram shown in Fig. 1. According to Griffiths et al. (2007), if the skin depth of the electromagnetic field in the sample is much larger than the thickness of the sample, ΔV is related to V as follows:

$$\Delta V/V = Q\omega\mu_0[\omega\varepsilon_0(\varepsilon_r - 1) - i\sigma] + R(\mu_r - 1) \quad (1)$$

where ω is the signal frequency, σ, ε_r, and μ_r are the electrical conductivity, relative permittivity, and relative permeability of the sample, respectively, ε_0 and μ_0 are the permittivity and permeability of free space, respectively, and Q and R are geometrical constants. Thus, the total signal ($V+\Delta V$) detected by the receiving coil lags the primary signal by an angle φ, which is approximately proportional to ω and σ.

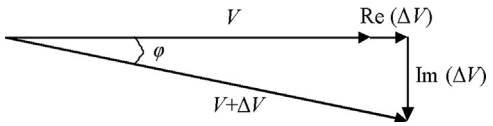

Fig. 1. Phasor diagram representing the primary signal V and the secondary signal ΔV detected. The total signal ($V+\Delta V$) lags the primary signal by an angle φ.

2.2. Design of the coil headgear

The structure of the coil headgear is shown in Fig. 2. One excitation coil (L_e, 20 windings, diameter=70 cm, 61.2 μH) and two receiving coils (L_{m1} and L_{m2}, each with 10 windings, diameter=30 cm, and 3.2 μH) are fixed in a square Perspex headgear, which is used as a coil fixator. L_e was first mounted on a Perspex and fixed to the headgear's upper surface. L_{m1} and L_{m2} were then fixed at the bottom of L_e such that the centers of the three coils are in a line and the distance between L_{m1} and L_{m2} is 55 mm. The coil sizes and the geometric location were tested to ensure that high sensitivity could be achieved. The mid-square hole (30×25 mm^2), mark line A, and mark point P provide references for the fixation of a rabbit's head: the bare skull area should be exposed in the square hole, the sagittal suture should align with line A, and the front halogen "cross stitch" should align with point P. The leads of the coils are twisted to avoid inductive pickup. Although not shown in Fig. 2, the headgear is 80 mm in height.

2.3. Experimental setup

The setup shown in Figs. 3 and 4 was used to experimentally detect cerebral hemorrhage in rabbits. An AFG3252 arbitrary signal generator (US Tektronix Company) outputs two sinusoidal signals at the same frequency and phase. One signal with an amplitude of 5 VPP is inputted into the excitation coil to generate an alternating magnetic field. The excitation field penetrates throughout the skull, and eddy currents are then generated in the left and right hemispheres and produce the secondary fields. Two identical receiving coils are placed at the same height from the surfaces of the two hemispheres and symmetrically along the sagittal suture to sense the two secondary fields created by the two hemispheres and the same primary field. The two receiving coils are separately connected to one terminal of a PXI5124

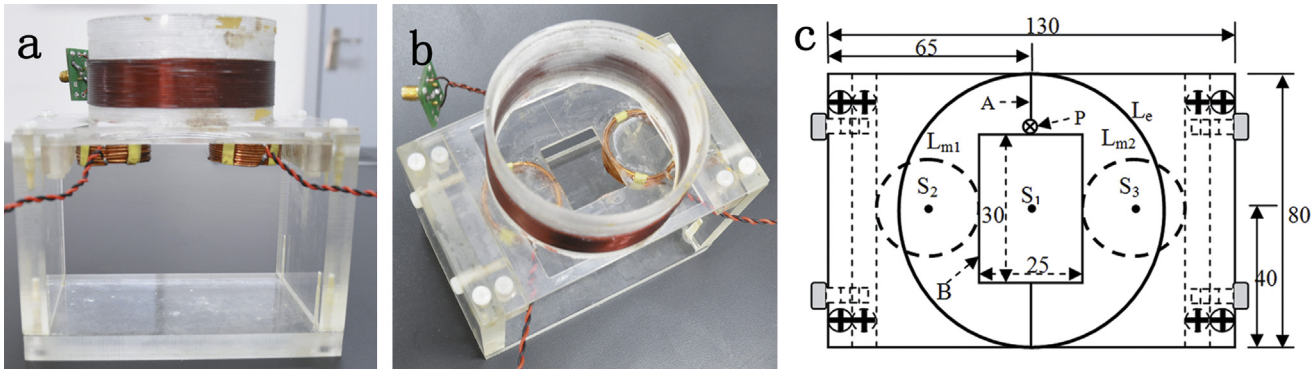

Fig. 2. Top view (a), front view (b), and schematic diagram (c) of the coil headgear. L_e: excitation coil; L_{m1} and L_{m2}: receiving coils; S1, S2, and S3: centers of L_e, L_{m1}, and L_{m2}, respectively; A: mark line; P: mark point; B: square hole.