Conceptual Design of a Miniature High Frequency Magnetic Fields Measurement System in a Human Body Phantom

Yoon-Myoung Gimm[#1], Sangwook Nam[#2], Ohyoung Kim [#3], Yongil Kwon [#4], Geun-Wu Gimm[#5]

[#1] *EMF Safety Inc., Eun-Go Building 7th Floor, Han-Nam-Dong 72-17, Yong-San-Gu, Seoul 140-886, So. Korea*
[1]gimm@dku.edu

[#2] *Seoul National University, San56-1, Sillim-Dong, Kwanak-gu, Seoul 151-742, So. Korea*
[2]snam@snu.ac.kr

[#3] *Dankook University, Jukjeon-Dong, Yongin-si, Gyeonggi-do 448-701, So. Korea*
[3]koybon@dankook.ac.kr

[#4] *Auto Wireless Inc., 413 SJ Technoville, 60-19 Gasan-dong, Geumcheon-gu, Seoul 153-769, So. Korea*
[4]yikwon@zenion-tech.com

[#5] *Seoul Science High School, Uamgil 63 (Hyehwa-dong 1-1), Jongro-gu, Seoul 110-530, So. Korea*
[5]gwgimm47@hanmail.net

***Abstract*—** A conceptual design of a microsystem is proposed for the measurement of high frequency magnetic fields in a human body. It incorporates a three-axes isotropic ferrite coil sensor, encoder, transmitter IC, and antenna into a small capsule. The structure of the magnetic sensor, link budget, and the structures of transmitter and receiver are described briefly. The experimental results will be provided later.

Key words: Three Axes Magnetic Field Sensor, Magnetic Field Measurement in a Phantom, Wireless Communication through the Human Body, In-body Communication, In-body Channel.

I. Introduction

The implanted medical devices in the human body have become more popular with the advent of the new electromechanical microtechnologies. They include implanted artificial cochleae, pace makers, insulin pumps, and devices for urinary urge incontinence.

Usually the devices need DC batteries inside for the operations of electronics. Some devices can operate for several years using the installed batteries, but other devices which consume much power have to be recharged frequently by external energy sources through wireless transmission. Most of them use divergent high frequency magnetic fields generated by current coils attached outside of the human body but nearby the internal devices[1][2]. The operating frequencies are in between 100 kHz and a few MHz.

When the influence of electromagnetic (EM) field on the human health was not an issue, it was not necessary to measure the driving EM fields. As people become to have more concerns on EM field in recent days, they are more satisfied when human body is as less exposed as possible by EM fields.

For doing this it is important to measure the field in the human phantom[3][4] at the relevant frequency, but it is not so easy to measure the fields about 1 MHz range with a sensor of the reasonable size giving rise to the negligible perturbation. Usual commercial magnetic sensors are neither proper in size nor in the sensing frequency bands.

We now propose a small (about a few cm^3) size magnetic field sensor for 1 MHz frequency range to measure the field in the human phantom. The digital informations of the measured field intensity are transmitted to outside of the body by wireless communication with a much higher carrier frequency than the magnetic field frequency to be measured and with a much less field intensity than the driving magnetic field for recharging. Fig. 1 shows the overall block diagram of the in-body magnetic field measurement system.

In section II, the structure and the operation of the magnetic sensor are described in detail. The wireless communication system is explained in section III and the proposed overall system design is described in section IV.

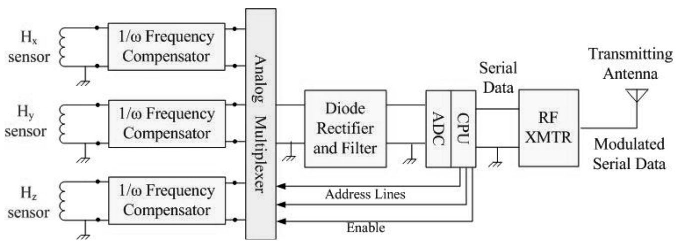


Fig. 1 Overall system block diagram for sensing and transmitting the magnetic field information inside the body.

II. THREE AXES MAGNETIC FIELD SENSOR

Fig. 2 shows the structure of the three axes magnetic field sensor used in this paper[5]. It is a typical isotropic ferrite coil sensor.

The induced ac voltage of each winding of the coil is proportional to the frequency of the field. To compensate the frequency response of the coil sensors, the 1/ω frequency compensator is inserted right after the analog multiplexer followed by three coil outputs.

The output AC voltages after the filter are,

$$V_i = \eta_i \, \psi_i \; H_i \; (i = \; x, y \text{ or } z) \qquad (1)$$

, where V_i is the induced voltage by the time-varying magnetic field, and η_i in [V/(A/m)] is the sensitivity of sensor coil i in air for the sensor aligned with the field vector, and ψ_i is the ratio of sensor response in air to response in the dielectric media (sometimes referred to as the conversion factor)[6], and H_i is the magnetic field component of i-axis direction of vector field $\overline{H}$ (x, y, z).

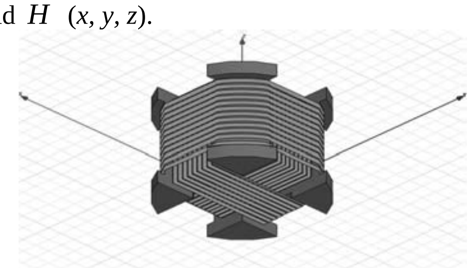


Fig. 2 A three-axes isotropic ferrite coil sensor.

The rms values of the sensed voltage V_i's are converted to DC values which are proportional to the intensity of the field. These values are digitalized and delivered to the CPU (communication assembly) and transmitted through the human body to the receiving block at outside body.

In the signal processor of the receiving block, H_i is retrieved from the delivered V_i and the magnitude of the magnetic field is calculated to be,

$$H = sqrt\left[\sum_{i}^{x,y,z} H_i^{\,2}\right] = \sqrt{H_x^{\,2} + H_y^{\,2} + H_z^{\,2}} \qquad (2)$$

In the signal processor, pre-determined η_i's and ψ_i's should be resident in the digital memory block of the receiving system.

η_i and ψ_i values are to be suggested in tabular form by measurement in the standard environment in Fig. 3.

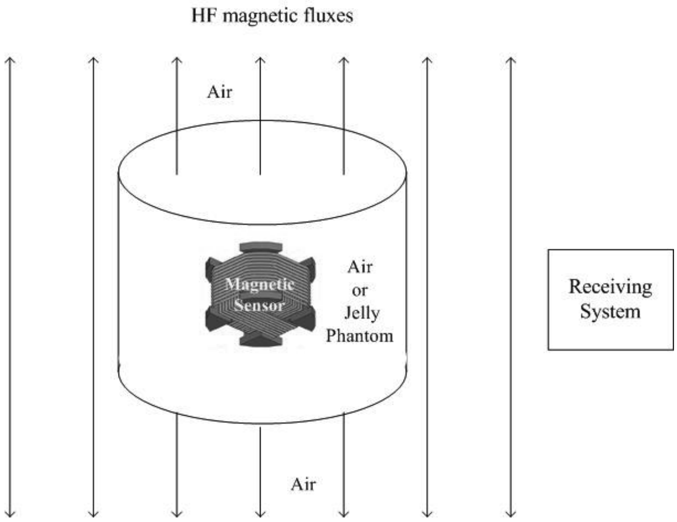


Fig. 3 Experimental setup for acquiring the sensor coil sensitivities and conversion factors. Each axis of the coil sensor is aligned in sequence to the external magnetic flux for coil sensitivity (in air) and conversion factor (in jelly).

III. WIRELESS COMMUNICATION SYSTEM IN HUMAN BODY

A. *Simplified Channel model*

In order to design a wireless communication system in human body, it needs to estimate the channel characteristics of human body. Since it is quite complex to model a human body precisely, we simply assume that human body consists of a simple medium and the distance between transmitter and receiver is less than 15 cm, as shown in Fig. 4.

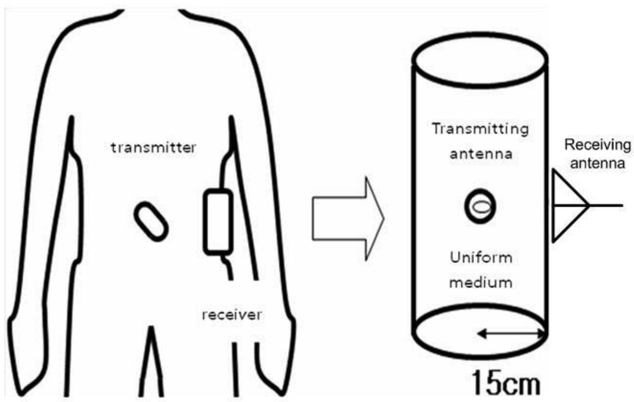


Fig. 4 Human body and its simplified model for wireless in-body communication. The signal at the receiving antenna is delivered to electronic circuits shown in Fig. 5(b).

The property of uniform medium is chosen according to the dielectric properties of body tissues provided by FCC(Federal Communication Commission)[7]. It is similar to the human body model for SAR measurement[8]. The transmission loss can be calculated by the Friis formula.

$$\frac{P_{RX}}{P_{TX}} = G_T G_R \left(\frac{\lambda}{4\pi R}\right)^2 e^{-2|\mathrm{Im}[k]|R} \qquad (3)$$

, where P_{TX}, P_{RX} are the transmitting and receiving power and G_T, G_R are the gain of transmitting and receiving antenna. R is the distance between transmitter and receiver, λ is the wavelength and k is propagation constant. The G_R value is assumed to be 1 because the receiving antenna is placed at outside of the body and its size is not limited. Therefore, the transmission loss is obtained using the transmitting antenna gain G_T, radiation loss $(\lambda/4\pi R)^2$ and propagation loss ($e^{-2|Im[k]R|}$). The transmitting antenna gain is determined by efficiency (*eff*) and directivity (*D*).

$$G_T = eff \cdot D \qquad (4)$$

The directivity (*D*) is assumed 1.5, because the antenna in human body is small. The efficiency of the antenna is directly related to each spherical mode efficiency since the electro-magnetic field of the antenna can be represented by sum of spherical modes. The efficiency of the transmitting antenna is approximated by the efficiency of the TE_{01} mode. TE_{01} mode is given by [9].

$$eff \approx eff_{TE01} = \frac{\mathrm{Re}[\eta]}{\mathrm{Re}\left[\eta + \dfrac{1}{j\omega\varepsilon a}\right]} e^{-2|\mathrm{Im}[k]|a} \qquad (5)$$

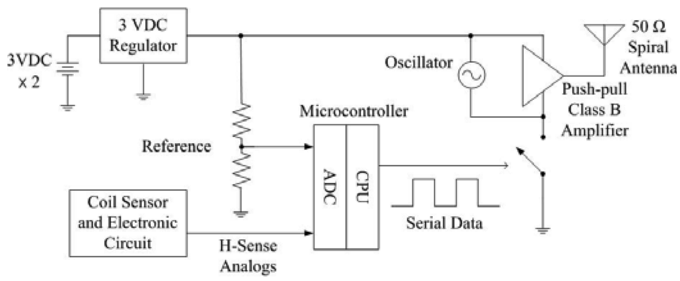
, where η is the characteristic impedance of medium, ω is the angular velocity, ε is the permittivity of medium, and a is the radius of the sphere which encloses the antenna. Using Eqs.(4) and (5), the total transmission loss can be estimated and it is found that the frequency range of 400 MHz to 600 MHz seems to be optimum for the in-body communication[10].

B. Design of Transmitter and Receiver

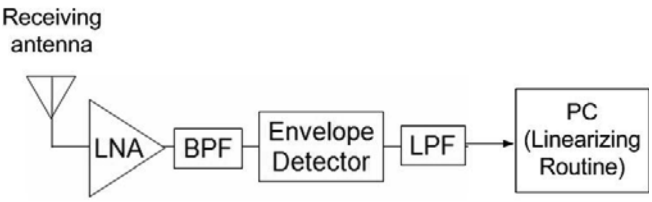
450 MHz is selected for the center frequency of the communication system. The OOK(On Off Keying) is used as modulation scheme and envelope detection as demodulation, which are suitable for miniaturization and high efficiency. The calculated link budget is shown in Table I. The stable performance of communications in human body is to be achieved by the enough link margin of 19 dB. Fig. 5a and 5b show block diagrams of the transmitter and receiver. The sensor coils and the following electronic circuits for sensing the magnetic field in Fig. 1 are enclosed in the transmitter structure in Fig. 5a.

TABLE I
COMMUNICATION SYSTEM LINK BUDGET

Parameter	Value
Tx power	0 dBm
Antenna gain (G_T)	-10 dB
Loss (Radiation and Propagation)	-53 dB
Estimated Receiving Power	-63 dBm
Link margin	19 dB
Minimum Receiving Power	-82 dBm
Receiver NF	5 dB
SNR(Env. detection, BER=10^{-5})	14 dB
Thermal Noise Power (BW = 20 MHz)	-101 dBm



(a) The transmitter structure in a capsule.



(b) The receiver block diagram.

Fig. 5 Block diagram of the transmitter and the receiver structure for wireless in-body communication.

IV. OVERALL SYSTEM DESIGN

The H-field sensors in Fig. 1 are made by winding coils around the machined cylindrical ferrite core. So it is not quite easy to make each inductance the same. Because of this nonidentical axis characteristics three different frequency compensators of the coils are designed separately for each H-field sensor. It is possible to decide RC values in the $1/(\omega+\omega_o)$ frequency compensators to give the nearly flat frequency characteristics of the rectified DC voltages in the interested frequency band (400 ~ 1,200 kHz). f_o ($=\omega_o/2\pi$) can be set near to 100 kHz by adjusting R and C values in the low pass filter.

A low barrier Schottky diode rectifier is utilized in Fig. 1 to convert the AC signal to DC. The diode has a square law characteristic in the weak magnetic field and shows a linear characteristic in the strong magnetic field. To expand the measurable dynamic range of the magnetic field intensity of Eq.(1), it needed a mathematical linearizing software routine in the PC receiving the demodulated baseband data from the receiving module through the wire in Fig. 5(b).

The DC power operating the whole sensing and transmitting structure in Fig. 5(a) is regulated 3 VDC from the two 3 VDC (= 6 VDC) batteries. The reason for the regulation is that reference voltage of the ADC in the microcontroller should be without regard to the battery voltage which can be variable during the system operation.

In Fig.5(a), the electronic circuits in the capsule contains an 8-bit microcontroller with a 10-bit resolution ADC to which the rectified analog signal is input as shown in Fig. 1. The reference voltage of the ADC in Fig. 5(a) is divided by software to give the proper resolution of the sensor output analog signal.

The timer in the microcontroller is set to control the date rate. The serial data for OOK modulation at RF transmitter are supplied to the RF transmitter via GPIO port of the microcontroller in the UART transferring format.

The control sequence of the whole system by the microcontroller is shown in Fig. 6.

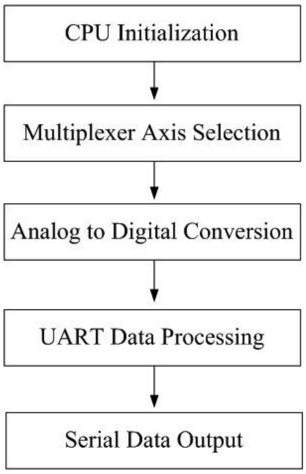


Fig. 6 Flow chart of the system control by the microcontroller.

The RF oscillator and an amplifier are integrated together in the capsule in Fig. 5(a). The spiral monopole antenna in the capsule is wound on the dielectric.

It is tested in a phantom filled with the muscle equivalent jelly with ε_r=2,336, σ= 0.485 [S/m] at 800 kHz range[11] for the sensor module calibration.

V. CONCLUSION

To measure the high frequency magnetic field intensity in the human size volume, it needs a relatively small magnetic field sensor. For transmitting the informations of the magnetic field to the outside region, it is plausible that the informations are delivered by wireless rather than through the wire.

The magnetic field sensing devices and communication systems which correspond to this purpose are designed in this study. The system can be calibrated in the human muscle simulating phantom afterwards.

ACKNOWLEDGMENTS

This research has been supported by the Intelligent Microsystem Center (IMC; http://www.microsystem.re.kr), which carries out one of the 21st century's Frontier R&D Projects sponsored by the Korea Ministry Of Commerce, Industry and Energy.

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