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Needle type GMR sensor in biomedical applications

Abstract. *This paper presents two biomedical applications in which a needle-type GMR sensor with spin valve was proposed to estimate magnetic fluid weight density inside large tumors and to detect magnetic field generated by low current signals in medicine. First application is hyperthermia therapy, a form of cancer treatment. In second application GMR needle-type sensor was used for measurement of magnetic field generated by nervous systems.*

Streszczenie. *Artykuł prezentuje dwie propozycje zastosowania czujnika GMR z zaworem spinowym: do estymacji ciężaru właściwego płynu magnetycznego doprowadzanego do komórek nowotworowych oraz do pomiaru pola magnetycznego generowanego przez przepływ prądu o małym natężeniu, zachodzący w zjawiskach medycznych. Pierwsza aplikacja dotyczy leczenia nowotworów metodą hipertermiczną. Druga aplikacja jest związana z pomiarem pól magnetycznych generowanych przez układ nerwowy. (**Zastosowanie czujnika GMR typu igłowego w aplikacjach biomedycznych**).*

Keywords: GMR sensor, magnetic fluid, low magnetic field measurement.
Słowa kluczowe: czujnik GMR, płyn magnetyczny, pomiary małych pól magnetycznych.

Introduction

A novel giant magnetoresistive (GMR) probe with a needle consisting of a sensing element at the tip is proposed for accurate estimation of magnetic fluid weight density inside the body and detection of magnetic field generated by low current signals. The novel needle probe is characterized by high sensitivity and small size. Along with these features it is also minimally invasive. These features make it an attractive tool to be used in a wide variety of medical applications such as hyperthermia therapy and the diagnosis and treatment of nerve tissue diseases.

Hyperthermia therapy [1, 2] exploits the fact that cancer cells are more sensitive to heat than healthy cells. Generally all parameters except the magnetic fluid weight density are known in the specific heat equation which governs the heat given in hyperthermia therapy to destroy cancer cells. Hence, accurate estimation of magnetic fluid weight density inside the body is critical for successful treatment. A unique needle-type SV-GMR sensor is fabricated for inserting into the body in a low-invasive way.

In diagnosis of nerve tissue diseases the measurement methods are based on: electrical potentials measurement (i.e. electrocardiogram) and magnetic field measurement (i.e. SQUID magnetometer [4]). Both methods are non-invasive, but they give measurement results from specific part of body. Using unique needle type SV-GMR sensor, nerve signals can be detected locally.

Needle type SV-GMR sensor

A novel the fabricated needle-type GMR sensor as shown in Fig. 1 is the key feature of this research.

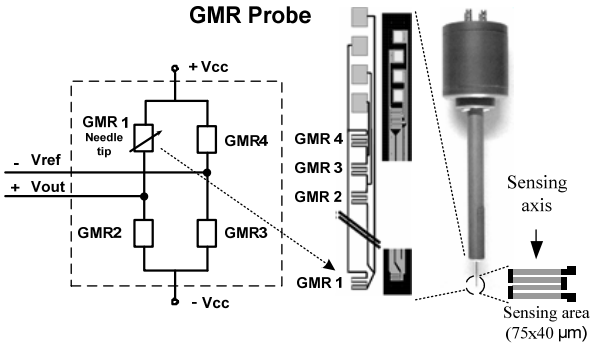

Fig.1. Needle-type GMR sensor

The needle is made of a compound of Aluminium Oxide and Titanium Carbide (Al_20_3/TiC) and 20 mm in length (15 mm is available to be inserted inside the body). Referring to Fig. 1 it can be seen that the GMR sensors in the needle probe are designed as a bridge circuit. At the tip of the needle there is a sensing area of 75 × 40 μm. The three other GMR sensors of the bridge circuit are placed near the bonding pads. The constant current of 0.5 mA is applied to the SV-GMR sensor. There are two output signals from the probe. Value of first one depends on magnetic field surrounding bonding pads part of the needle and value of the second is connected with value of magnetic field at the tip of the needle. The sensitivity of the needle-type GMR sensor in its sensitive axis was measured. It is approximately 13 mV/mT.

Hyperthermia treatment

One of the possible technical solutions to the hyperthermia treatment is injecting magnetic fluid (nanoparticles) into the affected area. Magnetic nanoparticles are safe to be used inside the body since for example dextran-magnetite has no measurable toxicity index. Also a localized magnetic field gradient can be used to keep the magnetic nanoparticles in a chosen site inside the body until it is utilized for a given procedure and then removed. Cancer treatment stands to benefit immensely from these favorable features since magnetic nanoparticles can be used as contrast agents for imaging tumors, in hyperthermia therapy to destroy cancer cells and in site-specific drug delivery to reduce the side effects associated with chemotherapy. Hyperthermia therapy utilizes heat to destroy tumor cells. Magnetic nanoparticles can be used as self heating agents. Magnetic fluid injected near the affected area inside the body is more readily taken up and hence easily entered into tumor cells compared to healthy cells. An external magnetic field is applied and heat is produced due to hysteresis loss of the magnetic nanoparticles. Tumor cells are more sensitive to heat than healthy cells [5]. However, the weight density of the magnetic nanoparticles is an important parameter for giving heat in such a way that it does not affect other healthy cells. Furthermore, magnetic fluid weight density along with applied magnetic flux density amplitude and exciting frequency is directly proportional to the specific heat capacity [6].

Magnetic nanoparticles are assumed to be spherical in shape and uniformly distributed in the fluid. The relative permeability of magnetic nanoparticles and liquid are assumed to be infinity and one respectively. The magnetic nanoparticles have a cluster structure. Since there is space

between the magnetic nanoparticles the space factor of spherical magnetite is considered and an equation defining the relationship between the relative permeability and weight density is obtained [6], as shown below.

$$\mu = 1 + C_d D_w / h_s \gamma_f, \qquad D_w << 1 \tag{1}$$

where: C_d – theoretical coefficient (4.0), D_w – weight density, h_s – space factor of spherical magnetite (0.523), γ_f - specific gravity of magnetite (4.58).

From (1) it can be seen that the relative permeability is proportional to the magnetic fluid weight density. It is worth to note there is no effect due to the shape or size of the cavity.

Fig. 2 shows a uniform magnetic flux density applied to a spherical cavity filled with magnetic fluid. The magnetic flux lines converge at the magnetic fluid filled cavity. This gives rise to a difference between the applied magnetic flux density B_0, and the magnetic flux density inside B_1, the cavity. The magnetic flux density at the center of the cavity can be expressed as shown below.

$$B_1 = \mu B_0 /(1 + N(\mu - 1)) \qquad \mu \approx 1 \tag{2}$$

where: *N* - demagnetizing factor of the cavity.

Equation (1) is substituted into (2) to obtain the difference *δ*, between the magnetic flux density inside and outside a magnetic fluid filled cavity.

$$\delta = B_1 - B_0 / B_0 = (C_d (1 - N) D_w) /(h_s \gamma_f), \qquad D_w << 1 \tag{3}$$

It can be seen from (3) that the magnetic fluid weight density can be calculated from the difference between the magnetic flux density inside and outside a magnetic fluid filled cavity. Magnetic fluid weight density is also proportional to the change ratio of magnetic flux density. The demagnetizing factor N, which depends on the shape and size of the cavity, influences the estimation of magnetic fluid weight density.

Consider the event where the tip of the needle is inserted into the centre of a cavity (as presents Fig. 2), under a uniform magnetic flux density (B_0).

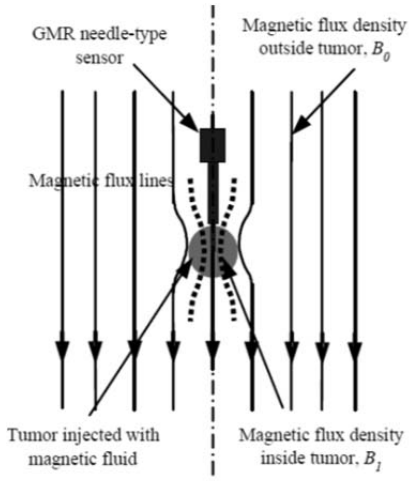

Fig.2. Spherical magnetic fluid filled cavity under the influence of a uniform magnetic flux density

The four GMR sensors are exposed to B_0, assuming that the cavity is empty (permeability of 1 can be assumed inside and outside the cavity). So there is no change in magnetic flux density inside and outside the cavity since B_1 is equal to B_0. However, when the cavity is filled with magnetic fluid the permeability inside is greater than outside the cavity. Hence, the GMR sensing area at the tip of the needle is exposed to a magnetic flux density B_1, which is higher than the applied magnetic flux density B_0. However, since the other three sensors are located further up near the bonding pads, and hence outside the magnetic fluid filled cavity, they will still be exposed to the applied magnetic flux density. This way the magnetic flux density inside a magnetic fluid filled cavity (B_1) and outside the cavity (B_0) can be measured simultaneously.

Helmholtz tri-coil (Fig. 3) produces a uniform magnetic flux density of 100 μT at 100 Hz for experiments. The fluctuation of the magnetic flux density is 0.01%, 0.03 m in the axial and radial direction from the midpoint. The demagnetizing factor N, of a cavity depends on the aspect ratio s. The aspect ratio of a cavity s = (long axis/diameter). Cylindrical agar cavities of diameter 63 mm (s = 1) were injected with magnetic fluid.

In previous experiments done on smaller cavities [7], equation (3) has been verified for a range of sizes. It was shown that the change in magnetic flux density did not vary so much between different size cavities as long as s and hence N remained the same. The change in magnetic flux density only increased with increasing magnetic fluid weight density.

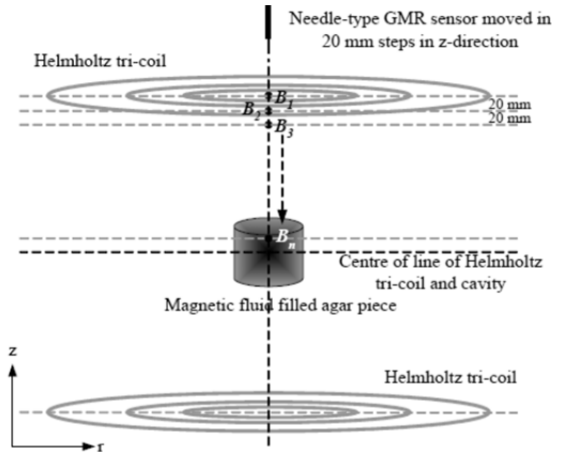

Fig.3. Experimental method for large cavities

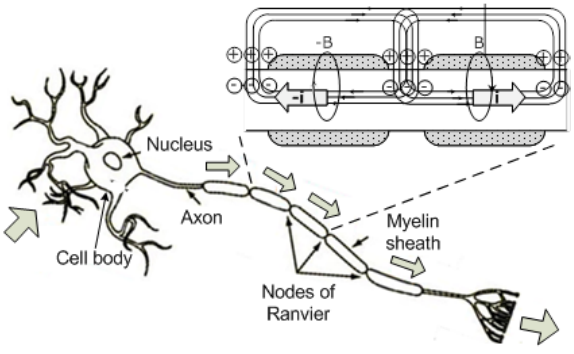

Fig.4. Stimulation of nervous cell

Magnetic field generated by nervous system

In previous work [3] straight wire as nerve model was used. It was developed a new model in order to simulate process of generation of magnetic field in myelinated axon (see Fig. 4). Stimulation of neuron causes electrical signal (often called action potential) propagation along the axon. The

neuron then increases its internal potential, setting off a chain of events which is repeated for each Node of Ranvier as the nerve impulse "jumps" down the axon. There is a movement of K and Na ions between outside and inside of axon membrane and it can be seen as current dipoles with opposite direction in adjacent segments [4]. These currents generate magnetic field. This phenomenon was implemented in model presented in Fig. 5. Model was built using silicon plate where holes were made (diameter of hole s = 200 μm, distance between holes l = 1000 μm) and through which wire with diameter 20 μm was knitted.

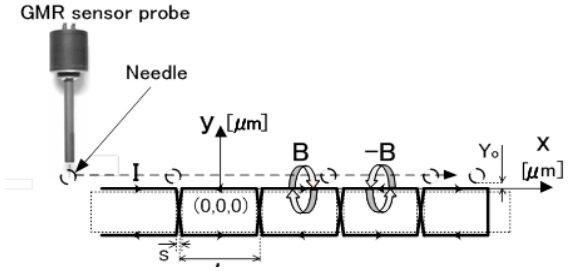

Fig.5. GMR sensor and nerve model

Nerve model was supplied with action current (10 μA_{P-P}) modeled as pulse signal with 10% duty cycle at frequency 1 kHz, which resembles signal appearing in real nerve. Position of a sensor was changed during experiments. Sensor was moved along x axis of the model with distance between tip of the sensor needle and model (y axis direction) approximately 15 μm (see Fig. 5). An example of measured signal from the sensor after 10000 times amplification is illustrated in Fig. 6. Pulse output is in phase with input action current.

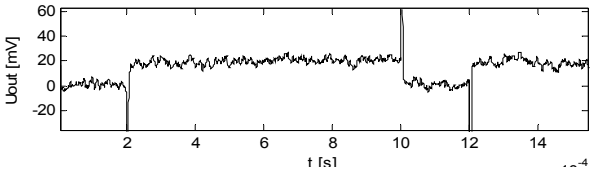

Fig.6. Measured sensor output signal after 10^4 times amplification

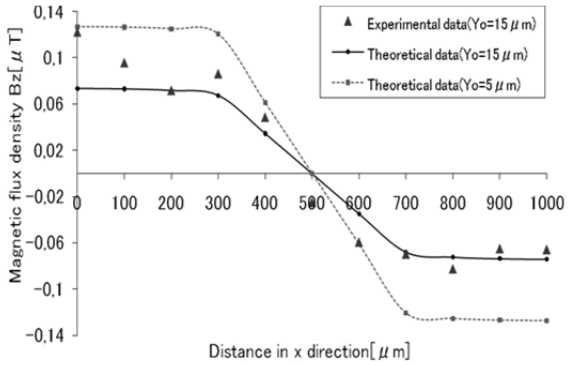

Fig.7. Detected signal and calculated magnetic flux density versus distance in x direction

Comparison between experimental data and theoretical data calculated in case of 15 and 5 μm distance between nerve model and sensor is presented in Fig. 7. Experimental data agrees quite well with the theoretical results, but some inaccuracy is caused by problems with accurate measurement of distance between sensor and model. Another reason is that in this configuration sensitivity characteristic of GMR probe is slightly nonlinear with changing value of measured magnetic field.

The measurement performed by GMR sensor showed that magnetic field generated by small current signals in some distance from the sensor can be detected as well as through the layer covering the nerve. Needle type GMR probe can be successfully used everywhere where minimal invasive measurement is required or operations with using large size magnetometers is difficult.

Summary

The needle-type GMR sensor has a good potential to be used in other medical applications such as targeted drug delivery where the needle-sensor can be used to confirm if the magnetic particles and drugs are present at a given site by detection as well as estimation of the content density for supplying heat to initiate a chemical reaction. Due considerations should also be given to improve some aspects of the proposed methods for successful implementation in clinical applications.

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