


2015

Methods of magnetic field switching for biomedical and power applications

Narimdinda Robert Bouda
Iowa State University

Follow this and additional works at: <https://lib.dr.iastate.edu/etd>

 Part of the [Electrical and Electronics Commons](#), and the [Electromagnetics and Photonics Commons](#)

Recommended Citation

Bouda, Narimdinda Robert, "Methods of magnetic field switching for biomedical and power applications" (2015). *Graduate Theses and Dissertations*. 14347.
<https://lib.dr.iastate.edu/etd/14347>

This Thesis is brought to you for free and open access by the Iowa State University Capstones, Theses and Dissertations at Iowa State University Digital Repository. It has been accepted for inclusion in Graduate Theses and Dissertations by an authorized administrator of Iowa State University Digital Repository. For more information, please contact digirep@iastate.edu.

Methods of magnetic field switching for biomedical and power applications

by

Narimdinda Robert Bouda

a

A thesis submitted to the graduate faculty

in partial fulfillment of the requirements for the degree of

MASTER OF SCIENCE

Major: Electrical Engineering

Program of Study Committee:
Jiming Song, Co-Major Professor
Mani Mina, Co-Major Professor
Udoyara Tim

Iowa State University

Ames, Iowa

2015

Copyright © Narimdinda Robert Bouda, 2015. All rights reserved.

TABLE OF CONTENTS

LIST OF FIGURES	iv
LIST OF TABLES	v
ACKNOWLEDGEMENTS	vi
ABSTRACT	vii
CHAPTER 1 INTRODUCTION	1
1.1. What is transcranial magnetic stimulation (TMS)?	1
1.2. TMS fundamentals and technical Aspects	2
1.2.1. Circuit requirements	2
1.2.2. Coil requirements	4
1.2.3. Human anatomy	5
1.3. Conclusion	5
CHAPTER 2 CONCEPTUAL OVERVIEW	6
2.1. A common understanding	6
2.2. How to design a high current magnetic field generator	7
2.3. Conclusion	9
CHAPTER 3 HIGH CURRENT MAGNETIC FIELD GENERATOR	10
3.1. HCMFG	10
3.2. Topology, power supply, energy storage and switches	11
3.3. Control circuit, coil considerations and energy efficiency	12
3.4. Conclusion	15
CHAPTER 4 EXPERIMENTAL SETUP	16
4.1. Electronic System	16
4.2. Practical considerations	18
4.3. Conclusion	20
CHAPTER 5 TESTING AND VERIFICATION	21
5.1. Rectifier Circuit	21
5.2. Gate driver	22
5.3. Magnetic field measurements	22
CHAPTER 6 CONCLUSIONS AND FUTURE CONSIDERATIONS	24
6.1. Summary	24
6.2. Contributions of work	24
6.3. Accomplishments	24
6.4. Meeting the standards	25
6.5. Bibliography	26
APPENDIX A. TECHNICAL SPECIFICATIONS	32
APPENDIX B. HARDWARE	36
APPENDIX C. SOFTWARE	43

LIST OF FIGURES

Figure 1 The TMS applied field strength gradient to human brain and example of monophasic (top) and biphasic (bottom) pulse profile [22].....	3
Figure 2 H bridge and half modify bridge topology using MOS or IGBT switches [22].....	7
Figure 3 Taxonomy of the design process	8
Figure 4 Topology of the IGBT circuit (HCMFG ₂).....	11
Figure 5 Topology of the MOSFET circuit (HCMFG ₁)	11
Figure 6 Gate driver for the HCMFG ₂	13
Figure 7 Control circuit for NMOS (HCMFG ₁)	13
Figure 8 Control circuit for PMOS HCMFG ₁	14
Figure 9 Rectifier circuit of the HCMFG ₂	21
Figure 10 Rectifier circuit of the HCMFG ₁	22
Figure 11 HCMFG ₂ Topology	36
Figure 12 HCMFG ₁ Topology	37
Figure 13 1300A pulse.....	38
Figure 14 -650A pulse	38
Figure 15 -590A pulse	39
Figure 16 -1300A pulse	40
Figure 17 -1300A pulse	40
Figure 18 [+250, - 460] A pulse.....	41
Figure 19 [+700, - 890] A pulse.....	41
Figure 20 [+700, - 1100] A pulse.....	42
Figure 21 TMS GUI.....	43

LIST OF TABLES

Table 1 Equipment and components used in the experimental setup	32
Table 2 Ports available to the user	32
Table 3 Description of buttons on the stimulator.....	32
Table 4 Device 1 current consumption	33
Table 5 Device 2 current consumption	34
Table 6 Coil type A.....	34
Table 7 Coil type B.....	34
Table 8 Load specification for HCMFG ₁ and HCMFG ₂	35

ACKNOWLEDGEMENTS

I would like to take this opportunity to express my gratitude to everyone who supported me throughout the course of my Master's program. I am thankful for their aspiring guidance, invaluable constructive criticism and friendly advice during the work. I am sincerely grateful to everyone for sharing their truthful and illuminating views on numerous issues related to the project.

I would not have contemplated this road if not for my parents, my guide Dr. Mani Mina, Dr. John Pritchard, and Dr. Robert Weber who instilled within me a love of creative pursuits, life – long learner, science, all of which finds place in this thesis. With their enthusiasm, inspirations and their great efforts in explaining things, I have garnished my personal repertoire giving me the opportunity to learn, think, challenge, and grow.

I am grateful to all my advisors Ali Soltanshahi, Tony Moore, and Vicky Thorland-Oster for assisting me in many different ways. Words fail me to express my appreciation for Eurydice Ulysses for her support and generous care. Last but not the least Neelam Prabhu Gaunkar for her help with the thesis formatting and suggestions for improvement. Finally Thank you to everyone I have interacted with directly and indirectly and in particular the mentoring team, senior design Team (Greg Bulleit, Matt Stegemann, Joshua Abbott, Spencer Ulven, Z. Wang, K Zheng, S. Chiey)

ABSTRACT

Surrounded by electromagnetic (EM) fields humans continuously interact and utilize EM fields for domestic devices, telecom systems, power systems and different medical applications including MRI (Magnetic Resonance Imaging) and TMS (Transcranial Magnetic Stimulation). During the last century, numerous new electronic devices and technologies have been developed leading to an exponential increase in exposure to Electromagnetic Fields. Although all of these applications are byproducts of Maxwell's experimentation, there are considerable differences in the way they interact with us.

In the treatment of noninvasive treatment of the human brain, the majority of the research has been mainly focused on realizing systems that can produce gigantic current and magnetic fields. Transcranial Magnetic Stimulation (TMS) in this application has provided numerous opportunities and possibilities. It has shown more promise in the therapeutic role for treatment of neurological disorders. TMS researchers are working towards improving the technical developmental tools for modeling, and magnetic field generation study for deep body penetration. The development stage of these devices with associated risks in manufacturing, tight medical tolerances, cost and reliability pose important research challenges which are addressed in this research.

The objective of this research is to understand, develop and offer alternative methods for the implementation of several magnetic field generators that satisfy the requirement for magnetic field applications in medical, and other fields. This work presents a continual progression of the magnetic field generators from larger scale to small scale with variable energy consumption, high speed, and deployable systems. Additionally challenges and practical designs are presented

CHAPTER 1 INTRODUCTION

The functioning of the human mind has been one of the main incentives for philosophers throughout the ages. The need to examine and measure different aspects of the brain anatomy has enhanced the recent developments in neuroimaging techniques. As these techniques have become more affordable and accessible for research, they have allowed an increasingly questioning attitude in making use of neuroimaging methods. Several neuroimaging techniques have provided correlational maps of cognitive processes in the adult human brain at different levels of temporal and spatial detail. Moving beyond a correlational description of the relationship between brain and the behavior was the fresh approach offered by transcranial magnetic stimulation [1], [2], [3], [4], [5-11]. This chapter gives a brief introduction on how high current magnetic field generators contribute to advanced electromagnetic models and medical applications. The chapter also introduces the ideology behind transcranial magnetic stimulation and circuit requirements.

1.1. What is transcranial magnetic stimulation (TMS)?

Electromagnetic induction first described by Michael Faraday in 1831 is defined by: the act of producing a current in a conductive object by using a time-varying magnetic field. It is one of the most relevant experimental observation for magnetic stimulation [12], [13], [14], [15], [21]. Faraday wound two coils on an iron ring and ultimately observed when the coil on one side was connected or disconnected from a power source, an electrical current passed through the coil on the other side. The iron ring acted as a medium linking the magnetic field from the first coil to the second. A change in the magnetic field, related to the changing current in the first coil, induced a current in the second coil. The experiment was enhanced due to the coupling effect given by the iron ring between the two coils. The same principle is applied in non-invasive magnetic stimulation in which the stimulating coil acts as one coil, space as the medium for the flow of the magnetic field and the electrically conductive matter in the human body as the second coil. Over the years there has been an expansion in the development and use of magnetic stimulators worldwide in medical applications. Commercial stimulators produce pulse rates as high as 1250 Hz with pulse stimulation intervals as low as 1 μ s (microsecond) with complete digitization. A growth in research interest has produced a marked spread into the clinical areas of diagnosis, and therapy. The ability of magnetic stimulation to induce electrical currents within

the body tissue allows influencing of many of the functions in the human body. Additionally it allows to reach deep neural structures such as the motor cortex and spinal nerve without pain and non-intrusion [12], [13], [14], [15], [16-21]. TMS is gaining great interest in psychiatry and is providing an extra boost to the therapeutical treatment methodologies provided by clinicians.

1.2. TMS fundamentals and technical Aspects

1.2.1. Circuit requirements

Magnetic stimulators consist of two different parts: a high current pulse generator producing discharge currents of 1000 A (amperes) or more; and a stimulating coil producing magnetic pulses with field strengths of approximately 1 tesla or more with a variable pulse duration. The discharge current flows through the stimulating coil to generate the necessary magnetic pulse. This pulse induces current in electrically conductive regions of the human body. If the induced current is of high amplitude and duration it will stimulate neuromuscular tissue in the same way as with conventional electrical stimulation. The first commercial magnetic stimulators originated from Sheffield in 1985 [16-21]. A typical magnetic stimulator consists of a capacitor charging or discharging alternatively with the appropriate control and safety electronics. Using the charging circuitry the energy storage system (capacitor) is charged to a level which can be up to a maximum of 3000 volts (kV) depending on the device. When the device receives an input signal as a trigger, the energy stored in the capacitor is discharged into the stimulating coil. The stored energy, with the exception of the energy lost in the wiring and capacitor, is transferred to the coil and then returned to the instrument to reduce coil heating. The discharge system consists of a switch and an electronic device, either a power metal oxide semiconductor field effect transistor (MOSFET) or insulated gate bipolar transistor (IGBT) or silicon controlled rectifier (Thyristor), and is capable of switching large currents in a few microseconds. Power MOSFETs, IGBTs and Thyristor conduct current only in one direction. As indicated in Figure 1 there are two types of waveforms: monophasic or biphasic which is commonly found in magnetic stimulation devices.

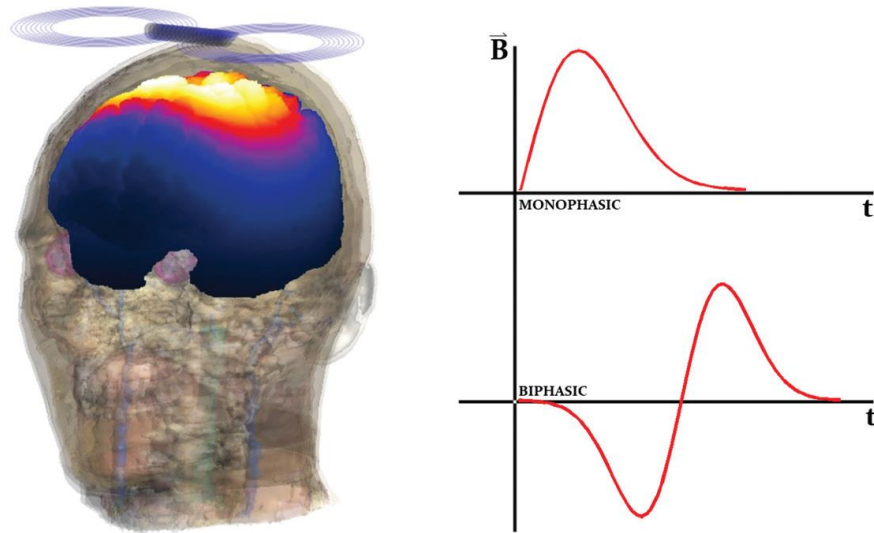


Figure 1 The TMS applied field strength gradient to human brain and example of monophasic (top) and biphasic (bottom) pulse profile [22]

Standard monophasic stimulators generate an over damped sine current pulse and dissipate the whole pulse energy as heat. In biphasic devices, the oscillation is underdamped, which leads to an almost sinusoidal waveform, and recovery of a substantial amount of the pulse energy [16-19]. Monophasic discharge currents reduce heat dissipation in the coil, and increase the accuracy in stimulation. Additionally monophasic pulse allows for a better understanding of the mechanisms involved in magnetic stimulation. The waveforms of coil current for a typical coil when used in monophasic and biphasic configuration are shown in Figure 1. In general the total energy stored in magnetic stimulators ranges from 500 J to 10 kJ, which is coherent with the standards for obtaining an effective magnetic stimulation based on maximizing the peak coil energy. The high power level can be achieved by using a large energy storage capacitor or by having an efficient energy transfer from the capacitor to the coil. Around 500 J of energy is needed to be transferred from the storage capacitor into the stimulating coil in few microseconds. The impulse power output of a typical magnetic stimulator during the discharge phase is approximately 10 MWatts. During the discharge, energy initially stored in the capacitor in the form of electrostatic charge is converted into magnetic energy in the stimulating coil in approximately 400 μ s or longer based on the device characteristics. A fast rate of energy transfer is necessary to achieve a rapid rate of rise of magnetic field. It also needs to be noted that since the magnetic field strength falls off with distance from the stimulating coil, the stimulus strength is at its highest close to the coil surface. The stimulation characteristics of the magnetic pulse,

such as depth of penetration, strength and accuracy, depend on the rise time, peak magnetic energy transferred to the coil and the spatial distribution of the field [20-21].

1.2.2. Coil requirements

The stimulating coil is generally enclosed in a plastic cover, which consists of multiple tightly wound, well insulated copper coils together with other electronic circuitry such as temperature sensors and safety switches. Up to now circular coils with a diameter of 80-100 mm have been the most widely used for magnetic stimulation. In the case of circular coils, the induced tissue current is zero on the central axis of the coil and increases to a maximum in a ring approximately under the diameter of the coil. Stimulation is likely to occur under the winding and not under the coil center. During the stimulating phase, when the magnetic field is increasing from zero to its maximum, the induced current generated in the tissue flows in the opposite direction to the coil current. Although the circular coil is a very useful general purpose coil, the site of stimulation is not well defined. The most notable advancement in coil design has been the double coil (also termed butterfly or figure of eight coil). Double coils utilize two windings normally placed side by side [23], [24], [25], [26-30].

Magnetic field strength is the most widely used figure of merit for the output of magnetic stimulators. Although an important parameter, magnetic field strength is not an accurate measurement of magnetic stimulator performance. Magnetic field strength is defined as the magnetic flux density, however does not reflect the total magnetic flux produced by the stimulating coil over its total area. In a small coil, the magnetic flux is concentrated in a small area, the magnetic field intensity will be higher comparatively in a larger coil, but the field decreases much more rapidly with distance. Hence a small coil is somewhat more powerful in the stimulation of superficial nerves and a large coil is more suitable for structures at depth. The amplitude, waveform and spatial characteristics of the induced current are major parameters in magnetic stimulation. A more accurate indicator of the stimulating power output is the induced charge density per phase defined as the integral of the induced current density during the rise time of the magnetic field. It provides a better indicator of output by taking into account the effects of both amplitude and duration of the induced stimulating current [26-30]. Different areas, such as bone structure (grey matter, fat, white matter) with differing conductivities affect the induced current and its path which in turn is based on a choice of coil on magnetic field strength and induced current with a suitability for its intended application must be taken into account [26-30].

1.2.3. Human anatomy

Magnetic stimulation's most attractive attributes are safety and ease in stimulating most neural structures, unimpeded by fat and bone and without discomfort. The majority of the clinical applications for the technique are for the non-invasive stimulation of the peripheral and central motor pathways. Other uses include stimulation of the left and right prefrontal cortex, visual cortex, language center, cerebellum and peripheral sensory nerves [26-30].

1.3. Conclusion

The main goal of this chapter is to explore through research and understanding of the necessary background information and concept required to design and offer different platform for magnetic stimulation.

CHAPTER 2 CONCEPTUAL OVERVIEW

In this chapter, the necessary concepts for understanding the operation of the HCMFG are presented. The proposed magnetic field generator and the experimental setup are presented. The chapter also gives the reader an inception of the experimental design being identified.

2.1. A common understanding

Transcranial magnetic stimulation (TMS) is one of the leading techniques in the noninvasive treatment of the human brain which offers the advantages of undergoing virtually no surgical exposure and little discomfort. TMS is widely facilitating the study of cerebral functions. Historically and up to date, TMS has offered advantages in the treatment of depression since its first demonstration [31]. In addition, it is also being used for the treatment of neurological disorders as well as post-traumatic stress disorder. TMS is becoming one of the more promising and available methods for diagnostic and therapeutic uses in medicine. Some of the leading research efforts in TMS is to enhance the technical developmental for modeling, and magnetic field generation study for deep body/brain penetration. Having the right kind of signal at the exact position of the brain is necessary for better treatment and analysis [32]. Currently, some of the main research and development need is to create and control the current, voltage, and pulse shapes of magnetic stimulators. The idea of being able to create a given pulse shape at a given depth inside the brain is of great interest to medical experts. In order to achieve this goal one needs to work on a versatile pulse shapes with practical variability of signal level, shape duration, and repetition rates. These challenges pose very important research questions, some of which will be addressed in the following chapters. The current development stage of these devices through traditional means carries risks in manufacturing, due to tight medical tolerances and associated costs and reliability. The medical requirements for these devices has been a real concern for most engineers and scientists. Consequently, there has been growing interest related to TMS in the field of bio-engineering with respect to its therapeutic applications. A pulsed current sent through a coil produces a magnetic field that, in turn, induces electric field in the brain, which can cause neurons to fire. Since the introduction of TMS in 1985 [31-34], the fundamental characteristic of the stimulators has remained intact. Therefore, the main goal is to offer design considerations or modifications for a new high current magnetic field Generator that results in enhanced control, reliability and pulse shaping with adjustable width for TMS.

2.2. How to design a high current magnetic field generator

To design an optimal high current magnetic field Generator for TMS applications, several conditions need to be taken into account: the current requirement, the shape of the pulse, and the component selection for the electronic driver or control circuit. For TMS stimulators, monophasic and biphasic signal shapes are usually the types used. A monophasic magnetic pulse does not change the polarity of the field throughout the duration of the pulse, in contrast, a biphasic pulse does change the polarity of the field. There are several methods used to create a monophasic and biphasic pulse: a full H-bridge and a modified H-bridge two such methods. In general for both circuitry, the switching devices are either power MOSFETs or Insulated Gate Bipolar Transistors (IGBTs). The popular use of IGBTs can be drawn from the fact that they can easily be controlled compared to current-controlled devices in high-voltage and high-current applications [22]. The entire topology is shown in Figure 2. Two types of TMS devices exist in the market, controllable pulse-parameter TMS (cTMS) and repetitive TMS (rTMS).

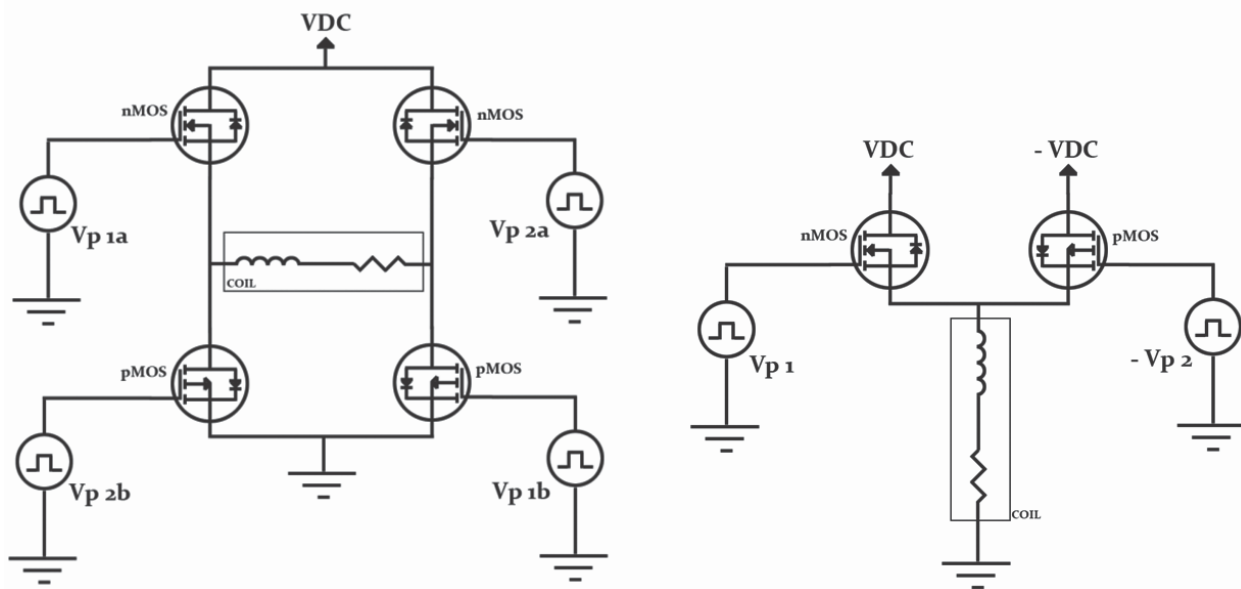


Figure 2 H bridge and half modify bridge topology using MOS or IGBT switches [22]

Based on experimental data presented through the years, rTMS with monophasic pulses have shown stronger and more selective effect than conventional rTMS with biphasic pulses [35-36]. Several topologies have been utilized, however they were not adequate to the current needs

and requirements of such applications which do not give control over the pulse parameters (duration, amplitude, polarity, etc) [35]. In the current research both systems previously enumerate offer flexibility, modularity (pulse duration, and waveform), and their evolving capacity due to an easy use of microcontroller. Both the experimental systems have been designed to allow an easy maintenance and enable any testing on animal living tissues. Figure 3 below describes the necessary steps that enable specific design specifications for the high current magnetic field generator (HCMFG₁ and HCMFG₂).

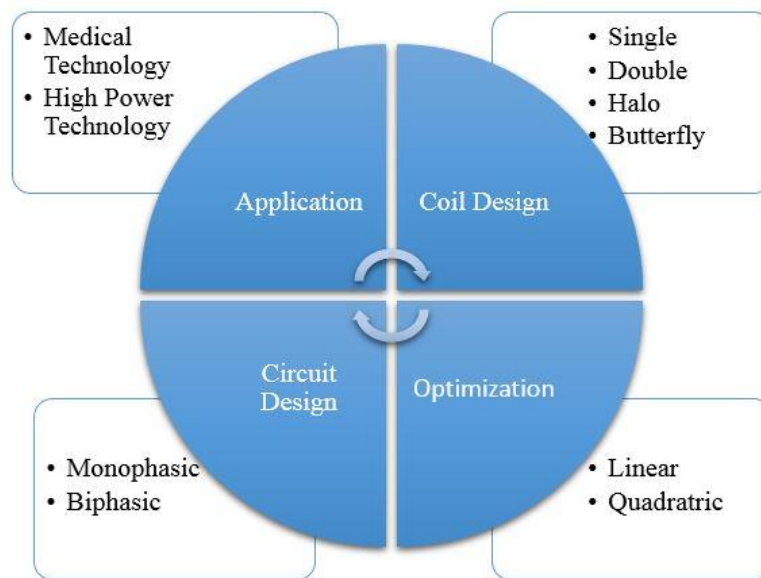


Figure 3 Taxonomy of the design process

In order to explore the necessary steps for the design process, it is helpful to have a classification system that divides the design process into parts that can be explored individually. First for the application, the HCMFG can be viewed as either a medical technology or power technology depending on the user, then the next step is to select the type of switching device for the circuit. MOSFET or IGBT were considered as candidates for the switching circuit. The choice will be governed by factors as such as which topology (monophasic or biphasic) or coil (single, double or halo) that is to be used. Once a switching architecture has been decided upon, the designer can then optimize conditions to tune the driver circuit.

2.3. Conclusion

A new paradigm for stimulation has been introduced and discussed in this chapter. It presents recent guidelines and covers recommended steps for implementing different platforms for high current magnetic field generators.

CHAPTER 3 HIGH CURRENT MAGNETIC FIELD GENERATOR

3.1. HCMFG

High current magnetic field generator consists of several subsystems. The HCMFG is controlled by a user-defined software. The conceptual basis for the operation of the High Current Magnetic Field Generator includes magnetization circuitry, signal processing, and magnetization behavior of a magnetic coil/body. These concepts are important not just in understanding the operation, but also in improving the performance. In general, for both monophasic and biphasic systems, the switching devices are either high power MOSFETs or insulated gate bipolar transistors (IGBTs). IGBTs are popular since they offer ease of control. They also offer more flexible solutions compared to current-controlled devices in high-voltage and high-current applications. When comparing the MOSFET and IGBT, it is found that IGBTs can have breakdown voltages above 1000 V, while the MOSFETs can have breakdown voltages below 250 V. The choice given to the designer is application specific and cost, size, speed, and thermal requirements should be considered. In certain high current applications, MOSFETs cannot be used or cascaded since MOSFET solutions can be costly. IGBTs, on the other hand, can more easily handle the required current with a single device and, as a result, be more cost-effective. Additionally, IGBTs are used in applications requiring low duty cycle, low frequency, and higher voltages. In contrast, MOSFETs usually have applications requiring higher frequency, higher duty cycles, and low-voltages. The goal is to demonstrate the two technologies given the same switching topology (e.g. same application) in order to reveal the utility of each. The first utilizes two bidirectional IGBTs (HCMFG₁) as seen in Figure 5 with two respective gate drivers. The second system shown in Figure 4 uses two power MOSFETs (HCMFG₂). Several types of commercial TMS devices are currently used, featuring controllable pulse-parameter TMS and repetitive TMS (rTMS) capabilities. Based on experimental data presented throughout the years, rTMS with monophasic pulses have shown stronger and more selective effects than conventional rTMS with biphasic pulses.

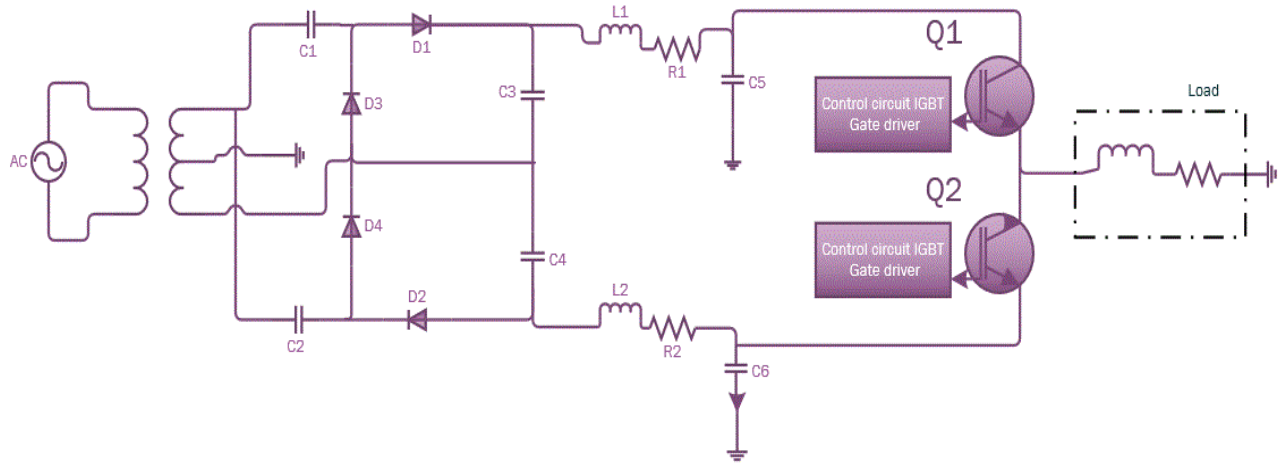


Figure 4 Topology of the IGBT circuit (HCMFG₂)

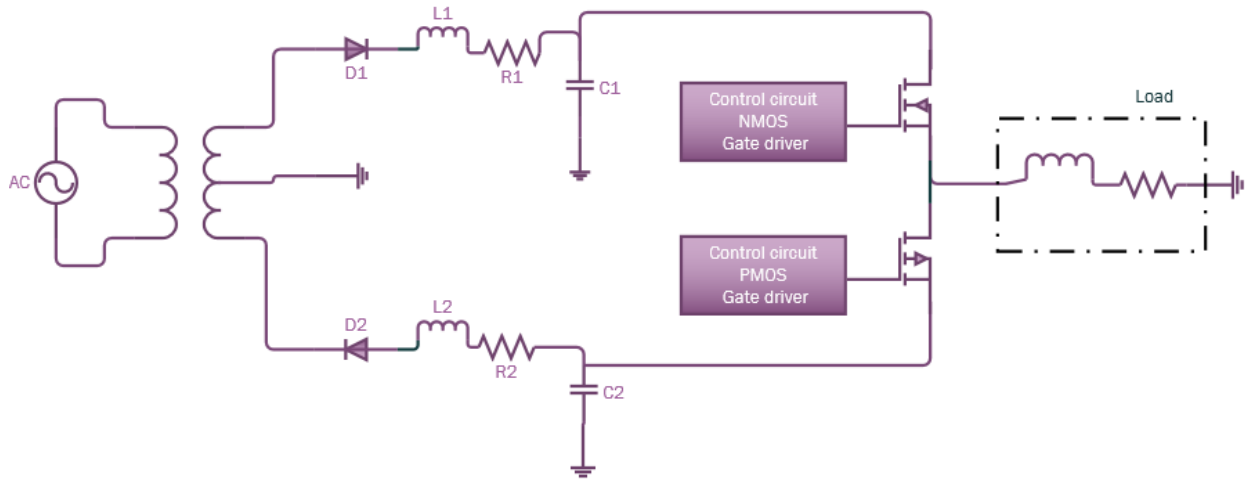


Figure 5 Topology of the MOSFET circuit (HCMFG₁)

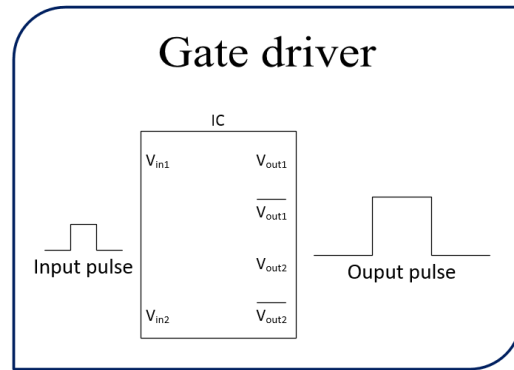
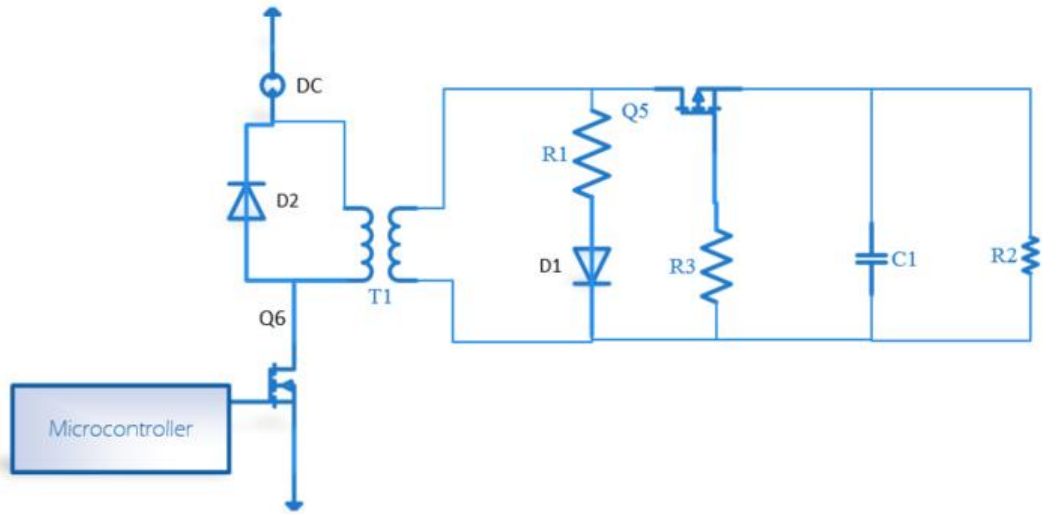
3.2. Topology, power supply, energy storage and switches

Figures 4 and 5 show simplified schematics of the pulse proposed pulse generators. Controlling the charge and discharge of a large amount of energy from capacitors through the system's coil is the essential idea of these designs. Each is composed of several functional subsystems: the power supply (120 VAC \pm 10%/60Hz), energy storage, gate driver control

circuitry, and the load. Capacitors C5 and C6 in Figure 4 and capacitors C1 and C2 in Figure 5 are respectively charged with a positive and negative voltage by the center-tapped transformer and a rectifying circuit. The stimulation coil is connected to one of the two energy storage capacitors C5 or C6 for the IGBTs in Figure 4 and C1 or C2 for the MOSFETs in Figure 4, depending on the desired direction of magnetization. The device is also composed of a fuse system and low power capacitor that is discharged in the coil through two bidirectional IGBTs. In order to properly turn these switches on (IGBTs or MOSFETs) for high power applications, special gate drivers are often required. However, to drive a high side switch (VDD connected to the collector of the IGBT or drain of the MOSFET), the output of the gate driver is connected directly to the gate of the switch with the ground of the driver connected to the emitter (for IGBTs) or the source (for MOSFETs). The energy storage capacitors are charged by the transformer through a rectifier circuit which quadruples the DC voltage after rectification. Two solid-state relays and a resistor bank allow safe discharge of the capacitor. In addition, the system is equipped with two capacitors of the same value ($C5 = C6 = 10000 \mu\text{F}$) and two bidirectional IGBTs with the same operational range. A capacitor bank, relay, and two MOSFETs (NMOS and PMOS) are used in case of the HCMFG₂.

3.3. Control circuit, coil considerations and energy efficiency

In Figures 6, 7 and 8 the control circuits are shown. They all utilize two gate drivers to actively control the IGBTs or MOSFETs which are specially designed to avoid unpredictable voltage spikes, oscillations or ringing, and false turn-ons. Usually these undesirable effects are a byproduct of an inadequate power supply, improper layout, and mismatch of driver specifications. A square pulse is sent from a microcontroller to the gate driver resulting in an output waveform with adjustable amplitude and width for the IGBTs (Q1 and Q2 for HCMFG₁) It should be noted that in the case of the HCMFG₂ a similar system is used for both MOSFETS, the gate driver shown in Figures 7 and 8 provide a high (dV/dt) immunity at the source, DC isolation and impedance matching.

Figure 6 Gate driver for the HCMFG₂Figure 7 Control circuit for NMOS (HCMFG₁)

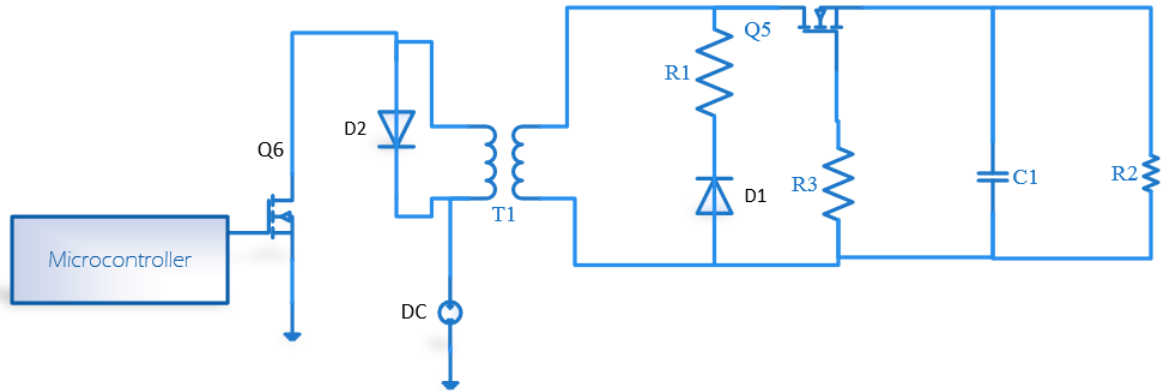


Figure 8 Control circuit for PMOS HCMFG₁

Different types of coils exist for TMS applications. Recent works have reported an enhancement on the focality and efficiency of magnetic stimulation in correlation with the depth of penetration, geometry of the coil, and the magnetic pulse duration [31],[32],[33],[35-37]. There is still an ambiguity regarding the optimal design for maximum efficiency. The energy delivered to a coil for a given electric field strength is written as

$$W = \frac{1}{2} \iiint \mathbf{B} * \mathbf{H} \, dv$$

where \mathbf{B} and \mathbf{H} are magnetic flux density and magnetic field intensity, respectively, and the energy is calculated over the volume . The maximum coil energy consumption can be expressed as

$$W = \frac{1}{2} L I_{max}^2$$

where I_{max} is the maximum current and L is the inductance of the coil. The coil induction depends on geometry, core permeability, and number of turns. However in order to improve the performance of the stimulation few parameters are to be considered: the amplitude of magnetic or electric field, charge density, excitation threshold, and the system's composition of resistance inductance, and capacitance (RLC). Parasitic RLC affects can distort the waveform of the magnetic pulse. In addition the RLC with a constant rise time can effect the waveform of the magnetic pulse [38]. Energy storage capacitor C keeps charging and then discharges rapidly

through the stimulating coil. The discharging loop of magnetic stimulator satisfies the following differential equation

$$L \frac{d^2 i}{dt^2} + R \frac{di}{dt} + \frac{i}{C} = 0$$

In configuring the appropriate coil, several parameters were taken into account: the length, the inductance, radius, and the number of turns. In both applications, a single coil configuration with N turns was utilized. The magnetic flux density B was approximated using the following equation [39]:

$$B = \frac{\mu N I}{\sqrt{L^2 + (4 * R^2)}}$$

where B is the magnetic flux (T), μ is the permeability (H/m), I is the current through the coil (A), N is the number of turns for the coil, L is the length of the coil (m), and R is the radius of the coil (m).

3.4. Conclusion

In this chapter, two different architectures for High Current Magnetic Field Generators was proposed, additionally the formulation for theoretical and practical approximation was reviewed. There are practical limits for generating pulse waveforms with arbitrary shapes involving high currents, voltages, and energies. However, the designs presented here i.e. High Current Magnetic Field Generator systems (HCMFG₁ and HCMFG₂) deploy a simple and practical solution to pulse generation with flexible adjustment parameters.

CHAPTER 4 EXPERIMENTAL SETUP

In this chapter, the experimental setup for the High Current Magnetic Field Generator is presented. The systematic development of the HCMFG, electronic control circuitry, and measurement methods are discussed.

4.1. Electronic System

The goal of the electronic subsystem is to supply sufficient driving current through the coil to generate a sufficiently high magnetic field. The electronic subsystem includes the coil, the driving circuitry, supporting electronic equipment, oscilloscope, and wires and cables which connect the separate components. In both topologies a pulse with adjustable width is precisely provided by a microcontroller. A monophasic pulse is obtained by turn on and off of the switches (IGBTs or MOSFETs). When switching on or off, the stimulation coil is connected to one of the capacitors that allows magnetic field to be generated around the coil whose energy and direction can be related to the voltage across the capacitors. A biphasic pulse is obtained by activating and introducing a delay to avoid any overlap between the switches (Figure 9). Additionally given a coil system of $30\text{ m}\Omega$ with $60\text{ }\mu\text{H}$ a corresponding current of 480 A is obtained for the NMOS when applying a 10 V signal to the gate driver (Figure 10, right). When a pulse with amplitude -10 V was then applied to the gate drive circuitry for the PMOS circuit, which resulted in an output current of approximately -400 A (Figure 10, left). Similar results were observed with IGBTs as shown in Figure 10.

The monophasic current waveforms for the HCMFG₁ is shown in Figure 10. The positive current waveform (Figure 11, right) was created by applying a positive pulse to IGBT Q1. The load current observed is a result of the released due to the charged capacitor C5. During turn-off, because due to the magnetic energy accumulated in the inductance L, the load current retains the same direction. The internal diode in Q2 starts to conduct until the load current reaches zero, allowing the current to change direction. When a pulse is sent to Q2 the load current changes direction and similar process occurs, resulting in a negative current waveform (Figure 11, left). A biphasic waveform (Figure 11) was obtained by introducing a time delay between the input pulses of both IGBT Q1 and Q2.

Figure 9 shows typical waveforms obtained by applying a positive and negative pulse to MOSFETs. Since a single IGBT can allow current to flow in either the forward or reverse direction between the collector and emitter pin, the gate driver circuitry can be minimized (only a positive gate pulse is required for current to flow in either direction for an IGBT, for example). MOSFETs require two different devices to achieve this effect (NMOS or PMOS). This complicates the driver circuitry since a positive and negative pulse generator is required to activate each FET.

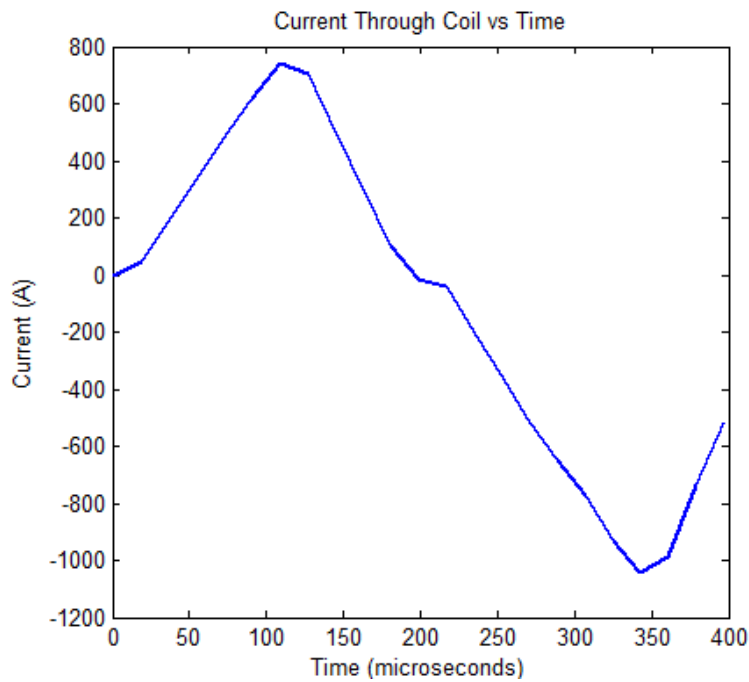


Figure 9 Current through the coil with gate voltage applied (Biphasic) to the reverse (left) and forward (right) IGBT

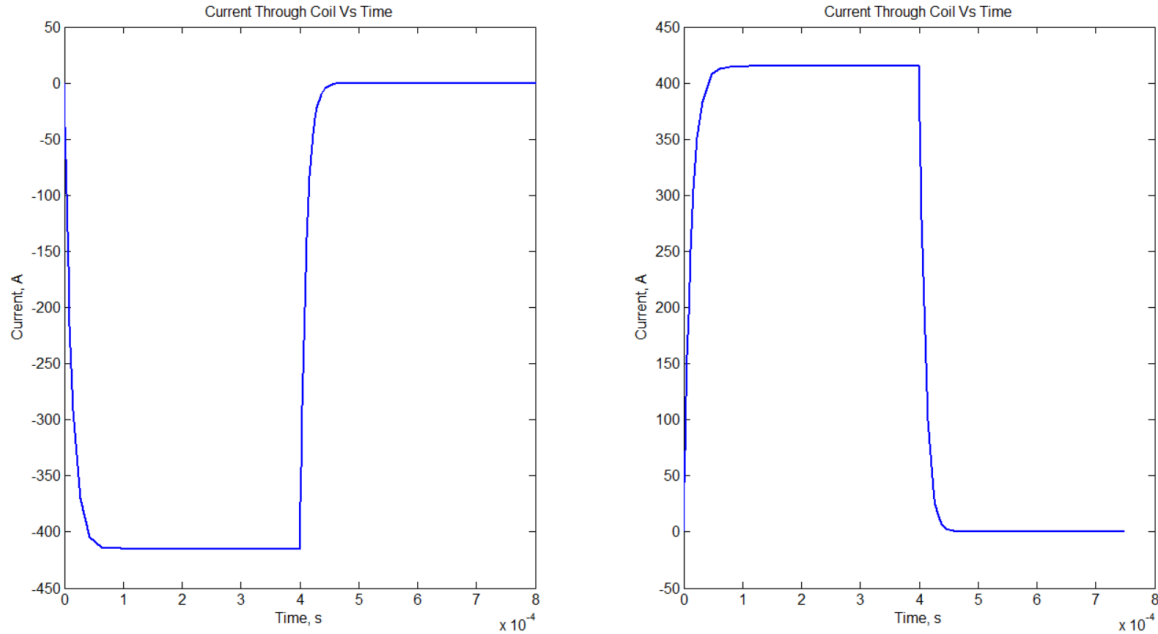


Figure 10 Current through the coil with the gate voltage applied to the PMOS (left) and NMOS (right)

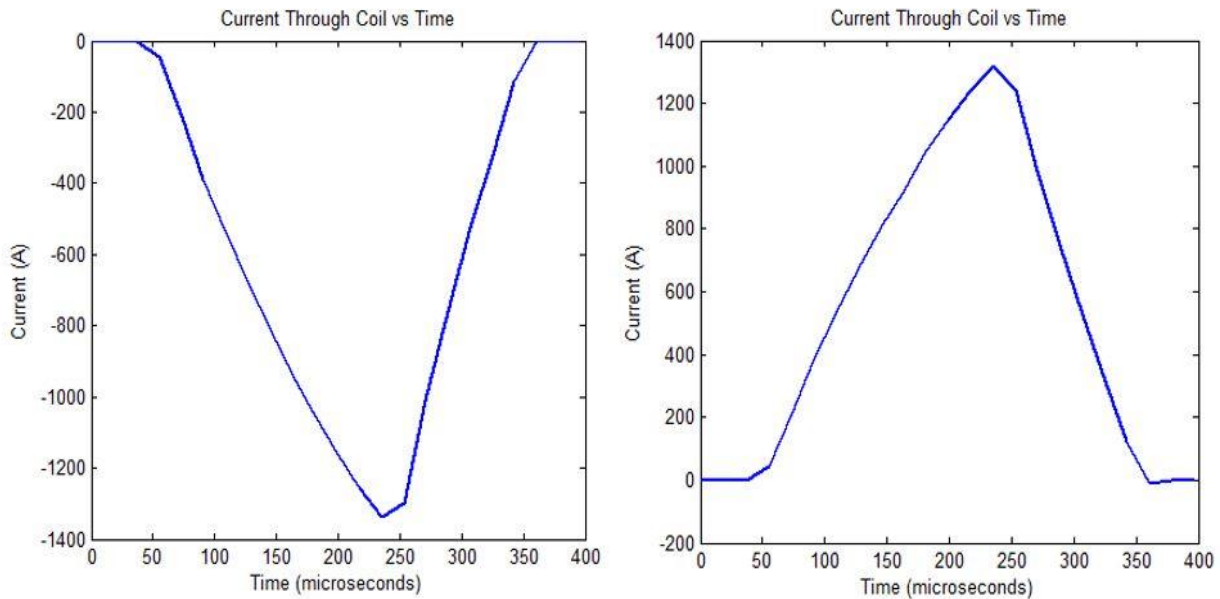


Figure 11 Current through the coil with gate voltage applied to the reverse (left) and forward (right) IGBT

4.2. Practical considerations

There are a number of practical issues in the experimental system which needs to be considered. In the event that the designer or experimenter would like to replicate the experiments of this document, the first step would be to construct the specialized rectifying

circuit which quadruples the DC output voltage. In this case, in building the rectifying circuit it is important that polarized capacitors be used to maintain stability of the DC output. TMS device uses large capacitor which are charged to high voltages. This energy is then applied across the patient's head or region of stimulation. Depending on how the discharge of the capacitor is achieved, different topologies can be utilized.

- Monophasic: the current does not change directions during discharge
- Biphasic: the current changes directions after half the time of discharge

Both circuits have advantages and disadvantages. The second circuit/ topology offers a reduction in the total amount of energy needed to stimulate. A lower amount of energy means that the size of the capacitor can be increased and power supply can be decreased. There are different ways of controlling the discharge in biphasic circuits. For a discharge, there are three main parameters that can be controlled, voltage (V), current (I), and time (T). It is also possible to control the total energy discharged, but it is limited by the charging voltage and capacitance of the capacitor. Both the proposed systems are energy efficient and have low weight, small size and low cost. The possibility to alter currents, voltages is given to the designer as a consideration. The capacitor acts as an energy storage device whose energy level can be calculated using the following formula:

$$W = \frac{C \cdot V^2}{2} \text{ Joule}$$

Where C is the capacitance (F) and V the voltage (V). When building the electronic driving circuit, an IGBT or MOSFET can be used and the connections short by nature to reduce inductance and bypass capacitors should be utilized. Additionally the following will need to be taken into consideration to prevent unwanted voltage spikes, oscillations or ringing effects and to avoid any false turn-on:

- Layout
- Power supply by-passing
- Mismatch of driver to the driven IGBT or MOSFET

A very crucial point is proper grounding. A very low-impedance path for current return to ground avoiding loops is a good design practice.

4.3. Conclusion

In this chapter, experimental results were presented. The adjustment range of the systems were shown to be limited mainly by energy dissipation, capacitance, the maximum pulse amplitude, and the pulse width.

CHAPTER 5 TESTING AND VERIFICATION

A verification process was applied to check that the system conforms to its specified requirements. The following chapter details how the measurements were obtained for different subsystem of the HCMFG.

5.1. Rectifier Circuit

The AC to DC converter circuit uses a transformer that has a center-tapped secondary for both systems (HCMFG₁ and HCMFG₂). Each half of the secondary works with a diode to form a half rectifier circuit. The center tap on the secondary winding effectively cuts the secondary in half. The peak voltage passed by each diode to the load is only half of the total peak secondary voltage. In the latter system (HCMFG₂) a voltage multiplier was used. It is very similar in many ways to the AC to DC converters however with the combination of rectifier diodes and capacitors the input peak voltage is multiplied to give a DC output equal to a multiple of the peak voltage of the AC input voltage.

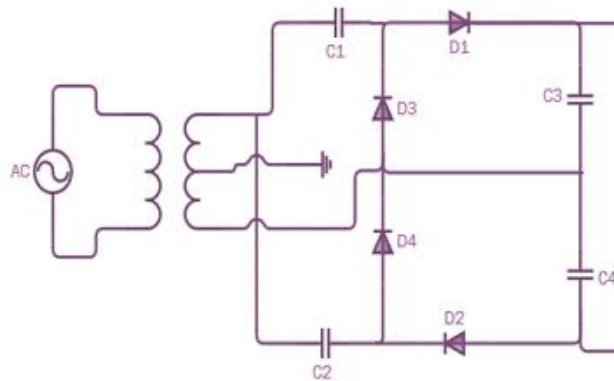


Figure 9 Rectifier circuit of the HCMFG₂

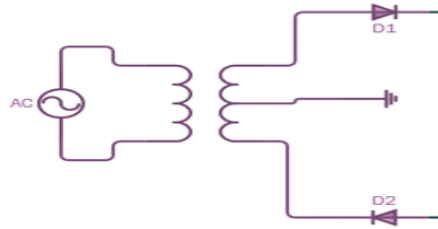


Figure 10 Rectifier circuit of the HCMFG₁

5.2. Gate driver

The microcontroller controls the switching device (MOSFET or IGBT); which determines what kind of signal will be generated, either monophasic or biphasic, and the duration of the pulse. The microcontroller receives an input from either the front panel or the GUI. The voltage pulse is first sent to an IGBT or MOSFET gate driver. The driver essentially isolates the 5V signal from the large power flowing through the IGBT or MOSFET. These gate drivers Figures 6, 7, and 8 respectively require a 15 V floating supply, which is achieved from the transformer. The transformer has two 18 VAC isolated supplies which are converted to 15 VDC. This voltage is controlled via a gate driver with a 5 V pulse. Unlike MOSFETs, IGBTs are turned on by the capacitance between the gate and emitter (not the gate and the source). The gate driver takes the emitter voltage of the IGBT and applies the 15 V across the gate and emitter to turn it on meanwhile 5V is being supplied through the Arduino.

5.3. Magnetic field measurements

The magnetic field generated within the coil was measured using a gaussmeter. To obtain the magnetic measurements due to the coil, a current was run through the coil, and the tip of a LakeShore gaussmeter was placed within the coil, with the tip being approximately perpendicular to the applied field. The resultant field was approximately 4000G. One suitable method used is to measure the voltage over the coil using the oscilloscope. In order to do this, the coil must be connected to traces on the circuit board which goes to a connector that can be connected to the oscilloscope for measurement. Three alternative methods were used to determine the current in the coil and approximate the magnetic field. These methods both indirectly measured the current in the coil. In one approach, a current sensor was used. In the

second approach, the wire forming the N turn coil was passed through a toroidal transformer with an N turn coil wrapped around it. This acted as an N: N transformer and the voltage was measured over a resistor on the N turn side. Knowing this voltage, the resistance, the turn ratio, and the permeability of the toroid, the current in the N turn coil could be calculated. This approach yielded a good time dependent measurement, but it added inductance to the coil and was difficult to implement on the circuit board. In the third approach, a resistor was added in parallel with the coil. The resistance was kept small to minimize the power dissipated by the resistor. The voltage over the resistor was measured with this oscilloscope and the current through the resistor could be calculated using the relationship $v = iR$.

CHAPTER 6 CONCLUSIONS AND FUTURE CONSIDERATIONS

6.1. Summary

Two different configurations of a High Current Magnetic Field Generator with adjustable pulse width for Transcranial Magnetic Stimulation and Power applications has been developed. An overview of the pertinent concepts was provided and a formulation describing the basic switch operation was presented. Results were provided about how the two systems may be designed with electrical efficiency greater than conventional devices. This was done by tuning the types of pulse and, reconsidering the topology. There are limits in practicality of generating pulse waveforms with arbitrary shapes which involve high currents, voltages, and energies. However, both High current magnetic field generator systems (HCMFG₁ and HCMFG₂) deploy a practical, and conceptually simple design approach to generate pluses with flexible adjustment parameters. The adjustment range of the systems are limited by the following factors: energy dissipation, capacitor voltage, the maximum electric field pulse amplitude (determined by the maximum storage energy capacitor voltages), and the pulse width which is limited by the increase in coil current and by the exponentially increasing capacitor voltage.

6.2. Contributions of work

There are many components which constitute the body of this work, but there are three major contributions which make this work valuable and unique. These contributions are:

- The design and successful implementation of a High Current Magnetic Field Generator as compared to existing technologies.
- The total system design which produced a deployable switching system.
- The development and implementation of a driving circuit that is used to extend the limitations of controllable pulse, interchangeable polarity, flexible capability and control which resulted in enhancing the High Current Magnetic Field Generation for similar applications in the future.

6.3. Accomplishments

The previous chapters presented the main work done on this project. The importance of each task, the reason for its attempt, and the method of achieving it were presented. This project has succeeded in delivering:

1. Two different High Current Magnetic Field Generators
2. Designing, implementing, and different electronic control circuits which include:
 - a. a series of control circuits,
 - b. tunable signal generators;
 - c. two different types of half bridge switching circuits

6.4. Meeting the standards

The initial idea presented in this master's thesis is developing several prototypes and mainly focused on investigating the possibilities of integrating them. However, the thesis presented is not primarily a study, but rather discusses development of different methods of magnetic field generator switching (HCMFG) for medical and high power applications.

While methods of magnetic field generator switching (HCMFG) improved significantly from its original proof-of-concept stage, there are a number of additional experiments which may be of interest in improving both designs. One such experiment would be to reduce the size of the entire system and the total cost. Another alteration which could be made is to change the capacitor with higher capacitance. Doing this would result in an increase of stored energy. The microcontroller used to control the entire design should be replaced with one that has a greater processing capabilities from the same family before the project develops further. Short-circuit protection circuitry needs to be added before the device is used clinically without the supervision of electrically competent people. Other safety factors that need to be considered are shielding of the high voltage parts since these parts of the device should be protected from human touch. During the design of the PCB's several compromises have been made in order to fit the layout onto the boards. This led to a layout that is not completely optimized when it comes to Electromagnetic Compatibility EMC and the disturbances generated on the PCB's need to be carefully investigated and minimized. Furthermore, no tests regarding sensitivity to disturbances were conducted and this is an important safety issue that needs to be considered when introducing the device in a clinical environment.

6.5. Bibliography

- [1] Chen, Robert, et al. "Involvement of the ipsilateral motor cortex in finger movements of different complexities." *Annals of neurology* 41.2 (1997): 247-254.
- [2] Chen, Robert, Leonardo G. Cohen, and Mark Hallett. "Role of the ipsilateral motor cortex in voluntary movement." *The Canadian journal of neurological sciences. Le journal Canadian des sciences neurologiques* 24.4 (1997): 284-291.
- [3] Gerloff, Christian, et al. "Stimulation over the human supplementary motor area interferes with the organization of future elements in complex motor sequences." *Brain* 120.9 (1997): 1587-1602.
- [4] Hucurs, David G., Raymond C. Tallis, and David G. Thompson. "The cortical topography of human swallowing musculature in health and disease." *Nature medicine* 2.11 (1996).
- [5] Imoniemi, Risto J., et al. "Neuronal responses to magnetic stimulation reveal cortical reactivity and connectivity." *Neuroreport* 8.16 (1997): 3537-3540.
- [6] Krings, T., et al. "Functional magnetic resonance imaging and transcranial magnetic stimulation Complementary approaches in the evaluation of cortical motor function." *Neurology* 48.5 (1997): 1406-1416.
- [7] Krings, Timo, et al. "Stereotactic transcranial magnetic stimulation: correlation with direct electrical cortical stimulation." *Neurosurgery* 41.6 (1997): 1319-1326.
- [8] Paus, Tomáš, et al. "Transcranial magnetic stimulation during positron emission tomography: a new method for studying connectivity of the human cerebral cortex." *The Journal of neuroscience* 17.9 (1997): 3178-3184.
- [9] Thompson, M. L., G. W. Thickbroom, and F. L. Mastaglia. "Corticomotor representation of the sternocleidomastoid muscle." *Brain* 120.2 (1997): 245-255.

- [10] Traversa, Raimondo, et al. "Mapping of motor cortical reorganization after stroke A brain stimulation study with focal magnetic pulses." *Stroke* 28.1 (1997): 110-117.
- [11] Walsh, Vincent. "Brain mapping: Faradization of the mind." *Current Biology* 8.1 (1998): R8-R11.
- [12] Jalinous, Reza. "Technical and practical aspects of magnetic nerve stimulation." *Journal of Clinical Neurophysiology* 8.1 (1991): 10-25.
- [13] Mouchawar, G. A., et al. "Closed-chest cardiac stimulation with a pulsed magnetic field." *Medical and Biological Engineering and Computing* 30.2 (1992): 162-168.
- [14] Devlin, J., P. Matthews, and M. Rushworth. "Semantic processing in the left inferior prefrontal cortex: a combined functional magnetic resonance imaging and transcranial magnetic stimulation study." *Cognitive Neuroscience, Journal of* 15.1 (2003): 71-84.
- [15] Kamen, Gary, and Graham E. Caldwell. "Physiology and interpretation of the electromyogram." *Journal of Clinical Neurophysiology* 13.5 (1996): 366-384.
- [16] Van der Kamp, Wim, et al. "Cortical excitability and response variability of transcranial magnetic stimulation." *Journal of clinical neurophysiology* 13.2 (1996): 164-171.
- [17] Krings, T., et al. "Functional magnetic resonance imaging and transcranial magnetic stimulation Complementary approaches in the evaluation of cortical motor function." *Neurology* 48.5 (1997): 1406-1416.
- [18] Mathis, J., and Ch W. Hess. "Motor-evoked potentials from multiple target muscles in multiple sclerosis and cervical myelopathy." *European Journal of Neurology* 3.6 (1996): 567-573.

- [19] Nielsen, Jørgen Feldbæk. "Logarithmic distribution of amplitudes of compound muscle action potentials evoked by transcranial magnetic stimulation." *Journal of clinical neurophysiology* 13.5 (1996): 423-434.
- [20] Rörich, Simone, et al. "Callosal and corticospinal tract function in patients with hydrocephalus: a morphometric and transcranial magnetic stimulation study." *Journal of neurology* 245.5 (1998): 280-288.
- [21] Noordhout, A. Maertens, et al. "Motor and somatosensory evoked potentials in cervical spondylotic myelopathy." *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section* 108.1 (1998): 24-31.
- [22] Pritchard, John W., Mani Mina, and Narimdinda Robert Y. Bouda. "Feel the Pulse." *Magnetics Technology International*. UKiP Media Events, 17 Dec. 2013. Web. 07 Jan. 2014.
- [23] Benecke, R. "Magnetic stimulation in the assessment of peripheral nerve disorders." *Bailliere's clinical neurology* 5.1 (1996): 115-128.
- [24] Kobayashi, Masahito, and Alvaro Pascual-Leone. "Transcranial magnetic stimulation in neurology." *The Lancet Neurology* 2.3 (2003): 145-156.
- [25] Yunokuchi, K., et al. "Estimation of focus of electric field in an inhomogeneous medium exposed by pulsed magnetic field." [Engineering in Medicine and Biology, 1999. 21st Annual Conference and the 1999 Annual Fall Meeting of the Biomedical Engineering Society] *BMES/EMBS Conference, 1999. Proceedings of the First Joint*. Vol. 1. IEEE, 1999.
- [26] Hess, C. W., et al. "Magnetic stimulation of the human brain: influence of size and shape of the stimulating coil." *Motor Disturbances II* (2012): 31-42.

- [27] Kraus, Karl H., et al. "The use of a cap-shaped coil for transcranial magnetic stimulation of the motor cortex." *Journal of Clinical Neurophysiology* 10.3 (1993): 353-362.
- [28] Mills, K. R., S. J. Boniface, and M. Schubert. "Magnetic brain stimulation with a double coil: the importance of coil orientation." *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section* 85.1 (1992): 17-21.
- [29] Bashir, Shahid, et al. "Integrated Technologies Like Noninvasive Brain Stimulation (NIBS) for Stroke Rehabilitation; New Hopes for Patients, Neuroscientists, and Clinicians in Iran." *Basic and Clinical Neuroscience* 1.4 (2010): 6-14.
- [30] Miranda, Pedro Cavaleiro, Mark Hallett, and Peter J. Basser. "The electric field induced in the brain by magnetic stimulation: a 3-D finite-element analysis of the effect of tissue heterogeneity and anisotropy." *Biomedical Engineering, IEEE Transactions on* 50.9 (2003): 1074-1085.
- [31] A. T. Barker, R. Jalinous, and I. L. Freeston, "Non-invasive magnetic stimulation of human motor cortex," *Lancet*, vol. 1, no. 8437, pp. 1106–7, 1985
- [32] Williams, P. I., et al. "New designs for deep brain transcranial magnetic stimulation." *Magnetics, IEEE Transactions on* 48.3 (2012): 1171-1178.
- [33] Amassian, Vahe E., et al. "Suppression of visual perception by magnetic coil stimulation of human occipital cortex." *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section* 74.6 (1989): 458-462.
- [34] McCarthy, Gregory, and Charles C. Wood. "Scalp distributions of event-related potentials: an ambiguity associated with analysis of variance models." *Electroencephalography and Clinical Neurophysiology/Evoked Potentials Section* 62.3 (1985): 203-208.

- [35] MagVenture A/S, Product Information Sheet: MagPro X100 with Option, Technical Data, 2007. [Online]. Available: <http://www.magventure.com>
- [36] Sommer, Martin, et al. "Neuronal tissue polarization induced by repetitive transcranial magnetic stimulation?." *Neuroreport* 13.6 (2002): 809-811.
- [37] Hosono, Yuki, et al. "Comparison of monophasic versus biphasic stimulation in rTMS over premotor cortex: SEP and SPECT studies." *Clinical Neurophysiology* 119.11 (2008): 2538-2545.
- [38] Deng, Zhi-De, Sarah H. Lisanby, and Angel V. Peterchev. "Electric field depth–focality tradeoff in transcranial magnetic stimulation: simulation comparison of 50 coil designs." *Brain stimulation* 6.1 (2013): 1-13.
- [39] J. A. Stratton, *Electromagnetic Theory*, New York: McGraw-Hill, pp. 233, 1941
- [40] Krings, Timo, et al. "Stereotactic transcranial magnetic stimulation: correlation with direct electrical cortical stimulation." *Neurosurgery* 41.6 (1997): 1319-1326.
- [41] Thompson, M. L., G. W. Thickbroom, and F. L. Mastaglia. "Corticomotor representation of the sternocleidomastoid muscle." *Brain* 120.2 (1997): 245-255.
- [42] Traversa, Raimondo, et al. "Mapping of motor cortical reorganization after stroke A brain stimulation study with focal magnetic pulses." *Stroke* 28.1 (1997): 110-117.
- [43] Wassermann, Eric M., and Jordan Grafman. "Combining transcranial magnetic stimulation and neuroimaging to map the brain." *Trends in cognitive sciences* 1.6 (1997): 199-199.
- [44] Iwata, Nobue Kobayashi, et al. "Facilitatory effect on the motor cortex by electrical stimulation over the cerebellum in humans." *Experimental brain research* 159.4 (2004): 418-424.

- [45] Hashimoto, M., and K. Ohtsuka. "Transcranial magnetic stimulation over the posterior cerebellum during visually guided saccades in man." *Brain* 118.5 (1995): 1185-1193.
- [46] Wessel, K.. "Enhancement of inhibitory mechanisms in the motor cortex of patients with cerebellar degeneration: a study with transcranial magnetic brain stimulation." *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control* 101.4 (1996): 273-280.
- [47] Catano, Antonio, Myriam Houa, and Pierre Noël. "Magnetic transcranial stimulation: clinical interest of the silent period in acute and chronic stages of stroke." *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control* 105.4 (1997): 290-296.
- [48] Brasil-Neto, Joaquim P. "Optimal focal transcranial magnetic activation of the human motor cortex: effects of coil orientation, shape of the induced current pulse, and stimulus intensity." *Journal of clinical neurophysiology* 9.1 (1992): 132-136.

APPENDIX A. TECHNICAL SPECIFICATIONS

Equipment used for experimental setup

Table 1 contains a list of the equipment and components used in the experimental setup and includes the manufacturer and model numbers.

Table 1 Equipment and components used in the experimental setup

Equipment/Component	Manufacturer		Models
	Electrical system		
DC Supplies	Agilent		34410 A
	Agilent		E3631A
Oscilloscope	Agilent		DSO-X 2024

Ports

The table below describes some of the ports available to the user.

Table 2 Ports available to the user

Inputs		
Type	Value	Description
Ext trig	USB	Connection between the computer and the prototype
Power source	120 V, 15 A	Wall outlet

Buttons

The table below describes the buttons on the stimulator

Table 3 Description of buttons on the stimulator

Buttons	
Name	Description
Connect / Disconnect	The microcontroller (Arduino) button for the GUI allows the user to connect/disconnect the Arduino to/from the MATLAB application. When the Arduino is connected or disconnected completely the text box will display that it has been safely connected or disconnected
Pulse and Record	The Pulse and Record button sends an input pulse to the Arduino for it to send its 5V pulse into the gate drivers for 200 microseconds to trigger our IGBT/MOSFET and creates a 400 microsecond current pulse. This button records the data by reading the output of the current sensor, which is attached to the inductive load. The Arduino reads the data from the sensor and outputs the current read through its digital pins.
Capacitor Discharge	The Capacitor Discharge button allow to send an input pulse to the microcontroller (Arduino) to control all four relays to discharge all the large capacitors through our high wattage resistors. The relays are on for two minutes to ensure the capacitors are fully discharged.
Pulse Type (Monophasic or Biphasic)	This Pulse Type button allow the user to choose between different options.
Pulse Length	This button determines the length of the pulse

Current consumption

The following table presents the current consumption of the system during different stages.

Table 4 Device 1 current consumption

Current Consumption Device 1	
Mode	Current
Idle	4A
Charging max	10A
Charging min	4A
Safety discharge	2A

Table 5 Device 2 current consumption

Current Consumption Device 2	
Mode	Current
Idle	3A
Charging max	7A
Charging min	4A
Safety discharge	2A

Coil parameters

Table 6 Coil type A

Parameter	Value
Coil Type A	Solenoid
Length	.1524 m
Radius	.0143 m
Number of Turns	50
Wire AWG	10
LRC Meter Inductance	22 μ H
LRC Meter Resistance	.07 Ω

Table 7 Coil type B

Parameter	Value
Coil Type	Solenoid
Length	.1524 m
Radius	.0143 m
Number of Turns	52
Wire AWG	10
LRC Meter Inductance	32 μ H
LRC Meter Resistance	.3 Ω

Load specifications

- Max Current =1800 A
- Max voltage= 200V
- Warm up time= 5 minutes
- Cool down time = 2 minutes
- Minimum Pulse Time =50 μ S
- Current Sensor Limit = 1200 A

Table 8 Load specification for HCMFG₁ and HCMFG₂

Resistance	Inductance				
	50 μ H	20 μ H	10 μ H	1 μ H	50nH
0.1					
0.05					
0.01					
0.005					
0.001					

APPENDIX B. HARDWARE

This appendix provides additional information regarding the schematics and PCB layouts used in the project. It also presents some plots of different types of discharges.

PCB devices1

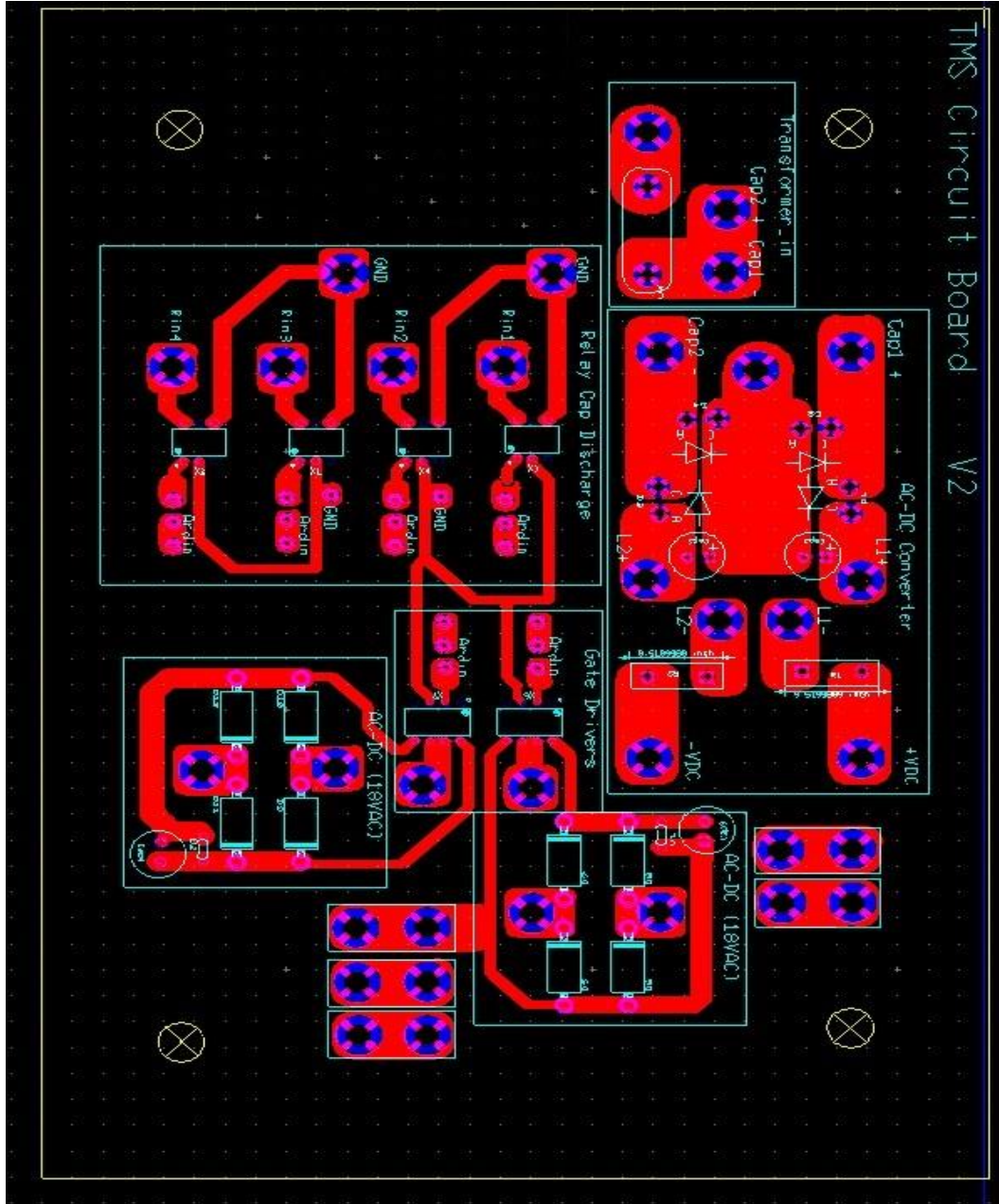
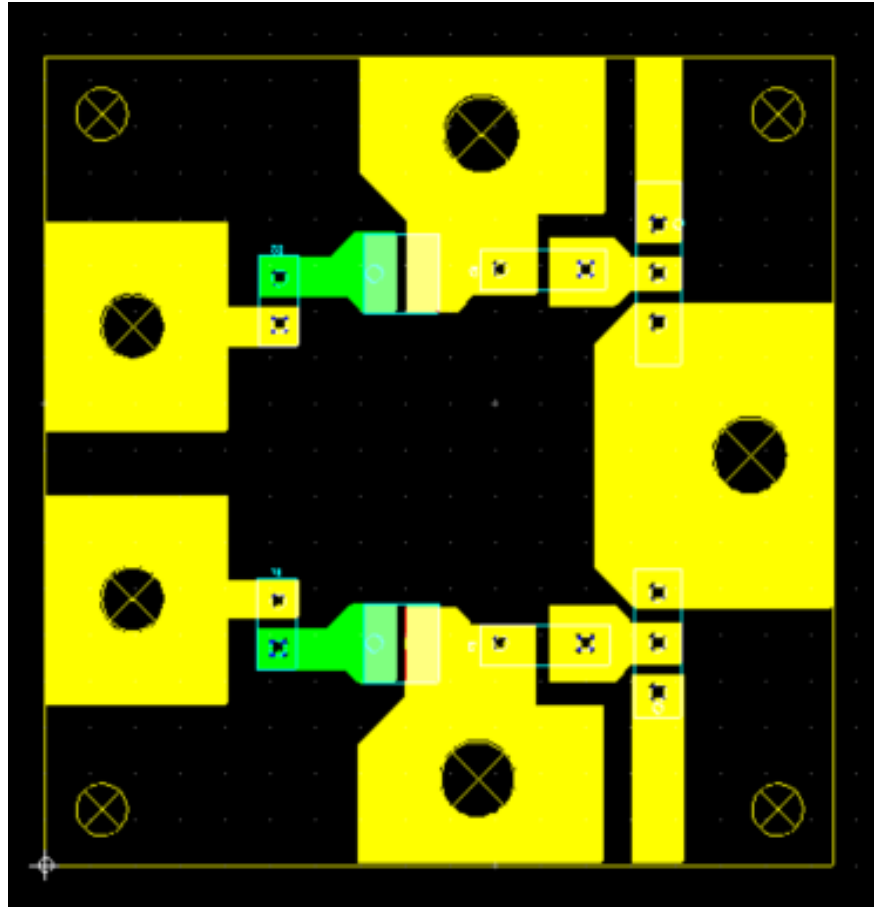


Figure 11 HCMFG₂ Topology

PCB devices 2

Figure 12 HCMFG₁ Topology

Monophasic graph

- Positive pulse
 - 195V 0.3ohm load

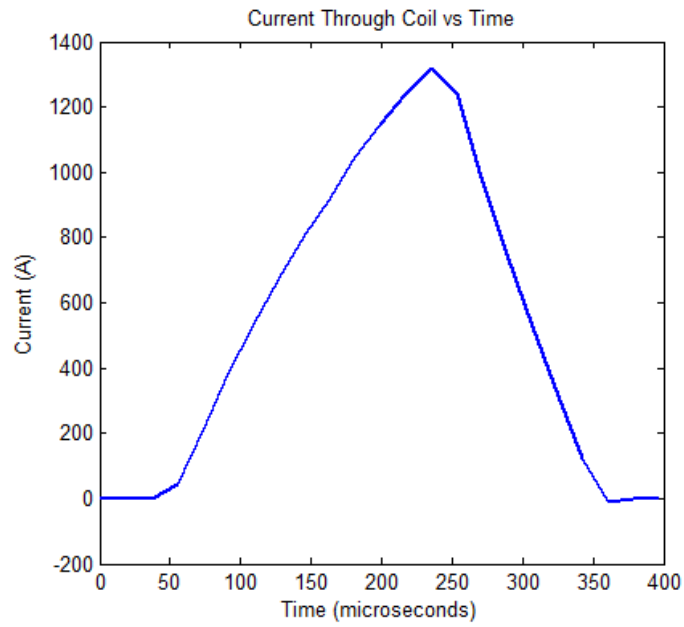


Figure 13 1300A pulse

- Negative pulse
 - 150 V 0.3ohm load

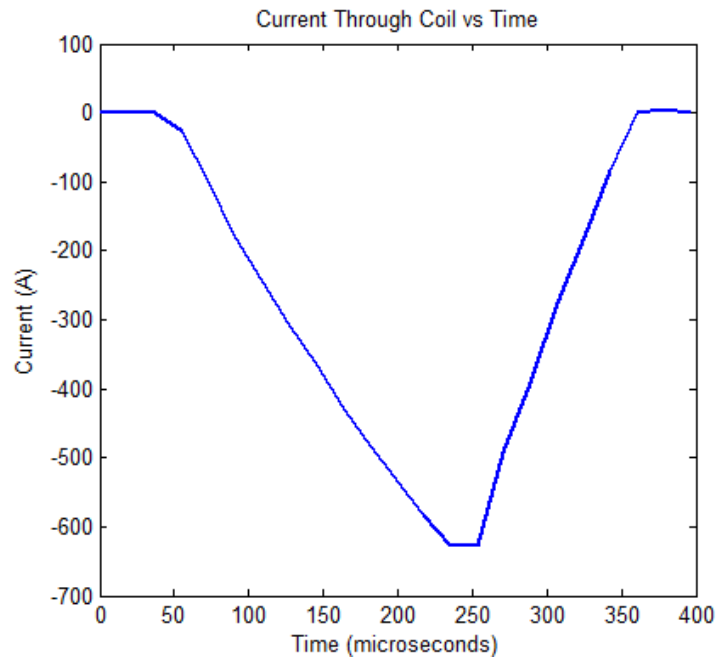


Figure 14 -650A pulse

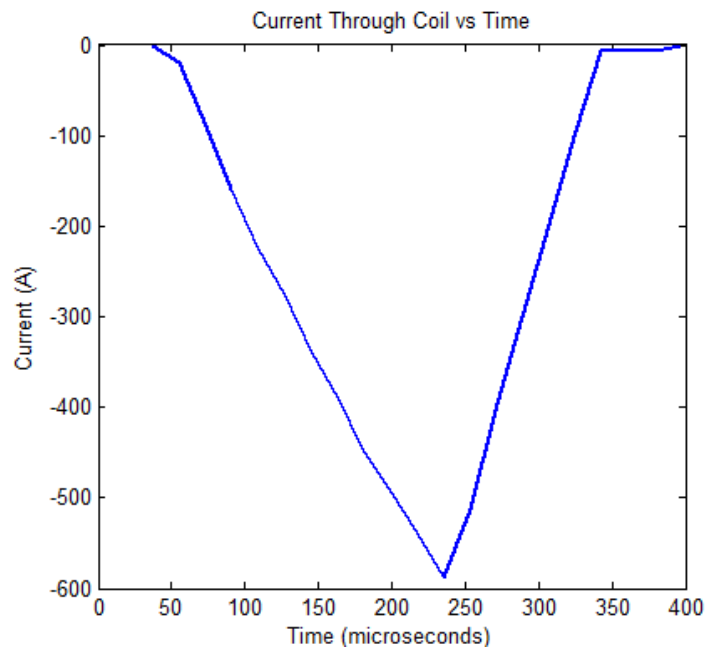
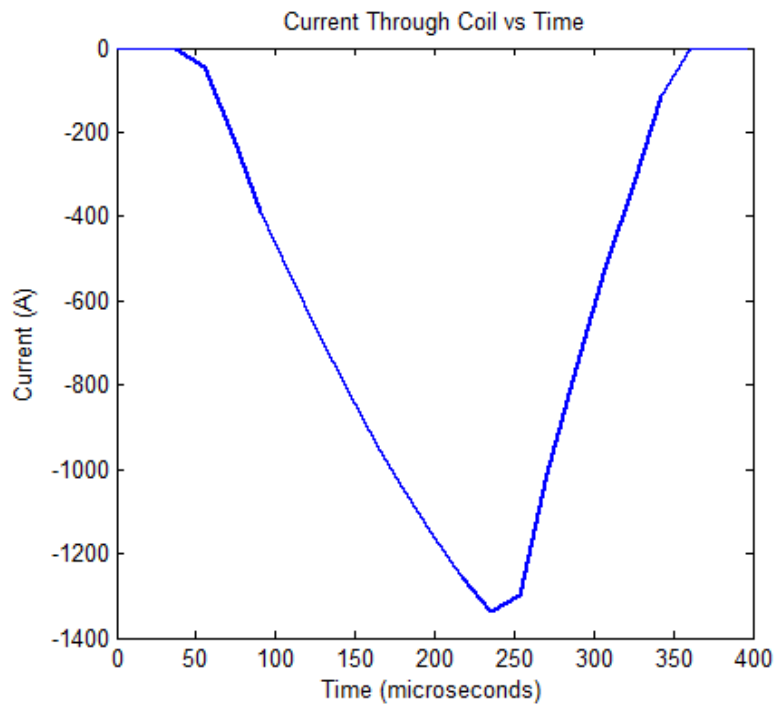


Figure 15 -590A pulse



- 190 V 0.3ohm load

Figure 16 -1300A pulse

- 195 V 0.3ohm load

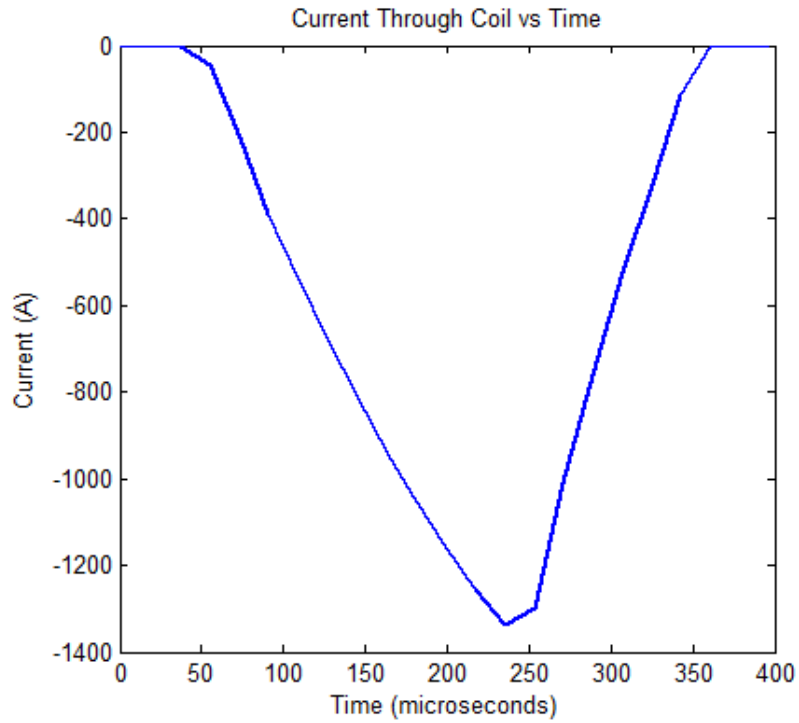


Figure 17 -1300A pulse

Biphasic graph

- 145V .3 ohm load

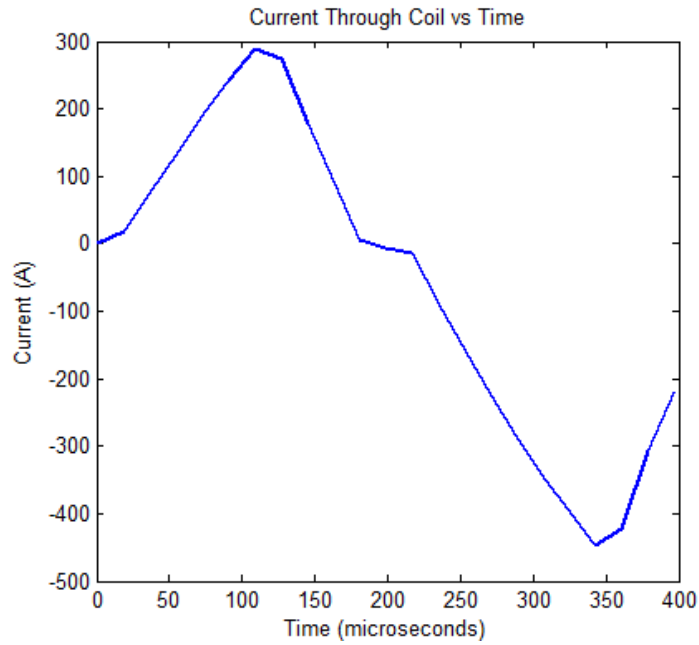


Figure 18 [+250, - 460] A pulse

- 173V 0.3ohm load

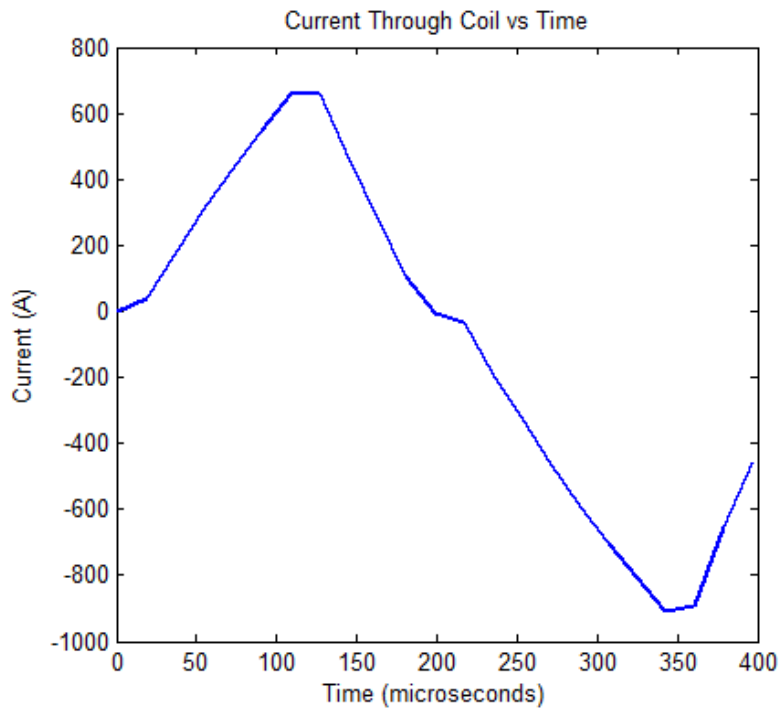


Figure 19 [+700, - 890] A pulse

- 190V 0.3ohm load

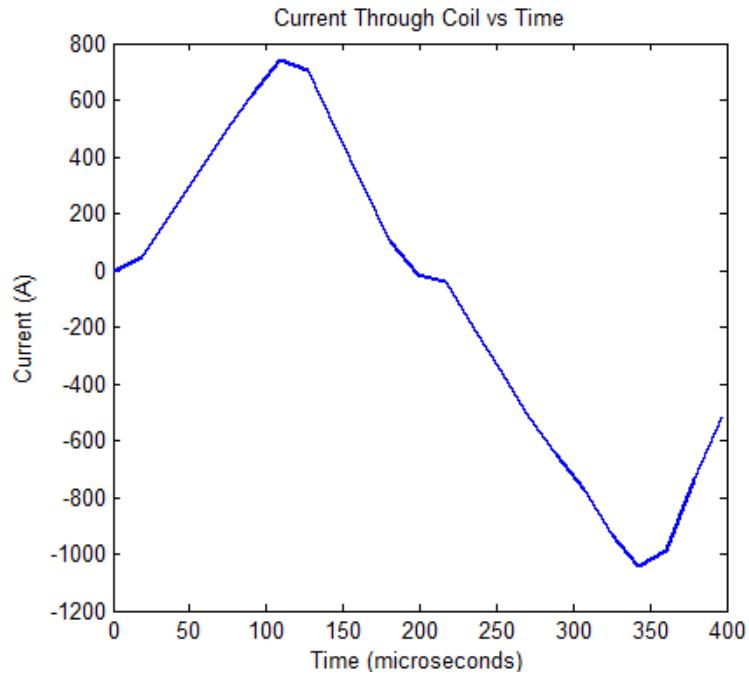


Figure 20 [+700, - 1100] A pulse

APPENDIX C. SOFTWARE

Graphical user interface

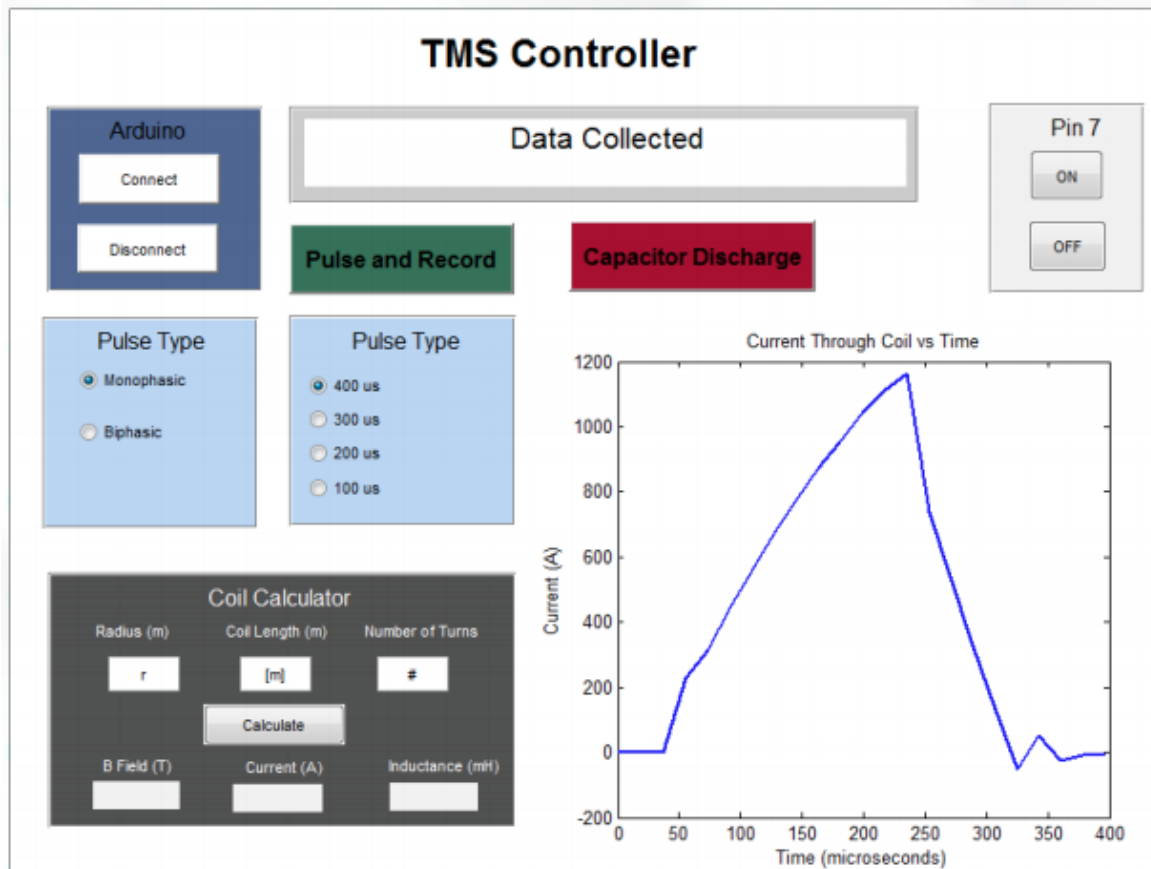


Figure 21 TMS GUI

Program listing

```
function varargout = LEDS(varargin)
% LEDS MATLAB code for LEDS.fig
%   LEDS, by itself, creates a new LEDS or raises the existing
%   singleton*.
%
%   H = LEDS returns the handle to a new LEDS or the handle to
%   the existing singleton*.
%
%   LEDS('CALLBACK',hObject,eventData,handles,...) calls the local
%   function named CALLBACK in LEDS.M with the given input arguments.
%
%   LEDS('Property','Value',...) creates a new LEDS or raises the
%   existing singleton*. Starting from the left, property value pairs are
%   applied to the GUI before LEDS_OpeningFcn gets called. An
%   unrecognized property name or invalid value makes property application
%   stop. All inputs are passed to LEDS_OpeningFcn via varargin.
```

```

%
% *See GUI Options on GUIDE's Tools menu. Choose "GUI allows only one
% instance to run (singleton)".
%
% See also: GUIDE, GUIDATA, GUIHANDLES

% Edit the above text to modify the response to help LEDES

% Last Modified by GUIDE v2.5 09-Dec-2014 09:42:48

% Begin initialization code - DO NOT EDIT
gui_Singleton = 1;
gui_State = struct('gui_Name',    mfilename, ...
                  'gui_Singleton', gui_Singleton, ...
                  'gui_OpeningFcn', @LEDES_OpeningFcn, ...
                  'gui_OutputFcn', @LEDES_OutputFcn, ...
                  'gui_LayoutFcn', [] , ...
                  'gui_Callback', []);
if nargin && ischar(varargin{1})
    gui_State.gui_Callback = str2func(varargin{1});
end

if nargout
    [varargout{1:nargout}] = gui_mainfcn(gui_State, varargin{:});
else
    gui_mainfcn(gui_State, varargin{:});
end
% End initialization code - DO NOT EDIT
handles.length=0;
handles.radius=0;
handles.turns=0;

end

% --- Executes just before LEDES is made visible.
function LEDES_OpeningFcn(hObject, eventdata, handles, varargin)
% This function has no output args, see OutputFcn.
% hObject    handle to figure
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
% varargin   command line arguments to LEDES (see VARARGIN)

```

```

% Choose default command line output for LEDS
handles.output = hObject;
handles.uno=0;
handles.Pulsetype=char('Monophasic');

% Update handles structure
guidata(hObject, handles);

% UIWAIT makes LEDS wait for user response (see UIRESUME)
% uiwait(handles.figure1);
end

% --- Outputs from this function are returned to the command line.
function varargout = LEDS_OutputFcn(hObject, eventdata, handles)
% varargout cell array for returning output args (see VARARGOUT);
% hObject handle to figure
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)

% Get default command line output from handles structure
varargout{1} = handles.output;
handles.uno=0;
set(findobj('Tag','Status'),'String','');
guidata(hObject, handles);

end

% --- Executes on button press in Connect.
function Connect_Callback(hObject, eventdata, handles)
% hObject handle to Connect (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
set(findobj('Tag','Status'),'String','Connecting to Arduino...');
handles.uno=serial('COM3');
fopen(handles.uno);
pause(1);
set(findobj('Tag','Status'),'String','Connected!');
handles.Pulselength=char('u400');
guidata(hObject, handles);

```

end

```
% --- Executes on button press in Send_Pulse.
function Send_Pulse_Callback(hObject, eventdata, handles)
% hObject    handle to Send_Pulse (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
fprintf(handles.uno,'a');
set(findobj('Tag','Status'),'String','Pulse LED');
guidata(hObject, handles);
```

end

```
% --- Executes on button press in Disconnect.
function Disconnect_Callback(hObject, eventdata, handles)
% hObject    handle to Disconnect (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
```

```
set(findobj('Tag','Status'),'String','Disconnecting...');
fclose(handles.uno);
pause(1);
set(findobj('Tag','Status'),'String','Finished');
guidata(hObject, handles);
```

end

```
% --- Executes on button press in read.
function read_Callback(hObject, eventdata, handles)
% hObject    handle to read (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
disp(handles.Pulsetype);
```

```
%MONOPHASIC
if(strcmp(handles.Pulsetype, 'Monophasic'))
```

```
if( strcmp(handles.Pulselength, 'u400'))
    set(findobj('Tag','Status'),'String',['+ Monophasic Selected 400 microns']);
```

```

    pause(1);
    fprintf(handles.uno,'m'); % this section reads the serial from the Arduino
end
if( strcmp(handles.Pulselength, 'u300'))
    set(findobj('Tag','Status'),'String',['+ Monophasic Selected 300 microns']);
    pause(1);
    fprintf(handles.uno,'h'); % this section reads the serial from the Arduino
end
if( strcmp(handles.Pulselength, 'u200'))
    set(findobj('Tag','Status'),'String',['+ Monophasic Selected 200 microns']);
    pause(1);
    fprintf(handles.uno,'j'); % this section reads the serial from the Arduino
end
if( strcmp(handles.Pulselength, 'u100'))
    set(findobj('Tag','Status'),'String',['+ Monophasic Selected 100 microns']);
    pause(1);
    fprintf(handles.uno,'k'); % this section reads the serial from the Arduino
end

    for i=1:1:24
        A=fscanf(handles.uno);
        B=sscanf(A,'%d');
        X{i}=B;
        disp(X);
    end
set(findobj('Tag','Status'),'String',['Pulse Sent']);
pause(1);
set(findobj('Tag','Status'),'String',['Data Collected']);
data=cell2mat(X);
voltage=cell2mat(data(2:end));
voltage = (voltage*(5.0/1023.0)); % converts A0 step data to voltage
voltage=round(voltage/.01)*.01; % converts Voltage to rounded voltage

current = (voltage-voltage(1))*640.0; % converts voltage to current
disp('your current is: ');
disp(current);
disp('your voltage is: ');
disp(voltage);
disp('your time is: ');
disp(data(1));
disp(length(voltage));
t=1:18:414;

```

```

%t=1:67:1541;
disp('lengths');
%disp(length(t));
disp(length(current)); % plots data
h=plot(t,current);
set(h,'linewidth',2);
title('Current Through Coil vs Time');
xlabel('Time (microseconds)'); % x-axis label
ylabel('Current (A)'); % y-axis label
end

%Negative Monophasic
if(strcmp(handles.Pulsetype, 'NegativeMonophasic'))
    if( strcmp(handles.Pulselength, 'u400'))
        set(findobj('Tag','Status'),'String',['- Monophasic Selected 400 microns']);
        pause(1);
        fprintf(handles.uno,'n'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u300'))
        set(findobj('Tag','Status'),'String',['- Monophasic Selected 300 microns']);
        pause(1);
        fprintf(handles.uno,'u'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u200'))
        set(findobj('Tag','Status'),'String',['- Monophasic Selected 200 microns']);
        pause(1);
        fprintf(handles.uno,'i'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u100'))
        set(findobj('Tag','Status'),'String',['- Monophasic Selected 100 microns']);
        pause(1);
        fprintf(handles.uno,'o'); % this section reads the serial from the Arduino
    end

    for i=1:1:24
        A=fscanf(handles.uno);
        B=sscanf(A,'%d');
        X{i}=B;
        disp(X);
    end
    set(findobj('Tag','Status'),'String',['Pulse Sent']);
    pause(1);

```

```

set(findobj('Tag','Status'),'String',['Data Collected']);
data=cell2mat(X);
voltage=cell2mat(data(2:end));
voltage = (voltage*(5.0/1023.0)); % converts A0 step data to voltage
voltage= round(voltage/.01)*.01; % converts Voltage to rounded voltage

current = (voltage-voltage(1))*640.0; % converts voltage to current
disp('your current is: ');
disp(current);
disp('your voltage is: ');
disp(voltage);
disp('your time is: ');
disp(data(1));
disp(length(voltage));
t=1:18:414;
%t=1:67:1541;
disp('lengths');
%disp(length(t));
disp(length(current)); % plots data
h=plot(t,current);
set(h,'linewidth',2);
title('Current Through Coil vs Time');
xlabel('Time (microseconds)'); % x-axis label
ylabel('Current (A)'); % y-axis label
end

%BIPHASIC
if(strcmp(handles.Pulsetype, 'Biphasic'))
    if( strcmp(handles.Pulselength, 'u400'))
        set(findobj('Tag','Status'),'String',['Biphasic 400 microns']);
        pause(1);
        fprintf(handles.uno,'b'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u300'))
        set(findobj('Tag','Status'),'String',['Biphasic 300 microns']);
        fprintf(handles.uno,'v'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u200'))
        set(findobj('Tag','Status'),'String',['Biphasic 200 microns']);
        fprintf(handles.uno,'c'); % this section reads the serial from the Arduino
    end
    if( strcmp(handles.Pulselength, 'u100'))
        set(findobj('Tag','Status'),'String',['Biphasic 100 microns']);

```



```

fprintf(handles.uno,'c'); % this section reads the serial from the Arduino
end
for i=1:1:24
    A=fscanf(handles.uno);
    B=sscanf(A,'%d');
    X{i}=B;
    disp(X);
end
set(findobj('Tag','Status'),'String',['Pulse Sent']);
pause(1);
set(findobj('Tag','Status'),'String',['Data Collected']);
data=cell2mat(X);
voltageset=data(2:end);
voltage = (voltageset*(5.0/1023.0));
voltageground= round(voltage/.01)*.01;

current = (voltageground-voltageground(1))*640.0;
disp('your current is: ');
disp(current);
disp('your voltage is: ');
disp(voltageground);
disp('your time is: ');
disp(data(1));
disp(length(voltage));
t=1:18:414;
%t=1:67:1541;
disp('lengths');
%disp(length(t));
disp(length(current));
h=plot(t,current);
set(h,'linewidth',2);
title('Current Through Coil vs Time');
xlabel('Time (microseconds)'); % x-axis label
ylabel('Current (A)'); % y-axis label
end
guidata(hObject, handles);
end

% --- Executes on button press in Monophasic.
function Monophasic_Callback(hObject, eventdata, handles)

```

```

% hObject handle to Monophasic (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
% Hint: get(hObject,'Value') returns toggle state of Monophasic
guidata(hObject, handles);
end

% --- Executes on button press in Biphasic.
function Biphasic_Callback(hObject, eventdata, handles)
% hObject handle to Biphasic (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)

% Hint: get(hObject,'Value') returns toggle state of Biphasic
guidata(hObject, handles);
end

% -----

% --- Executes when selected object is changed in uipanel4.
function uipanel4_SelectionChangeFcn(hObject, eventdata, handles)
% hObject handle to the selected object in uipanel4
% eventdata structure with the following fields (see UIBUTTONGROUP)
%   EventName: string 'SelectionChanged' (read only)
%   OldValue: handle of the previously selected object or empty if none was selected
%   NewValue: handle of the currently selected object

% handles structure with handles and user data (see GUIDATA)
handles.Pulsetype=get(hObject,'Tag');
disp(handles.Pulsetype);
guidata(hObject, handles);

end

% --- Executes when selected object is changed in uipanel8.
function uipanel8_SelectionChangeFcn(hObject, eventdata, handles)
% hObject handle to the selected object in uipanel8
% eventdata structure with the following fields (see UIBUTTONGROUP)
%   EventName: string 'SelectionChanged' (read only)
%   OldValue: handle of the previously selected object or empty if none was selected

```

```

% NewValue: handle of the currently selected object
% handles structure with handles and user data (see GUIDATA)
handles.PulseLength=get(hObject,'Tag');
disp(handles.Pulsetype);
guidata(hObject, handles);
guidata(hObject, handles);
end

```

```

function Turns_Callback(hObject, eventdata, handles)
% hObject handle to Turns (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of Turns as text
% str2double(get(hObject,'String')) returns contents of Turns as a double
handles.turns=str2double(get(hObject,'String'));
disp('value added');
disp(handles.turns);
guidata(hObject, handles);

end

```

```

% --- Executes during object creation, after setting all properties.
function Turns_CreateFcn(hObject, eventdata, handles)
% hObject handle to Turns (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
% See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
end

```

```

function Length_Callback(hObject, eventdata, handles)
% hObject handle to Length (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB

```

```

% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of Length as text
%    str2double(get(hObject,'String')) returns contents of Length as a double

handles.length=str2double(get(hObject,'String'));
disp('value added');
disp(handles.length);
guidata(hObject, handles);

end

% --- Executes during object creation, after setting all properties.
function Length_CreateFcn(hObject, eventdata, handles)
% hObject    handle to Length (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%    See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
end

function Radius_Callback(hObject, eventdata, handles)
% hObject    handle to Radius (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'String') returns contents of Radius as text
%    str2double(get(hObject,'String')) returns contents of Radius as a double

handles.radius=str2double(get(hObject,'String'));
disp('value added');
disp(handles.radius);
guidata(hObject, handles);

end

```

```

% --- Executes during object creation, after setting all properties.
function Radius_CreateFcn(hObject, eventdata, handles)
% hObject    handle to Radius (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: edit controls usually have a white background on Windows.
%       See ISPC and COMPUTER.
if ispc && isequal(get(hObject,'BackgroundColor'),
get(0,'defaultUicontrolBackgroundColor'))
    set(hObject,'BackgroundColor','white');
end
end

% --- Executes on button press in Calculate.
function Calculate_Callback(hObject, eventdata, handles)
% hObject    handle to Calculate (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
disp(handles.turns);
disp(handles.radius);
disp(handles.length);
L= 1.00e3*(4*pi*1.00e-
7*handles.turns*handles.turns*pi*handles.radius*handles.radius)/handles.length;
Lstring=num2str(L);
current=1000; %this is actually an input
B=(4*pi*1.00e-7*handles.turns*current)/handles.length;
Bfield=num2str(B);

currentstr=num2str(current);
set(findobj('Tag','Induct'),'String',[Lstring]);
% set(findobj('Tag','magfield'),'double',handles.turns);
set(findobj('Tag','currenttext'),'String',[currentstr]);
set(findobj('Tag','bfield'),'String',[Bfield]);
guidata(hObject, handles);

end

% --- Executes on button press in Discharge.
function Discharge_Callback(hObject, eventdata, handles)
% hObject    handle to Discharge (see GCBO)

```

```

% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
fprintf(handles.uno,'d');
set(findobj('Tag','Status'),'String','Discharging Capacitors');
pause(120);
set(findobj('Tag','Status'),'String','Circuit Safe!');

guidata(hObject, handles);
end

```

```

% --- Executes on button press in pushbutton9.
function pushbutton9_Callback(hObject, eventdata, handles)
% hObject handle to pushbutton9 (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
fprintf(handles.uno,'p');
set(findobj('Tag','Status'),'String','ON');

```

```

guidata(hObject, handles);
end

```

```

% --- Executes on button press in pushbutton11.
function pushbutton11_Callback(hObject, eventdata, handles)
% hObject handle to pushbutton11 (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
fprintf(handles.uno,'s');
set(findobj('Tag','Status'),'String','OFF');

```

```

guidata(hObject, handles);
end

```

```

% --- Executes during object creation, after setting all properties.
function NegativeMonophasic_CreateFcn(hObject, eventdata, handles)
% hObject handle to NegativeMonophasic (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles empty - handles not created until after all CreateFcns called
end

```

```
% -----  
  
% --- Executes during object creation, after setting all properties.  
function figure1_CreateFcn(hObject, eventdata, handles)  
% hObject    handle to figure1 (see GCBO)  
% eventdata  reserved - to be defined in a future version of MATLAB  
% handles    empty - handles not created until after all CreateFcns called  
end
```