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Magnetic Navigation of an Untethered Micro Device Using Four Stationary Coils

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***Abstract*—We introduce a magnetic navigation of a small magnet using four stationary coils. We used a Maxwell gradient coil to get magnetic propulsion force and three Helmholtz coils to control the moving direction of the magnet in the magnetic navigation. Using a three-channel coil driver with output capacity of 320A, we performed magnetic navigation of a small NdFeB magnet with the size of 10mm×10mm×12mm on a horizontal plane. When navigated with a slow speed of about 1 mm/s, the magnet kept track of any arbitrary navigational path. We expect the proposed magnetic navigation method can be easily incorporated into the system for human applications since it does not use any moving coils.**

***Index Terms*—Magnetic navigation, untethered device, Maxwell coil, Helmholtz coil, microrobot, stationary coil.**

I. Introduction

MAGNETIC navigation of an untethered micro device in a living system has gained utmost importance for biomedical applications such as coronary artery treatment or drug delivery to the cancerous tissues using a micro robot [1]–[3]. Due to the small size of a micro robot, it is very challenging to deliver energy to the robot especially when it is untethered in a living subject. Magnetic propulsion is known to be an effective way to navigate a micro device carrying a small magnet in it along blood vessels in a human body [4], [5]. Since human body is transparent to magnetic field, magnetic force can be easily generated by applying external magnetic field to it without any interaction between the human body and the external magnetic field.

Recently, gradient fields of an MRI system have been used to navigate a small spherical magnetic core against the blood flow in a porcine carotid [2]. When navigated by clinical MRI gradient coils, the magnetic core size should be greater than 1mm to gain the magnetic force enough to propel the core against the arterial blood flow due to the finite gradient field strength of 40-60mT/m [5]–[8]. It is thought that much greater gradient fields are necessary to navigate a small magnet that can be fit into a micro device designed for blood vessel applications [9].

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For magnetic navigation of an untethered micro device in a human body, stationary coils rather than rotating coils of previous studies [1] are much more desirable. Stationary coils can provide much faster navigation control and they are more easily incorporated into the system for human applications. Since a human body has a high aspect ratio, it is practically impossible to move the coils without moving the human subject together. In this paper, we propose a stationary coil system for navigating a micro device in any arbitrary direction in 3D space.

II. Stationary Coil System for Magnetic Navigation

A. Four Stationary Coils

A magnetic moment **M** in the magnetic field **B** experiences magnetic force given by [10]

$$\mathbf{F} = V(\mathbf{M} \cdot \nabla)\mathbf{B} \,. \tag{1}$$

V is the volume of magnetic material. Equation (1) implies that we need non-uniform magnetic field to gain magnetic force from the field. For better controllability of the magnetic navigation, it is desirable to make uniform gradient field. It is well known that Maxwell gradient coil, consisting of two circular coils with opposite current directions, makes most uniform gradient field at the center of the coil when the coil radius r and the coil height h satisfies $h = \sqrt{3}r$ [11]. Equation (1) can be rewritten component by component as follows:

$$\begin{bmatrix} F_x \\ F_y \\ F_z \end{bmatrix} = V \begin{bmatrix} M_x \dfrac{\partial B_x}{\partial x} + M_y \dfrac{\partial B_x}{\partial y} + M_z \dfrac{\partial B_x}{\partial z} \\ M_x \dfrac{\partial B_y}{\partial x} + M_y \dfrac{\partial B_y}{\partial y} + M_z \dfrac{\partial B_y}{\partial z} \\ M_x \dfrac{\partial B_z}{\partial x} + M_y \dfrac{\partial B_z}{\partial y} + M_z \dfrac{\partial B_z}{\partial z} \end{bmatrix}. \tag{2}$$

If we confine the navigation space to the central region of the Maxwell coil where the gradient field is uniform, the gradient field generated by the Maxwell coil can be approximated as follows [9]:

$$\begin{bmatrix} \dfrac{\partial B_x}{\partial x} & \dfrac{\partial B_y}{\partial y} & \dfrac{\partial B_z}{\partial z} \end{bmatrix}^{\mathrm{T}} = \begin{bmatrix} -0.5g & g & -0.5g \end{bmatrix}^{\mathrm{T}}, \tag{3}$$

where g is the factor that is proportional to the coil current. Therefore, the magnetic propulsion force exerted by the Maxwell coil is given by

$$\begin{bmatrix} F_x \\ F_y \\ F_z \end{bmatrix} = gV \begin{bmatrix} -0.5M_x \\ M_y \\ -0.5M_z \end{bmatrix}. \quad (4)$$

Equation (4) implies that we can generate the magnetic force in an arbitrary direction by changing the direction of the magnetic moment.

We use three identical orthogonal Helmholtz coils in the x-, y-, and z-directions to orient the magnetic moment along the desired directional unit vector $\mathbf{a}=[a_x, a_y, a_z]^T$. If the magnetic moment is to be oriented along the unit vector $\mathbf{a}$, i.e., the magnetic moment becomes $\mathbf{M}=M_o\mathbf{a}=[M_x, M_y, M_z]^T$ where M_o is the amplitude of the magnetic moment, we have to apply the following currents to the Helmholtz coils:

$$\begin{bmatrix} I_x & I_y & I_z \end{bmatrix}^T = I_o \begin{bmatrix} a_x & a_y & a_z \end{bmatrix}^T. \quad (5)$$

I_o is the scaling factor to the Helmholtz coil currents and it can be used to control the torque exerted on the magnetic moment, i.e., the bigger I_o is, the greater the torque is. When the Helmholtz coils exert torque on the magnetic moment, drag forces appear against the rotational motion. The drag forces may come from the frictions between the magnet and the blood or vessel wall. The magnetic torque, the drag force, and the rotational inertia of the magnet determine the behavior of the magnet's rotational motion. For complete understanding of the magnet motion under the external magnetic field, we have to derive the equation of motion which is governed by many physical factors. Here, we assume that any motion of the magnetic moment in the external magnetic field, either propulsion or rotation, appear in an equilibrium state of the magnetic force and the drag force for the sake of simple analysis.

Figure 1 shows the coil configuration used for computer simulations as shown in Fig. 2 and magnetic navigation experiments. To fit on the cubic frame, the diameters of the coils are all the same as the height of the cube. With this coil configuration, magnetic propulsion direction is the same as the magnetic moment direction only when the micro device is propelled in the z-direction. In other directions, the magnetic moment direction does not coincide with the propulsion direction, which might be troublesome when the micro device has an elongated shape for the use inside blood vessels. However, if flexible coupling is allowed between the magnet and the micro device, then this coil configuration can also be used for the vessel applications.

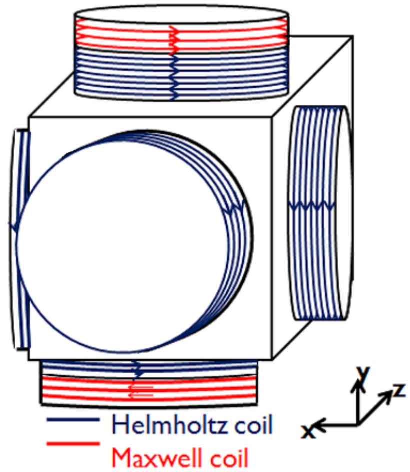

Fig. 1. The coil system consisting of three orthogonal Helmholtz coils and a Maxwell coil. The heights of the coils have been changed to fit on the cubic surface.

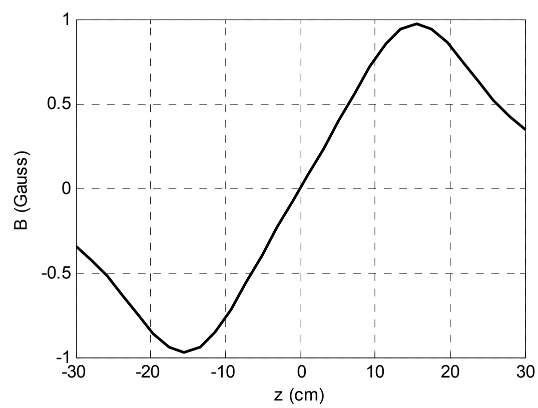

Fig. 2. A simulation result showing relationship between location and magnetic field by a Maxwell coil (h=2r=30cm).

Fig. 3. The fabricated coil set consisting of three Helmholtz coils and a Maxwell gradient coil.

B. Coil Driving System

We used a three-channel coil driving system (EP-GAMP-01001, Elico Power, Korea) which had been used as a gradient coil driver in a conventional MRI system. Each channel has the maximum output current of 320A with a current amplifier. We can control the output current using the control voltage with the gain of 20A/V. To match impedances between the current driver and the magnetic navigation coils, we adjusted the inductance of coils to 550μH.

III. Experimental Setups

We have made three Helmholtz coils and a Maxwell coil using 2mm × 3mm copper wires. The coils were wound on an acrylic frame which consists of a hollow cubic frame and six protruding cylindrical outlets from the cubic surface. The height of the cubic frame is 30cm and the height of the protruding outlet is 38cm. The number of turns of the Helmholtz coils and the Maxwell coil are 20 and 20, respectively. The inductance of the Helmholtz coils and the Maxwell coil are 552μH and 543μH, respectively. Figure 3 shows the fabricated coil set.

Since our coil driver has only three channels, we have performed the magnetic navigation experiments on a horizontal plane using the *x*- and *z*-directional Helmholtz coils and the Maxwell coil. In the experiment, we positioned a cylindrical NdFeB magnet with the size of 10mm × 10mm × 12mm on a Teflon plate. The maximum coefficient of static friction between the magnet and plate is 0.424, and coefficient of kinetic friction is 0.232.

For efficient experiments, we have developed a navigation control panel on a personal computer using LabView (National Instruments, USA) and a 16 bit digital-to-analog conversion board (NI 9263 4-channel, National Instruments, USA). On the control panel, we can synthesize coil current waveforms either from the predetermined waveform files or the joy stick outputs. Figure 4 shows the control panel in which we can control the direction of a magnet and the acceleration of movement.

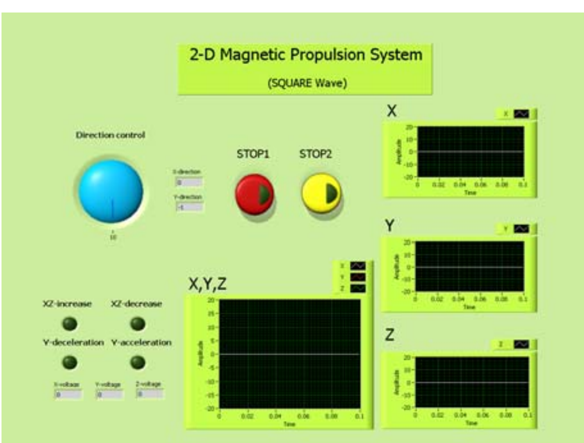

Fig. 4. Magnetic navigation control panel.

IV. Experimental Results and Discussions

We have performed magnetic navigation experiments on an acrylic plate on which desired navigation paths has been drawn. Firstly, we determined the maximum Helmholtz coil current amplitude which gave smooth 90 degree rotation

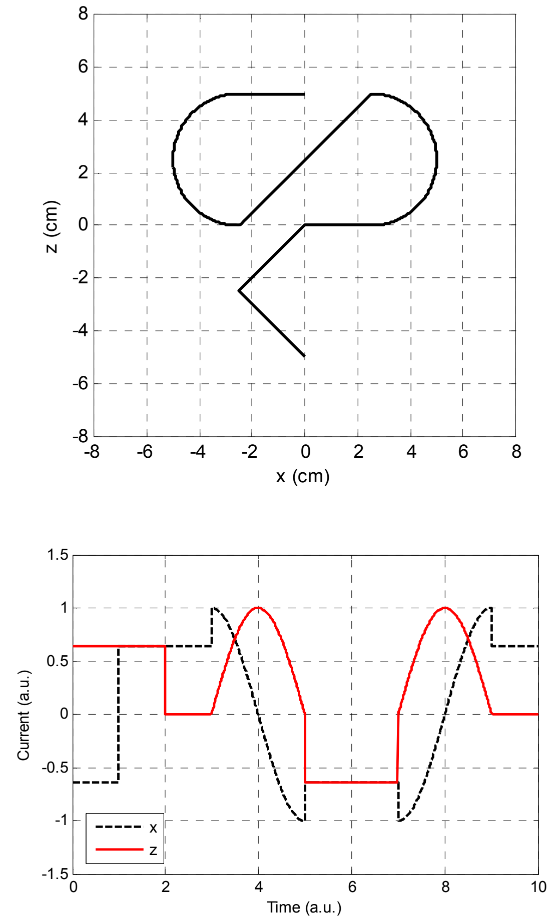

Fig. 5. An example of a desired navigation path (top) and corresponding Helmhotz coil current waveforms (bottom).

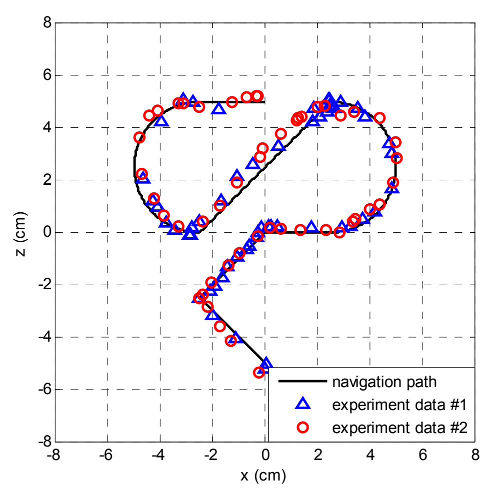

Fig. 6. The desired path and the average navigation path.

motion when rectangular pulses of 1s were switched between the *x*- and z-directional Helmholtz coils. When the Helmholtz coil currents were too high, the rotational motion was not smooth with overshooting response. With the maximum Helmholtz coil current, we normalized the Helmholtz coil currents for navigating the magnet along the desired paths.

Changing the Helmholtz coil currents according to the navigation direction, we propelled the magnet by applying current to the Maxwell coil. This time, we also determined the optimal Maxwell coil current by trial-and-error approach that gave most smooth propulsion motion along the desired path.

Figure 5 shows a desired path and corresponding Helmholtz coil current waveforms. On top of the desired path, we have shown the magnetic moment which changes its direction according to the navigation direction. Figure 6 shows the desired path together with the actual navigation path. We performed navigation experiments and we recorded the magnet motion using a video camera. From the video images, we tracked the actual navigation paths for each experiment.

The average velocity of magnet is nearly 0.917mm/s. The velocity of a straight course is faster than that of a curve since the moving direction and acceleration should be considered simultaneously in the condition of changing direction. The differences between the actual location and the desired location are acceptable levels within 3mm. Further studies are needed for the improvement of accuracy and velocity. Also, a bigger power system than our experimental system should be applied in order to use the navigation system by stationary coils of almost $1m^3$ in clinical human applications.

V. Conclusions

We have developed a magnetic navigation system consisting of four stationary coils and current drivers. With the system, we have successfully navigated a small magnet on a flat plane along the desired path. For more precise and stable navigation, we need to develop feedback-controlled navigation system.

References

[1] K. B. Yesin, K. Vollmers, and B. J. Nelson, "Modeling and control of untethered biomicrorobots in a fluidic environment using electromagnetic fields," *Int. J. Robot. Res.*, vol. 25, pp. 527–536, 2006.

[2] S. Martel, J.-B. Mathieu, O. Felfoul, A. Chanu, E. Aboussouan, S. Tamaz, and P. Pouponneau, "Automatic navigation of an untethered device in the artery of a living animal using a conventional clinical magnetic resonance imaging system," *Appl. Phys. Lett.*, vol. 90, pp. 114105-1–114105-3, 2007.

[3] J.-B. Mathieu, S. Martel, L. Yahia, G. Soulez, and G. Beaudoin, "Preliminary investigation of the feasibility of magnetic propulsion for future microdevices in blood vessels," *Bio-Med. Mat. Eng.*, vol. 15, pp. 367–74, 2005.

[4] J.-B. Mathieu and S. Martel, "Magnetic steering of iron oxide microparticles using propulsion gradient coils in MRI," in *Proc. 28th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, New York, 2006, pp. 472-475.

[5] J.-B. Mathieu, G. Beaudoin, and S. Martel, "Method of propulsion of a ferromagnetic core in the cardiovascular system through magnetic gradients generated by an MRI system," *IEEE Trans. Biomed. Eng.*, vol. 53, pp. 292–299, February, 2006.

[6] A. Chanu, O. Felfoul, G. Beaudoin, and S. Martel, "Adapting the clinical MRI software environment for real-time navigation of an endovascular untethered ferromagnetic bead for future endovascular interventions," *Magn. Reson. Med.*, vol. 59, pp. 1287–1297, 2008.

[7] O. Felfoul, J.-B. Mathieu, G. Beaudoin, and S. Martel, "In vivo MR-tracking based on magnetic signature selective excitation," *IEEE Trans. Med. Imaging*, vol. 27, pp. 28–35, 2008.

[8] J.-B. Mathieu and S. Martel, "Magnetic microparticle steering within the constraints of an MRI system: proof of concept of a novel targeting approach," *Biomed. Microdev.*, vol. 9, pp. 801–808, 2007.

[9] B. H. Han, S. Park, and S. Y. Lee, "Gradient waveform synthesis for magnetic propulsion using MRI gradient coils," *Phys. Med. Biol.* vol. 53, pp. 4639-4649, 2008.

[10] J. J. Abbott, O. Ergeneman, M. P. Kummer, A. M. Hirt, and B. J. Nelson, "Modeling magnetic torque and force for controlled manipulation of soft-magnetic bodies," *IEEE Trans. Robotics*, vol. 23, pp. 1247–1252, 2007.

[11] E. M. Haacke, R. W. Brown, M. R. Thompson, and R. Venkatesan, *Magnetic Resonance Imaging: Physical Principles and Sequence Design*, New York: Wiley, 1999.