MINIATURE WIRELESS GASTROSTIMULATOR

by

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ABSTRACT

MINIATURE WIRELESS STIMULATOR

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Gastroparesis is a common disorder caused to patients suffering from diabetes, cancer and Parkinson's disease. Gastric electrical stimulation (GES) has attracted significant attention in the treatment for Gastroparesis. GES inflicts electrical pulses to the stomach tissues to help regain normal motility and hence reduces Gastroparesis symptoms like vomiting, nausea, abdominal bloating etc. Conventional gastric stimulator needs a long surgery to be implanted and it is a big pacemaker like device that runs by a non-rechargeable battery. It has to be replaced once the battery gets exhausted. Hence another round of surgery is done when these devices are needed to be replaced and re-implanted. This takes the toll over the patients both physically and financially. Most of the times the insurance companies deny to bear the cost and hence the patient cannot afford a treatment.

In this work, two miniaturized wireless gastric stimulators for delivering GES as longterm implants have been designed and demonstrated. These devices are designed as such that they can be implanted through endoscopic implantation and the patients do not need to undergo any surgery at all for the implantation. Wireless telemetry for both devices is based on inductive coupling at a carrier frequency of 1.3 MHz from an external transmitter which delivers power. One design embodies a rechargeable battery with a circuitry to recharge the battery. The magnetic coupling is used to turn the device ON and OFF. The second device is completely batteryless with a circuitry to harvest the radio-frequency energy in real time. Both the circuits were made on printed circuit boards which consisted of microcontroller and many other discrete components. These were coated with biocompatible polymer to protect the implant circuitry from the gastric fluids or any other medium that is present in the stomach. The transmitter consists of a class-E amplifier model and resonance circuitry. An optimization procedure was investigated for achieving the maximum wireless energy transfer for the radio-frequency inductive coupling. The devices have been tested on an acute pig model and Electrogastrogram (EGG) signals of the stomach were recorded. The stomach motility was monitored based on the frequency and amplitude of the myoelectrical pulses and also their consistency. The output voltage was also noted to analyze the power delivery to the tissues. These experiments revealed favorable and significant impacts on the gastric electrical activity.

Once the devices were tested surgically, a series of animal experiments were performed on porcine model to demonstrate the feasibility of the device implantation through endoscopy. Several endoscopy procedures were investigated successfully to provide many possibilities of implantation for the doctors. Endoscopic implantations are outpatient procedures and thus by implanting it endoscopically any surgery need can be avoided.

TABLE OF CONTENTS

ACKNOWLEDGEMENTSiii
ABSTRACTv
LIST OF ILLUSTRATIONS xi
LIST OF TABLES
Chapter Page
1. INTRODUCTION1
1.1 Motivation1
1.2 Proposed Application2
1.3 Objective
1.4 Thesis Organization4
2. INTRODUCTION TO WIRELESS GASTROSTIMULATOR6
2.1 Background6
2.2 Gastroparesis6
2.3 Popular Treatments8
2.4 Gastric Electrical Stimulation Methods8
2.5 Stimulation System11
2.6 Discussion12
3. WIRELESS CIRCUIT DESIGN14
3.1 Implant Configuration14
3.1.1 Charge Pump and Implant Coil
3.1.2 Pulse Generator19
3.1.2.1 Oscillator 55519
3.1.2.1.1 Single Multivibrator in the Implant

3.1.2.1.2 Double Multivibrator in the Implant	23
3.1.2.2 PIC10F Series Microcontroller	24
3.1.2.1.1 Wireless Rechargeable Stimulator	24
3.1.2.1.2 Batteryless Wirelessly-Powered Stimulator	26
3.1.2.3 PIC12F Series Microcontroller	27
3.1.2.4 Comparative Study of Different Pulse Generators	28
3.2 Transmitter Configuration	31
3.2.1 Class-E Amplifier Optimization	32
3.2.2 Wireless Transmission Efficiency Optimization	36
3.3 Discussion	41
4. BENCHTOP EXPERIMENTS	43
4.1 Experiments with BWPS Circuit	43
4.2 Experiments with WRS Circuit	45
5. SURGICAL IMPLANTATION	48
5.1 Surgical Implantation	52
5.2 Porcine Experiments	52
5.2.1 Porcine Model-I	52
5.2.1.1 Results	53
5.2.1. 2 Discussions	57
5.2 Porcine Model-II	62
5.2.1 Results	62
5.1.2 Discussions	67
6. ENDOSCOPIC IMPLANTATION	70
6.1 Resolution Clip Method	70
6.1.1 Methods	70
6.1.2 Results and Discussions	72

6.2 Pinning Method	73
6.2.1 Method	73
6.2.2 Results	76
6.2.3 Discussions	77
6.3 Precutaneous Endoscopic Gastronomy	78
6.3.1 Method	78
6.3.2 Results	80
6.3.3 Discussions	82
6.4 Prototype Endoscopic Tack Method	82
6.4.1 Method	82
6.4.2 Discussions	84
6.5 Prototype Endoscopic Spring Coil Method	84
6.5.1 Method	84
6.5.2 Discussions	85
6.6 Submucosal Pocket Method	85
6.6.1 Method	85
6.6.2 Discussions	87
CONCLUSION AND FUTURE WORKS	89
7.1 Conclusion	89
7.2 Future Works	90
7.2.1 Transmitter Belt	90
7.2.1.1 Design	90
7.2.1.2 Results and Future Modifications	91
7.2.2 Miniature Wireless Implantable EGG Sensor	92
7.2.2.1 Design	92
7.2.2.2 Results and Future Modifications	94

7.

7.2.3 Folding Antenna Design for Stimulator	96		
7.2.3.1 Design	97		
7.2.3.2 Results and Future Modifications			
APPENDIX			
A. PROGRAM FOR STIMULATOR PULSE GENERATOR WITH PIC1	0F 101		
B. PROGRAM FOR STIMULATOR PULSE GENERATOR WITH PIC12F			
C. PROGRAM FOR TRANSMITTER CIRCUIT			
D. PROGRAM FOR OPTMIZATION			
REFERENCES	126		
BIOGRAPHICAL INFORMATION			

LIST OF ILLUSTRATIONS

Figure	Page
2.1 Myoelecytric activity of the stomach, (a) Stomach slow waves generated from the corpus, (b) Initial point of contraction due to slow wave transmission (c) Final point of contraction	7
2.2 Temporary stimulation of a Gastroparesis patient with Enterra Therapy	10
2.3 Stimulation system	12
3.1 Resonating peak at 1.3MHz observed at the receiving end	15
3.2 Implant model with detailed charge pump layout	15
3.3 Pulse train definition	16
3.4 Generated pulse (a) Low setting, (b) Medium setting and (c) High setting	17
3.5 Stimulator circuit layout with one 555 timer placed in the implant circuitry	20
3.6 Astable mode operation circuitry of 555 multivibrator	21
3.7 Stimulator circuit layout with two 555 timers placed in the implant circuitry	23
3.8 WRS stimulator (a) Charging mode (b) Stimulation mode	25
3.9 BWPS stimulator during stimulation	26
3.10Figure 3.10 Change of settings (a) Low to Medium, (b) Medium to High and (c) High to Low doses due to magnetic interrupt with PIC12F series	28
3.11Output voltage response with single multivibrator pulse generator, double multivibrator pulse generator, PIC10F series pulse generator and PIC12F series pulse generator.	29
3.12Size comparison of the single multivibrator implant circuit, double multivibrator implant circuit, WRS device and the BWPS device.	30
3.13Class-E amplifier schematics (a) Without the ESRs of the components and (b) Including ESRs of the components	34
3.14 Equivalent circuit of the inductive coupling resonant circuit	36
3.15 Transmitter efficiency with transmitter radius (a) 4cm, (b) 5cm, (c) and (d) 7cm	40
4.1 Attenuation experiment with the salt solution	43

4.2 PDMS insulation testing in extremely (a) Alkaline and (b) Acidic solutions	44
4.3 Output response of the stimulator dipped in salt solution.	44
4.4 Discharging characteristics of the stimulators operated at Low, Medium and High settings	46
4.5 Charging characteristic of the WRS devices	46
5.1 Stimulator leads inserted in the animal experiment	48
5.2 The animal setup during the surgical implantation	49
5.3 Stomach tissue layer structure	50
5.4 Stimulator testing scenarios.	51
5.5 EGG signal recorded from the Mucosa	52
5.6 Implantation procedure of (a) Enterra [®] neurostimulator in the human body for GES and (b) During the tests for WRS and BWPS stimulators	56
5.7 Average power delivered during the "on" time in the cycle to the(a) Serosa and (b) Mucosa by different stimulators	60
5.8 The stimulator compared with the neurostimulator Enterra [®] . Both WRS and BWPS stimulators have the same size	61
5.9 Variation of FAR ratio with power delivered per cycle for (a)Serosal stimulation and (b) Mucosal stimulation	65
5.10Form factor comparison of Gen-2 wireless stimulator and the commercial Enterra device	68
6.1 Device delivery of the wireless stimulator to the stomach with the endoscope	71
6.2 Endoscopic Implantation with resolution clips. (a) Clip inserted through working channel of the endoscope, (b) Resolution grasps the ear electrode and the Mucosal tissue, (c) Resolution clip deployed and (d) Device attached to the Mucosa with both the ear electrodes attached to the Mucosa	72
6.3 Block diagram for endoscopic implantation with pinning method	73
6.4 Gen-2-New wireless stimulator	74
6.5 Endoscopic implantation of wireless Gastrostimulator through endoscopic pinning method (a) Endoscopic esophageal reintubation and deployement of the stimulator to the stomach, (b) Wire electrodes pinned to gastric Mucosa, (c) Temporary transvenous cardiac pacing lead screwed in the Mucosa and (d) Temporary transvenous cardiac pacing lead secured to Mucosa	75

6.6 Block diagram for endoscopic implantation with PEG method	79
6.7 Device Implantation by PEG method (a) Temporary transvenous cardiac pacing lead screwed and secured in the Mucosa, (b) PEG endoscopic needles punched into the stomach through the skin, (c) Endoscopic hook grasping the metal wire inserted through the needle, (d) Metal wire pulled out through the mouth, (e) Insertion of the second metal wire through the second PEG needle and (f) Stimulator device attached to the Mucosa wall by the tension of the metal wires created by pulling outward	80
 6.8 Device Implantation by endoscopic tack method (a) The catheter containing the endoscopic tack pushing into the gastric Mucosa (b) Deployment of the endoscopic tack into the Mucosa while attaching the ear electrode to the gastric tissue and (c) The magnified image of the attachment of the endoscopic tack to the ear electrode 	83
6.9 Device Implantation by endoscopic spring coil method (a) The release of the spring coil after grasping the ear electrode to the Gastric Mucosa (b) Deployment of the endoscopic spring coil into the Mucosa while attaching the ear electrode to the gastric tissue and (c) The magnified image of the attachment of the endoscopic spring coil to the ear electrode.	85
6.10Illustration of wireless stimulator implantation by SPDI method	87
6.11Gen-2 device with circular wire wrapped electrodes	87
6.12Endoscopic pictures of wireless stimulator implantation by SPDI method	88
7.1 Block diagram of portable transmitter	91
7.2 Stimulator at setting Low operating with the portable transmitter.	92
7.3 Circuit diagram of the EGG sensor	93
7.4 Frequency shift with respect to the applied voltage at the instrumentation amplifier	95
7.5 Fabricated EGG sensor on PCB.	96
 7.6 Folding antenna and Umbrella design structure (a) Folded antenna of 4 cm in diameter, (b) Deformation shown for 4 cm diameter antenna, (c) Foldded antenna of 6 cm and (d) Deformation shown in 6 cm diameter antenna 	97
7.7 Folding antenna and planar spiral coil model (a)Designing and fabrication of folding spiral coil with PDMS and AWG wires, (b) Stimulator with coil loop antenna and planar coil antenna, (c) Folded planar coil antenna around the stimulator and (d) Deformation of planar coil antenna	98
7.8 Experimental setup of stimulator with planar spiral folding antenna	99

LIST OF TABLES

Table	Page
3.1 Pulse and Cycle Specifications for the Stimulators	16
3.2 Resistance and Capacitance values for the Multivibrators	23
3.3 Symbol Representation for Design Optimization	33
5.1 Currents Delivered by the Stimulators at Low, Medium and High Dose Settings in the Serosa Region	53
5.2 Currents delivered by the Stimulators at Low, Medium and High Dose Settings in the Mucosa Region	54
5.3 EGG signal summary	55
5.4 Energies and Powers Deliveries by different Stimulators to Serosa and Mucosa	58
5.5 Voltage and Current Characteristics for Second Animal Experiment	63
5.6 EGG Signal Summary and Energy Calculation	64
5.7 Variation of Output Voltage with the Distance Between the Stimulator and Transmitter in Different Mediums	66
6.1 EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through Pinning Method	77
6.2 EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through PEG Method	81
6.3 EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through Prototype Endoscopic Tack Method	84
7.1 Frequency Shift due to Different Input Voltages	95

CHAPTER 1

INTRODUCTION

1.1 Motivation

In the last few years Gastroparesis has become not only a national concern but more like an international epidemic. In recent advances many medicinal cures and gastric electrical stimulation techniques have been tried and utilized for the treatment of the disorder but none of these were successful enough to eradicate this endemic completely. A non-rechargeable, gastric electrical stimulator, Enterra Therapy (Medtronics Inc, MS) has shown very promising results. It has not been even couple of decades when wireless technology was not very humdrum solution in the widespread field of medical devices. They could barely live upto the expectations in terms of clinical stipulations and performance requirements. The reliability and cost effectiveness was always thought to be questionable when it was about wireless technology associated with medical instruments.

On the contrary, the recent progression in wireless have metamorphosed the scenario and attracted enormous interest from healthcare industry. People realized that the yardstick that can be achieved by wireless technology cannot be achieved by any other approaches. The wireless approach offers another aspect for therapy, diagnosis and cure since wires often prove to be too bulky, unsafe, uncomfortable or even impossible to be employed in some circumstances. Sometimes a big battery adds a lot of extra space, weight and volume to the device which limits the implantation and usage in many cases. Moreover, the contingency of infection due to exposure of wires with the tissue is always an important issue that limits the usage of non-wireless devices for medical usages. On the other hand wireless technology

1

provides a platform for the second level of advancement to the medical device industries since most of the concerns related to non-wireless technology are annihilated.

Considering both the idiosyncrasies, minimal advancement in providing a less painful solution to Gastroparesis and designing a wireless medical device, furnished the idea in us to develop a wireless micro gastric stimulator which can be implanted in a non-invasive way into the stomach.

1.2 Proposed Application

There are many implantable devices that we may see around us starting from pacemakers, neurostimulators, cochlear implants and many others. Since Gastroparesis is a common disorder seen these days as discussed earlier hence gastric stimulators are also one such emerging applications from the group of implantable devices.

The most popular device which is used for curing the dysrhythmias of the stomach is Enterra Therapy as mentioned in the introduction section. This is a non rechargeable battery device that delivers current pulses to the tissue and this helps the gastric tissues to regain the myoelectrical activity. It also improves the symptoms like vomiting, nausea and abdominal pain caused due to Gastroparesis [1.1-1.5].

In this thesis, two new wireless ways to stimulate the gastric muscles has been demonstrated. The first one utilizes a miniature, wirelessly rechargeable, battery for powering the pulse generator while the second one employs a novel design by which a wireless, batteryless, gastric stimulator harvests energy wirelessly so as to power a microcontroller and therefore delivers electrical stimulation. Energy has been transferred from the transmitter to the stimulator through inductive coupling. A series of animal experiments have been performed to demonstrate the feasibility of the stimulators.

In order to provide a much reduced size, simple and cost effective, non-surgical solution in this thesis, the medical stimulator has been also implanted by several endoscopically procedures. This has eliminated the complete surgical hassle.

2

1.3 Objective

The dissertation exhibits an enumerated design of the stimulators and the experimental demonstrations. The prototype operation is tested to manifest the feasible way of human implantation.

The objectives behind the development of the micro gastric stimulator were as follows:

- To illustrate of a new way of delivering pulses to the gastric muscles for the treatment of Gastroparesis
- To design such a stimulator that is very small in comparison to the available solution (Enterra Therapy).
- To fabricate a stimulator that should be able to be operated by wireless power from outside the body so that future replacement is not vital.
- To evince the initial possibility of endoscopic implantation of the gastric stimulator.
 This can be useful for many other wireless medical devices including Gastrostimulator.

1.4 Thesis Organization

Chapter 2 states the severity of Gastroparesis and the popular treatments available in the market. It also elucidates about the commercial gastro-stimulating device that is widely used in the market. It also introduces the proposed wireless stimulator for Gastroparesis. The overall system design is divided into two critical building blocks, (1) Transmitter circuit telemetry and (2) Implant circuit and telemetry.

The transmitter and the stimulator implant systems are introduced in chapter 3. An overview of inductive coupling circuit is presented. This chapter also affirms the configuration of each block in the transmitter system. It includes the class E amplifier, the signal generator and the transmitter antenna design. A simplified system model for the transmitter circuitry, with design optimization techniques based on components characterization results are described.

The stimulator implant circuitry components are also explained in chapter 3. There are two types of implant circuitry that is mainly demonstrated in this research. The first contains a battery and the second is batteryless. A design of a radio frequency (RF)-to-direct current (DC) power converter including a charge pump is presented. This chapter recites the detailed description of both the models. This chapter also provides a detailed comparative study between various pulse generator modules used for the implant.

Chapter 4 specifies the bench top experiments and the initial tests of the implant circuitry. This chapter also states the charging and discharging characteristics for the re-chargeable stimulator model. Both the re-chargeable and the batteryless model is configured in three different settings and this chapter explains the benchtop stimulation performance of each.

While testing the device on an animal model it was tested both surgically and through endoscopic implantation procedure. The surgical implantation is demonstrated in chapter 5. This chapter expresses the surgical implantation techniques and measurements for the stimulator devices tested on two porcine models. Both *in vivo* and *in* vitro experiments were carried out and have been illustrated in the chapter. This chapter replicates the research results and draws conclusion on the gastric activity after and before stimulation.

Chapter 6 presents the endoscopic implantation techniques. Six different endoscopic techniques have been demonstrated in this chapter and that includes Resolution Clip Method, Precutaneous Endoscopic Gastronomy, Fish Hook Tack Method, Spiral Hook Method and Submucosal Pocket Method. Results are shown after the stimulation done through endoscopic implantation; the data is presented and analyzed.

Finally, Chapter 7 recapitulates the research results, draws conclusions, and outlines the future work. It mainly encompasses various models that can be explored in future in the field of gastrostimulator. In future work fragment, the first model that has been illustrated is a portable transmitter that the patient can carry at every place. Secondly the feasibility and proof

4

of concept for designing a new EGG device has been stated. Last but not the least two new designs for folding antennas for gastric devices have been introduced.

CHAPTER 2

INTRODUCTION TO WIRELESS GASTROSTIMULATOR

2.1 Background

Gastroparesis is a significant global health problem which is considered to be disorder caused due to diseases like cancer, diabetes and Parkinson's disease. It is one of the most common reasons that individuals fetch medical care, and the number is alarmingly rising each year. It has been observed that the large population who suffers from Diabetes, amongst them 40-50% suffers from Gastroparesis sooner or later [2.1]. According to the survey of World Health Organization (WHO) 171 million people in the world were suffering from Diabetes in 2000 and the number will increase to 366 million by 2030 while in United States the number of diabetic patients was 18 million in 2000 and this will supposedly shoot up to 30 million in 2030 [2.2].

2.2 Gastroparesis

Gastroparesis is a gastric dysrhythmia and is characterized by delayed emptying of a solid meal. This leads to many symptoms like nausea, early satiety, fullness, anorexia, severe weight loss, abdominal pain, vomiting, and bloating. It is also associated with abnormal gastric myoelectrical activity i.e. abnormal slow-wave frequency, low slow-wave amplitude, and slow-wave uncoupling. This is also known as Bradygastrias or abnormal gastric motility i.e. gastric hypo-motility and uncoordinated gastric or duodenal contractions sometimes referred as Tachygastrias [2.3].

Gastric peristaltic contractions are the nexus for emptying of solids from the stomach and it mainly originates from the Corpus region of the stomach and spread down to the Pyrolic region and thus pushes the food through the Pyrolic canal. These peristaltic movements are the resultant of the slow waves generated by the Interstitial Cells of Cajal (ICC). The ICCs form a dense network of electrically coupled cells between the circular and longitudinal smooth muscle cells. The Muscularis Externa contains an inner circular layer of smooth muscles and a less developed outer longitudinal layer. On the other hand Serosa is a layer of simple Squamous Epithelium on top of that. Mucosa is the inner layer of the stomach and is composed of Epithelium, Lamina Propria, and a Muscularis Mucosa. Muscularis Mucosa has smooth muscle cells in inner circular and outer longitudinal layers similar to Serosa. The slow waves conduct to the smooth muscle cells through the ICCs. Fig. 2.1(a) shows a schematic diagram which depicts how the slow waves are generated in the pacemaker point of the stomach and how it travels throughout. Fig. 2.1(b) and (c) depict the peristaltic contractions and relaxations of the gastric muscles due to the slow wave transmission through the ICC and smooth muscle cells network [2.4]. Thus when a gastric electrical stimulation (GES), has to be given to the stomach to regain the peristaltic movements and myoelectrical activity, it has to be given either to the smooth muscle cells of the Serosa or the Mucosa.





There are a number of conditions that can lead to Gastroparesis but amongst those two most common factors are gastric surgery, diabetes mellitus and cancer. Gastroparesis may be also distinguished as idiopathic, it is classified as such when the factor that have preceded to Gastroparesis cannot be identified. It has been observed that in this group a number of these patients have had previous gastrointestinal (GI) tract viral infections [2.3].

2.3 Popular Treatments

At the initial step many medical treatments are given which involves prokinetic combined with antiemetic agents, due to slow wave transmission [2.4] and also nutritional support such as oral supplementations high on calories. Feeding tube with a jejunostomy, and pain management drugs are also considerable alternatives. Prokinetic agents are those which generally stimulate gastric muscles and hence reduce nausea and vomiting. These drugs include antibiotics like Metoclopramide, Erythromycin, Cisapride, and Domperidone though Metoclopramide and Erythromycin are only such drugs, which are commercially available in the United States. It has been discerned that both of them equally have bad side effects and often intolerable for more than 40% of patients. Partial or complete gastrectomy is also considered to be a solution but it is regarded as a last solution due to the associated morbidity during the process [2.3].

2.4 Gastric Electrical Stimulation Methods

Several major approaches have been used for the treatment of Gastroparesis, including (a) antibiotic medication, the most common preference by the patients at the preliminary stage; (b) feeding tube to feed the patients since the stomach stops moving and food cannot be engulfed without an external help and (c) surgical implantation of gastrostimulator is considered to be the last resort. As an attractive option but not yet been implemented in wide clinical uses, GES is a therapeutic technique in which either high frequency/low energy (HFLE) or low frequency/high energy (LFHE) electric pulses are delivered to the stomach tissues for gastric emptying and regaining the normal stomach motility for the

patients, who are suffering from Gastroparesis. In past experiments, the currents for the stimulation pulses varied from 2 to 6 mA [2.5–2.12] and the stimulators need to be operational continuously or in long periods. Providing energy to the stimulators has thus become a primary challenge for device designs. For fully implantable devices, the battery needs to be included in the stimulator. As a result to include a large capacity battery, the device size is large so that a surgery of implantation with general anesthesia and hospitalization up for 4 days is needed. Because the implantation is a dramatic step, most of the patients will first receive a temporary treatment with stimulator electrodes attached to the stomach and wires transnasally through esophagus connected to an external stimulator. During temporary stimulation the patient carries in a pouch in front of the chest which contains the stimulator [2.7].

Fig. 2.2 shows a detailed snap shot of temporary stimulation in a patient. This temporary stimulation is carried between 1-6 months while the patient is kept under observation during this period. Due to the lack of specially-designed smaller wireless stimulators for GES applications, presently a FDA-approved neurostimulator Enterra[®] (Medtronic, Inc.) is used for the HFLE-GES applications in humans. The device, originally designed for neurostimulation, is implantable and can be remotely controlled by a handheld module. The device produces variable pulse trains with different voltage and current levels to the output electrodes. One of the frequently used stimulation specification is stimulating pulses delivered at 14 Hz frequency with an output current level of 5 mA. The device dimensions are 60 mm x 55 mm x 10 mm and it contains a non-rechargeable lithium-ion battery which has to be replaced each 2 to 8 years as it gets exhausted [2.8, 2.13].

With the intention to reduce the stimulator size, development of a device that operates on rechargeable batteries have been demonstrated, and also a novel device that does not contain any battery at all but can harvest electromagnetic energy wirelessly to generate stimulation currents was made. Wireless energy transmission through body has been proposed by Schuder *et al.* who detailed the theoretical illustration. In their research inductive coupling between a

pancake-shaped coil on the skin surface was used, that can transfer power efficiently to a coil placed inside the body [2.14], Others such as Andren *et al.*, Meyers *et al.* and Newgard *et al.* have suggested to use ferromagnetic cores in inductive coupling for efficient transfer of magnetic energy through the skin [2.15-2.17].



Figure 2.2 Temporary stimulation of a Gastroparesis patient with Enterra Therapy

Energy transfer through the inductive coupling for medical devices can overcome many issues like contact failure, the need for replacing the battery each time it is exhausted and hence infection due to repeated surgeries. However significant power delivery is the biggest challenge for devices operating with inductive coupling. These devices generally have to be operated at a very short distance [2.18] due to limited power. Since then many medical applications have been demonstrated for transmission of energy in neuron recording [2.19, 2.20], pressure monitoring in stems [2.21] and gastro-esophageal acid reflux sensing [2.22] applications. For medical devices many such applications focused on providing energy into the body to operate sensors and wireless transmitters to transponder bio-signals. With the

advances in low-power integrated circuits, the energy required to operate such devices provided by inductive coupling proved to be considerable after taking special measures.

Electric stimulation on neurons in cochlear and retinal implants to enable propagation of neural signals usually require less energy than the energy required by Cajal or smooth muscle cells to enable tissue motility. Magnetic fields through inductive coupling at lower frequencies that can penetrate deeper inside the human body without electromagnetic interference to the electronics in the implant, provide advantages in neurostimulation applications [2.23–2.25]. However, delivering a significant amount of electromagnetic power to the implant still remains the main technical challenge.

2.5 Stimulation System

The stimulation system consists of an external transmitter and an implant. The external transmitter transfers electromagnetic energy through the body to provide the operating power for the implant whereas the implantable stimulator generates voltage pulses and have protruded electrodes anchored into the stomach tissues. In the animal experiments, fixed helix tip bipolar electrodes with a 0.9 mm diameter and ring electrodes with a diameter of 1.1 mm (Temporary Transvenous Pacing Lead 6416-200 cm, Medtronic, Inc.) were used however several different kinds of electrodes were used during the endoscopic implantation procedures. Fig. 2.3 shows the complete outlook of the stimulation system including the transmitter and the implant.

In practical uses, the stimulators will be implanted by either surgery which is the present implementation method for existing large stimulators or with an endoscope noninvasively. The stimulator models needed to be miniaturized enough that they can be implanted in the endoscopic manner non-invasively. The miniaturization has been illustrated in the later chapters.

With our stimulators, electromagnetic wave energy is transmitted wirelessly through the body by inductive coupling between the transmitter antenna and the implant antenna to either charge the rechargeable battery when not stimulating or power up the stimulator directly in real time. The stimulation pulse trains were preprogrammed in a microprocessor placed in the implant. In our experiments, three different stimulation settings (Low, Medium and High) were used. These settings were designed according to previous human studies using GES for Gastroparesis treatment. The devices were tested in an animal model and EGG signals were recorded to identify the significance in modulation of stomach motility.



Figure 2.3 Stimulation system

2.6 Discussions

After the basic reviews considering all the solutions and treatments available in the market for curing Gastroparesis, coming up with a more innovative and simple solution seemed to be a very vital need. It will be beneficial to the society and the humanitarian community. Henceforth in this work, development of a new method for gastric electric stimulation has been

aimed. The first design is an incremental improvement to present stimulation method with a rechargeable battery and a wirelessly charging transmitter but the size is way smaller than the existing device. Since the battery can be recharged and does not need to have a large capacity to last more than four years, therefore the size of the stimulator could be dramatically decreased. The second design involves a completely wireless model without any battery. This device is operated at real time whenever electromagnetic energy will be supplied to the implant through the transmitter. In this method, eliminating the battery eliminates the need for future implant replacement due to limited battery recharging lifetime.

Further miniaturization of both the models was performed to ensure that the device is capable of getting implanted with an endoscope. This will eliminate the need for a major surgery, general anesthesia and hospitalization as well and it will also reduce the treatment costs.

CHAPTER 3

WIRELESS CIRCUIT DESIGN

3.1 Implant Configuration

The implant circuit was designed on a 2-layer printed circuit board (PCB). A coil antenna with the inductance value of 15 μ H was made from a AWG-24 magnet wire (Belden Wire & Cable) wound around the PCB. The use of a thicker magnetic wire would have escalated the maximum magnetic flux (ψ) linkage since increasing the inductance per turn ratio will aggravate the magnetic flux as it can be demonstrated in equation 3.1. [3.1]. Lowering the gauge also makes the wire thicker and thus restricting the flexibility. AWG-24 was chosen because of the moderate flexibility and for winding restrictions around the PCB board. Also switching to lower gauges would have increased the implant size. For the endoscopically implantable stimulator the maximum permissible thickness is 8 mm and therefore 15 numbers of turns with AWG-24 was a suitable choice.

$$\psi = \frac{LI}{N} \tag{3.1}$$

The operating frequency (f) was 1.3 MHz and was chosen since it has been observed in many previous works that frequency range from 1-30 MHz is the best for implantable medical devices and provides maximum permissible exposure (MPE) of magnetic field to the internal tissues [3.2]. Absolute resonance is very crucial hence the resonance frequency was verified by the spectrum analyzer connected to the receiving antenna as shown in Fig. 3.1. The resonating capacitor of the implant side was chosen as 1070 pF based on the equation (3.2) as follows:

$$f = \frac{1}{2\pi\sqrt{L_2C_2}} \tag{3.2}$$

Figure 3.1 Resonating peak at 1.3MHz observed at the receiving end.



Figure 3.2 Implant model with detailed charge pump layout.

The energy harvesting circuit consists of a series of diodes and capacitor bank. This whole capacitor and diode assembly resembles Dickson's charge pump [3.3] and gives a DC amplified output from the received RF energy. The circuit diagram of the implant illustrating the details of the charge pump is shown above in Fig. 3.2.



Figure 3.3 Pulse train definition.

	T g	T _o	T on	T off
Low	330 µs	14 Hz/71.4 ms	0.1 s	5 s
Medium	330 µs	28 Hz/35.7 ms	1 s	4 s
High	330 µs	55 Hz/18.2 ms	4 s	1 s

Table 3.1 Pulse and Cycle Specifications for the Stimulators

The definitions for the pulse train parameters are illustrated in Table 3.1 and the parameters can also be contemplated from Fig. 3.3. Three different stimulation settings, based on previous works [3.4 – 3.6] were implemented in the pulse generators as shown in Fig. 3.4. Pulse frequency is defined within cycle "on" time period T_{on} and the N is the number of pulses during the cycle "on" time period T_{on} . These settings were proposed based on the Low, Medium and High dose needed to retain the stomach motility. Higher settings have faster pulses and

less T_{off} time. T_{off} depicts the "off" period between each cycle of stimulation while T_o shows the "off" period between each pulse.



Figure 3.4 Generated pulse (a) Low setting (b) Medium setting and (c) High setting.

As demonstrated in Fig. 3.4 each cycle operating in the Low setting encompasses 2 pulses in 0.1 s with a cycle length of 5.1 s, Medium setting includes 28 pulses in 1 s with a total

cycle length of 5 s whereas High settings incorporates 216 pulses in 4 s time period with a cycle length of 5 s.

3.1.1 Charge Pump and Implant Coil

The capacitors used in the charge pump measures $0.68 \ \mu$ F. The capacitance value was calculated based on the current requirement at output of the charge pump. Since the current required that needs to be delivered to the tissue was quite high, that led to such higher value of the capacitances. Equation (3.3) demonstrates the formula required to calculate the capacitance values based on the current and voltage requirement.

$$C = \frac{Q}{V} = \frac{It}{V}$$
(3.3)

For the calculation of capacitance values "C" the voltage value of 2.5 V and current of 5 mA was incorporated in the equation (3.3). The value of time "t" was considered as 330 µs since all the pulse widths for Low, Medium and High resembles that value. While choosing the capacitors it has to be taken into consideration that, the lower the equivalent series resistance (ESR), the better is the ripple noise handling capacity. Therefore low ESR multi layer ceramic capacitors were chosen.

Schotkky diode BAT54SWT1G (Fairchild Semiconductor) was used in the diode bank due to their low forward voltage drop (0.35 V) as compared to the normal PN junction diodes. Schotkky diodes are also a better bet due to their fast switching speed. The implant coil was chosen depending on size of the implant, the required flexibility, inductance per coil length and the current requirement at the output. For endoscopic implantation the size of the implant can be no more than 11 mm in width and 10 mm in height. Therefore considering 2 mm allowance for insulation purpose, the size of our implant printed circuit board (PCB) can be not be bigger than 9 mm in width and 8 mm in thickness. The coil needs to be flexible enough so that it can be winded around the PCB. Coil diameters are inversely proportional to their gauge numbers. Choosing a very thick coil will make it non flexible whereas choosing very thin coil might not generate sufficient current that is required at the output. It can be seen from equation (3.4) that the output current is inversely proportional to the inductance per unit length "L" of the coil.

$$I \propto \frac{1}{L} \tag{3.4}$$

Inductance per unit length "L" of the coil can be depicted from equation (3.5).

$$L = 0.1257a \left[2.303 \log_{10} \left(\frac{16a}{d} - 2 \right) \right] \mu H$$
 (3.5)

Inductance per coil is inversely proportional to cross sectional area therefore thicker coil provides less inductance per unit length and thereby increasing the maximum current output. Optimizing both the equations and considering the flexibility of the coil, AWG-24 was chosen for our implant coil.

3.1.2 Pulse Generator

The rate of the stimulation pulses required at the output is significantly high (55 Hz for High settings) hence different pulse generators were explored depending on the size, the power consumed and the response time. The subsections below describe in detail the performances of each pulse generator circuitry.

3.1.2.1 Oscillator 555

Our first model involves the design of the pulse generator circuitry with a 555 oscillator. There are two different circuit scenarios that have been considered and demonstrated in subsub-sections 3.1.2.1.1 and 3.1.2.1.2.

3.1.2.1.1 Single Multivibrator in the Implant

The first design with 555 oscillator includes one 555 timer (Maxim Integrated Products-ICM7555ISA+) in the implant which generates a continuous pulse of 14Hz for Low setting, 28 Hz for Medium setting and 55 Hz for High setting with pulse width of 330 µs each. This timer is referred as *"faster"* pulse generator since significantly fast pulses are generated as compared to the modulation pulses. The second 555 oscillator pulse generator was placed in the transmitter circuitry and this provides the modulation pulses at 0.1 s with an "off" period of 5 s for Low setting, 1 s with an "off" period of 4 s in Medium setting and 4 s with an off period of 1 s for High setting. This pulse generator/multivibrator circuit is referred as *"slower"* pulse generator. An AND logic gate was used with the modulation 555 multivibrator and the frequency generator generating carrier frequency of 1.3 MHz at 50% duty cycle. The implant circuitry receives RF energy for the modulated period and this RF energy gets transferred into amplified DC signal through the charge pump as discussed in sub-section 3.1.1. The DC signal powers the faster pulse generator which gives the final output to the tissue through stainless steel electrodes. Fig. 3.5 shows the circuit diagram for the wireless stimulator circuit with one 555 multivibrator implanted in the stimulator circuit.



Figure 3.5 Stimulator circuit layout with one 555 timer placed in the implant circuitry.

Both the multivibrators are operated at astable mode for pulse generation. An astable multivibrator is a timing circuit whose 'Low' and 'High' states are both unstable. As such, the

output of an astable multivibrator toggles between 'Low' and 'High' continuously, in effect generating a train of pulses. This circuit is therefore also known as a 'pulse generator' circuit.

In this circuit, the charging capacitor "Ca" gets charged through Ra and Rb which are the charging and discharging resistances for Ca. Thus eventually it builds up enough voltage to trigger an internal comparator to toggle the output flip-flop. Once toggled, the flip-flop discharges Ca through Rb into pin 7, and this is the discharge pin of the integrated circuit (IC). When Ca's voltage becomes low enough, another internal comparator is triggered to toggle the output flip-flop. This once again allows Ca to get charged till the input voltage through Ra and Rb and the cycle starts all over again. Then the capacitor charges up to 2/3Vcc (the upper comparator limit) which has been determined by the 0.693(Ra+Rb)×Ca combination and discharges itself down to 1/3Vcc (the lower comparator limit) determined by the 0.693(Rb×Ca) combination. Fig. 3.6 shows the astable circuitry of the multivibrator and the connection specifications.



Figure 3.6 Astable mode operation circuitry of 555 multivibrator.

The charge and discharge time can be easily obtained from equation (3.6). The charging time resembles the "on" time (T1) of the pulses whereas the discharging resembles the "off" time (T2). This can be predicted from equation (3.6) too. For the faster pulse generator T1 is 330 μ s and T2 is 71.4 ms, 34.5 ms and 18.2 ms for Low, Medium and High settings. For the slower pulse generator T1 is 0.1 ms, 1 s and 4 s while T2 is 5 s, 4 s and 1 s for Low, Medium and High setting respectively.

$$T1 = 0.693(Ra + Rb) \times Ca$$

$$T2 = 0.693Rb \times Ca$$
(3.6)

When connected as an astable multivibrator, the output from the 555 Oscillator will continue indefinitely charging and discharging until the power supply is removed. The duration of one full cycle is therefore equal to the sum of the two charging and discharging periods together and is given in equation (3.7) as below:

$$T = T1 + T2 = 0.693(Ra + 2Rb) \times Ca \tag{3.7}$$

The output frequency of oscillations can be found by inverting the equation above for the total cycle time. Hence the final frequency equation can be obtained from equation (3.8) as follows:

$$f = \frac{1.44}{(Ra + 2Rb) \times Ca} \tag{3.8}$$

By altering the time constant of just one of the RC combinations, the duty cycle can be accurately set and is given as the ratio of resistor Rb to resistor Ra. The Duty Cycle for the 555 Oscillator can be predicted as below, which is the ratio of the "on" time divided by the "off" time as shown in equation (3.9):

$$DutyCycle = \frac{T1}{T1+T2} = \frac{Ra+Rb}{Ra+2Rb} \times 100\%$$
(3.9)
By connecting the diode, D between the trigger input and the discharge input, the timing capacitor is charged up directly through the only resistor Ra, while resistor Rb is effectively shorted out by the diode. The capacitor discharges as normal through resistor Rb. The

	Ra _{faster} (kΩ)	Rb _{faster} (ΜΩ)	Ca _{faster} (µf)	Ra _{slower} (kΩ)	Rb _{slower} (kΩ)	Ca _{slower} (µf)
Low	49.25	10.65	0.01	10	500	15
Medium	49.25	5.33	0.01	100	400	15
High	49.25	2.72	0.01	400	100	15

Table 3.2 Resistance and Capacitance values for the Multivibrators

previous charging time of T1 = 0.693(Ra + Rb)×Ca is modified to take account of this new charging circuit and is given as: 0.693(Ra × Ca). The duty cycle is therefore given as Ra/(Ra + Rb). To generate a duty cycle of less than 50%, resistor Ra needs to be less than resistor Rb. Without the diode it is not possible to generate a duty cycle below 50%. Table 3.2 shows the values of Ra and Rb in both the multivibrator for producing the signal continuous pulses and the modulated pulses.

3.1.2.1.2 Double Multivibrator in the Implant



Figure 3.7 Stimulator circuit layout with two 555 timers placed in the implant circuitry.

This circuit involves two multivibrator implanted in the stimulator circuit. The transmitter emits an un-modulated continuous RF signal at 1.3 MHz with 50% duty cycle which is received by the implant tuned circuitry and amplified after rectification by the charge pump. The multivibrator block containing dual multivibrators (one slower and the other one faster) gets powered by a 2.5 V DC voltage from the charge pump and the regulator assembly together.

This circuit does not require any logic gate for the operation but the double multivibrator implant is much bigger in size compared to the single multivibrator implant circuitry. Fig. 3.7 shows the circuit layout of the double multivibrator implant circuitry. The values of Ra, Rb and Ca of the faster and the slower pulse generator is similar to the single multivibrator implant circuitry as demonstrated in Table 3.2

3.1.2.2 PIC10F Series Microcontroller

Peripheral interface controllers (PIC) are a comparatively new genre of microcontroller family that has a different storage and control line. The PIC series was chosen due to the inbuilt oscillator with selectable speed and due to the low power consumption. PIC10F series is the lowest power PIC available in the market with the smallest surface area and hence perfect for our application. Two different designs have been implemented with the PIC series: the first is a wirelessly rechargeable stimulator (WRS) containing a rechargeable, lithium-ion battery, while the second design employs a batteryless wirelessly-powered stimulator (BWPS) that harvests electromagnetic energy from an external transmitter too.

3.1.2.2.1 Wireless Rechargeable Stimulator

The WRS system consists of a stimulator and an external wireless transmitter which charges a 3 V, 11 mAh rechargeable lithium-ion cell battery in the implant through inductive coupling, as shown in Fig. 3.8. It contains a charge pump to convert the received RF energy from the transmitter to a DC current, and to amplify the voltage for recharging the battery. Resonant capacitors for the coil antenna were used to capture the RF energy.



Figure 3.8 WRS stimulator (a) Charging mode (b) Stimulation mode.

The external energy was transmitted with 4 watts of electromagnetic power at 1.3 MHz. The transmitter coil antenna measures 6 cm in radius. A magnetic reed switch (KSK-1A80-1015) was used to connect the rechargeable battery to the recharging driver circuit, or to a PIC. It is activated by an external magnet so as to shift the circuitry from the charging mode (Fig. 3.8(a)) to the stimulation mode, or vice versa (Fig. 3.8(b)). The PIC was pre-programmed to generate a pulse train with specific frequency and duty cycle. The different stimulation settings, are similar to Fig. 3.4 as show above prior in this chapter.



3.1.2.2.2 Batteryless Wirelessly-Powered Stimulator

Figure 3.9 BWPS stimulator during stimulation.

The BWPS system consists of a batteryless stimulator and an external transmitter which transmits RF electromagnetic energy into the body so as to power up the stimulator. The

transmitter circuitry is exactly similar to the WRS transmitter that is used for charging the WRS device. The stimulator contains similar circuits as WRS devices and multivibrator implants for harvesting electromagnetic energy and generating simulation pulses, as shown in Fig. 3.9. In the transmitter block a function generator (Fluke PM5139, 0.1mHz-20MHz) generates a 1.3 MHz square waveforms at 50 % duty cycle and 0 offset voltage. The signal gets amplified by a class-E amplifier, and is fed to a transmitter antenna. The signals were converted into electrical currents by a receiving coil antenna and rectified with a charge pump to power the preprogrammed PIC, also in three settings shown as Fig. 3.4. There is no battery implanted in the circuit and hence it is more of a real time transmission stimulation operation. This generated electrical pulse trains similar to those in the WRS devices. When the transmitter is moved away from the body, the PIC device in the BWPS turns off and the pulse trains stop instantaneously due to lack of sufficient drivable voltage.

Both WRS and BWPS designs were packaged in a soft polydimethylsiloxane (PDMS) coating, extended two thin wire leads served as electrode connections and both the designs measures 9 mm x 37 mm x 8 mm. With PIC10F series since three programs cannot be incorporated in a single microcontroller hence one device is made for each Low, Medium and High setting. The program for each setting has been demonstrated in Appendix A.

3.1.2.3 PIC12F Series Microcontroller

PIC12F series is also as small as PIC10F series however they consume more power than PIC10F microcontrollers. This series was used so that all the three settings can be incorporated in a single PIC microcontroller and fabrication of different devices for different settings is not needed. This device was mainly modeled and programmed for the batteryless application of the wireless stimulator. Hence the stimulation mode is very similar to the BWPS device with PIC10F series. The switching mechanisms of the stimulator from one setting to another have been performed by an interrupt low signal to the interrupt pin of the PIC12F683. This setting change occurs due to swapping of setting program memory locations as elucidated in Fig. 3.10. If the PIC microcontroller is set at setting-Low initially then after the first interrupt is given, it goes to setting-Medium and stays in there until another interrupt signal comes up. This is demonstrated in Fig. 3.10(a). If the patient needs stimulation in High setting then a second interrupt needs to be given by the doctors as illustrated in Fig. 3.10(b). Fig. 3.10(c) shows the scenario if the patient is in hi-setting mode and needs to go back in setting-Low. The change of settings operation can be performed by the doctors under clinical observations. The program written for combining three settings in a single device has been illustrated in Appendix B.



Figure 3.10 Change of settings (a) Low to Medium, (b) Medium to High and (c) High to Low doses due to magnetic interrupt with PIC12F series

3.1.2.4 Comparative Study of Different Pulse Generators

Fig. 3.11 shows the output voltage response under High settings with different pulse generators descried in the previous sections. The voltage response was obtained with 800 Ω load attached to the output of each stimulator device with different pulse generators and thus mimicking the stomach tissues. Tissue impedance of the internal layer of the stomach ranges from 200 Ω -800 Ω and therefore the highest limit of impedance was considered to be on the safe side. Output current was calculated by dividing the output voltage with the output load. The energy harvesting and the charge pump circuit was kept similar for all the designs while



Figure 3.11 Output voltage response (a) Single multivibrator pulse generator, (b) Double multivibrator pulse generator, (c) PIC10F series pulse generator and (d) PIC12F series pulse generator.

fabricating the implants. Only the pulse generator was replaced by different pulse generating ICs. The implant circuitry with single 555 multivibrator implanted provided 1.25 mA output current as shown in Fig. 3.11(a). The stimulator implant with two multivibrator provided pretty low value of output current (625 mA) depicted in Fig. 3.12(b). The output current was highest when PIC10F206 was used as the pulse generator in the stimulator and we could obtain approximately 3.5 mA at the output (Fig. 3.12(c)) could be obtained. The output current decreased by 0.8 mA while using microcontroller PIC2F683 than in comparison to PIC10F206 (3.12(d)).



Figure 3.12 Size comparison of the single multivibrator implant circuit, double multivibrator implant circuit, WRS device and the BWPS device.

The output voltage and current varied with different pulse generator circuitry due to the power dissipation in the components. ICM7555 which was used in both the multivibrator pulse generator circuits dissipate 780 mW of power during operation therefore the multivibrator circuit where one multivibrator was used gave more output current than the one with two multivibrators. The circuit with two multivibrators dissipated 1560 mW (780 mW + 780 mW) in the pulse generating block. The PIC10F206 dissipates 800 mW at absolute maximum settings (when the input voltage is 5.5 V however in this case a 2.0 V voltage was obtained from the

charge pump) while in our operating range, 350 mW power was dissipated into it and thus ramping the output voltage and current. Similarly PIC12F683 also dissipates 800 mW power. However in our operating conditions only 420 mW power was dissipated into it due to the DIP package used instead on the SMD package. Fig. 3.12 shows the size comparison of the different stimulator designs. The multivibrator designs are much thicker than the design with the PICs. Due to the smaller size of the PIC ICs significant size reduction was possible. Also many passive components (capacitors and diodes), required for generating pulse timing cycles in Multivibrators circuits can be abandoned by using PIC devices in the pulse generator due to their programmable nature. The passive components in the multivibrator design of the implant also resulted in the power dissipation which was avoided by using PIC microcontroller. The PIC WRS and BWPS device provided more voltage output by transmitting same amount of power for the similar distance when compared with the multivibrators. Therefore the devices used for the animal experiments to demonstrate the performance of the stimulator had PIC10F0206 as the pulse generator in it.

3.2 Transmitter Configuration

The transmitter block consists of a signal generator, a class-E amplifier and a transmitter coil antenna resonating at the desired carrier frequency. The carrier frequency was purposely kept low at the 1.3 MHz public band for low transmission losses occurred in tissues [3.1–3.2]. There is a tradeoff between the harvested voltage by inductive coupling and the carrier frequency. As the frequency increases, the induced voltage also increases. However, the transmission loss may also increase hence considering the whole scenario, the coupling efficiency decreases as the frequency increases after 10 MHz of the application involves implantable devices.

31

3.2.1 Class-E Amplifier Optimization

A class-E power amplifier was utilized so that in the ideal case the voltage and current of the amplifier would be 90° out of phase and hence the power consumption would be considered zero [3.7–3.8]. Class-E power amplifier has been contemplated for transcutaneous power delivery for numerous such previous works [3.9]. Other models of amplifiers like class-A, class-B, class-AB or class-C are either very power inefficient or produce very high distortion, though they have much less complex structure. While amplifiers like class-D, produce ten times or even higher frequency output signal than the input signal. Producing higher frequency limited its use in our circuitry. It would have made the transmitted signal susceptible to transmission losses while traveling through the biological tissues. Class-D amplifiers also contain many inaccurate spectral components like harmonics. The carrier frequency was supplied through a rectangular waveform of 9V peak-to-peak from a function generator. A circular coil antenna with a radius of 6 cm was made from AWG-24 wires. A quality factor (Q) of 80 and an inductance (L₁) of 48 μH were measured. A MOSFET (IRF540, Fairchild Semiconductor) was considered in the class-E amplifier for its low threshold voltage. The inductor L_E in the class-E amplifier works as part of a current source from the 9V_{DC} power supply and provides the current to the resonant circuit. Table 3.3 shows the symbol representation of the components for the transmitter block and the implant block including the class-E amplifier. The value of L_E was chosen as 300 μ H which is more than six times the inductance of the transmitter antenna. The tuning C_1 of 330 pf were utilized for transmitter resonance at 1.3 MHz and a shunt capacitor C_E of 10 nf was used for discharging. Absolute resonance is a very necessary factor for maximum power transmission through the class-E amplifier and the transmitter antenna. The circuit diagram is shown in Fig. 3.13.

The current through L_1 and C_1 follows a sinusoidal nature with a frequency equal to that of the input frequency. When the MOSFET is closed, L_1 and C_1 supply the current back to the switch and no current flows through *C*. The voltage observed across the switch is zero at this instance. Then the MOSFET turns on and L_1 and C_1 start to supply current once again. The current continues to follow in the same opposite direction as it was flowing however now it flows through C_1 . This makes a positive voltage drop across the MOSFET. The current in L_1 and C_1 is reversed; it follows in the positive direction. The capacitance *C* supplies the current to L_1 and C_1 and this reduces the voltage across the MOSFET and finally drops it to zero. At this moment the MOSFET closes itself. Ideally the voltage and current should be 90 degree out of phase [3.10–3.11]. The circuit should be operated at a high-Q and also at a particular frequency and duty cycle combination [3.8]. It was observed that our circuit delivers the highest power at 50% duty cycle. It was chosen to minimize the direct DC power consumption. Troyok et al. have shown their class-E amplifier operating in minimized loss mode when the frequency of operation was considered to be 1 MHz. [3.12]

Symbol	Description	Symbol	Description
,	D		
L_E	Power inductor	V _{dc}	Input voltage to the Class-E amplifier
C_E	Shunt capacitor	R_{C1}	ESR of the tuning capacitor
R_{DS}	ESR of the MOSFET IRL540	R_{CE}	ESR of the shunt capacitor
R_{LE}	ESR of the power inductor	η_E	Class-E amplifier efficiency
η	Total efficiency	$\eta_{ au}$	Transmitter efficiency
L_1	Transmitter inductance	R ₁₀	Single turn resistance of the transmitter
L_2	Implant inductance	R_{20}	Single turn resistance of the implant
C_1	Tuning capacitance of transmitter	C_2	Tuning capacitance of the implant
L ₁₀	Transmitter single turn inductance	L ₂₀	Implant single turn inductance
R₁	ESR for the transmitter	k	Coupling coefficient, $0 \le k \le 1$
R_2	ESR for the implant	V _{in}	Input voltage to the transmitter coil
М	Mutual inductance between coils	V _{out}	Output voltage at the load
Н	System efficiency	A_{v}	Voltage gain
$R_{AC \ load}$	Load impedance of the implant	r ₁	Radius of the transmitter coil
W ₀	Angular frequency	<i>r</i> ₂	Radius of the implant coil
N_1	Number of turns in the transmitter	P _{in}	Input power to the transmitter antenna
N ₂	Number of turns in the implant	Pout	Output power at the load
σ	Conductivity of copper	δ	Skin depth for the AWG 24 wire
Х	Distance between the two coils	V _{induced}	Induced voltage in the secondary
Z _{ref}	Reflected impedance	1	Current flowing through the primary
P_{Loss}	Power loss due to ESRs	$P_{R_{DS}}$	Power loss due to EST of IRF540
$P_{R_{LE}}$	Power loss due to ESR of R_{LE}	$P_{R_{CE}}$	Power loss due to ESR of R_{CE}
$P_{R_{C1}}^{}$	Power loss due to ESR of R_{C1}	P_{R_1}	Power loss due to ESR of R_1
Φ	Phase of class-E amplifier	D	Duty ratio

Table 3.3 Symbol Representation for Design Optimization.

The efficiency equation of the class-E amplifier can be derived as follows shown by equation 3.10

$$\eta_E = \frac{P_{in}}{P_{in} + P_{Loss}} \tag{3.10}$$

The efficiency of the amplifier can be calculated by incorporating all the values of the ESRs in the power calculation, the phase and the duty ratio in the equation (3.10). The phase can be calculated as specified in equation (3.11).

$$\phi = \tan^{-1} \left[\frac{\cos(2\pi D) - 1}{2\pi (D - 1) + \sin(2\pi D)} \right] + n\pi$$
(3.11)



Figure 3.13 Class-E amplifier schematics (a) Without the ESRs of the components and (b) Including ESRs of the components.

$$P_{Loss} = P_{R_{DS}} + P_{R_{LE}} + P_{R_{CE}} + P_{R_{C1}} + P_{R_{1}}$$

$$P_{Loss} = \frac{P_R}{R} \begin{cases} R_1 + R_{C1} + R_{CE} + [R_{LE} + R_{DS}(2 - D) + R_{C1}] \left[\frac{\cos(2\pi D + \phi) - \cos(\phi)}{\pi \sqrt{2}(1 - D)} \right]^2 \\ + (R_{DS} - R_{C1}) \left[D + \frac{1}{4\pi} \sin(2\phi) - \sin(4\pi D + 2\phi) \right] \end{cases}$$

$$P_{Loss} = \left(\frac{P_R}{R}\right) \left\{ \begin{pmatrix} R_{CE}(D-1) + R_{LE} + R_{DS}(2-D) + \pi(1-D)^2 \\ 2\pi[DR_{DS} + (1-D)R_{CE} + R_{C1} + R_1] \\ \frac{(R_{CE} - R_{DS})\cos(2\pi D + 2\phi)\sin(2\pi D)}{[\cos(2\pi D + \phi) - \cos(\phi)]^2} \end{pmatrix} \right\}$$

After incorporating the value of P_{Loss} into the class-E amplifier efficiency equation (3.10) the final efficiency equation obtained is as depicted in equation (3.12).

$$\eta_{E} = \frac{1}{1 + \left(\frac{R_{C1} + R_{1} + R_{CE} + (R_{DS} - R_{CE})\left[D + \frac{1}{4\pi}\sin(2\phi) - \sin(4\pi D + 2\phi)\right]}{R}\right)}{+ \left(\frac{[R_{LE} + R_{DS}(2 - D) + R_{CE}][\cos(2\pi D + \phi) - \cos(\phi)]^{2}}{2\pi^{2}(1 - D)^{2}R}\right)}$$

$$\eta_{E} = \frac{1}{1 + P_{R} \left(\frac{R_{DS}(2-D) + R_{CE}(D-1) + R_{LE}}{V_{dc}^{2}} + \frac{1}{\pi (1-D)^{2} \left[2\pi \left\{ \frac{DR_{DS} + (1-D)R_{CE}}{+R_{1} + R_{C1}} + \cos(2\pi D + 2\phi)\sin(2\pi D) \right]} \right] (3.12)$$

$$\frac{1}{V_{dc}^{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}} \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}}} \right] \left[\frac{1}{\sqrt{\frac{1}{2} \left[\cos(2\pi D + \phi) - \cos(\phi) \right]^{2}$$

The values of the ESRs for each component of the class-E amplifier were taken from the commercial datasheet available. The value of R_{DS} was 0.075 Ω , R_{CE} was 0.8 Ω and R_{LE} was 2.9 Ω . However the value of R_{C1} was 0.2 and r_1 was measured as 2.3 respectively. D was considered as 0.5 and the value of Φ was calculated from equation (3.11) as a lag of 32.5 degree. The calculated efficiency from equation (3.12) of the class-E amplifier is 67%. This efficiency can be increased by using capacitors with less ESR like in tantalum capacitors and by using less resistive coils than AWG-24. However using commercial coils would be more beneficial.

3.2.2 Wireless Transmission Efficiency Optimization

To consider maximum power transfer through the inductive link, an optimization procedure of the transmitter antenna was carried out. In this procedure, the main variables to be optimized are the turn numbers of the transmitter coil for a particular coil gauge and also the transmitter size. To predict the correct turn number and size of the transmitter a series of calculative programming and experiments were performed. Fig. 3.14 shows the equivalent circuit of our inductive coupling system with symbols defined in Table 3.3.



TransmitterImplantFigure 3.14 Equivalent circuit of the inductive coupling resonant circuit

The mutual inductance M between two coil antennas is represented as

$$M = k\sqrt{L_1 \cdot L_2} \tag{3.13}$$

The system efficiency, η_T , is defined as the ratio of output power at the load P_{out} and the input power supplied to the transmitter P_{in} :

$$\eta_{T} = \frac{P_{out}}{P_{in}} = \frac{\frac{V_{out}^{2}}{R_{ACload}\sqrt{2}}}{\frac{V_{in}^{2}}{R_{1}\sqrt{2}}} = \left(\frac{V_{out}}{V_{in}}\right)^{2} \frac{R_{1}}{R_{ACload}}$$
(3.14)

The relationship between V_{out} and V_{in} as voltage gain can be obtained [3.13 – 3.14] as

$$V_{induced} = -j\omega M \frac{V_{in}}{R_1 + Z_{ref}}$$
$$V_{induced} = -j\omega M \frac{V_{in}}{R_1 + \frac{(\omega M)^2}{R_2 + \frac{(\omega L_2)^2}{R_{ACload}}}}$$

$$V_{out} = \frac{j\omega L_2 V_{induced}}{R_2 + \frac{(\omega L_2)^2}{R_{ACload}}}$$

It is considered that

$$\frac{1}{R_{ACload}} \ll j\omega C_2$$

$$V_{out} = \frac{\omega_0^2 M L_2 V_{in}}{R_1 R_2 + R_1 \frac{(\omega_0 L_2)^2}{R_{ACload}} + (\omega_0 M)^2}$$

The gain can be calculated as follows:

$$A_{V} = \frac{V_{out}}{V_{in}} = \frac{\omega_{0}^{2}ML_{2}}{R_{1}R_{2} + R_{1}\frac{(\omega_{0}L_{2})^{2}}{R_{ACload}} + (\omega_{0}M)^{2}}$$
(3.15)

where $\omega_{_0} = 2 \pi f_{_0}$ and $f_{_0}$ is the resonant frequency.

By The system efficiency η_T is

$$\eta_{T} = \left(\frac{\omega_{0}^{2}ML_{2}}{R_{1}R_{2} + R_{1}\frac{(\omega_{0}L_{2})^{2}}{R_{ACload}} + (\omega_{0}M)^{2}}\right)^{2}\frac{R_{1}}{R_{ACload}}$$
(3.16)

 R_1 , R_2 , L_2 , and M has been derived as follows [3.15 – 3.16].

$$L_1 = N_1^2 L_{10} \qquad \qquad L_2 = N_2^2 L_{20} \qquad (3.17)$$

$$R_1 = N_1 R_{10} \qquad \qquad R_2 = N_2 R_{20} \tag{3.18}$$

The total magnetic field along the axis (B) is

$$B = \frac{\mu_0 N_1 r_1^2}{2(r_1^2 + x^2)^{\frac{3}{2}}} I$$

Hence the magnetic flux (φ_2) through the implant loop is

$$\phi_2 = B.Area(implant) = BN_2\pi r_2^2 = \frac{\mu_0\pi N_1 N_2 r_1^2 r_2^2}{2(r_1^2 + x^2)^{\frac{3}{2}}}I$$

And since $\phi_2 = MI$

$$M = \frac{\mu \pi N_1 N_2 r_2^2 r_1^2}{2 \sqrt{\left(r_1^2 + x^2\right)^3}}$$
(3.19)

The system efficiency, η , in terms of the physical parameters have been evinced in equation 3.20.

$$\eta_{T} = \left(\frac{\omega_{0}^{2} \frac{\mu \pi N_{1} N_{2} r_{2}^{2} r_{1}^{2}}{2 \sqrt{\left(r_{1}^{2} + x^{2}\right)^{3}}} N_{2}^{2} L_{20}}{N_{1} R_{10} N_{2} R_{20} + N_{1} R_{10} \frac{\left(\omega_{0} N_{2}^{2} L_{20}\right)^{2}}{R_{ACload}} + \left(\omega_{0} \frac{\mu \pi N_{1} N_{2} r_{2}^{2} r_{1}^{2}}{2 \sqrt{\left(r_{1}^{2} + x^{2}\right)^{3}}}\right)^{2}}\right)^{2} \frac{N_{1} R_{10}}{R_{ACload}} \quad (3.20)$$

Since the total transmitter efficiency involves both the class-E amplifier and the transmitter antenna hence to find the total efficiency of the transmitter block both the efficiencies need to be combined together and can be represented as follows. Total efficiency of the circuit derived from equation (3.12) and (3.20) is:

$$(\eta) = \eta_E \times \eta_T$$

Hence from equation (3.12) and (3.20) after incorporating the value of η_E and η_T we obtained the total efficiency η as shown in equation (3.21)

$$\eta = \begin{pmatrix} \frac{1}{R_{DS}(2-D) + R_{CE}(D-1) + R_{LE}} + & & \\ \frac{1}{V_{dc}^{2}} + & & \\ \pi(1-D)^{2} \begin{bmatrix} 2\pi \begin{cases} DR_{DS} + (1-D)R_{CE} \\ +R_{1} + R_{C1} \end{cases} + \cos(2\pi D + 2\phi)\sin(2\pi D) \\ \frac{R_{C1} - R_{DS}}{(R_{C1} - R_{DS})} \end{bmatrix} \end{pmatrix}^{2} \\ \begin{pmatrix} \frac{\omega_{0}^{2} \frac{\mu\pi N_{1}N_{2}r_{2}^{2}r_{1}^{2}}{2\sqrt{(r_{1}^{2} + x^{2})^{3}}} N_{2}^{2}L_{20} \\ \frac{\omega_{0}^{2} \frac{\mu\pi N_{1}N_{2}r_{2}^{2}r_{1}^{2}}{2\sqrt{(r_{1}^{2} + x^{2})^{3}}} N_{2}^{2}L_{20} \\ \frac{N_{1}R_{10}N_{2}R_{20} + N_{1}R_{10} \frac{(\omega_{0}N_{2}^{2}L_{20})^{2}}{R_{ACload}} + \left(\omega_{0} \frac{\mu\pi N_{1}N_{2}r_{2}^{2}r_{1}^{2}}{2\sqrt{(r_{1}^{2} + x^{2})^{3}}}\right)^{2} \end{bmatrix}^{2} \frac{N_{1}R_{10}}{R_{ACload}} \end{pmatrix}$$
(3.21)

Since we obtained $\eta_{\scriptscriptstyle E}$ as 67% (0.67) hence η is:

$$\eta = 0.67 \times \left(\frac{\omega_0^2 \frac{\mu \pi N_1 N_2 r_2^2 r_1^2}{2\sqrt{(r_1^2 + x^2)^3}} N_2^2 L_{20}}{N_1 R_{10} N_2 R_{20} + N_1 R_{10} \frac{(\omega_0 N_2^2 L_{20})^2}{R_{ACload}} + \left(\omega_0 \frac{\mu \pi N_1 N_2 r_2^2 r_1^2}{2\sqrt{(r_1^2 + x^2)^3}} \right)^2 \right)^2 \frac{N_1 R_{10}}{R_{ACload}}$$
(3.22)

In optimization of the efficiency, several practical limitations in the implant size were considered. Increasing the turn number N_2 and the radius of the coil r_2 will inherently increase the implant size. This thus limits the coil inductance L_2 . Increasing the tuning capacitor C_2 however also present a size increase due to the physical size of the capacitor. After iterations of trade-off between the size and output power P_{out} , a turn number N_2 of 15 was selected with the AWG-24 copper wire considering suitable size of the implant.



Figure 3.15 Transmitter efficiency variation with transmitter radius (a) 4cm, (b) 5cm, (c) 6 cm and (d) 7cm.

A MATLAB was written to detect the theoretical plot and was demonstrated in Appendix D has been written. It followed the theoretical efficiency calculation for various turns and radii of the transmitter while the implant parameters were kept constant as 15 turns and 1.03 cm radius in the MATLAB program. To verify the theoretical plot experiments were done where the size of the implant coil antenna was determined by the maximum size allowed for future endoscopic implantation procedure. The cross section of the implant coil was 3.33 cm² which is calculated from the area contemplated as 35 mm ×10 mm. This was equivalent to a coil radius of 1.03 cm when a circular coil is considered for convenience instead of a rectangular one. A single turn circular shape coil antenna with a radius of 1.02 cm was made and measured the respective parameters as R_{20} = 0.118 Ω , L_{20} = 0.178 μ H and quality factor Q=28. The load resistance R_{AC_Load} was used as 800 Ω which replicates more or less the tissue impedance. The N_2 =15 turns of coil antenna was made wrapping around the implant printed circuit board. The distance between the transmitter coil antenna and the implant was kept at the distance of 6 cm.

$$P_{output} = \frac{(V_{rms})^2}{R} = \frac{(V_{PP})^2}{2R}$$
(3.23)

During the experiments the output peak-to-peak voltage at the secondary coil side was measured with a load of 800 Ω . This voltage was considered for numerous (transmitter coil diameters as 4 cm, 5 cm, 6 cm and 7 cm. Forty eight different combinations were made by varying the number of turns from 1 to 12 respectively for each radius. The maximum output voltage response for each radius at a particular turn number. The root mean square (RMS) value of the voltage was calculated and the output power was calculated as shown in the equation (3.23) after obtaining the maximum voltage.

$$\eta = \frac{P_{output}}{P_{input}} = \frac{\left(V_{pp}\right)^2}{2RP_{input}}$$
(3.24)

The efficiency was calculated as demonstrated in equation (3.24) by dividing the calculated output power from the measurement by the input power supplied from the power

supply. The input power was 4 W and it was kept constant during all the experiments. Fig.3.15 shows the comparison between the simulated theoretical plots obtained from final efficiency equation (3.22) and the values obtained by measurement efficiency from equation (3.23).

3.3 Discussion

For designing the main transmitter circuitry the experimental plot was followed which suggested that 12 turns for a 6 cm radius are the best parameters for that produced maximum voltage amongst all the experiments done with the different transmitter size and radius combinations. Increasing the number of turns more than 12 increased the efficiency as we observed from the theoretical calculation. But it did not increase the efficiency as significantly as compared to the aggravating bulkiness due to the increment of the number of turns. It was calculated that the efficiency keeps on increasing very gradually till the number of turns reaches around 600. The intention was to keep the transmitter smaller and less bulky so that it could be worn by the patient as a belt.

CHAPTER 4

BENCHTOP EXPERIMENTS

4.1 Experiments with BWPS Circuit

Benchtop experiments were performed for verifying the performance of the wireless stimulator devices before testing them on animals. Human body consists of plenty of salts which may provide the risk of signal attenuation by a significant amount. The experiment also illustrated the attenuation of the wireless signal with the distance, till which the stimulator can provide pulses effectively. The NaCl solution replicated the true human body condition. 0.9% NaCl solution has been used in many prior works for such benchtop *in vivo* experiments [4.1]. 2% NaCl solution was used instead of 0.9% considering the extreme worst case scenario. When the patients have very high blood pressure the salt content in the body increases and therefore we have considered extreme conditions. The distance vs. attenuation results have been illustrated in the chapter 5.



Figure 4.1 Attenuation experiment with the salt solution

Therefore it is very important to observe that how the wireless stimulator device behaves in the salt solution. Benchtop experiments are conducted with 2% NaCl solution. The implant was hung in the water completely inserted in it with a water secure tape inside a transparent container while the transmitter antenna was fixed to the outside wall of the container with the antenna coils coaxially oriented. The distance of the transmitter coil from the edge was kept constant on the other hand the distance of the implant inside the water was varied as shown in Fig. 4.1.



Figure 4.2 PDMS insulation testing in extremely (a) Alkaline and (b) Acidic solutions.



Figure 4.3 Output response of the stimulator dipped in salt solution.

The PDMS insulation was tested by dipping each stimulator in extreme acidic and alkaline solutions and by keeping it for 3 Hrs in the solution. This process has been shown in Fig. 4.2. This experiment was done to test the survivability of the devices in the stomach fluids which can be highly acidic and along with foods which can be highly acidic and alkaline both. Alkaline solution was made from KOH and distilled water. The resultant pH was 11.6 and it was tested with a pH meter. For the acidic solution HCL was diluted with distilled water and the pH ramification was measured to be 2.5, it was confirmed with a pH meter. The measurement electrodes were attached to the output of the stimulator and measurements were taken with the National Instrument 6210 DAQ card to authenticate that the devices are capable of withstanding such extreme pH values.

Distance variation between the transmitter and the implant during thebenchtop experiments was done from 2.5 cm, considering the minimum muscle thickness possible in patients. It is observed from the Benchtop experiment shown in Fig. 4.3 that the stimulator devices can operate till more than 8 cm distance from the transmitter. The output voltage across the device was recorded with a load of 800 Ω . It has been seen that the attenuation in the voltage varies in cubical order as the distance was increased gradually.

4.2 Experiments with WRS Circuit

The discharging characteristics with the device set under each setting were analyzed and the curve was observed till the device battery got discharged from 3 V to 1.8 V. The lower limit was set to 1.8 V because after this level the PIC microcontroller that is responsible for pulse generation stops working. Fig. 4.4 shows that the stimulator operating under "High" setting discharges lot faster than the device operating at "Medium" and "Low" setting. It took approximately around 20 hrs for the device operating at "Low" setting to get discharged however it took around 12 hrs when the device was operating at "Medium" settings. It took just 8 hrs to get the device discharged from 3 V to 1.8 V when operated at "High" setting and therefore the device needs to be charged more frequently than when operated in "Medium" and "Low" settings.



Figure 4.4. Discharging characteristics of the stimulators operated at Low, Medium and High settings.



Figure 4.5 Charging characteristic of the WRS devices.

Fig. 4.5 demonstrates the charging characteristics of the WRS devices. The time taken for each device to get recharged is similar since the battery of all the WRS devices have the same rating. It took around 30 mins for recharging the devices from 1.8 V to 3 V with the devices placed at a distance of 8 cm from the transmitter antenna. It was contemplated that the recharging time is very little compared to the discharging time and can be easily achieved by the patients during watching television or doing some other household chores. Thus it is very convenient.

CHAPTER 5

SURGICAL IMPLANTATION

5.1 Surgical Implantation

For animal experiments two different modes of implantations were followed: Surgical implantation and endoscopic implantations. In this chapter surgical implantation has been concentrated upon. Animal tests were performed on porcine model following protocols designed by the Animal Control Board of the University of Mississippi Medical Center (Animal Protocol Number-1265). During each animal experiment, normal healthy pig weighing between 100–110 lbs was anesthetized. The output of different stimulators were connected to the Temporary Transvenous Pacing Lead 6416-200cm one at a time as shown in Fig.5.1.



Figure 5.1 Stimulator leads inserted in the animal experiment.

The two sets of unipolar, temporary, myocardial stainless steel pacing leads, and the leads were attached either serosally or mucosally. Before implantation abdominal hair was removed and the epigastric area was sterilized. After that laparotomy was performed and a ventral midline incision was made between the xyphoid process and umbilicus of the pig. The incision measured around 12-15 cm. Two pairs of pacing wires were implanted on the outer layer of the stomach (Serosa) along the greater curvature and the distance between 2 electrodes in each pair was about 0.5 to 1 cm. An Enterra[®] (Medtronic Inc.) neurostimulator was initially connected to the leads and a session of recording was conducted. Then each of the WRS and BPWS devices were tested, respectively. EGG recordings with all the stimulation settings were made. Once the Serosal recoding was done then the stomach was cut by a very small incision (2 cm). During this procedure the temporary myocardial leads attached to the stimulators were inserted into the inner layer of the stomach (Mucosa).



Figure 5.2 Animal setup during the surgical implantation

While the stomach's myoelectrical activity was recorded with the Electrogastrogram (EGG) device, the electrical pulses were also observed with a Data Acquisition Card (DAQ)

(USB 6210 National Instrument). This was connected in parallel to the output of the implantable stimulators. The tissue impedance was measured from the electrodes with the Enterra Therapy programmer. Fig. 5.2 describes the animal experimental setup when the Enterra Therapy and the stimulators were tested each at a time.



Figure 5.3 Stomach tissue layer structure.

For surgical implantation two animal experiments were carried out. Both the electrical parameters and Electrogastrogram (EGG) parameters were recorded The first experiment mainly helped us to specify that the EGG parameters varied with the stimulations. The second porcine experiment was conducted to establish a relationship between the powers delivered to the tissues and the comparative change in the EGG parameters.

During temporary stimulation with the commercial device since the temporary leads are inserted through the nose and mouth to the stomach hence they are implanted in the inner layer Mucosa thus giving mucosal stimulation. But when the commercial Enterra device is implanted permanently in a patient, the doctors implant it in the Serosa. This is because the Serosal layer is easily accessible after the upper skin layer is cut during the surgery. Fig. 5.3 shows the cross section of the stomach tissues and the internal gastric system layers. During surgical implantation our devices were implanted both on the Serosa and the Mucosa. The detailed description of this structure is provided in section 2.1 of chapter 2. The Mucosal layer is higher on ICC count and therefore stimulating the Mucosa provides more effect for the treatment of Gastroparesis than stimulating the Serosa. Fig. 5.4 defines the various scenarios and conditions under which the stimulator devices were tested during surgical and endoscopic implantation which will be described in chapter 6 in details.



Figure 5.4 Testing conditions of the stimulator.

Experimenting with all the stimulators (WRS and BWPS Low, Medium and High) were implanted on both Mucosa and Serosa were necessary to illustrate a comparative study and effectiveness of our WRS and BWPS stimulators and the commercial device. *In vitro and in vivo* tests were conducted. During the *in vitro* tests the stimulators were placed outside the body with two thin electrodes stimulating the stomach tissues (Mucosa or Serosa). The WRS devices

were operated by a rare-earth magnet while the BWPS devices were operated by the transmitter module. During this process of stimulation the medium between the stimulator and the transmitter is air. *In vivo* conditions represent when the stimulator was placed inside the porcine body with the skin layer closed after implantation. By observing the voltage drop in the *in vivo* condition with respect to the *in vitro* condition, it is easier to calculate the attenuation coefficient of the tissues. It also helped in analyzing the voltage drop that would take place as the muscle thickness of the patient will increase.

5.2 Porcine Experiments

5.2.1. Porcine Model-I



Figure 5.5 EGG signal recorded from the Mucosa.

The amount of electrical current delivered into the tissues was computed from the voltage readings measured from the data acquisition (DAQ) card and the measured DC impedances of tissues. For first Porcine experiment the Mucosal and Serosal EGG recordings were analyzed by signal averaging for mean frequencies and amplitudes, as well as for frequency-to-amplitude ratios (FARs). Fig. 5.5 shows the EGG signal from Mucosal recording. This indicates the frequency and amplitudes of the peaks. The FAR is calculated by dividing the mean frequency by the mean amplitude.

5.2.1.1 Results

Table 5.1 summarizes the delivered currents to the Serosal region by WRS and BWPS stimulators at three different settings (Low, Medium, High) in both *in vivo* and *in vitro* conditions. The non-rechargeable-battery based Enterra[®] device was also tested in the same way at Low and High dose settings *in vitro* and *in vivo* so as to compare results. Results obtained from the Enterra[®] device were compared with those from our WRS and BWRS stimulators. The measured currents delivered to the tissues were obtained from the peak values of the rectangular waveform pulses. Initially the WRS devices were tested on the Serosal tissues. The delivered to the tissues were 1.7 mA, 1.93 mA and 1.93 mA from Low, Medium and High settings respectively.

		5	ereen regier				
WRS							
Dose	Low		Medium		High		
	In vitro In vivo		In vitro	n vitro In vivo		In vivo	
Current	1.70 mA	1.70 mA	1.93 mA	1.93 mA 1.93 mA		1.93 mA	
BWPS							
Dose	Low		Medium		High		
	In vitro	In vivo	In vitro	In vivo	In vitro	In vivo	
Current	2.93 mA	2.26 mA	2.93 mA	2.26 mA	2.93 mA	2.26 mA	
Enterra®							
Dose	Low	I			High		
	In vitro	In vivo			In vitro	In vivo	
Current	5 mA*	5 mA*			8.9 mA	8.9 mA	

Table 5.1 Currents Delivered by the Stimulators at Low, Medium and High Dose Settings in the Serosa Region.

*: The Enterra[®] device behaves as a current source when used at the pre-set Low setting. The device works as a voltage source, like our stimulators, when used at a High setting.

For WRS device in both *in vitro* and *in vivo* conditions, the currents were measured to be the same value. Since the pulses were generated by battery, the currents stayed the same

as long as the tissue impedances were the same. The similar experiments then were repeated for Mucosal stimulation for WRS and BWPS stimulators, at every settings of the stimulators, *in vivo* and *in vitro*, along with the Enterra[®] device at Low and High settings. Mucosal stimulation results are shown in Table 5.2.

			-				
WRS							
Dose	Low		Medium		High		
	In vitro In vivo		In vitro	In vitro In vivo		In vivo	
Current	3.45 mA	3.45 mA	3.63 mA	3.63 mA	3.63 mA	3.63 mA	
BWPS							
Dose	Low		Medium		High		
	In vitro	In vivo	In vitro	In vivo	In vitro	In vivo	
Current	6.56 mA	5 mA	6.56 mA	5 mA	6.56 mA	5 mA	
Enterra®							
Dose	Low				High		
	In vitro	In vivo			In vitro	In vivo	
Current	5 mA*	5 mA*			17.7 mA	17.7 mA	

Table 5.2 Currents delivered by the Stimulators at Low, Medium and High Dose Settings in the Mucosa Region.

*: The Enterra[®] device behaves as a current source when used at the pre-set Low setting. The device works as a voltage source, like our stimulators, when used at a High setting.

The currents delivered to the Serosal areas were lower than those to the Mucosal areas. This was due to the 1179 Ω and 594 Ω impedances of the Serosal and Mucosal tissues, respectively. The currents were inversely proportional to the impedances, since our stimulators generated pulses based on voltage source.

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Stimulator	Setting	Rhythm	Mean Freq. (bpm)	Freq. Range (bpm)	Wave Amp. (V)	Mean Amp. (V)	Amp. Range (V)	FAR (bpm/V)	NOTE
Control	Off	RR	3.67	3.0-4.0	EA	0.16	0.15- 0.18	22.94	
Enterra [®] Low	Low	RR	3.67	3.0-4.0	EA	0.18	0.18- 0.19	20.39	
WRS Low	Low	RR	3.50	3.0-4.0	EA	0.20	0.20- 0.21	17.50	tion
WRS Medium	Medium	RR	3.08	3.0-3.3	EA	0.17	0.15- 0.20	18.10	Stimula
WRS High	High	RR	3.75	3.5-4.0	EA	0.14	0.10- 0.18	26.80	rosal S
BWPS Low	Low	EGG was	s not rec	orded					Se
BWPS Medium	Medium	RR	3.08	3.0-3.3	EA	0.07	0.05- 0.09	44.00	
BWPS High	High	RR	3.00	3.0-3.0	EA	0.14	0.12- 0.15	21.43	
Control	Off	RR	3.00	3.0-3.0	EA	0.09	0.08- 0.10	33.33	c
BWPS Low	Low	EGG was	GG was not recorded						
BWPS Medium	Medium	RR	3.00	3.0-3.0	EA	0.06	0.06- 0.07	50.00	sal Stir
BWPS High	High	RR	3.83	3.0-5.0	EA	0.1	0.08- 0.13	38.30	Muco

Table 5.3 EGG Signal Summary

(bpm: beat per minute; RR: regular rhythm; EA: equal amplitude; Freq.: frequency; Amp.: amplitude; FAR: frequency amplitude ratio)

The delivered currents for BWPS under *in vitro* and *in vivo* conditions differed due to attenuation of the electromagnetic waves, as they penetrated through the tissues from the transmitter to the implant. In the *in vivo* experiment, the pig muscle thickness was estimated to be 6 cm. To assist comparison with the *in vivo* case, the distance between transmitter and implant for the *in vitro* experiments was also kept at 6 cm in air. The attenuation of radio frequency (RF) energy reduced the amount of electrical currents that could be harvested for

real-time generation of stimulation pulses. The currents were reduced to 77% from 2.93 mA to 2.26 mA, and to 76% from 6.56 mA to 5 mA for Serosal and Mucosal stimulations, respectively. Peak transmitted RF power is estimated to be proportional to I^2R where *I* is the current and *R* is the impedance. Since the distance between the transmitter and the BWPS stimulator in air was kept the same as the thickness of the tissue, the attenuation of electromagnetic fields with distance was then normalized. A 41% (3.8 dB) power loss occurred in the pig tissues (6cm thick).



(a)

(b)

Figure 5.6 Implantation procedure of (a) Enterra[®] neurostimulator in the human body for GES and (b) During the tests for WRS and BWPS stimulators.

Table 5.3 shows the summary for recorded EGG signals for the first Pig experiment in various conditions. Fig. 5.3 shows the surgical implantation and testing of the commercial stimulator and our wireless stimulator. EGG measurements were not obtained to compare abnormal and normal signals, but to confirm stimulation effects on such gastric activities as the rhythm (regular or irregular), mean frequency, frequency range, amplitude (equal or unequal), mean amplitude, and amplitude range of the EGG signals.

Control measurements were obtained with the stimulators turned off. During this period no stimulations are given to the pig. Frequencies and mean amplitudes of the EGG signals did vary at different stimulation settings. Moreover, frequency to amplitude ratios (FARs) indicated significant changes in gastric activities during stimulation as compared to the control measurements. The FAR change was not very significantly high between the Low and the Medium settings of the stimulation. However the FAR change was drastic when the settings were switched from Medium to High, during both Mucosal and Serosal stimulation as it can be observed from Table 5.3. In clinical practice, gastric electric stimulation effects on solid/liquid emptying, as determined by scintigraphic measurement, are usually evaluated in GP patients for a period of 2–4 weeks. Our studies, however, were performed on a healthy pig under anesthesia so as to demonstrate the feasibility of our systems.

5.2.1.2 Discussion

In both Tables 5.1 and 5.2, the current supplied by the WRS device designed at the Low setting was lower than that supplied by the Medium or High settings. This difference was not due to its setting, since the PICs were set in all three devices with the same output amplitude of pulses, although they had different frequencies and duty cycles. Instead, the difference was likely owing to the fact that, even though all the rechargeable lithium batteries used were rated at 3 V when purchased, not all of them showed an exact 3 V output when fully charged. After the animal experiments, we tested and found that the maximum voltage of the battery used in the Low-setting WRS device was 2.7 V when fully charged instead of 3 V. In the Serosal stimulation, the 2.7 V battery, which was at the 90% level of a 3 V battery, produced a proportional output current of 1.7 mA (that is, 90% of the 1.93 mA current generated by a 3 V battery). Findings at Mucosal stimulation indicated a similar situation. Thus, our studies identified a possible issue with effects posed by the voltage level of lithium batteries. One possible solution would be including a voltage limiter at the output of a higher-voltage rechargeable battery; only a portion of available voltage could then be used for stimulation, so

that currents would be held at a constant level until the battery voltage would fall out of range and requires charging again.

Stimula tor	In vitro/ in vivo	I (mA)	P _p (mW)	P _c (mW)	Ε _ρ (μJ)	Ε _c (μJ)	P _{av_p} (μW)	P _{av_on} (μW)	P _{av_c} (µW)	
Enterra [®] at Low	Both	5.00	29.48	58.95	9.73	19.45	135.60	194.54	3.81	
WRS Low	Both	1.70	3.41	6.81	1.12	2.25	15.68	22.49	0.44	(C)
WRS Medium	Both	1.93	4.39	122.97	1.45	40.58	40.22	40.58	8.12	=1179
WRS High	Both	1.93	4.39	948.60	1.45	313.04	78.21	78.26	62.61	sa (R⁼
BWPS	In vitro	2.93	10.12	20.24	3.34	6.68	46.57	66.80	1.31	eros
Low	In vivo	2.26	6.02	12.04	1.99	3.97	27.70	39.74	0.78	Li
BWPS	In vitro	2.93	10.12	283.40	3.34	93.52	92.70	93.52	18.70	nted
Medium	In vivo	2.26	6.02	168.61	1.99	55.64	55.15	55.64	11.13	npla
BWPS	In vitro	2.93	10.12	2186.26	3.34	721.47	180.26	180.37	144.29	-
High	In vivo	2.26	6.02	1300.72	1.99	429.24	107.24	107.31	85.85	
Enterra [®] at Low	Both	5.00	14.85	29.70	4.90	9.80	68.32	98.01	1.92	
WRS Low	Both	3.45	7.07	14.14	2.33	4.67	32.53	46.66	0.91	94 Ω)
WRS Medium	Both	3.63	7.83	219.16	2.58	72.32	71.69	72.32	14.46	(R=5(
WRS High	Both	3.63	7.83	1690.65	2.58	557.91	139.39	139.48	111.58	lcosa
BWPS	In vitro	6.56	25.56	51.12	8.44	16.87	117.60	168.71	3.31	ער ע Mר
Low	In vivo	5.00	14.85	29.70	4.90	9.80	68.32	98.01	1.92	ted i
BWPS	In vitro	6.56	25.56	715.73	8.44	236.19	234.12	236.19	47.24	plan
Medium	In vivo	5.00	14.85	415.80	4.90	137.21	136.01	137.21	27.44	<u></u>
BWPS	In vitro	6.56	25.56	5521.38	8.44	1822.06	455.23	455.51	364.41	
High	In vivo	5.00	14.85	3207.60	4.90	1058.51	264.46	264.63	211.70	

Table 5.4 Energies and Powers Deliveries by different Stimulators to Serosa and Mucosa
Typically, in patients it takes at least 4 hours for EGG signals to stabilize before significant changes in them can be observed that may indicate improved stomach motility. Our animal experiments did not permit a wait of four hours for each measurement. However, significant EGG signal changes indicated that myoelectrical activities were modulated by our stimulators, and that the modulation varied with the stimulation parameters.

The stimulation energies and powers of the WRS and BWPS devices were calculated from the measured parameters and compared with the battery-based Enterra[®] device. The power per stimulation pulse $P_p = I^2 \times R$, the power per cycle $P_c = I^2 \times R \times N$, the energy delivered per pulse $E_p = P_p \times T_p$ and the energy delivered per cycle $E_c = E_p \times N$ can be calculated from the delivered current *I*. The average power for each pulse can be calculated as the energy for each pulse divided by the pulse period:

$$P_{av_p} = \frac{E_p}{T_p + T_o}$$
(5.1)

The average power delivered to the tissues is the measured total energy during the stimulation "ON" time. The average power during stimulation period indicates the intensity of stimulation.

$$P_{av_on} = \frac{E_c}{T_{on}}$$
(5.2)

The average power for the whole cycle is

$$P_{av_c} = \frac{E_c}{T_{on} + T_{off}}$$
(5.3)

Table 5.4 summarizes the results. At the Low setting, Enterra[®] behaved as a current source independent of tissue impedance and delivered a current of 5 mA to both Serosa and Mucosa. The average powers during pulse delivery to the Serosa and Mucosa tissues were 194.54 μ W and 98.01 μ W, respectively. The powers and energy for the Enterra[®] was specified



Figure 5.7 Average power delivered during the "on" time in the cycle to the (a) Serosa and (b) Mucosa by different stimulators.

in the Low setting because this is generally used for patients in medical purposes. For high impedance tissues in the Serosa, both WRS and BWPS devices delivered lower average power intensities during stimulation compared to Enterra[®], as shown in Figure 5.7(a). However, the FARs (17.50, 18.10, 26.80, 44.00, 21.43) of these devices, despite their delivery of lower powers, did vary compared to that (22.94) of the control case, just as the FAR (20.39) at the low setting in Enterra[®] also varied. The variations of FAR were not in a linear or semi-linear relationship with the delivered average power intensity during stimulation. As mentioned, since the stomach of pig is healthy so as it is impossible to observe the changes for FARs from

abnormal to normal conditions, it is difficult to find a direct relationship from the stimulation powers to the EGG FAR however change in FAR takes place due to artificial pulses delivered to the tissues. Further investigation in stimulation parameters and tissue characteristics of course is needed to address the variations.



Figure 5.8 The stimulator compared with the neurostimulator Enterra[®]. Both WRS and BWPS stimulators have the same size. The thickness of our stimulators is 9mm.

As seen in Fig. 5.7(b), for the low-impedance Mucosal tissues, Low and Medium settings of WRS delivered lower average power intensities, but the High setting delivered a higher one. While all three settings possess the same amount of current, the High setting one delivered 216 pulses during the stimulation time, compared to only 2 or 28 pulses for the Low and Medium settings, respectively. The BWPS devices delivered higher average power intensities during stimulation for both *in vitro* and *in vivo* conditions. The powers were lower for the *in vivo* condition compared to the *in vitro* one due to the tissue attenuation of electromagnetic energy. Even when penetrating through a 6 cm thick tissue, the BWPS device delivered power intensities that were higher than those by the battery-based Enterra[®]. The

currents in all three settings of BWPS were the same; however, a higher frequency of pulses during stimulation permitted more power intensities to be achieved. These findings indicate that high power intensity stimulation is possible with our batteryless, wirelessly-powered stimulators, despite electromagnetic energy losses in the tissues when penetrating the body and irrespective of the physical size of the stimulators.

Fig. 5.8 shows the form factors of the commercial Enterra device and the first generation (Gen-1) wireless stimulator. This first design of wireless stimulator (both WRS and BWPS) measures 37 mm \times 12 mm \times 9 mm whereas the commercial stimulator measures 55 mm \times 60 mm \times 10mm.

5.2.2. Porcine Model-II

While designing the second generation (Gen-2) wireless stimulator, it was designed as such so that it fits the form factor requirement for endoscopic implantation. The device size was reduced further. The devices were tested first in the second porcine experiment with surgical implantation procedure, before exploring endoscopic implantation technique. That ensured that whether the Gen-2 stimulator can deliver sufficient amount of power to the tissues to control the EGG signals after size reduction.

5.2.2.1 Results

In the second porcine experiment the EGG parameters were recorded similar to the first porcine experiment and the voltage readings were also noted with DAQ card as discussed in section 5.1.1 of this chapter. Table 5.5 shows the voltage and current value for each implant during stimulation of the Mucosal and Serosal region. *In vitro* tests were conducted for this experiment to illustrate the change in the EGG signals by stimulator voltage delivery. The electrodes were inserted in the region between the anterior wall and the greater curvature of the stomach. The Mucosa and Serosa tissue impedance was measured to be 707 Ω and 820 Ω

respectively once the electrodes were inserted into the specified positions. The impedance measurement was done with the accessory kit of Enterra Therapy device similar to the prior porcine experiment.

From Table 5.5 it is evident that BWPS stimulators could stimulate the stomach tissues by higher current pulses than the Enterra setting Low while stimulating the Mucosa. EGG result summary is expressed in Table 5.6 for the second porcine experiment and the FAR rations are calculated. This ratio is compared to the power and energy delivered per pulse and cycle in the tissues demonstrated in Table 5.5 and Fig. 5.9. This table also shows the various energy and power delivered by different stimulators, per pulse and per cycle.

Stimulator	Electrodes	Impedance	Current
		(Ω)	(mA)
Baseline		820	0
WRS Low	Ę	820	2.8
WRS Medium	tatio	820	2.8
WRS High	lant	820	2.8
BWPS Low		820	8.0
BWPS Medium	sal	820	6.5
BWPS High	Serc	820	6.5
Enterra Low	•,	820	5.0
Baseline		820	0
WRS Low	tion	707	3.8
WRS Medium	anta	707	3.8
WRS High	hpla	707	3.8
BWPS Low	all	707	9.0
BWPS Medium	ICOS	707	6.7
BWPS High	ML	707	6.7
Enterra Low		707	5.0

Table 5.5 Voltage and Current Characteristics for Second Animal Experiment.

The experiments were done first for the Serosa and then on the Mucosa. For both the regions baseline reading was recorded before stimulating the muscles. The order from lower to higher settings was followed during stimulation so that the muscles get accustomed to the stimulation pulses gradually without getting a sudden shock. Once all the WRS devices were tested and their respective FARs were recorded, then BWPS devices were tested in the same manner. The comparison of the FAR was done with the baseline reading.

Type of Stim.	odes ed	Power Delivered	Power Delivered	Energy Delivered	Energy Delivered	FAR
	Electr Plac	per Pulse (mW)	per Cycle (mW)	per Cycle (uJ)	per Pulse (uJ)	(bpm/V)
Baseline						27.24
WRS Low	c	6.43	12.86	2.12	4.24	11.36
WRS Medium	tatio	6.43	180.01	2.12	59.40	10.75
WRS High	olant	6.43	1388.62	2.12	458.24	5.36
BWPS Low	l m	52.48	104.96	17.32	34.64	13.15
BWPS Medium	osa	34.65	970.06	11.43	320.12	6.90
BWPS High	Ser	34.65	7483.32	11.43	2469.50	5.67
Enterra Low		20.50	41.00	6.77	13.53	4.18
Baseline	_					14.89
WRS Low	tion	10.32	20.63	3.40	6.81	13.63
WRS Medium	inta	10.32	288.87	3.40	95.33	25.00
WRS High	npla	10.32	2228.43	3.40	735.38	12.34
BWPS Low	al Ir	57.27	114.53	18.90	37.80	12.06
BWPS Medium	soor	31.74	888.64	10.47	293.25	10.34
BWPS High	M	31.74	6855.24	10.47	2262.23	10.23
Enterra		17.68	35.35	5.83	11.67	12.68

Table 5.6. EGG Signal Summary and Energy Calculation.



Figure 5.9 Variation of FAR ratio with power delivered per cycle for (a) Serosal stimulation and (b) Mucosal Stimulation

The propagation constant for the stomach tissue

$$V_0 = V_x e^{\gamma x} \tag{5.4}$$

Table 5.7 represents the output voltage response and the attenuation when the BWPS stimulators were placed inside the stomach and the transmitter-implant separation was varied along with the wireless energy travelling medium. The electrodes were implanted in the Mucosa region. During this process the implant was fixed to the stomach through an incision and the incision was closed after the implantation. The transmitter was fixed on the skin.

Table 5.7 Variation of Output Voltage with the Distance between the Stimulator and Transmitter in Different Mediums.

Device	Impedance	Air Thickness	Muscle Thickness	Stomach Tissue Thickness	Output Voltage	Voltage at the receiving end
	(Ω)	(cm)	(cm)	(cm)	(v)	(V)
BWPS	707	3	0	0	4	0.431
BWPS	707	3	0	4	3.2	0.379
BWPS	707	0	3	4	2.2	0.181

The measurement data was obtained by placing the implant 3 cm deep into the skin and was kept open while the output voltage at the output of the microcontroller and current were recorded. The skin thickness measured approximately 3 cm. As the skin incision was left open hence the transmission medium that was considered for calculation was air. The voltage obtained at the output was 4 V. The stimulator antenna voltage was calculated from the output voltage by Benchtop experiment mimicking the scenario of the animal experiment shown in Table 5.7. The output voltage is the voltage after the amplification of the antenna voltage considering all the power drops and amplification across the discrete components, hence the relationship is non-linear. Therefore for exact calculation of the attenuation voltage, the calculation has to be performed by the antenna voltage. This is the voltage measured at the receiving antenna before the charge pump and not the output voltage used for stimulating the tissues. In the benchtop experiment the case was observed by varying the transmitter and implant distance and observing the output of the microcontroller as 4 V, 3.2 V and 2.2 V. At that instant the voltage at the legs of the implant antenna was measured for the calculation of attenuation coefficient.

For the second series of data the implant was shifted down to the other end of the stomach (measuring 4 cm in thickness) and therefore the medium of energy transmission was considered as 3 cm of air and 4 cm of stomach tissue (x_{i}) . The voltage recorded was 3.2 V with the BWPS device. The 0.8 V drop in the voltage was due to the 4 cm of stomach tissue. The attenuation constant of the stomach tissues was calculated from equation (5.4). V_0 represents the initial voltage when the stimulator was not placed on the other side of the stomach and this is the voltage at the receiving end of the stimulator antenna which is 0.431 V as shown in Table 5.7. The propagation constant for stomach tissue was calculated as 0.0321.

In the third case the skin incision was closed and the transmitter was fixed on the closed skin and the voltage was recorded in the similar way as the previous cases. This time the medium of energy transmission was more like 3 cm of skin along with fat tissue and 4 cm of stomach thickness. The voltage dropped to 2.2 V. This drop from the previous case was due to the skin thickness considered. In this case the muscle attenuation was calculated to be 0.698 which is much higher than the stomach tissues. This is due to the fat content in the belly muscles. As we observed that our stimulator can be implanted even much deeper than 7 cm into the tissues.

5.2.2.2 Discussion

When the devices were in the Serosa region the currents delivered were less than when the devices were tested on the Musosa region. This is due to the fact that serosal impedance was much higher than the mucosal impedance and hence less current could be delivered to the Serosa tissues. BWPS-Low produced more current compared to Medium and High as anticipated from Table 5.5. The fact being that the Low setting contained couple of more turns in the receiver antenna compared to the Medium and High settings. This is due to fabrication anomaly. It also increased the size of the device and added 1 mm more to the thickness. Since the goal was to keep the size as small as possible hence for the other two designs (BWPS Medium and High setting) the number of turns was reduced so that the thickness will fit the yardstick of our measurement for endoscopic implantation in future. The turn number was decreased to15 instead of 17.

Power delivered per pulse was calculated from the output current and impedance. While the power delivered per cycle was calculated by multiplying the power delivered per pulse with the number of pulses as discussed in equations (5.1) and (5.3). These calculations were done to illustrate and analyze the relationship of the FAR to the power delivered to the tissues. It has been observed that the decrement of FAR is proportional to the increment in power delivered per cycle with each setting of stimulation.

This trend of decrement was noticed for all the BWPS devices when placed in both Mucosa and Serosa and also for the WRS devices placed in Serosa. While stimulating the Mucosa with WRS stimulator operating at Medium setting the FAR was very high. The specific reason for this deviation could not be predicted. It may be the result of post effect of the previous stimulation or sudden movement of the animal or any other external factor. It might be also be a resultant of any motion artifact or sudden increase in respiration rate. EGG signals are very susceptible to motion artifacts [5.1]. The motion artifacts does not stay for very long time and the signal gets back to the real state however due to lack of time the long term monitoring could not be done.

There was a waiting time for 2 mins before the next device was implanted. Ideally the waiting time has to be much longer before the stomach completely retains its normal myoelectrical activity and behaves independently of any stimulation. The waiting time was kept short so that all the devices could be tested on a single pig under the same conditions. The waiting time did not give the stomach enough time to retain the normal activity hence the myoelectrical activity change due to the next implantable stimulator could not be recorded in a

completely independent way from the previous stimulation. But even after all the unfavorable timing gaps, still the FAR emphasized that significant change has taken place in the gastric myoelectrical activity due to the stimulation and this change is dependent to the power delivered per cycle.



Figure 5.10 Form factor comparisons of Gen-2 wireless stimulator and the commercial Enterra device

Fig. 5 shows the form factor comparison of our device with the tradition stimulator (Enterra Therapy model). The Enterra Therapy device measures 60 mm X 55 mm X 10 mm while our devices measured 9 mm x 37 mm x 8 mm after the device size reduction from Gan-1 device. Our device is less than one tenth of the size of the traditional stimulator and hence much more comfortable in implanting. This made the device suitable for endoscopic implantation as discussed in earlier sections so that the patients do not have to undergo any surgery at all. In real life application the PDMS coating has to be replaced by a much rigid epoxy so that it can withstand the acidic juices of the stomach for a longer time.

CHAPTER 6

ENDOSCOPIC IMPLANTATION

Implanting a gastric stimulator by endoscopic implantation can be a boon to the patients who are suffering from chronic Gastroparesis due to Diabetes and Cancer. It can completely eradicate the surgical implantation hassle. This non-survival endoscopic study protocol, approved by the Institutional Animal Care and Use Committee of the University of Mississippi Medical Center (UMC), was conducted at its Animal Endosuite. This chapter describes the various endoscopic implantation procedures. These procedures have been done for both the temporary stimulation and the permanent stimulation methods.

6.1 Resolution Clip Method

The gen-2 device was tested on a 5-months-old male pig, approximately 100 lb with a cardiac rate of approximately 115 bpm and a respiratory rate of 12 bpm. The ear electrodes of Gen-2 device were initially designed so that it can be hooked up to the Mucosal tissue.

6.1.1 Method

The Resolution Clip (Boston Scientific, MA) are often use for clip placement within the Gastrointestinal (GI) tract for controlling mucosal/sub-mucosal defects or bleeding ulcers. It can be opened and closed up to five times prior to deployment. It has got a jaw span of 11 mm which is intended to grasp ample amount of Mucosal tissue. The handle is designed to actuate opening and closing of the jaws. This resolution clip model is compatible with Endoscopic Gastroscopes and can be inserted through working channels of the endoscope.

Fig. 6.1 shows the delivery procedure of the wireless stimulator with the endoscope. A suture was threaded to the one of the ear electrodes of the device and then hooked with the delivery device inserted through the working channel of the Endoscope. The endoscope is then tucked in to the stomach through the esophagus. Once the endoscope is inserted into the stomach the thread is released and hence the device falls into the stomach. The resolution clip is then inserted through on of the endoscopic channels as shown in Fig. 6.2(a). The resolution clip was opened by actuating from outside and it was made to grasp one of the ear electrode along with some Mucosa tissue as shown in Fig. 6.2(b). The resolution clip is deployed was demonstrated in Fig. 6.2(c) and thus attaching one of the ear electrodes to the Mucosa. Fig. 6.2(d) shows that final step of device attachment after the other ear electrode was also attached to the Mucosa in the same procedure.



Figure 6.1 Device delivery of the wireless stimulator to the stomach with the endoscope.



Figure 6.2 Endoscopic Implantation with resolution clips. (a) Clip inserted through working channel of the endoscope, (b) Resolution grasps the ear electrode and the Mucosal tissue, (c) Resolution clip deployed and (d) Device attached to the Mucosa with both the ear electrodes attached to the Mucosa.

6.1.2 Results and Discussions

No electrical or EGG data was recorded during the procedure to ensure the stimulation. However this process demonstrated the first successful placement of the wireless device without a surgery. In later sections several new endoscopic procedures has been demonstrated. The procedure of endoscopic implantation of the device with resolution clip method was very easy without damaging the Mucosa at all. The process was very quick. It took less than 10 mins to complete procedure to attach both the ends of the wireless stimulator to the gastric Mucosa. This procedure is the first endoscopic implantation of the wireless stimulator. It has been observed in many previous experiments of medical history that resolution clips can stay attached to the tissue for 1 month–6 months. Therefore this procedure of endoscopic implantation can be used for patients who do not need stimulation for long term and can be treated with temporary implantation procedure.

6.2 Pinning Method

The experiment was performed on a 7-month-old male pigs, approximately 110 lb each, with a cardiac rate approximately 110 cycles/min and a respiratory rate of 12 cycles/min.





Figure 6.3 Block diagram for endoscopic implantation with pinning method.

A diagnostic Gastroscope (Olympus GIF-Q160, Olympus America, Center Valley, PA) was used for endoscopic visualization and device attachment. Fig. 6.3 shows the block diagram of the wireless stimulator device after implantation. A new wireless device was fabricated similar to the dimensions and specifications of the Gen-2 devices except that the two ears were not

electrodes. These ears were fixtures that can be used for attachment of the device and were not connected either to the ground or the output of the microcontroller. However two small wire electrodes protruded from each side were introduced for stimulating the Mucosa unlike the previous version of Gen-2. This new version of Gen-2 device was named as "Gen-2-New" and it has been shown in Fig. 6.4.

For implanting it an Overtube (US Endoscopy, Mentor, OH) was gently pushed into the esophagus over the endoscope for subsequent, endoscopic esophageal reintubation (Fig. 6.5(a)). After endoscopic examination of the stomach, the miniature stimulator was easily deployed through the Overtube into the esophagus, and pushed by endoscope into the stomach. The wire electrodes (Fig. 6.5 (b)) were pinned into the gastric mucosa at the distal body with an endoscopic rat tooth forceps (Olympus America), and secured by Endoclip (Resolution clip, Boston Scientific, Natick, MA) (Fig. 6.5(b)). The device also is secured by another pair of Endoclips by one clip arm through the electrode's built-in 'ear' loop (not shown in the Figure). A temporary transvenous pacing lead is then inserted through the working channel



Figure 6.4 Gen-2-New wireless stimulator.

for measuring the EGG signal (Fig 6.5(c)). Once the temporary lead is screwed in the Mucosa, the lead is secured with resolution clips so that the temporary lead does not open up during the stimulation procedure (Fig. 6.5 (d)).



Figure 6.5 Endoscopic implantation of wireless Gastrostimulator through endoscopic pinning method (a) Endoscopic esophageal reintubation and deployement of the stimulator to the stomach, (b) Wire electrodes pinned to gastric Mucosa, (c) Temporary transvenous cardiac pacing lead screwed in the Mucosa and (d) Temporary transvenous cardiac pacing lead secured to Mucosa.

Once the cardiac pacing lead is attached to the Mucosa, the outside end is connected to the EGG recorder (Sandhill Scientific, Highlands Ranch, Colorado, USA) and thus the myoelectric signal of the stomach is recorded. The myoelectric signals are quantitatively analyzed by averaging mean frequencies and amplitudes, as well as the FARs of gastric slow waves.

Mucosal DC impedance was measured to be 1751Ω with the Enterra controller. A data acquisition (DAQ) system (DAQ USB 6210, National Instruments, Austin, TX) that interfaced between stimulator electrode wires and the computer was used to measure voltage outputs. The electrical current delivered to the tissues was computed from voltage readings and DC impedance measured.

6.2.2 Results

The EGG recordings were obtained in three steps, as follows. The first reading was recorded with the stimulator turned OFF ('Baseline-1'). The second reading was recorded with the stimulator turned ON as termed 'Stimulation'. The third reading was recorded with the stimulator turned OFF again ('Baseline-2'). Between each reading, an interval of 4 mins without EGG measurement was implemented to ensure separation between the readings. Slow wave frequency was analyzed qualitatively as regular (RR) or irregular rhythm (IRR), amplitudes were analyzed as equal or unequal, during recordings with and without stimulation. The FAR ratio, depicting the stability of slow wave signals, is shown in Table 6.1. A twenty-minute recording was taken for each EGG measurement.

Endoscopic attachment of the miniature stimulator was successfully performed. Measured output voltage of the pulses was 1.9 V indicating an electric current of 1.097 mA delivered to the tissues. Baseline-1 EGG signals showed IRR, with unequal amplitudes as rhythm and amplitude changed during stimulation. The FAR ratio seen during Baseline-1 was lower than the FAR observed during stimulation. Rhythmic activity worsened during Baseline-2, with IRR and unequal amplitudes observed. The FAR, during stimulation was 25.955 bpm/mV, increased to 42.657 bpm/mV.

Castrostinulator through r mining method.								
Devices	Impedance	Rhythm	Frequency	Mean	Amplitude	Amplitude	Mean	FAR
			Range	Frequency	of wave	Range	Amplitude	(bpm/mV)
	(Ω)		(bpm)	(bpm)		(mV)	(mV)	
Baseline1	1751	IRR	2.00-4.00	3.00	UA	0.0625- 0.525	0.1831	16.382
WRS Low	1751	RR	3.00-4.00	3.33	EA	0.1100- 0.1500	0.1283	25.955
Baseline2	1751	IRR	3.00-5.00	3.67	UA	0.0500- 0.1375	0.0858	42.657

Table 6.1. EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through Pinning Method.

The magnet, held at a 3 cm distance from the skin to facilitate the operation of the reed switch (itself in the stimulator situated beneath muscle, fat tissues and the stomach wall) to turn ON the stimulation, resulted in a total thickness from magnet to stimulator of approximately 9 cm.

6.2.3 Discussion

The prototype of a miniature, wirelessly gastric electrical stimulator for endoscopic implantation was successfully tested by pinning method in our experiment. The miniature stimulator was easily attached endoscopically (Fig. 6.5(b)). The device with the wire electrodes pinned into mucosa and 'ears' clipped was very secure. We were able to modulate the myoelectrical activities of the porcine stomach with this endoscopically implanted stimulator.

The miniature stimulator was designed for human stomach, where mucosal impedance varies between 200 Ω –800 Ω . The delivered current measured in the pig model, at 1.9 mA, was lower than the anticipated value of 5 mA value for human. This is also verified by the measured impedance of 1751 Ω in the pigs, higher than that expected for human stomach.

In the Baseline-2 recording, a significant increase in FAR (by 16.702 bpm/mV) was noted, as compared to the FAR during stimulation. Slow wave amplitude was impaired, and at times unequal peaks were observed, during the Baseline-2 recording. Slow wave rhythmic activity had also been irregular prior to stimulation, at the Baseline-1 reading, perhaps in response to the insertion procedure and/or minor scratches in the mucosa occasioned during insertion.

Our results predict the feasibility of an endoscopically implantable, miniature wireless system that can be operated and recharged in patients. Future system will deliver different pulses to stomach tissues according to the severity of gastric anomaly. Future applications in patients will not include such measurement wires. A significantly longer delay between measurements is suggested for obtaining more accurate EGG recordings, independent of prior stimulation effects. Due to the physical condition of the pig, and to permit maximum possible data capture, we limited the time interval between recordings to 4 mins. In future studies, we will obtain more data, but allow longer intervals between experimental observations.

6.3 Precutaneous Endoscopic Gastronomy Method

6.3.1 Method

For this method Gen-2 devices were used for the experiment with two ear electrodes in each end. The block diagram of the miniature stimulator device implanted through Percutaneous Endoscopic Gastrostomy (PEG) is illustrated in Fig. 6.6. A Temporary transvenous cardiac pacing lead was screwed to the Mucosa and secured by by two Endoclips (Fig. 6.7 (a)). PEG needles +/- endoscopic T-tag fasteners were used to punch into the skin and then the stomach of the pig. The stomach was accessed transmurally with PEG and metal wires were inserted into the stomach through these needle chambers as shown in Fig. 6.7 (b) and (c). Once the wire is inserted into the stomach they are pulled outside through the mouth by holding it with a metal hook operated through the endoscopic channel as demonstrated in Fig.

6.7(c) and (d). Another metal wire was inserted in the stomach in the similar way, through the PEG needle at 1 cm apart from the first wire. The metal wire was grasped by the endoscopic hook in a similar way as the previous scenario and pulled out of the mouth (Fig. 6.7 (e)). The insulation of both the metal wires hanging out from the mouth of the porcine model were removed and soldered with the stimulator ear electrodes. Once the device was attached with the wires both the wires were pulled through the skin. The tension created in the wires moved the device into the stomach. By pulling the wires to the maximum possible length through the PEG needle in the skin, the stimulator was fixed to the walls of the Mucosa as shown in Fig. 6.7 (f). The bare metal wires and the ear electrodes touching the tissue gave the required stimulation. Baseline reading with no stimulation was recorded for 10 mins before starting the stimulation process. Each device was used to stimulate the mucosal tissue for 8 mins followed by another baseline for 10 mins.



Figure 6.6 Block diagram for endoscopic implantation with PEG method.





Figure 6.7 Device Implantation by PEG method (a) Temporary transvenous cardiac pacing lead screwed and secured in the Mucosa, (b) PEG endoscopic needles punched into the stomach through the skin, (c) Endoscopic hook grasping the metal wire inserted through the needle, (d) Metal wire pulled out through the mouth, (e) Insertion of the second metal wire through the second PEG needle and (f) Stimulator device attached to the Mucosa wall by the tension of the metal wires created by pulling outward.

6.3.2. Results

The Mucosal impedance was measured as was 780 Ω with the Enterra recorder. EGG data was measured as in a similar way like all previous experiments and table 6.2 demonstrates the EGG results. The current was measured by taping two wires from the metal wires attached with the stimulator device hanging out of the porcine skin. While all the WRS and BWPS devices were tested on the Mucosa through the PEG method it was observed that BWPS devices delivered more current than WRS devices.

Stimulation	Current	Rhythm	Frequency Range	Mean Frequency	Amplitude	Amplitude Range	Mean Amplitude	FAR
	(mA)		(bpm)	(bpm)	.)p c	(mV)	(mV)	(bpm/V)
Baseline1		RR	2.0 - 4.0	2.667	UA	0.625 – 0.600	0.146	18.288
WRS Low	3.06	RR	5.0 - 6.0	5.333	EA	0.122 – 0.280	0.201	26.561
Baseline2		RR	4.0 - 5.0	4.667	UA	0.625 – 0.875	0.717	6.509
WRS Medium	3.06	RR	4.0 - 4.0	4.000	EA	0.250 – 0.300	0.267	14.998
Baseline3		RR	4.0 - 5.0	4.667	EA	0.128 – 0.240	0.196	23.811
WRS High	3.06	RR	4.0 - 5.0	4.333	EA	0.200 – 0.250	0.223	19.388
Baseline4		RR	6.0 – 5.0	5.667	UA	0.050 – 0.125	0.807	70.223
BWPS Low	3.71	RR	4.0 - 4.0	4.000	EA	0.060 – 0.063	0.061	65.573
Baseline5		RR	4.0 - 4.0	4.000	EA	0.175 – 0.20	0.187	21.352
BWPS Medium	3.71	RR	3.0 – 3.0	3.000	EA	0.145 – 0.180	0.159	18.861
Baseline6		RR	4.0 - 4.0	4.000	EA	0.375 – 0.450	0.400	10.000
.BWPS High	3.71	RR	3.0 – 3.0	3.000	EA	0.320 – 0.375	0.348	8.620
Baseline7		RR	3.0 – 4.0	3.667	EA	0.370 – 0.380	0.375	9.779

Table 6.2. EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through PEG Method.

The pig being healthy showed very regular EGG results during the stimulations and the baseline recordings. However the amplitude of the EGG signals tends to be unequal mostly during the baseline recordings but not all baseline recordings showed irregular rhythm. All stimulations showed equal amplitudes of the peaks of the EGG signal. The FAR changed during stimulation after placing the device endoscopically. However no definite pattern in the change of FAR was observed. Initially the FAR was higher but gradually the gastric muscles responded very faintly to the stimulations with the devices at the later part.

6.3.3. Discussions

Initially when the first baseline was recorded the FARs 18.288 bpm/mV and the signal had unequal amplitudes. The signal got stabilized when the first stimulation was given with WRS Low setting and the FAR changed dramatically by 8.272 bpm/mV. The initial instability of the signal amplitude was comprehended to be a resultant of the slight scars in the tissue due to puncturing by PEG needles. After stimulating with WRS Low setting the stimulator was taken out and WRS Medium stimulator was placed and this made the signal unstable in amplitudes again. The signal stabilized with WRS Medium stimulation. The FAR recordings were pretty stable during the third baseline, which was recorded after taking out the WRS Medium stimulator was taken out and the WRS High setting stimulator was placed. The wait period between each recordings was just 2 mins which was too short than required. The aftermath of the Medium stimulation might be present which helped the gastric tissues to retain the amplitude stability. The baselines recorded during BWPS stimulations were much stable except the first one. This might be due to the fact that the gastric muscles reserved a lot of energy from the previous WRS stimulations to make the EGG signal stable even during the baseline due to the lack of required time gap between stimulations.

6.4 Prototype Endoscopic Tack Method

6.4.1 Method

While endoscopic stimulation by this method Prototype endoscopic tacks (Cook Medical, Winston-Salem, NC) was inserted in the catheter by pressure. The catheter was maneuvered from outside through working channel. Once the device was deployed through the Overtube into the stomach, the catheter end was inserted through one of the ear electrodes and pushed hard into the gastric mucosal tissue (Fig. 6.8(a)). The tacks were pushed out of the delivering catheter when it seemed to pass through the ear electrode and pinched into the Mucosa. The device was attached transmurally to the gastric wall once the endoscopic tack was

deployed. Both the ends of the tack were released, the first hooked up to the external wall of the stomach while the second end attached the ear electrode to the Mucosa (Fig. 6.8(b). Fig. 6.8(c) shows a magnified picture of the tack attached to the Mucosal wall and holding the stimulator ear electrode. The block diagram of the device after attachment is similar to the Fig. 6.6. The step was repeated for attaching the other end of the stimulator device.



Figure 6.8 Device Implantation by endoscopic tack method (a) The catheter containing the endoscopic tack pushing into the gastric Mucosa (b) Deployment of the endoscopic tack into the Mucosa while attaching the ear electrode to the gastric tissue and (c) The magnified image of the attachment of the endoscopic tack to the ear electrode.

6.4.2 Results

Table 6.3. EGG Data Obtained after Endoscopically Implanting the Miniature Wireless Gastrostimulator through Prototype Endoscopic Tack Method.

Stimulation	Current (mA)	Rhythm	Frequency Range (bpm)	Mean Frequency (bpm)	Amplitude Type	Amplitude Range (mV)	Mean Amplitude (mV)	FAR (bpm/m V)
Baseline1		RR	5.0-5.0	5	UA	0.9-1.40	1.1	4.545
WRS Medium	3.06	RR	5.0-6.0	5.667	EA	0.15-0.35	0.233	24.320
Baseline2		RR	4.0-6.0	5.333	UA	0.8-1.0	0.933	5.713

It was observed from the surgical implantations, and endoscopic implantations through PEG method as demonstrated in earlier sections that the wireless stimulator can deliver enough power to change the EGG signal of the stomach. In this experiment only one stimulator device (WRS Medium setting) was tested to ensure that the connection between the ear electrodes with the gastric Mucosa was proper with the endoscopic tack method. Initially the EGG signal was showing unequal amplitude and the amplitude became more equally distributed after the stimulation. The wave was more rhythmic. The FAR ratio shot up very high during stimulation and decreased again during the second baseline reading. The second baseline was recorded with the device turned OFF however the device was not yet removed during this recordig. The second baseline also showed unequal amplitudes in the EGG signal.

6.4.3 Discussion

The high value of FAR ratio during stimulation was due to a sudden voltage shock in the gastric tissue. Ideally for the stimulation signal to get stabilized there should be a waiting period of 2hrs – 4 days. However the change in FAR during stimulation and making the EGG waves more equally distributed was quite promising outcome. The comprehensive device deployment and endoscopic implantation time took approximately 8 mins in the porcine model.

6.5 Prototype Endoscopic Spring Coil Method

6.5.1 Method

This method apparently resembles the prototype endoscopic tack method in grasping the gastric Mucosal tissue and affixing the ear electrode to it. However spring coils (Cook Medical, Winston-Salem, NC) were used instead of endoscopic tacks. The spring coil was forcefully inserted into the catheter. The catheter was maneuvered from outside through endoscopic working channel. Once the stimulator device was deployed into the stomach the catheter front end was inserted into one of the ear electrodes of the device and pushed into the Mucosa. The spring coil was gradually released encircling the spring coil as shown in Fig. 6.9(a). Eventually the coil was pushed out and attached the device to the stomach transmurally (Fig. 6.9(b). The Fig. 6.9(c) shows a magnified image of the spring coil attachment to the gastric Mucosa and the ear electrode. The same procedure was followed for attaching the ear electrode on the other end of the device.



Figure 6.9 Device Implantation by endoscopic spring coil method (a) The release of the spring coil after grasping the ear electrode to the Gastric Mucosa (b) Deployment of the endoscopic spring coil into the Mucosa while attaching the ear electrode to the gastric tissue and (c) The magnified image of the attachment of the endoscopic spring coil to the ear electrode.

6.5.1 Results and Discussion

Endoscopic attachment of the Gastrostimulator was successfully performed with the spring coil method. The duration of the attachment procedure was 10 mins. The spring coil could grasp the Mucosal layer and the coil went till the Submucosal layer allowing a very secure and firm attachment. EGG measurements could not be taken during this experiment however the voltage was measured at the output and thus ensuring the electrical connection between the tissue and the stimulating electrode.

6.6 Submucosal Pocket Method

6.6.1 Method

For this procedure a special type of Gen-2 wireless stimulator was fabricated. The ears in this device were designed for holding the device by the endoscopic forceps. The ears were not connected to the ground or the output of the microcontroller hence they do not behave as electrodes. However two circular wires were made to encircle the device as demonstrated in Fig. 6.10(a) and Fig. 6.11 with 1 cm apart from each other. These two wires were connected to the ground and output of the stimulator and thus behaved as electrodes. The upper insulation of the circular wires was removed so that the bare wires can be exposed and can touch the pocket muscle and provided the stimulation after inserting the stimulator in the submucosal pocket.



Figure 6.10 Illustration of wireless stimulator implantation by Submucosal pocketing and device implantation method.



Figure 6.11 Gen-2 device with circular wire wrapped electrodes.

For device implantation with Submucosal pocketing and device implantation (SPDI) following procedures were carried out. At first a Submucosal injection with diluted epinephrine (1:10,000) was given and an entry cut was made with a needle knife (Olympus America, Center Valley, PA). An endoscopic biliary stone extraction balloon (Cook Medical) was inserted through

the entry cut into the submucosal space to create an initial Submucosal pocket. Optionally, the entry opening can be dilated with an endoscopic biliary or through-the-scope dilating balloon (Cook Medical). The Gastroscope was advanced into the submucosal packet and limited Submucosal dissection with biopsy forceps was performed to enlarge the pocket based on the stimulator size. The wireless stimulator was brought near the opening by holding the device ear with the rat tooth forceps (Fig. 6.12(a)). The device was then implanted into the created Submucosal pocket (Fig. 6.10(b) and 6.12(b)). These pictures show the partial insertion of the device into the pocket. The device can be secured in the pocket by placing Endoclips at the opening. With tissue healing, the device is firmly embedded contacting the muscular layer of the gastric wall. Fig. 6.10(c) and 6.12(c) shows the wireless stimulator full insertion in the submucosal pocket.



Figure 6.12 Endoscopic pictures of wireless stimulator implantation by SPDI method.

6.6.2 Results and Discussion

Submucosal injection with dissection is an established clinical method to removing mucosal neoplasms. To the best of our knowledge, SPDI has not been reported previously in the literature for gastric device implantations. SPDI mimics surgical approach and methods. With SPDI, the stimulator was secured implanted and surface electrodes were in direct contact with the mucosal and submucosal layers. Depending on the size and depth of the Sub-Mucosal pocket, the stimulator can be partially, sub-totally or completely implanted or embedded tailoring

to the clinical indications and needs for device exchange. The complete implantation and insertion procedure by submucosal pocket method took approximately 20 minutes to be demonstrated.

These attachment devices and methods can be used in combination as well. Once endoscopic suturing devices and methods are available, the stimulator can also be attached to the gastric wall by suturing. It has been demonstrated that the clinical feasibility and utility of temporary GES using external surgical Gastrostimulator and endoscopic attachment of its transnasal electric leads to the gastric wall using Endoclips [6.1] is quite feasible. The treatment of gut dysmotility disorders is facing a new paradigm shift toward minimally invasive methods and natural evolution with these novel miniature Gastrostimulators and aforementioned new attachment methods and devices [6.2].

CHAPTER 7

CONCLUSION AND FUTURE WORKS

7.1 Conclusion

In this work, a completely new way of gastric stimulation has been investigated. The designed devices for gastric stimulation are wireless and miniaturized. The size was reduced by 90% and the current device measures less than 10% of the commercial stimulator. It will be a blessing for the patients who are suffering from Gastroparetic disorders due to diseases like diabetes and cancer. The wireless stimulator for Gastroparesis have been designed, fabricated and validated. A thorough optimization procedure was carried out while designing this prototype so that maximum power can be transferred from the transmitter to the stimulator to operate it efficiently. By carrying out the optimization procedure a 5mA of current could be delivered to the tissue over 8 cm of distance. The miniaturization has lead to the possibility of endoscopic implantation of the Gastrostimulator. The prototype was tested in vivo with anesthetized pigs. The device is small enough to be implemented in freely moving animals. The system was also tested by six different endoscopic procedures both for temporary implantation and the primary implantation. The results have shown that the gastric motility responds very well to the stimulation provided by the miniaturized wires stimulator. When used commercially it would abolish the hassle of re- surgical implantation of a stimulator due to battery exhaustion.

This can be a radical breakthrough and would provide a lot of comfort to the patients who do not want to undergo the surgical procedure. The design methodology for this endoscopic procedure of implantation of gastric devices can be also used for other gastric device applications in future.

89

7.2 Future Works

7.2.1 Transmitter Belt

The urge to miniaturize the complete wireless stimulation system lead to the designing of the portable transmitter which can be embedded in a thin belt that the patient can wear all the time.

7.2.1.1 Design

The transmitter was fabricated on a planar PCB. An 8 cm × 8 cm transmitter antenna was made with AWG-24 wire and 12 turns. The class-E amplifier design was kept similar to the used for the non-portable transmitter. Since exactly same components were used in the class-E amplifier and therefore the efficiency was 67% too as demonstrated earlier in the optimization calculation in chapter 3. The function generator (fluke PM5139) was replaced by a pre-programmed PIC12F683 module. This microcontroller was pre-programmed at 1.3 MHz and the program is illustrated in Appendix C. Two 3.7 V (TENERGY, 1250mAh) lithium-ion batteries put together in series were used for powering up the class-E amplifier. For the PIC12F683 microcontroller, a 3.7 V Lithum Ion battery was plugged in series with a 1.5V AA battery since the maximum voltage input for the microcontroller is 5.5 V and therefore putting two 3.7 V batteries in series would have crossed the maximum voltage limit.. The block diagram of the portable reader circuit is demonstrated in Fig. 7.1.

The complete transmitter was inserted into an handmade paper box to demonstrate the size. However the batteries were kept outside. The transmitter antena was glued in the inner upper edge of the box so that it can transfer the power to the stimulator without any hinderance in the magnetic field.



Figure 7.1 Block diagram of portable transmitter.

7.2.1.2 Results and Future Modifications

The experiment was performed with a Gen-2 BWPS stimulator device operating at the Low setting. The distance between the stimulator and the upper edge of the transmitter box was 8 cm.

It was observed on the computer screen in the LabView software that the transmitter induced a voltage of 1.8 V to the stimulator output. The BWPS stimulator was connected to a 800 Ω load mimicking the gastric tissue and hence the output current of 2.25 mA could be obtained across the load. Due to small size of 8 cm ×8 cm of the transmitter coil and the planar structure of the lithium-ion batteries in future the whole transmitter system can be inserted in a small belt which the patient can wear during their daily life.



Figure 7.2 Stimulator at setting Low operating with the portable transmitter.

7.2.2 Miniature Wireless Implantable EGG Sensor

EGG sensing is very important to observe the variation in the stomach myo-electric activity. It would be innovative if we can accompany out gastrostimulator along with an wireless EGG measuring device.

7.2.2.1 Design

A new wireless device for sensing the EGG signals have been demonstrated. For this the implant was designed as shown in Fig. 7.3. The EGG sensor design included a transistor. This transistor was used for turning OFF the modulation when the DC voltage from the regulator was too low. Every time it was turned off, it also gave the storage capacitor more time to harvest

the power from the reader. The minimum voltage that TLV3012 can operate was 2V and hence a 2V shunt regulator was used. For EGG signals a differential voltage was obtained from the gastric tissues and to transform this voltage to frequency signal, a modulation of the frequency generator has been used for our wireless sensing configuration.



Figure 7.3 Circuit diagram of the EGG sensor

The square wave signal generated from the relaxation oscillator depends on the input voltage level to the comparator. This change in the differential voltage in the EGG signals was eventually was given at the non-inverting input of the comparator and this also changed the output frequency. An op-amp (OPA349, Texas Instrument) was used as a buffer between the input signal and the modulating circuit.

The reader circuitry required for the EGG signal was replicated from the same reader used for Gastro Esophageal Reflux Disease (GERD) sensor demonstrated by Thermpon Ativanichayaphong [7.1]. The EGG signal lies in the range of 0.1 mV – 0.8 mV. Since the expected input voltage is very low for the sensing capability of the sensor circuit therefore an amplifier needed to be introduced in the module. An instrumentation amplifier of gain of 1000 was used in the instrumentation amplifier so that it amplified the voltage obtained from the EGG signal enough that it can be sensed by the modulating circuit producing a significant frequency shift at the reader side. The operating distance between the Reader and the EGG sensor was 8 cm for benchtop experiments.

For testing the functionality of the device a voltage of 0.1 V was given from a signal generator at a frequency of 0.05 Hz (mimicking the frequency of the EGG waves which is 3 - 4 cycles per min). A voltage divider of 1:1000 was used so that the input voltage droped down to 0.1 mV from 0.1 V similar to the EGG wave voltage. The voltage from the signal generator was varied from 0.1 V to 1 V and thus varying the input voltage to the instrumentation amplifier (INA333, Texas Instruments) from 0.1 mV to 1 mV.

7.2.2.2 Results and Future Modifications

Table 7.1 illustrates the frequency shift in kHz level due to the change in voltage at the input. In real life there will be no voltage divider or signal generator. The input has to come directly from the gastric tissues and needs to be fed into the instrumentation amplifier.

Signal Generator Voltage (V)	Voltage at the input of Instrumentation Amplifier (mV)	Frequency Shift (kHz)	Signal Generator Voltage (V)	Voltage at the input of Instrumentation Amplifier (mV)	Frequency Shift (kHz)
0	0	19.200	0.6	0.6	24.250
0.1	0.1	20.400	0.7	0.7	24.400
0.2	0.2	21.350	0.8	0.8	24.950
0.3	0.3	22.400	0.9	0.9	25.300
0.4	0.4	23.300	1.0	1.0	25.900
0.5	0.5	23.900			

Table 7.1 Frequency Shift due to Different Input Voltages


Figure 7.4 Frequency shift with respect to the applied voltage at the instrumentation amplifier.

Fig. 7.4 shows the frequency shift output response obtained at the reader end due to the frequency modulation of the signal from the EGG sensor side transferred wirelessly. A steady change in the frequency has been observed due to even very small change in the input voltage. Fig. 7.5 shows the fabricated EGG sensor on a double layer PCB board. The sensor measures 42 mm × 8 mm × 5 mm. The size can be further reduced by using a 4 layer PCB board for the fabrication. By observing the pace of the frequency shift the frequency of the slow wave in the EGG signal can be detected and by observing the amount of frequency shift the amplitude of the slow waves can be calculated. The FAR ratio of the EGG signal can be derieved from these two entities. The frequency shift due to a steady output change was pretty significant, however a thorough analysis still needs to be done to affirm if this device is suitable for EGG measurements in real life. The EGG signals are susceptible to respiratory signals and stomach contractions. Some more filtering might need to be induced in the sensor to eradicate the high frequencies due to respiration or body contractions.



Figure 7.5 Fabricated EGG sensor on PCB.

7.2.3 Folding Antenna Design for Stimulator

The implant circuitry and performance of the wireless gastric stimulator is completely dependent on the wireless energy harvested by the implant antenna circuitry from the transmitter. Induction of more power in the circuit can definitely generate more output voltage for the gastric tissues. It will provide enormous scope of introducing new features along with the stimulation method. One can think of combining the EGG sensor with the stimulator in future and incorporate a relay switch model, that whenever the EGG signal of the patient has abnormal frequency or amplitude in the myo-electric activity the stimulator can get turned on. There are various applications that can be thought about, and if there is more power available to work with.

7.2.3.1 Design



Figure 7.6 Folding antenna and Umbrella design structure (a) Folded antenna of 4 cm in diameter, (b) Deformation shown for 4 cm diameter antenna, (c) Foldded antenna of 6 cm and (d) Deformation shown in 6 cm diameter antenna.

To augment the power in the implant, if the size of the antenna can be increased then the power harvested can be inseased significantly. But again for most implantable devices the size is limited, thereby a new design of folding antenna have been illustrated. Fig. 7.6 demonstrates a new umbrella structure for folding antennas. The tension and strength was provided by kapton thin films cut into strips as shown in the figure. Once the antenna was folded, the kapton stips embedded in the PDMS, helped with the reflexes to get it back in shape. The small implant circuit can be placed in the middle if the antenna. The atenna can be folded as shown in Fig 7.6 (a) and (c). Fig 7.6 (b) and (d) shows the deformation and reformation of the antenna structure. Smaller diameter antenna produced less deformation.





Figure 7.7 Folding antenna and planar spiral coil model (a)Designing and fabrication of folding spiral coil with PDMS and AWG wires, (b) Stimulator with coil loop antenna and planar coil antenna, (c) Folded planar coil antenna around the stimulator and (d) Deformation of planar coil antenna.

The second folding antenna model involves the designing of a planar circular coil antenna as demonstrated in Fig. 7.7(b). In Fig 7.7(a) the PDMS coating fabrication procedure to make the circular coil is shown. The coil always have a tendency to bounce back and fall apart. So to make it absolutely planar a weight has to be given from the top during curing of the PDMS. Fig. 7.7(c) shows the form factor comparison of a Gen-2 stimulator and the spiral coil folding antenna stimulator. The length of the folding antenna stimulator is exactly similar to the Gen-2 device. The width at the opened condition is 3 cm. For the Gen-2 device after PDMS coating it was 9 mm. Once it is folded the stimulator with the folding antenna measured 11 mm

and this is less than the Gen-1 device width and a little bigger (2 mm) than the Gen-2 device width.



7.2.3.2 Results and Future Modifications

Figure 7.8 Experimental setup of stimulator with planar spiral folding antenna.

Benchtop experiments were performed with the Gen-2 Stimulator and the folding antenna stimulator. The Gen-2 stimulator had 15 turns and the planar spiral folding antenna has 7 turns. The setup for testing the folding antenna stimulator is shown in Fig 7.8. The experiment was done with a 800 Ω load attached at the output of both the stimulators. The Gen-2 stimulator generated 3.0 V at the output across the load from a distance of 8 cm however the stimulator with the folding antenna generated 3.5 V from a distance of 18 cm.

As it has been observed the folding antenna has given a huge leverage to play with the distance between the transmitter and the implant for generating the same and even more voltage at the implant output end with the same input power from the transmitter side. Converting the whole PCB design on a planar Kapton surface with copper deposited on it can also be explored in future. The spiral folding antenna can be made on the other side of the Kapton and thus combining the whole design in a single package.

APPENDIX A

PROGRAM FOR STIMULATOR PULSE GENERATOR WITH PIC10F WRITTEN IN ASSEMBLY LANGUAGE

PIC10F206 Setting 1

; Notes: Use GP1 for output						
; Rest a		S ********	******	****	******	
,	list #include radix he CON	e <p10f2 ex IFIG. 0F</p10f2 	p=10f20 06.inc> FEB	6	;list directive to define processor ;processor specific variable definitions	
;******** ;Dofino	*********	******	******	*******	***************************************	
,Denne:	S :*******	******	*****Variable Defi	nitions********	***************************************	
,						
delay1 delay2 delay3			EQU EQU EQU	0x10 0x11 0x12	;delay counter used for 1 s timer ;delay counter used for 4 s timer ;delay counter used for 35.7ms	
•******* ,	*******	******	******	******	********	
.*******	org	*****	0x1FF	****	;Processor reset vector for pic 10F206	
, ; Interna ;as mov	al RC ca /lw k, wh	libration here the	value is placed a k is a literal value	t location 0x1	F by Microchip	
	org movwf	OSCCA	0x000 IL		;load cal value into oscal	
	nop		;nop		;nop here for debug reasons	
	bsf		OSCCAL,0		;enable fosc/4 output on GP2	
.*	goto	MAIN				
,; ;subrou	tines	******	*********Insert Su		****	
, wait1						
	nop btfss goto retlw	TMR0,7 wait1 0x00	,			
;<5S de wait5S	elay					
longer f	movlw	0x99			;load 5 sec counter: higher number =	
ionger t	movwf	delay2				
loop3	movlw	0x01			;load timer0	
loon	movwf	TMR0				
1000	btfss	TMR0,7	,			

	goto decfsz goto	loop delay2,1 loop3				
loop4	movlw movwf	0x22 TMR0	;load timer0			
	btfss goto	TMR0,7 loop4				
	movlw movwf	0x41 delay1				
10005	decfsz goto	delay1,f loop5				
·* _	retlw	0x00				
, MAIN :init stut	ff		;Program begins here			
,init ota	movlw	b'11110100'	;set comparator to output not external			
startun	clrf movlw	GPIO b'11111101'	;clear the data latch ;set GPIO 1 to be write, rest read on			
Startup	tris cIrf	GPIO GPIO	;turnoff outputs			
;*						
;Main L	oop movlw movwf	b'00000010' delay1	;load counter this is 0.1sec counter ; higher value=longer time for pulser			
main	movlw	b'11000001'	;PSA 1:4			
	movlw movwf bsf call	0x30 TMR0 GPIO,1 wait1	;higher number = shorter time ;load TIMER0 ;make output high ;delay for 330us			
	movlw	b'01000111'				
	option bcf	GPIO,1	;PSA 1:256 ;make output low			
doloy74	movlw movwf	0x06 delay3	;repeate 6 times until ~71mS is riched			
20.097	movlw movwf call	0x52 TMR0 wait1	;load TIMER0 counter ;higher number = shorter time ;delay (71 / 6)mS			

d g	lecfsz joto	delay3,1 delay71mS	
n delay71m d g n n	novlw novwf nSExtra lecfsz goto de nop	0xDC delay3 a delay3 lay71mSExtra	;extra delay to complete 71.4mS
d g	lecfsz joto	delay1,1 main	;dec seconds counter: if1s has expired ;do it again if 1 sec hasn't expired
n n n n n	100 100 100 100 100 100		
С	all	wait5S	
n	юр		
g E	joto END	MAIN	; directive 'end of program'

PIC10F206 Setting 2

•*******	.**************************************					
; Notes ; Rest a	: Use GF are input	P1 for output s	*****	****	*****	
,	list #includ radix he CON	e <p10f206.inc> ex FIG. 0FEB</p10f206.inc>	p=10f206		;list directive to define processor ;processor specific variable definitions	
;******** :Dofino	*********	******	*****************	*****	***************************************	
,Denne	J *********	************Var	able Definitions'	*******	******	
delay1 delay2		EQU EQU		0x10 0x11	;delay counter used for 1 s timer ;delay counter used for 4 s timer	
delay3		EQU		0x12	delay counter used for 35.7ms	
.*******	******	*****	*****	*****	*****	
, •*******	org	0x1FF	****	****	;Processor reset vector for pic 10F	
, : Interna	al RC ca	libration value is	placed at location	on 0x1Fl	F by Microchip	
;as mov	/lw k, wh	ere the k is a lite	eral value			
	ora	0,000				
	movwf	OSCCAL			;load cal value into oscal	
	nop	;nop			;nop here for debug reasons	
	bsf	OSCCA	AL,0		;enable fosc/4 output on GP2	
•*=====	goto	MAIN				
, ;subrou	itines					
.******** , 	*******	*******	*Insert Subs here	9*******	***************************************	
Walt'i	non					
	btfss	TMR0,7				
	goto	wait1				
	retlw	0x00				
wait2						
	decfsz	delay3,1				
	goto	wait2				
·* <u>-</u>	retlw	0x00				
, MAIN					;Program begins here	
;init stu	ff .					
	movlw movwf	D'11110100' CMCON0			;set comparator to output	

clrf GPIO ;clear the data latch movlw b'11111101' ;set GPIO 1 to be write tris GPIO GPIO ;turnoff outputs clrf ;Main Loop movlw b'00011100' ;load counter this is 1sec counter movwf delay1 main movlw b'11000001' ;PSA 1:4 option movlw 0x30 ;higher number = shorter time movwf TMR0 :load TIMER0 GPIO,1 bsf ;make output high call wait1 ;delay for 330us movlw b'01000111' ;PSA 1:256 option GPIO,1 bcf ;make output low movlw 0x00 ;load TIMER0 counter movwf TMR0 ;higher number = shorter time call wait1 ;delay for 35.7ms movlw 0x76 movwf TMR0 wait1 call decfsz delay1,1 ;dec seconds counter: if 1s expires ;do it again because 1s hasn't expired goto main ;movlw b'11000111' ;make sure PSA = 1:256 ;option movlw b'01111010' ;load 4 sec counter: higher number = longer time movwf delay2 loop3 ;load timer0 movlw 0x00 movwf TMR0 loop btfss **TMR0,7** loop goto decfsz delay2,1 goto loop3 MAIN goto END ;directive 'end of program'

PIC10F206 Setting 3

·*******	*******	*****	*****	*******	******
; Notes ; Rest a	: Use GF	P1 for output S	*****	****	*****
,	list #include radix he CON	e <p10f206.inc> ex FIG. 0FEB</p10f206.inc>	p=10f206		;list directive to define processor ; processor specific variable definitions
;******** :Define	·*********	******	********************	*******	***************************************
,Denne,	J :********	************Var	able Definitions'	*******	******
delay1 delay2 delay3		EQU EQU EQU		0x10 0x11 0x12	;delay counter used for 1 s timer ;delay counter used for 4 s timer ;delay counter used for 35.7ms
.******** , .********	org	0x1FF	***************************************	*********	;Processor reset vector for pic 10F
; Interna is a liter	al RC ca ral value	libration value is	placed at locati	on 0x1Fl	F by Microchip;as movlw k, where the k
	org 0x000 movwf OSCCAL				;load cal value into oscal
	nop	;nop			;nop here for debug reasons
	bsf	OSCCA	AL,0		;enable fosc/4 output on GP2
·*=====	goto	MAIN			
, ;subrou	tines				
;********* wait1	*******	*****	*Insert Subs her	e*******	***************************************
	nop btfss goto retlw	TMR0,7 wait1 0x00			
wait2	decfsz	delay3,1			
·*	goto retlw	wait2 0x00			
, MAIN	-			=	; Program begins here
,iiii Siu	movlw movwf	b'11110100' CMCON0			;set comparator to output

clrf GPIO ;clear the data latch movlw b'11111101' ;set GPIO 1 to be write GPIO tris GPIO clrf ;turnoff outputs ;Main Loop movlw 0xD8 ;216 times (4S) movwf delay1 main movlw b'11000001' ;PSA 1:4 option movlw 0x30 ;higher number = shorter time movwf TMR0 ;load TIMER0 GPIO,1 bsf ;make output high ;delay for 330us call wait1 movlw b'01000111' ;PSA 1:256 option GPIO,1 bcf ;make output low movlw 0x39 ;load TIMER0 counter movwf TMR0 ;higher number = shorter time call wait1 ;delay for ~18.192ms ;nops - 18.2 mS fine tuning nop nop nop nop nop nop nop decfsz delay1,1 ;dec seconds counter: if clear 1sec has expired ;do it again when 1s expires goto main ;make sure PSA = 1:256 ;movlw b'11000111' ;option movlw 0x1F ;load 1 sec counter: movwf delay2 loop3 movlw 0x3 :load timer0 movwf TMR0 loop TMR0,7 btfss goto loop decfsz delay2,1 goto loop3 movlw 0x63 ;load timer0 movwf TMR0 call wait1

loon4	movlw movwf	0x45 delay2		
10001	decfsz goto	delay2, f loop4		
fine tun	nop ing nop			;1S
	goto END	MAIN	; directive 'end of program'	

APPENDIX B

PROGRAM FOR STIMULATOR PULSE GENERATOR WITH PIC12F WRITTEN IN ASSEMBLY LANGUAGE

Pulse Generator Program with Change of Settings.

; ************************************					
;======* End notes	*=========	*******	***************************************		
LIST P=12F #INCLUDE <p12f6 radix hex</p12f6 	-683 83.INC>	; list dir ; proce	ective to define processor ssor specific variable definitions		
; CONFIG. & _MCLRE_ON & _WI	_FCMEN_ON & _IESO_()T_OFF & _PWRTE_ON	OFF & _ & _INTI	CP_OFF & _CPD_OFF & _BOD_OFF RC_OSC_NOCLKOUT		
;*************************************	*****	*******	***************************************		
.*************************************	*********Variable Definitio	********** NS ^{*******}	***************************************		
delay1 delay2 delay3 CONFIGADDR EC	EQU EQU EQU 2U 0x10 EQU EQU EQU	0x60 0x61 0x62 0x63 0x64 0x65	;delay counter used for 1s ;delay counter used for 4s ;delay counter used for35.7ms ; set the EEPROM Address as 20 hex ;check_present_setting ; 01h setting1, 02h setting2, 04h set3		
·****	- ~ ~	*******	; 1011 for setting2,0111 for setting3;		
; org	0x1FF		;Processor reset vector for pic 10F206		
; Internal RC calibration is a literal value org 0x00 goto START_ org 0x04 goto INTERF	n value is placed at locati _PROGRAM RUPT_SERVICE	on 0x1F	F by Microchip as movlw k, where the k		
; movwf OSCC	AL		;load cal value into oscal		
START_PROGRAM nop	;nop		;nop here for debug reasons		
bsf	OSCCAL,0		;enable fosc/4 output on GP2		

MOVLW 07h ;Set GP<2:0> to MOVWF CMCON0 ;digital I/O BANKSEL ANSEL; CLRF ANSEL ;digital I/O BANKSEL GPIO ; clrf GPIO ;clear the data latch movlw b'11111101' ;set GPIO 1 to be write, rest read on startup ;tris GPIO BANKSEL TRISIO; movwf TRISIO BANKSEL GPIO ; ; turnoff outputs clrf GPIO movlw b'10010000' ; set GIE and INTE bit as 1 ; to enable external interrupt on movwf INTCON GP2/INT BANKSEL OPTION REG bsf OPTION_REG,6 ; rising edge enabled interrupt -----**______ ;; Read the EEPROM address and take decision according to that which program to run BANKSEL EEADR ; ; commenting all banksel MOVLW CONFIG._ADDR; MOVWF EEADR :Address to read BSF EECON1,RD ;EE Read MOVF EEDAT.W :Move data to W BANKSEL TMR0; movwf program; program is to test which setting btfss program,1; goto MAIN1; btfss program,2; goto MAIN2; btfss program,3; goto MAIN3 ; goto MAIN1; ;test purpose for now 10-13-10 ; read it carefully INTERRUPT_SERVICE bcf INTCON,1; clrf INTCON; call wait5S setting1; ; when get interrupt wait for 5s movlw b'10010000' ; set GIE and INTE bit as 1

movwf INTCON ; to enable external interrupt on GP2/INT ;clear timer0 at interrupt to stop ovrflw clrf TMR0; btfss check_present_setting,0 goto next1 movlw b'11111011' movwf DATA_EEPROM call EEPROM_WRITE_MODULE goto MAIN2 next1 btfss check present setting,1 goto next2 movlw b'11110111' movwf DATA EEPROM call EEPROM_WRITE_MODULE goto MAIN3 next2 btfss check_present_setting,2 nop movlw b'11111101' movwf DATA_EEPROM call EEPROM_WRITE_MODULE goto MAIN1 EEPROM WRITE MODULE BANKSEL EEADR ; MOVLW CONFIG. ADDR; MOVWF EEADR ;Address to write BANKSEL TMR0 ;; very important movf DATA EEPROM,0 BANKSEL EEDAT ; movwf EEDAT ; move W to EEDAT ;;begin write -- the below sequence must be executed BANKSEL EECON1 ; **BSF EECON1,WREN** :Enable write **BCF INTCON, GIE** :Disable INTs ;See AN576 **BTFSC INTCON, GIE** GOTO \$-2 : MOVLW 55h ; Unlock write MOVWF EECON2; MOVLW 0xAA; MOVWF EECON2 ; BSF EECON1,WR ;Start the write **BSF INTCON, GIE** :Enable INTS ;;; test_write_is_done btfsc EECON1,WR goto test_write_is_done

;goto M return ;	IAIN2 0x00	; when write is done go to MAIN2		
;;**************** ;subroutines	*****************	*Program for setting 1 starts here **********************************		
wait1_setting1 nop btfss goto retlw	TMR0,7 wait1_setting1 0x00	Insert Subs here***********************************		
;<5S delay wait5S_setting ² movlw movwf loop3_setting1	l 0x99 delay2	;load 5 sec counter:		
loop_setting1 btfss goto decfsz goto	0x01 TMR0 TMR0,7 loop_setting1 delay2,1 loop3_setting1	;load timer0		
movlw movwf loop4_setting1 btfss goto	0x22 TMR0 TMR0,7 loop4_setting1	;load timer0		
movlw movwf loop5_setting1 decfsz goto retlw	0x41 delay1 delay1,f loop5_setting1 0x00			
;*========== ;;setting_2_wait sbsf OF nop btfss goto retlw	TION_REG,6; TMR0,7 wait1_setting2 0x00			
wait2_setting2 decfsz aoto	delay3,1 wait2 settino2			

retlw 0x00 wait1_setting3 nop **TMR0,7** btfss wait1_setting3 goto retlw 0x00 wait2_setting3 decfsz delay3,1 goto wait2_setting3 retlw 0x00 MAIN1 ; Program begins here ;interrupt stuff bsf OPTION_REG,6; BANKSEL check_present_setting ; Reset the memory bank to file registers movlw b'10010000' ; Set GIE and INTE bit as 1 movwf INTCON ; To enable external interrupt on **GP2/INT** movlw 0x01; movwf check present setting ; Contains info about present set bsf INTCON,1 ;for checking of change of setting ;init stuff ;movlw b'11110100' ;set comparator to output not external ;movwf CMCON0 :Main Loop movlw b'00000010' ;load counter this is 0.1sec counter movwf delay1 main_setting1 movlw b'11000001' ;PSA 1:4 option movlw 0x30 ;higher number = shorter time ;clrf TMR0 ; BANKSEL TMR0; movwf TMR0 ;load TIMER0 bsf GPIO,1 ;make output high call wait1 setting1 ;delay for 330us movlw b'01000111' option ;PSA 1:256

GPIO,1 bcf ;make output low movlw 0x06 ;repeate 6 times until ~71mS is riched movwf delay3 delay71mS movlw 0x52 ;load TIMER0 counter movwf TMR0 ;higher number = shorter time call wait1_setting1 ;delay (71 / 6)mS decfsz delay3,1 goto delay71mS movlw 0xDC movwf delay3 delay71mSExtra ;extra delay to complete 71.4mS decfsz delay3 goto delay71mSExtra nop nop decfsz delay1,1 ;dec seconds counter: if clear 1sec has expired goto main_setting1 ;do it again because 1s hasn't expired nop nop nop nop nop nop call wait5S_setting1 nop MAIN1 goto MAIN2 bsf OPTION REG,6; BANKSEL check_present_setting ; Reset the memory bank movlw 0x02; movwf check_present_setting ; ;movlw b'10010000' ; set GIE and INTE bit as 1 :movwf INTCON ; to enable external interrupt on **GP2/INT** ;goto MAIN2 .* _____ ;Main Loop movlw b'00011100' ;load counter this is 1sec counter movwf delay1

main_setting2 movlw b'11000001' :PSA 1:4 option movlw 0x30 ;higher number = shorter time BANKSEL TMR0 movwf TMR0 ;load TIMER0 bsf GPIO,1 ;make output high wait1_setting2 ;delay for 330us call movlw b'01000111' ;PSA 1:256 option bcf GPIO,1 ;make output low movlw 0x00 ;load TIMER0 counter BANKSEL TMR0 movwf TMR0 ;higher number = shorter time call wait1 setting2 ;delay for 35.7ms movlw 0x76 BANKSEL TMR0 movwf TMR0 call wait1_setting2 decfsz delay1,1 ;dec seconds counter: if clear 1sec has expired ;do it again because 1s hasn't expired goto main setting2 ;movlw b'11000111' ;make sure PSA = 1:256 :option movlw b'01111010' ;load 4 sec counter: higher number = longer time movwf delay2 loop3_setting2 movlw 0x00 BANKSEL TMR0 ;load timer0 movwf TMR0 loop_setting2 btfss TMR0,7 loop setting2 goto decfsz delay2,1 goto loop3 setting2 goto MAIN2 MAIN3 bsf OPTION_REG,6; BANKSEL check_present_setting ; Reset the memory bank movlw 0x04; movwf check_present_setting ; movlw b'10010000' ; set GIE and INTE bit as 1 movwf INTCON ; Enable external interrupt on GP2/INT ; goto MAIN3 •*______

;Main Loop movlw 0xD8 ;216 times (4S) movwf delay1 main setting3 movlw b'11000001' :PSA 1:4 option movlw 0x30 BANKSEL TMR0 ;higher number = shorter time movwf TMR0 ;load TIMER0 bsf GPIO,1 ;make output high ;delay for 330us call wait1_setting3 movlw b'01000111' :PSA 1:256 option GPIO,1 bcf ;make output low movlw 0x39 BANKSEL TMR0 ;load TIMER0 counter movwf TMR0 ;higher number = shorter time call wait1_setting3 ;delay for ~18.192ms ;nops - 18.2 mS fine tuning nop nop nop nop nop nop nop decfsz delay1,1 ;dec seconds counter: clear 1s expires main_setting3 ;do it again because 1 sec has not goto expired ;movlw b'11000111' ;make sure PSA = 1:256 ;option movlw 0x1F ;load 1 sec counter: higher number = longer time movwf delay2 loop3_setting3 ;BANKSEL TMR0 movlw 0x3 :load timer0 movwf TMR0 loop_setting3 ;BANKSEL TMR0 btfss TMR0,7 goto loop setting3 decfsz delay2,1 loop3 setting3 goto movlw 0x63 ;BANKSEL TMR0 ;load timer0

movwf TMR0 call wait1_setting3 movlw 0x45 movwf delay2 loop4_setting3 decfsz delay2, f loop4_setting3 goto ;1S fine tuning nop nop MAIN3 goto ; directive 'end of program' END

APPENDIX C

PROGRAM FOR TRANSMITTER CIRCUIT WRITTEN IN ASSEMBLY LANGUAGE

: Notes: Use GP1 for output ; Rest are inputs LIST P=12F683 ; list directive to define processor CLUDE <P12F683.INC> ; processor specific variable definitions #INCLUDE <P12F683.INC> radix hex ; _CONFIG. _FCMEN_ON & _IESO_OFF & _CP_OFF & _CPD_OFF & _BOD_OFF & MCLRE ON & WDT OFF & PWRTE ON & INTROOSC NOCLKOUT org 0x00 ;goto START_PROGRAM ;START_PROGRAM nop ;nop here for debug reasons ;nop nop movlw b'01110001' BANKSEL OSCCON ; OSCCON,0 bsf ; movwf OSCCON ; osc run on 8 MHz. MOVLW 07h ;Set GP<2:0> to BANKSEL CMCON0; MOVWF CMCON0 ;digital I/O BANKSEL ANSEL; CLRF ANSEL ;digital I/O BANKSEL GPIO ; clrf GPIO :clear the data latch movlw b'11111101' ;set GPIO 1 to be write, rest read on startup ;tris GPIO BANKSEL TRISIO; movwf TRISIO BANKSEL GPIO ; GPIO clrf bsf GPIO,1 MAIN ; 8/6 = 1.3 MHz with 50% duty cycle bcf GPIO,1 bcf GPIO,1 bcf GPIO,1 bsf GPIO,1 goto MAIN END

Signal Generator, Generating Square Wave Pulses of 1.3MHz with 50% Duty Cycle.

APPENDIX D

PROGRAM FOR OPTIMIZATION

Matlab Program for Optimization of Parameters in Inductive Coupling

clc clear all Xmin= input('Enter the value for Xmin(The minimum distance between two coils it can be .5 cm % the minimum distance considered to be .5 cm or to 6 cm = '; variable Xmax= input('Enter the value for Xmax(The maximum distance between two coils it can be .5 % the maximum distance to be consider is 6 cm(cm to 6 cm) = ');written in slide) period = linspace(Xmin,Xmax,3); % breaking the distance X in to 10 equal parts dx=period(2)-period(1);% This is the value of delta X derived from the maximum distance considered W0=8.1681e+006: % value of womega not pi=3.141: Myu = 0.000001256; % value of myu r1min=input('Enter the value for r1min(The minimum radius of the primary coil it can be between 1 cm to 10 cm or variable) = '); % In lab we measured the minimum radius was taken 1 cm which can be variable r1max=input('Enter the value for r1max(The maximum radius of the primary coil it can be between 1 cm to 10 cm or variable) = '); % we consider the max r1 be 10 cm r2min=input('Enter the value for r2min(The minimum radius of the secondary coil it can be between 1 cm to 10 cm or variable) = '; % We consider the r2 min which is also 1 cm r2max=input('Enter the value for r2max(The maximum radius of the secondary coil it can be between 1 cm to 10 cm or variable) = ': % r2 max considered to be 10 cm t=linspace(r1min.r1max.3); % breaking the distance r1 in to 10 equal parts. we can change the parts just changing number. % This is the value of delta r1 derived from the dr1=t(2) - t(1);maximum distance considered s=linspace(r2min,r2max,3); % breaking the distance r2 in to 10 equal parts % This is the value of delta r2 derived from the dr2 = s(2) - s(1);maximum distance considered etanot = input('Enter the value for etanot which is = '); % value for Etan not N1max= input('Enter the value for N1max(maximum number of turn for N1) = '); % input for number of turn in coil 1 N2max= input('Enter the value for N2max(maximum number of turn for N2) = '); % input for number of turn in coil 2 RAC Load min = input('Enter the value for RAC Load min (the minimum is 100 ohm) = '); % the minimum value from 100 - 4000 ohm RAC_Load_max = input('Enter the value for RAC_Load_max (the maximum is 4000 ohm) = '); % the maximum value from 100 - 4000 ohm label = linspace(RAC Load min,RAC Load max,3); % RAC Load varied from 100 to 4000 in to 10 prts drac load=label(2)-label(1); for i=1 for RAC Load = RAC Load min:drac load:RAC Load max; % Considering the min distance as RAC Load min RAC Load

if RAC Load < RAC Load max

```
X=Xmin;
for X=Xmin:dx:Xmax;
  if X < Xmax
 r2=r2min;
 read = xlsread('measurement.xlsx');
newdat=r2*100;
R20=read(round(newdat),3);
L20=read(round(newdat),4);
L20=L20*10^(-6);
   for r2 = r2min:dr2:r2max;
   if r_2 < r_{2max}
 r1=r1min;
 read = xlsread('measurement.xlsx');
newdat=r1*100;
R10=read(newdat,3);
 for r1 = r1min:dr1:r1max
   if r1 < r1max
for b=1:N2max;
  if b < N2max
    for a=1:N1max;
       if a < N1max
    M=W0*W0*Myu*pi*a*b*b*b*r2*r2*r1*r1*L20;
    N=(r1*r1+X*X)*(r1*r1+X*X)*(r1*r1+X*X);
    P=2*sqrt(N);
    A=M/P;
    B=a*R10*b*R20;
    C=W0*W0*b*b*b*b*L20*L20*a*R10/RAC_Load;
    D=(W0*W0*Myu*Myu*pi*pi*a*a*b*b*r2*r2*r2*r2*r1*r1*r1*r1)/(P*P);
    E=(A/(B+C+D))^{2*a*R10};
                                     %For Efficiency percentage
    Eta=E*100/RAC_Load;
                                     %plot(Eta)
       if Eta > etanot
  doit = [r1, r2, R10, R20, L20, X, RAC_Load, a, b, Eta ];
  datam(i,:)=doit;
  i=i+1;
 a=a+1;
    else
       a=a+1;
   end
        else
b=b+1;
```

```
end
     end
  else
r1=r1+dr1;
read = xlsread('measurement.xlsx');
newdat=r1*100;
R10 = read(round(newdat),3);
  end
end
    else
         r2=r2+dr2;
         read = xlsread('measurement.xlsx');
newdat=r2*100;
R20=read(round(newdat),3);
L20=read(round(newdat),4);
L20=L20*10^(-6);
    end
 end
    else
    X = X + dx;
    end
 end
  else
   RAC_Load=RAC_Load+ drac_load;
  end
end
  else
     'stop'
  end
end
end
programend=xlswrite('imported',datam)
```

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BIOGRAPHICAL INFORMATION

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132