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A method of nerve electrical stimulation by magnetic induction

Guanghao Zhang, Yue Li, Xiaolin Huo, and Tao Song

Abstract—**The implantable electrical stimulator is usually not suitable for long term use because of its large size and short battery life, besides the magnetic stimulation can not provide deep nerve stimulation. Therefore, this paper developed a method of the electrical stimulation by using the magnetic induction. We implanted a small inductive coil which was connected with nerve electrodes. When the stimulation was applied, a coaxial coil containing a pulsed current was placed outside. So the electrical field evoking the nerve was formed because of the generation of inductive electromotive force in the inductive coil. Finite element analysis was used to analyze the electric field in the nerve and fiber model was used to predict the generation of action potential. This innovative method was applied on the sciatic nerves of rats. EMG was successfully recorded after the electrical stimulation by the magnetic induction. The results demonstrated that this new method was effective to stimulate the deep nerve.**

I. Introduction

RECONSTRUCTION of motor and sensory functions after spinal cord injury (SCI) is one of the most challenging tasks in neuroscience. With the development of biomedical engineering and neuroscience, rehabilitation engineers have designed devices and systems that can apply electrical currents to neural tissues for the purpose of restoring absent body functions, a technique referred to as functional electrical stimulation (FES) [1,2]. In addition, research on the application of oscillating field stimulation (OFS) to enhance axonal regeneration has existed for more than 15 years [3,4,5]. OFS, which imposes a weak electric field of about 500μV/mm on the nerve, seems to be safe and effective in enhancing neurological outcome after SCI. However, implantable FES and OFS systems are invasive interventions and multiple risks (surgical technique, wound infection, breakage of wires, battery life, and control device) are associated with theses kinds of therapy [6].

Comparing to the implantable FES and OFS systems, non invasive magnetic stimulation is better accepted by patients. Transcranial magnetic stimulation was first introduced by Barker in 1985 [7]. Recently, this technology was widely used in evaluating corticospinal tract functionality after SCI, which made spectacular progress in research of SCI [8,9]. In contrast to electrical stimulation, the application field of magnetic stimulation is relatively small, because the magnetic field and inductive electric field reduce rapidly as the distance increases. For example, it is not possible to stimulate the descending motor tracts of spinal cord by magnetic stimulation [10]. Moreover, magnetic stimulation can not produce direct current in biological tissue, and it is unlikely to use magnetic stimulation to substitute OFS. Fortunately, some researchers had proved that superthreshold electrical stimulation (3V, 0.1ms at 20Hz) was effective on peripheral nerve regeneration in rats [11,12]. Therefore, magnetic stimulation can be used as a therapeutic method as long as the stimulating depth is deep enough.

This paper proposes a method of deep nerve electrical stimulation by magnetic induction (ESMI). A small inductive coil connecting to the nerve electrodes was implanted under the skin in priority. When the stimulation was applied, a coaxial stimulating coil containing a pulsed current was placed outside. So the electric field, which makes the nerve excited, is formed on the nerve because of the generation of inductive electromotive force (IEMF) in the inductive coil. In this paper, we first proved that ESMI could produce much stronger electric field on the nerve than direct magnetic stimulation. Then finite element analysis was used to analyze the electric field on the nerve, and fiber model was used to predict the generation of action potential. Finally, the compound action potential was successfully recorded in the muscle when ESMI was applied on rats.

II. Principles of ESMI

The key components of ESMI were two coaxial coils. The position of coils and nerve was shown in Fig. 1. A N turns circular coil containing a pulsed current $I(t)$ was fixed on XOY plane. The radius of each turn of the coil was a_i. Nerve electrodes with a distance of d were connected with a single turn coaxial inductive coil with a radius of b. The distance between the two coils was h. The deep nerve was not necessary to be parallel to the coils.

The current $I(t)$ in the stimulating coil was predicted by a series RLC model. The current pulse was generated when a capacitor C, initially charged to a voltage V_0, discharged through a coil whose inductance was L and resistance was R. Let $\beta=R/2L$. In the damped case (β^2-1/LC<0), the current was given by the following expression:

$$I(t) = \frac{V_0}{\omega L} e^{-\beta t} \sin(\omega t) \quad (1)$$

where $\omega^2=1/(LC)-\beta^2$. Actually, a diode was used to reduce oscillations. So the real waveform of the current pulse contained only the first positive peak.

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Guanghao Zhang is with the Institute of Electrical Engineering, Chinese Academy of Sciences. He is also with the graduate university of Chinese Academy of Sciences (e-mail: zhangguanghao@ mail.iee.ac.cn).

Yue Li is with the Institute of Electrical Engineering, Chinese Academy of Sciences (e-mail: liyue@ mail.iee.ac.cn).

Xiaolin Huo is with the Institute of Electrical Engineering, Chinese Academy of Sciences (e-mail: huoxl@ mail.iee.ac.cn).

Tao Song is with the Institute of Electrical Engineering, Chinese Academy of Sciences (corresponding author: phone: +86-010-8254-7164; fax: +86-010-8254 -7164; e-mail: songtao@ mail.iee.ac.cn).

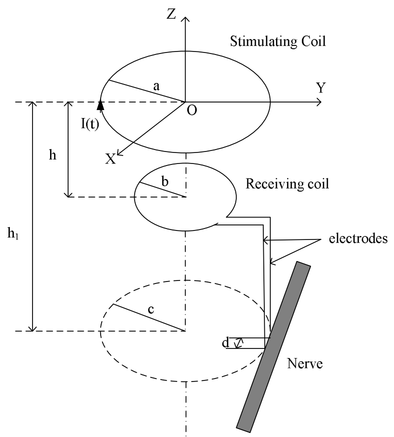

Fig. 1. Position of nerve and coils

The magnetic vector potential **A** in arbitrary point P (ρ, α, z) produced by the stimulating coil was [13]

$$\mathbf{A}(\rho,\alpha,z)=\frac{\mu_0 I(t)}{2\pi}\cdot\sum_{i=1}^{N}\sqrt{\frac{a_i}{\rho}}[(\frac{2}{k_i}-k_i)K(k_i)-\frac{2}{k_i}E(k_i)]\mathbf{e}_\alpha \tag{2}$$

where μ_0 is the permeability of vacuum, $K(k)$ and $E(k)$ were complete elliptic integral of the first and second kind, $\mathbf{e}_\alpha$ was the unit vector on α direction, and

$$k_i=\sqrt{\frac{4a_i\rho}{(a_i+\rho)^2+z^2}} \tag{3}$$

From (2), the magnetic vector potential could be divided into spatial term $A_0(\rho, z)$ and temporal term $I(t)$. Due to Maxwell equation, the electric field $\mathbf{E}_\mathrm{i}$ in arbitrary point was

$$\mathbf{E}_\mathbf{i}(\rho,z)=-A_0(\rho,z)\frac{dI}{dt}\mathbf{e}_\alpha-\nabla\varphi \tag{4}$$

where φ was the accumulating charge on the interface. Then the IEMF E_{IEMF} was equal to the integral along the inductive coil direction. Hence,

$$E_{IEMF}=\int_{coil}\mathbf{E}_i\cdot\mathbf{dl}=2\pi bA_0(b,-h)\frac{dI}{dt} \tag{5}$$

If the distance between the two electrodes was small enough, the electric field between the electrodes E_N was

$$E_N=E_{IEMF}/d \tag{6}$$

In order to study whether ESMI was more effective than direct magnetic stimulation, we defined the effectiveness coefficient η to be the ratio between electric field produced by ESMI and that produced by direct magnetic stimulation. We supposed that the stimulating coil was parallel to the nerve and the interface when magnetic stimulation was applied, because the largest longitudinal electric field gradient, which made the nerve excited, would be formed on the nerve and the accumulating charge could also be neglected. The distances from stimulating coil and Z axis to the nerve were h_1 and c as shown in Fig. 1. The electric field on the nerve produced by direct magnetic stimulation could be expressed as the product of $A_0(\rho, z)$ and dI/dt. Hence,

$$\eta=\frac{E_N}{E_i}=\frac{2\pi bA_0(b,-h_1)}{dA_0(c,-h)} \tag{7}$$

If the inductive coil and the deep nerve were on the same plane and the distance between the nerve and Z axis was equal to the radius of the inductive coil, in other words, h_1=h, c=b, (7) can be written as

$$\eta=2\pi b/d \tag{8}$$

According to (8), the electric field on the nerve was greatly improved after implanting the inductive coil. For example, if b=5mm and d=2mm, the electric field was about 16 times of that produced by direct magnetic stimulation.

III. MODEL OF EXTRACELLULAR STIMULATION

A. Cuff Electrodes

Bipolar silicon cuff electrodes (2 parallel ring stainless steel electrodes) with a diameter of 2mm were used. The two ring electrodes were connected with the two terminals of the inductive coil. Thus, the voltage between the two electrodes was equal to the IEMF. In Fig. 2, the inhomogeneous volume conductor was shown schematically. The length of the volume conductor was 500mm, and the height was 100mm. The potential at the upper, left, and right boundary was set to zero, whereas the normal current density at the lower boundary was zero. The model consisted of a nerve bundle (fascicle) with a radius of 0.7mm, a perineurium with a thickness of 50μm, and a cuff with a length of 16mm, an inner diameter of 2mm and a thickness of 1mm. The electrodes were 4mm from the edge of the cuff. The conductivities were shown in table I [14].

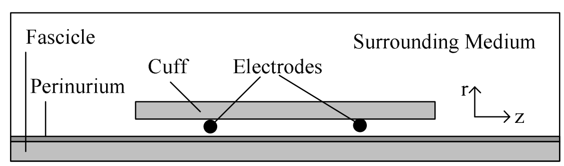

Fig. 2. Volume conductor mode

B. Fiber Model

The fiber model was a cable model, identical to the widely used Mcneal model [15] for extracellular electrical stimulation. The membrane current at node n was equal to the sum of the incoming axial currents and to the sum of the capacitive and ionic currents through the membrane. Hence,

$$C_m\frac{dV_n}{dt}+I_{i,n}=G_a\cdot(V_{n-1}-2V_n+V_{n+1}+V_{e,n-1}-2V_{e,n}+V_{e,n+1}) \tag{9}$$

where C_m, G_a, $V_{\mathrm{e,n}}$, V_n, and $I_{\mathrm{i,n}}$ were the nodal capacitance, axial nodal resistance, extracellular potential at node n, membrane potential at node n, and total ionic current at node n. For stimuli near or greater than threshold, the ionic current at the excitation node was given by

$$I_{i,n}=i_{Na}+i_K+i_P+i_L\ . \tag{10}$$

where the terms on the right side were the individual component of ionic current specified by Schwarz [16], since our experiment animals were Sprague-Dawley rats. The extracellular potential $V_{e,n}$ could be acquired from section A. The waveform of stimulus could be obtained form the temporal part of IEMF. Integration of the differential equations was performed with a simple Euler integration with a step size of 0.1μs.

IV. Experiments and Results

A Effectiveness of ESMI

The magnetic stimulator used in the experiment was designed by our lab. A 10 turns coil with an inner diameter of 30mm and an outer diameter of 60mm was used. The inductance and resistance of the coil were 7.7μH and 0.06Ω. The capacitor was initially charged to 1000V. The radius of single turn inductive coils ranged from 10mm to 60mm. We placed the coaxial inductive coils of different diameters at a distance of 6mm away from the stimulating coil. The two electrodes with a distance of 2mm were placed in saline water. Voltage between the electrodes was measured as IEMF. Electric field produced by direct magnetic stimulation was equal to IEMF divided by the perimeter of inductive coil. Electric field produced by ESMI was equal to IEMF divided by the distance between electrodes. Fig.3 showed the measured values of IEMF in different inductive coils and those calculated by (5). Fig. 4 compared the effectiveness coefficient acquired from experiment data and those calculated by (8). Results showed that (8) was correct and ESMI was more effective than direct magnetic stimulation.

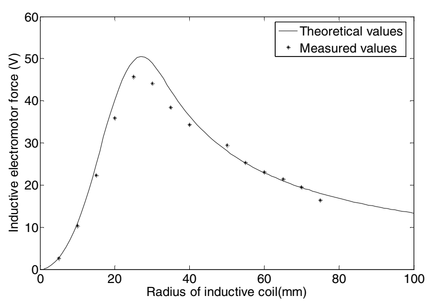

Fig. 3. IEMF in the inductive coils

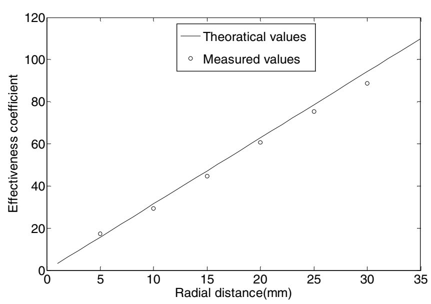

Fig. 4. Effectiveness coefficient as a function of radial distance

B Electric field in the cuff electrodes

The electric field on the nerve was calculated by COMSOL. When the voltage between two electrodes was 1V, the electric field on Z axis was shown in Fig. 5. Results showed that the largest longitudinal electric field gradient appeared just under the electrodes, where action potential would be evoked.

TABLE I
Conductivities

Subdomain		Conductivity (S/m)
Fascicle	Transverse	0.083
	Longitudinal	0.6
Perineurium		0.034
Cuff		1×10^{-12}
Surrounding Medium		1
Electrodes		1.1×10^{6}

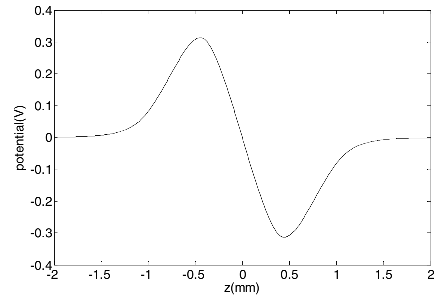

Fig. 5. Electric field on Z axis

C Fiber model

A myelinated fiber with a diameter of 5μm, which located in the axis of the nerve, was studied. Results showed that when the IEMF exceeded 1.4V, the fiber was excited at the cathode. Since our stimulating waveform was biphasic, action potentials would be evoked at each electrode. At the first phase, action potential was evoked at one electrode and it propagated bidirectionally. After about 60μs, action potential was triggered at the other electrode, and then it propagated bidirectionally, too. Since an action potential spent much more time than 60μs to travel form one electrode to the other, the two action potentials would collide at a point between the two electrodes. Finally, there was only one action potential propagating in one direction. The simulated single fiber action potential was described in Fig. 6. Results showed that action potential could be evoked by ESMI successfully.

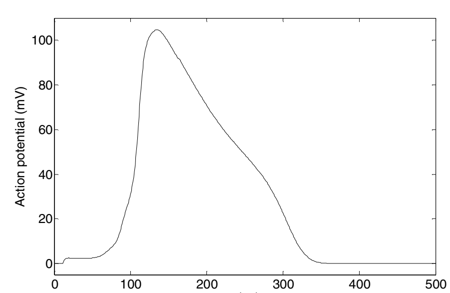

Fig. 6. Action potential as a function of time

D Sciatic nerve stimulation

Five Sprague-Dawley rats were studied. In Fig. 7, the rat and experiment equipment were shown schematically. Bipolar silicon cuff electrodes described in section III were implanted around the sciatic nerves. The two ring electrodes were connected with the two terminals of the single turn inductive coil with a diameter of 14mm. Silicon, a kind of biological compatible material, was used to insulate the implanted components. The inductive coil was fixed under the skin. When stimulation was applied, the coaxial stimulating coil was placed outside. The distance between the two coils was variable. The capacitance in the stimulator was charged to 1000 volts, and then stimulation was applied to all these rats. The compound action potential could be recorded on the muscles of hind limbs when the distance between the two coils was less than 55mm. Direct magnetic stimulations were also applied on these rats, but compound action potential could not be recorded when the distance from stimulating coil to sciatic nerve was 55 mm. Fig. 8 showed the compound action potential evoked by ESMI in the experiment. Thus, ESMI could stimulate deeper nerves than direct magnetic stimulation.

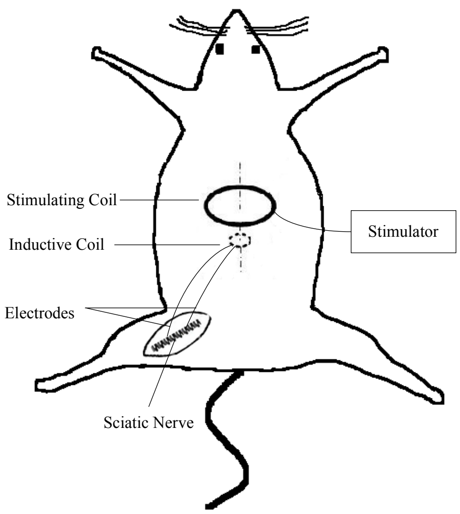

Fig. 7. The rat and experiment equipment

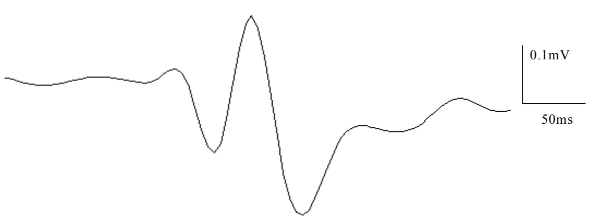

Fig. 8. Compound action potential recorded in the experiment

V. CONCLUSION

This paper proposes a method of nerve electrical stimulation by using the magnetic induction, which abandons the implanted battery and reduces the size of implantable stimulator. Various kinds of nerves, especially deep nerves, such as sciatic nerve, spinal cord and brain, can be effectively excited by this method. However, the waveform of the IEMF is relatively simple, and the frequency of stimulation is constrained by the magnetic stimulator. So it is hard to use ESMI to restore complex motor function.

Although superthreshold pulsed electrical stimulation was effective on peripheral nerve regeneration [11,12], the stimulation was applied only one hour or 2 weeks after injury. It is logical to deduce that longer time electrical stimulation is beneficial. Because implanted batteries could not provide enough energy for long time stimulation, the method of ESMI, which requires external power source and much less implanted components, will be an ideal substitution. Therefore, our future work will focus on the therapeutic effects of daily ESMI on peripheral nerve and spinal cord regeneration.

REFERENCES

[1] K. T. Ragnarsson, "Functional electrical stimulation after spinal cord injury: current use, therapeutic effects and future directions," *Spinal Cord,* vol. 46, pp. 255-274, 2008.

[2] S. F. Giszter, "Spinal cord injury: Present and future therapeutic devices and prostheses," Neurotherapeutics, vol. 5, pp. 147-162, 2008.

[3] R. B. Borgens, J. P. Toombs, A. R. Blight, M. E. McGinnis, M. S. Bauer, W. R. Widmer, and J. R. Cook, "Effects of applied electric-fields on clinical cases of complete paraplegia in dogs," Restorative Neurology and Neuroscience, vol. 5, pp. 305-322, Sep 1993.

[4] R. B. Borgens, J. P. Toombs, G. Breur, W. R. Widmer, D. Waters, A. M. Harbath, P. March, and L. G. Adams, "An imposed oscillating electrical field improves the recovery of function in neurologically complete paraplegic dogs," Journal of Neurotrauma, vol. 16, pp. 639-657, Jul 1999.

[5] S. Shapiro, R. Borgens, R. Pascuzzi, K. Roos, M. Groff, S. Purvines, R. B. Rodgers, S. Hagy, and P. Nelson, "Oscillating field stimulation for complete spinal cord injury in humans: a Phase 1 trial," J Neurosurg Spine, vol. 2, pp. 3-10, 2005.

[6] S. Hamid and R. Hayek, "Role of electrical stimulation for rehabilitation and regeneration after spinal cord injury: an overview," European Spine Journal, vol. 17, pp. 1256-1269, Sep 2008.

[7] A. T. Barker and R. Jalinous, "Non-invasive magnetic stimulation of human motor cortex," Lancet, vol. 1, pp. 1106-1107, 1985.

[8] P. H. Ellaway, M. Catley, N. J. Davey, A. Kuppuswamy, P. Strutton, H. L. Frankel, A. Jamous, and G. Savic, "Review of physiological motor outcome measures in spinal cord injury using transcranial magnetic stimulation and spinal reflexes," Journal of Rehabilitation Research and Development, vol. 44, pp. 69-75, 2007.

[9] H. Nollet, L. Van Ham, P. Deprez, and G. Vanderstraeten, "Transcranial magnetic stimulation: review of the technique, basic principles and applications," Veterinary Journal, vol. 166, pp. 28-42, 2003.

[10] I. J. Scivill, A. T. Barker, and I. L. Freeston, "Finite element modelling of magnetic stimulation of the spine," Proceedings of the 18th Annual International Conference of the Ieee Engineering in Medicine and Biology Society, Vol 18, Pts 1-5, vol. 18, pp. 393-394, 1997.

[11] A. A. Al-Majed, C.M. Neumann, T.M. Brushart, and T. Gordon, "Brief electrical stimulation promotes the speed and accuracy of motor axonal regeneration," J. Neurosci. vol. 20, pp. 2602–2608, 2000.

[12] M. Vivó, A. Puigdemasa, L. Casals, E. Asensio, E. Udina, and X. Navarro, "Immediate electrical stimulation enhances regeneration and reinnervation and modulates spinal plastic changes after sciatic nerve injury and repair," Experimental Neurology, vol. 211, pp. 180–193, 2008.

[13] Y. Lei, Magnetic field of axial symmetry coil, China metrology publishing house, 1991

[14] J. J. Struijk, "The Extracellular Potential of a Myelinated Nerve Fiber in an Unbounded Medium and in Nerve Cuff Models," Biophysical Journal vol. 72 , pp. 2457-2469, 1997.

[15] D. R. McNeal, "Analysis of a model for excitation of myelinated nerve," IEEE Transactions on biomedical engineering vol. BME-23, pp. 329-337, 1976.

[16] J. R. Schwarz and G. Eikhof, "Na currents and action potentials in rat myelinated nerve fibres at 20 and 37°C," Pflügers Archiv European Journal of Physiology, vol. 409, pp. 569-577, 1987.