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The Characteristics on the Change of Cerebral Cortex using Alternating Current Power Application for Transcranial Magnetic Stimulation

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A transcranial magnetic stimulation device is a complicated appliance that employs a switching power device designed for discharging and charging a capacitor to more than 1 kV. For a simple transcranial magnetic stimulation device, this study used commercial power and controlled the firing angle using a Triac power device. AC 220V 60 Hz, the power device was used directly on the tanscranial magnetic stimulation device. The power supply device does not require a current limiting resistance in the rectifying device, energy storage capacitor or discharge circuit. To control the output power of the tanscranial magnetic stimulation device, the pulse repetition rate was regulated at 60 Hz. The change trigger of the Triac gate could be varied from 45° to 135°. The AVR 182 (Zero Cross Detector) Chip and AVR one chip microprocessor could control the gate signal of the Triac precisely. The stimulation frequency of 50 Hz could be implemented when the initial charging voltage Vi was 1,000 V. The amplitude, pulse duration, frequency stimulation, train duration and power consumption was 0.1-2.2T, 250~300 μs, 0.1-60 Hz, 1-100 Sec and < 1 kW, respectively. Based on the results of this study, TMS can be an effective method of treating dysfunction and improving function of brain cells in brain damage caused by ischemia.

Keywords : magnetic, stimulation, alternating current, triac, AVR182

1. Introduction

Transcranial Magnetic Stimulation can stimulate the focal cerebral cortex using an artificial magnetic pulse artificially without cutting the head when there is brain disease. The device was developed to examine the brain spinal nervous system by observing the muscle reaction and stimulating the mobility areas of the brain. It has been used in clinics for treatment and nervous system disease diagnosis since the 2,000s. The device is also used in bone setting treatment in muscle pain, neuralgia, fracture patients in a magnetic low frequency physical therapy device using magnetic stimulation and is also used for urinary incontinence treatment by stimulating the pelvic nerve. Magnetic stimulation produces less pain on stimulation compared to electrical stimulation and can stimulate the brain area and spinal nerves in areas where it is difficult to insert electrodes. One advantage is that the stimulation process is non-contact but the stimulation device is large and it is difficult to stimulate areas precisely. Generally, a magnetic stimulation device consists of a capacitor that stores the energy needed for stimulation and a coil forming a strong magnetic field by discharging. A magnetic field of pulse form is formed by injecting short, strong currents into a coil and the neighboring magnetic field stimulates the nerves by inducing an eddy current within the human tissues. An eddy current employs the same principle as a current being injected through electrodes in electrical nerve stimulation. A magnetic field of sufficient intensity should be generated in a pulse width of hundreds of μs.

The energy storage capacitors of the discharge circuit and stimulation coils form a L-C resonant circuit so that a sufficiently strong current can flow in the stimulation coils. In general, to charge the capacitor of a magnetic stimulation device to more than 1 kV, the power voltage should be increased or a switching power supply device designed for a discharge device should be used. Approximately 50 Hz is needed to induce the continuous contraction of muscles and the power loss occurring during discharge should be considered. To reduce the power supplied by the power supply, a method for collecting the

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magnetic energy accumulated in the stimulation coil to a capacitor again is used but this method has the disadvantage of excessive currents flowing in the circuit momentarily because the capacitor voltage becomes converse and accumulates in the direction of the increased charging voltage. As a result of a connecting power circuit and a capacitor directly, the power output is shorted if the excessive current flow and capacitor voltage become 0 V. To solve these problems, a large inductor between the charging power and capacitor is connected for buffering. In addition, the problems of a volume increase and price rise are encountered. To compensate for the many problems of the general magnetic stimulation device, this study applied normal power, and controlled the firing angle using a Triac power device for transcranial magnetic stimulation device.

2. Circuit Topology and Operational Principle

The designers of transcranial magnetic stimulation device always consider the output power maximization of various transcranial magnetic stimulation devices against general constraints, such as size, cost, efficiency and stability. Since a transcranial magnetic stimulation device employs a rectifying device, energy storing capacitor or alternating (60 Hz) current directly, electronic parts, such as the internal impedance of leakage transformer and current limiting resistor of discharging circuit, are not needed. Therefore, the transcranial magnetic stimulation device has many advantages in terms of cost, size and efficiency compared to a typical DC power supply. On the other hand, the output power of the general transcranial magnetic stimulation device can be controlled by the voltage regulator of a slide type. Controlling the output power of the transcranial magnetic stimulation device precisely is difficult in this way. To solve this problem, a simple transcranial magnetic stimulation device was introduced using a Triac (TG35C) semiconductor switch by being connected to the voltage regulator instead of the main transformer. The output power of desired transcranial magnetic stimulation device could be controlled by controlling the trigger angle of the exchange line and the number of pulses in the first of the leakage transformer. The characteristic of this method is that the resistance of the current limit function and energy storing condenser, rectifying device and separate power supply device are not required. Since magnetic fields are not harmless to humans and they can impact nerve tissues and muscle tissues by penetrating deeply into the body, the treatment effect is quite good. Moreover, depression, insomnia, bipolar disorder *etc.* induced by stress can be treated if the spinal nerves and peripheral nerves are treated by the magnetic field.

To treat by penetrating deeply into the body, a waveform with a narrow pulse width and rapid rise time is best. As there is little noise and heat, it may be most suitable for human magnetic treatment. Therefore, the appropriate waveform can be a Biphasic type and all domestic and foreign PMF (Pulsed Magnetic Field) types employ the Biphasic type. The existing method has disadvantages in that the energy transfer efficiency decreases rapidly with increasing pulse repetition rate and the pulse shapes cannot be changed variously. Since the Triac can be triggered on and off, the pulse rise time can be shortened and various pulse shapes can be made. On the other hand, there is some loss of energy transfer according to the Pulse repetition rate rise. The entire system can be divided largely into a stimulating coil, pulse power supply, AVR 182 leased circuit, and control part using AVR one-chip-microprocessor. To stimulate nerves with a magnetic pulse, the size of the electric field induced by a neighboring magnetic field should be large enough to stimulate the nerves. An electric field more than tens of V/m should be induced at the nerve part and to induce with this electric field, a 1-2Tesla magnetic field should be switched from the epidermis and the shape of stimulating coil to the nerves should be ~200 μSec. To create this a magnetic field, a current of thousands of A should be shed instantly to the stimulating coil, whose diameter is approximately 0.5 and there are number of coils. As a way of flowing a large current instantly to the stimulating coil, a method to charge an electric charge on a capacitor and discharge this to stimulating the coil temporarily is used. If the current flows in Coil wire, according to Fleming's right hand rule, a magnetic field forms around the coil wire and an electric field is induced in the direction perpendicular to the magnetic field. Eddy currents occur within conductors if a neighboring magnetic field (Time varying magnetic field) flows around the electrical conductors, such as metals. The average electrical conductivity of the human body is approximately 0.3 Simens/m and the electrical conductivity is quite low compared to high quality conductors, such as metals but it has the properties of conductors. Therefore, if a neighboring magnetic field flows around a human body, an eddy current can be induced within a human body similar to general electric conductors. The eddy current induced by the neighboring magnetic field shows the same effect as the current injected directly using electrodes inside the body. As shown in Fig. 1, the device was constructed using AVR 182 (Zero Cross Detector) Chip, Triac control unit, AVR one-chip-microprocessor (AT90S8535) Units and stimulating coil. For the coils,

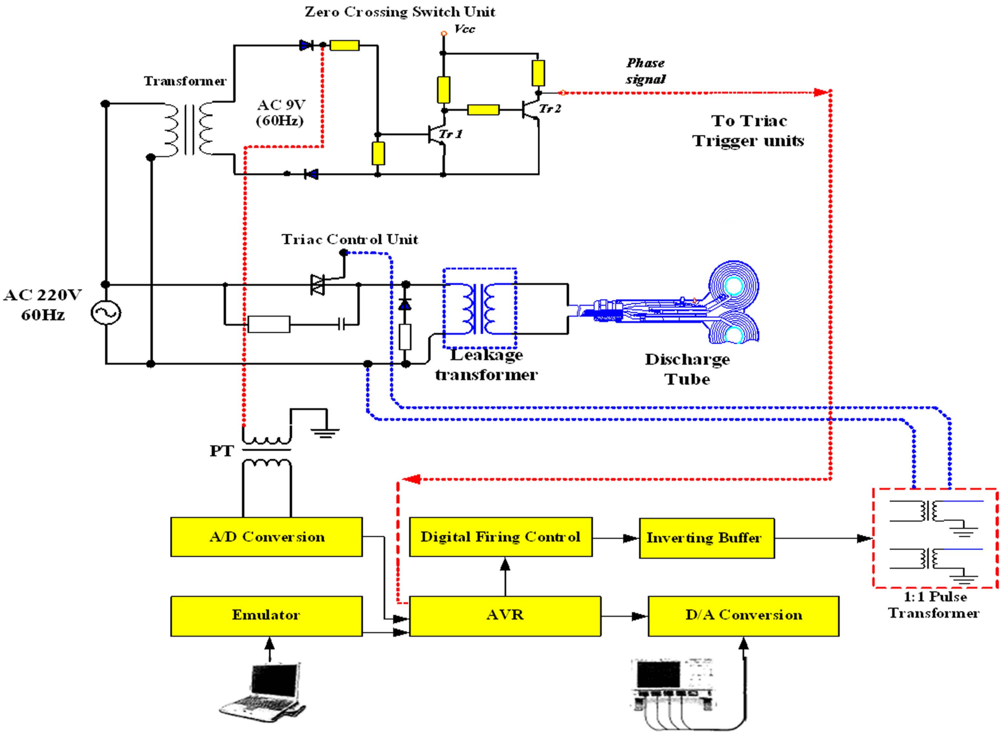

Fig. 1. (Color online) AVR 182 Chip, Triac control unit, AVR one-chip-microprocessor Units, stimulating coil.

copper pipes with an inside caliper and an external diameter of 8 mm and 10 mm, respectively, were used. To prevent interference, such as interference with wires, glass fiber with a 1,500 V withstand voltage and 0.2 mm thickness (t) was used to coat the copper wires, and for complete shielding, materials with a 5,000 V withstand voltage and 0.2 mm thickness (t) were used. The full length of the stimulating coil was approximately 3.5 m, and the pre stage and post stage were pressed approximately 12 cm and 60 cm, respectively, and final 6 cm end was pressed. For various purposes, the materials durability was increased by heat treatment (tempering) of 650ºC at the end for 5 minutes and a total of 10 minutes. When pressing, a 5.4 cm panel followed by a 6 cm panel was added.

3. Experiment and Results

A large stimulating coil with the Helix type was made from Litz wire (number of leading wires: 280 strands). The total number of turns were 10 turns and the diameter of it was 150 mm. The inductance of this coil was approximately 9 μH and the maximum magnetic field strength was 1.2Tesla. The stimulating coil was designed to treat large parts, such as the head, back, shoulders, waist, belly etc. A small stimulating coil was made with the Helix type using Litz wire (number of leading wires: 180 strands). The total number of turns were 10 turns and the diameter of it 100 mm. The inductance of the stimulating coil was approximately 9 μH and the maximum magnetic field strength was 2.2Tesla. This stimulating coil can be used for young children or small parts, such as the neck or elbows, ankles etc. A figure-of-eight coil probe was made in the form of a figure of 8 shape using Litz wire (number of leading wires: 150 strands) and connecting 2 Helix types. The total number of turns were 10 turns and the diameter of it were 80 mm. The inductance of the stimulating coil was approximately 9 μH and the maximum magnetic field strength was 1.2Tesla. The stimulating coil can be used mainly when treating heads or certain areas concentrically. In particular, care should be taken when designing transcranial magnetic stimulation device because the magnetic field strength, frequency, Train Time, Pause time and treatment time etc. need to be adjusted. The power device can control the output of the transcranial magnetic stimulation device by changing the pulse repetition rate as a constant pulse width. The repetition rate was designed to be varied from 5-60 Hz and the maximum pulse voltage that can be obtained was approximately 2 kV. Figure 2 shows the one chip microprocessor used to control the signal to drive the Triac to operate the transcranial magnetic stimulation device using commercial power. After a decompressing voltage was flown to the switching part and an AC voltage (220 V) of same-phase

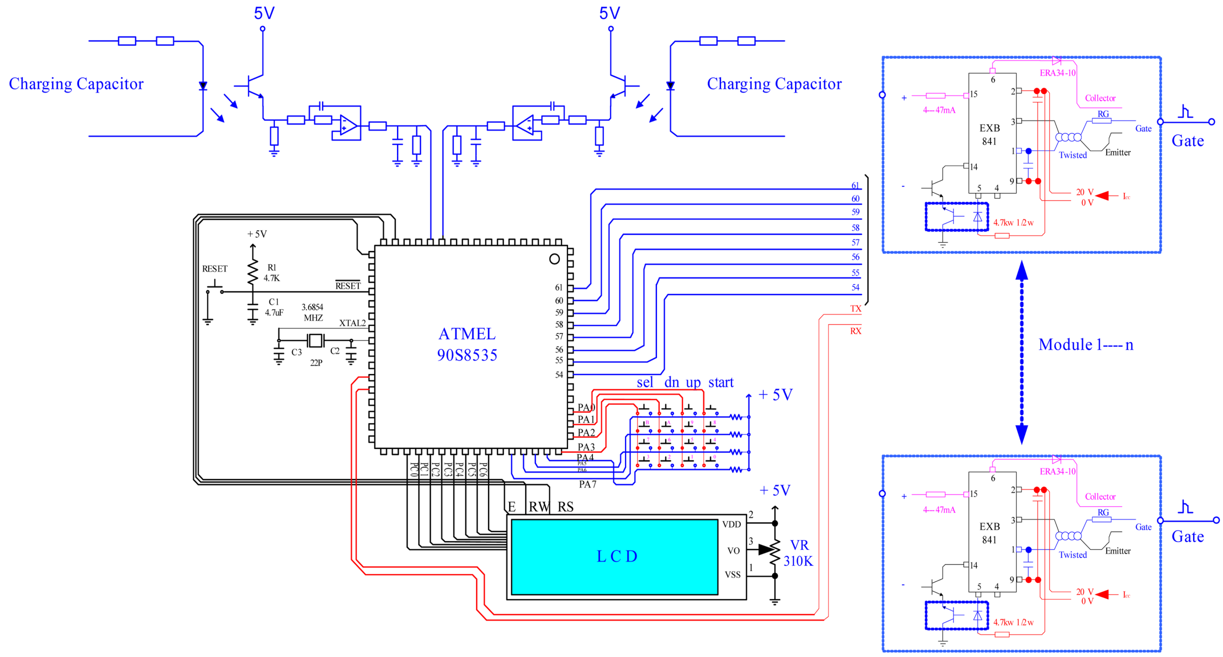

Fig. 2. (Color online) One Chip microprocessor for control and the signal to drive the Triac to operate the transcranial magnetic stimulation device using commercial power.

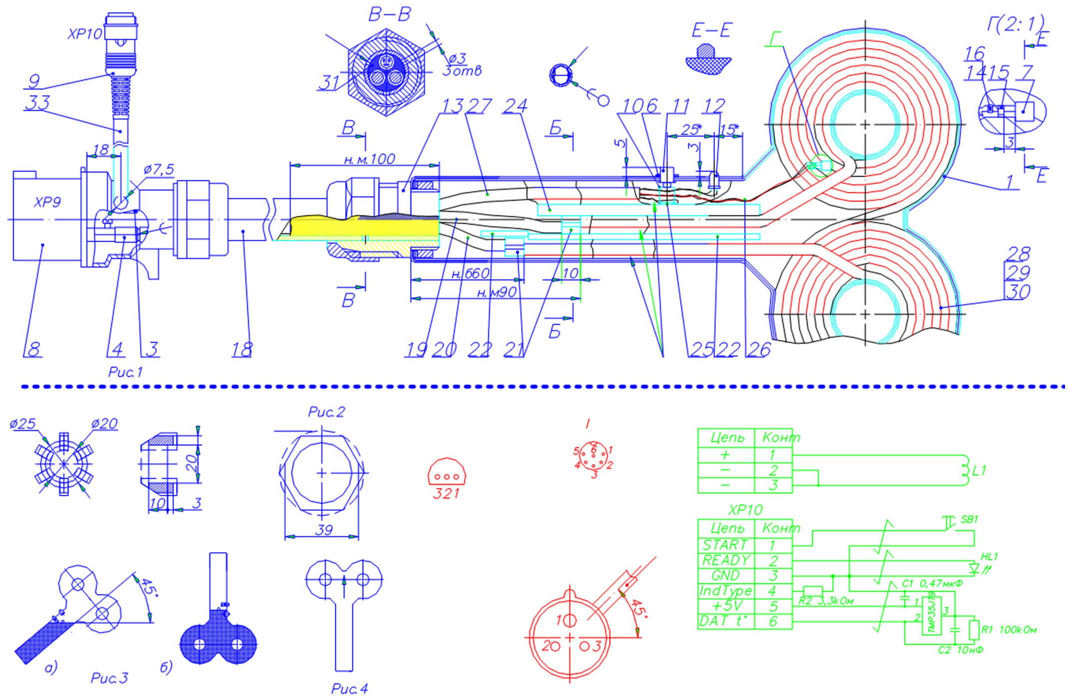

Fig. 3. (Color online) It is listed by site stimulation coil with the actual design.

as 9 V through a small transformer, the current was flown to the base of a transistor Tr1 through a diode and resistor. The signal through a differentiator and amplifier at the point where the voltage flown to the base of Tr1 becomes 0, which generates a pulse width of approximately 500 μs with a 5 V square-wave pulse at the collect of T2. In Fig. 3, It is listed by site stimulation coil with the actual design. In Fig. 4, In experiments, we show a block diagram of a complete magnetic stimulation. In Fig. 5, Checking for zero signal waveform coming from the AVR chip

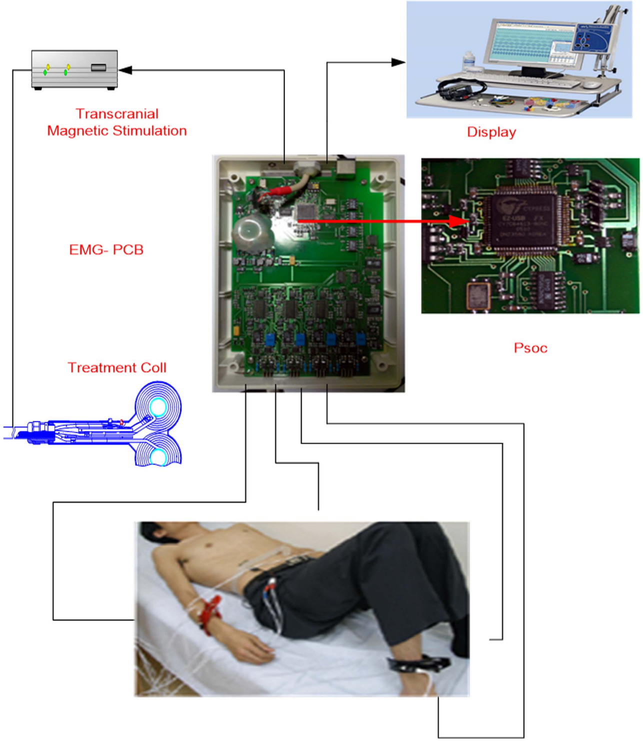

Fig. 4. (Color online) In experiments, we show a block diagram of a complete magnetic stimulation.

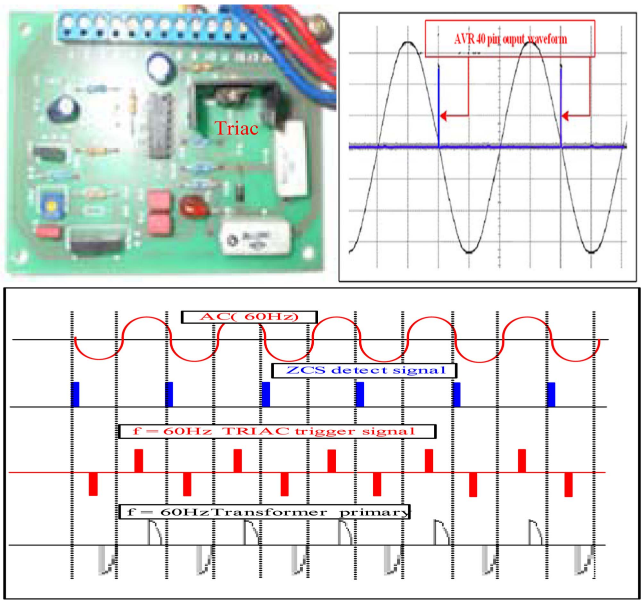

Fig. 5. (Color online) In-output signal of the ZCS according to the commercial power and operating waveform of the Triac trigger signal PCB board.

(182) terminal 40. Input/output signal of the ZCS according to the commercial power and operating waveform of the Triac trigger signal pcb board. Figure 5 shows the checking for a zero signal waveform coming from the AVR chip terminal 40. In this study, a photocoupler was not used in the existing trigger method configuration and the input exchange voltage was flowed into the main controller using an input voltage sensor, which was changed to a synchronized coordinate system voltage. The phase angle used for the coordinate transformation was a synchronous coordinate transformation and PI controller, which synchronizes the phase angle with the power voltage. The phase angle determines the firing angle of the Triac compared to the command signal and makes a pulse shape to trigger the Triac in the arithmetic value. The concept is to measure the input power using the voltage sensor and voltage information coming through the bridge diodes in commercial power. The pulse method to directly trigger the phase angle inside the AVR-Chip to SCR has a very simple circuit and might have strong characteristics in the external environment, such as the reliability improvement, noise of the power voltage, secular changes, temperature characteristics as well as linear control, and the characteristic of an existing analog method. In Fig. 5, Input/output signal of the ZCS according to the commercial power and operating waveform of the Triac trigger signal PCB board. The high leakage transformer switching part consists of a Triac, high voltage leakage transformer and resistance, capacitor as shown in Fig. 1. A Triac was used as the switching device to switch the voltage from the AC and high voltage leakage transformer (1^{st}: 220 V, 2^{nd}: 5 kV, road transport) for a neon transformer to a flowing switching pulse with a low voltage by changing it into a high voltage to a discharge tube. To prevent saturation of the magnetic flux of the transformer by discharging the energy accumulated in the inductive load of a transformer, the resistance and return current diode was connected to the first side of the transformer in parallel. In particular, the Zero Cross Detector circuit senses the zero voltage of the exchange line and transmits information to the AVR one chip microprocessor. A 60 pulse per second occurs in at zero voltage of the AC line of the AVR 182 circuit. The desired trigger pulse repetition rate of the Triac gate is given by the ratio of the input pulse in a 40 output pulse terminal of the AVR. To trigger the influence of the initial value of input voltage on the transcranial magnetic stimulation device output power, the AVR-One-chip-Microprocessor controls the trigger angle from 45° to 135°. In Fig. 6, IGBT drive module's EXB841 of FUJI is used for delay, repetition and various pulses. As a transformer for a neon sign, there is a large leakage reactance that limits the high current due to the sudden impedance drop of the secondary circuit load Since the discharge initial voltage Vi and the discharge complete voltage Ve is 1,100 V and 760 V, respectively,

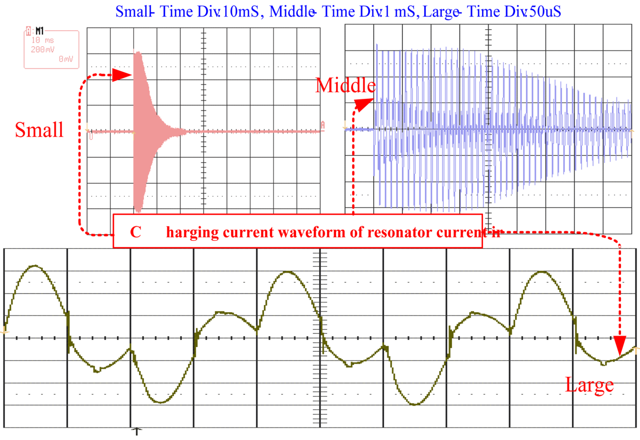

Fig. 6. (Color online) EXB841 of FUJI is used for delay, repetition and various pulses.

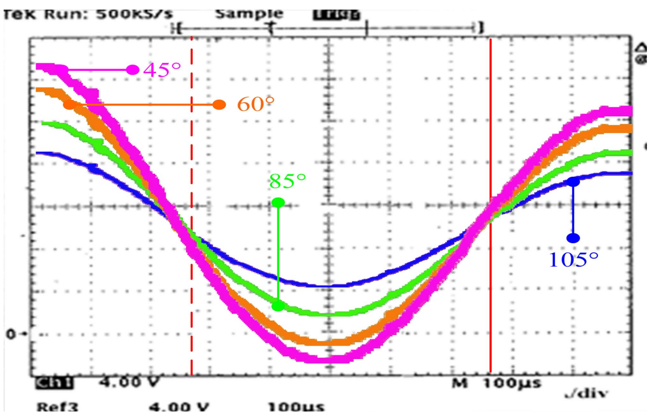

Fig. 7. (Color online) Discharge voltage waveform of the discharge circuit according to the firing angle.

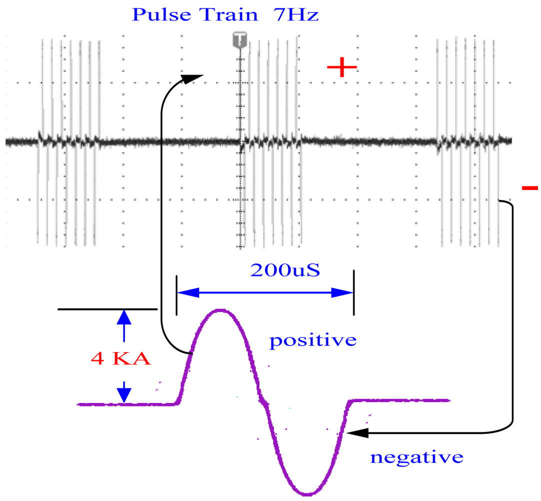

Fig. 8. (Color online) Stimulating coil discharge current waveform in a repetition pulse.

the power capacity can be calculated using the following formula:

$$\mathrm{Pd} = \frac{1}{2}(Cf_c)(V_i^2 - V_e^2),$$

$$[C = 80 \text{ uF}, Fc = 50 \text{ Hz}, Vi = 1{,}000, Ve = 760]$$

Pd becomes 844W. Figure 7 shows the transcranial magnetic stimulating coil discharge voltage waveform of the discharge circuit according to the firing angle. Within 20mS after the discharge, it can be charged with 1,000 V and discharge the initial voltage (Vi), and a stimulation frequency of 50 Hz can be employed if Vi is 1,000 V. In

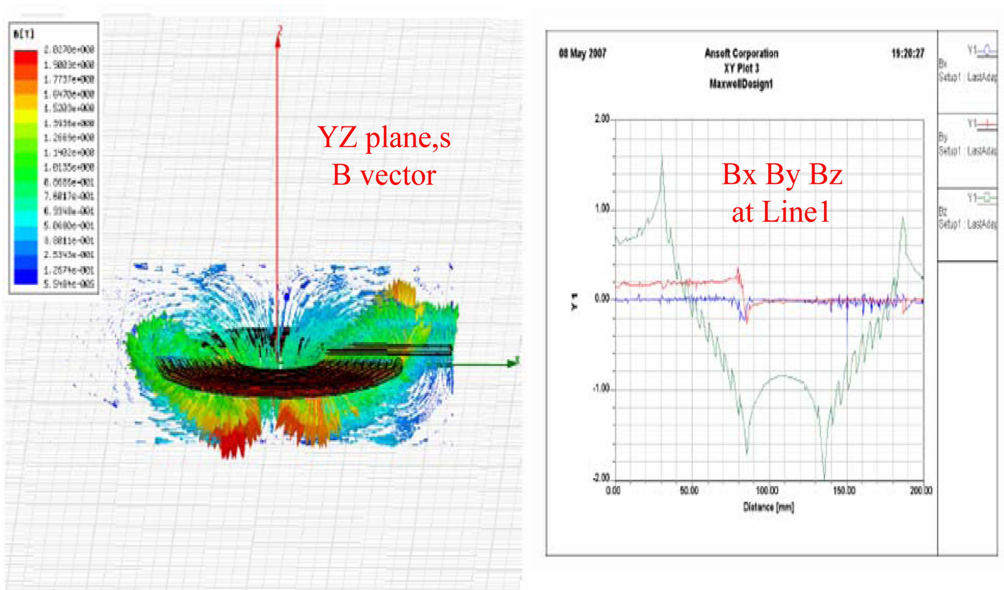

Fig. 9. (Color online) It shows the results of simulation Helix type Coil analysis of the stimulation coil with the Ansoft Tool.

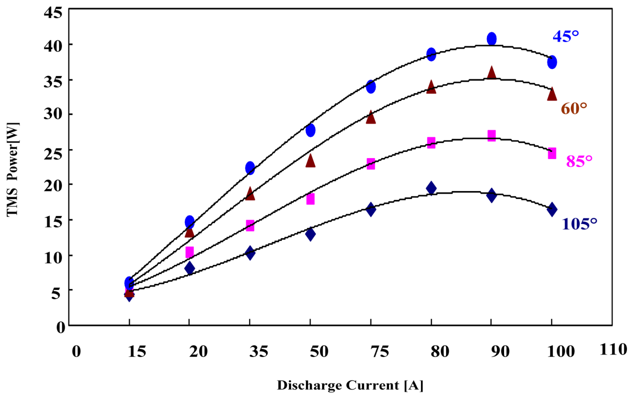

Fig. 10. (Color online) Transcranial magnetic stimulating coil discharge voltage waveform of the discharge circuit according to the firing angle.

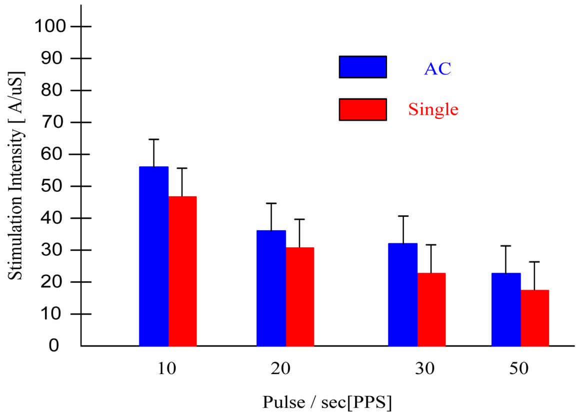

Fig. 11. (Color online) Output comparison of the proposed circuit and general control method.

Fig. 8, Stimulating coil discharge current waveform in a repetition pulse. In Fig. 8, Output relationship according to the radius depending on the discharge current. When the initial voltage discharge was 1000 V, a neighboring magnetic field of 1 Tesla formed at a distance 50mm from the stimulation occurring coil. Even if the IGBT is switched during the discharge operation period, the rating is not exceeded and each part of the voltage and current of the IGBT peripheral circuits is stable. In Fig. 9, It shows the results of simulation (Maxwell 3D: YZ plane B vector, Bx, By, Bz magnitude) Helix type Coil analysis of the stimulation coil with the Ansoft Tool. In Fig. 10, Transcranial magnetic stimulating coil discharge voltage waveform of the discharge circuit according to the firing angle. In Fig. 11, Output comparison of the proposed circuit and general control method. In Fig. 12, Experimental waveforms according to load (no-load). Fig. 13 the implementation of the experimental apparatus, In this study. In Fig. 14, show the waveform of each area along the stimulation coil as a whole.

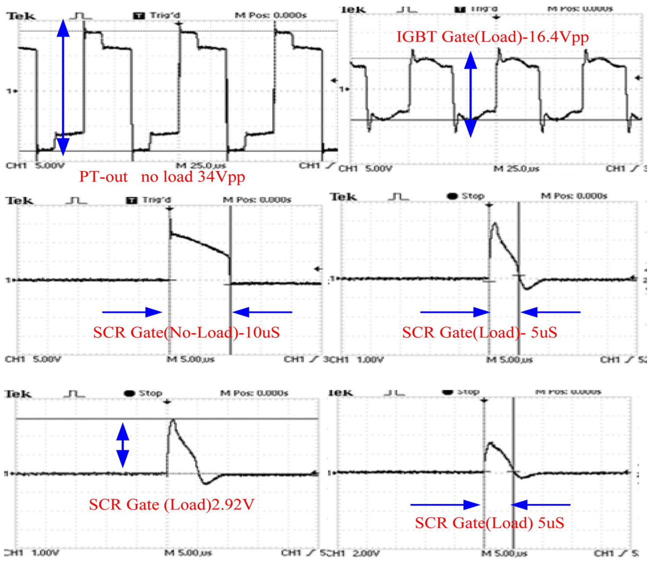

Fig. 12. (Color online) Experimental waveforms according to load.

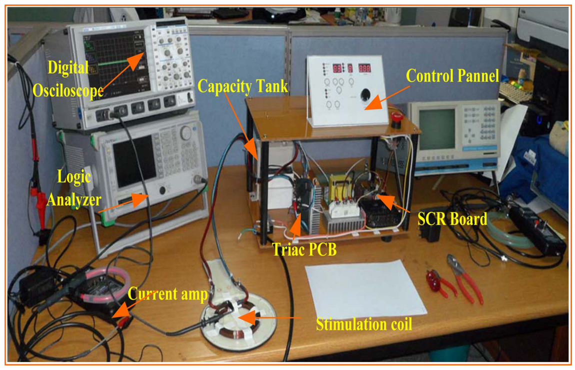

Fig. 13. (Color online) Implementation of the experimental apparatus.

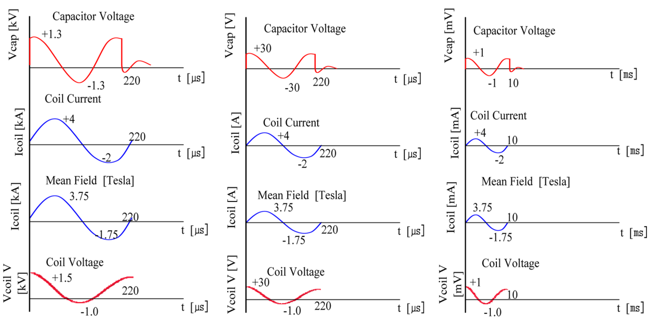

Fig. 14. (Color online) Show the waveform of each area along the stimulation coil as a whole.

4. Conclusion

TRIAC power devices were used in a simple transcranial magnetic stimulation device and examined by applying them to commercial power to control the firing angle. The voltage was switched directly from the voltage of commercial power to the power device on transcranial magnetic stimulation device. The power supply device does not require a rectifying device, energy storage capacitors or current limiting resistor on discharge circuit. The pulse repetition rate was adjusted to 60 Hz to control the transcranial magnetic stimulation device at the output power. The trigger of the Triac gate can be changed from 45° to 135°. The AVR 182 Chip circuit and AVR one chip microprocessor can control the gate signal of the Triac exactly. A 50 Hz stimulating frequency could be implemented when the initial charge voltage Vi was 1,000 V. The amplitude, pulse duration, frequency, stimulation train duration, pauses between trains and power consumption were 0.1-2.2T, 250-300 μS, 0.1-60 Hz, 1-100 Sec, and < 1 kW, respectively. Based on the results of this study, TMS can be an effective method of treating dysfunction and improving function of brain cells in brain damage caused by ischemia.

Acknowledgement

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