Wireless Gastric Stimulators

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***Abstract* — Gastric dysmotility disorders including gastroparesis cause chronic debilitating symptoms. Gastric electrical stimulation (GES) has shown to be effective in stomach pacing, gastric emptying and treating gastric dysrhythmia. However, the current battery-based GES devices are bulky yet with limited lifetime. The device implantation requires a major surgery. In this paper, we present miniature wireless gastric electrical stimulators to overcome the existing shortcomings. Rechargeable battery based and batteryless stimulators were tested in benchtop experiments and live porcine animal models. The battery-based system utilizes wireless charging while the batteryless device harvests electromagnetic energy to power a microcontroller that delivers electrical pulses to tissues. Both prototypes were tested at three different stimulation settings for demonstration. Recorded electrogastrograms (EGG) in stomach were analyzed in terms of signal amplitude, frequency, stimulation power and energy. EGG frequency variations, amplitude ratios, and gastric rhythmic activities during GES were analyzed. The results showed that the wireless miniature stimulators modified gastric rhythmic activities. Utilization of these miniature devices may eliminate the need of repeated surgeries for battery replacement. The small size of the stimulators also makes it feasible to implant them through endoscopic surgeries.**

***Index Terms* — Endoscopy, gastro-stimulator, gastric electrical stimulation (GES), gastroparesis, stimulator.**

I. Introduction

Gastric dysmotility disorders such as gastroparesis (GP) cause chronic debilitating symptoms such as nausea, vomiting, early satiety, weight loss and abdominal pain [1]. It was estimated that GP affects more than 1.5 million Americans with over 100,000 suffering from severe forms of the disorder (2012). At least 20% of people with type-1 diabetes develop GP and it can occur in type-2 diabetes as well [1, 2]. Gastroparesis is characterized by the abnormal amplitude and frequency of the gastric slow waves [3] affecting myoelectrical activities in stomach. Drug therapies with prokinetics have been used for symptomatic control only and up to 40% of gastroparetic patients are intolerant to the prokinetics due to side effects [4, 5]. Recently, gastric electrical stimulation (GES) has been shown to be effective for symptomatic relief [6].

GES can be achieved with either surgical implantation of a commercially available neuro-stimulator [7]or temporary endoscopic stimulation (EndoStim) [8]. GES alters intrinsic gastric electrical activities by stimulating the stomach at a frequency generally higher than the normal electrophysical frequency of 3 cycles per minute [9]. The required electrical current ranges from 2 to 6mA, likely affecting the interstitial cells of Cajal and smooth muscle cells [10]. Despite promising results, GES has not been widely adopted in clinical applications due to the requirement of major implantation surgeries or bulky temporary wires through the nostrils that hamper their daily activities in the EndoStim cases. Currently, permanent GES requires surgical implantation of a neuro-stimulator (Enterra®, Medtronic) approved by the FDA. The device produces an electrical pulse train at a frequency of 14 Hz and a current of 5 mA in its lowest setting. The non-rechargeable battery based device has a size of 60×55×10 mm^3 and is implanted during a 1-3 hour operation under general anesthesia and requires several days of post-operative hospitalization. The shortcomings of this method are the limited battery life and the large device size that requires a major surgery.

We developed miniature wireless gastro-stimulators that can be implanted endoscopically without the need for surgery or external tethered wires. Both wirelessly rechargeable-battery-based and batteryless devices were developed. The battery-based system utilizes wireless charging while the batteryless device harvests electromagnetic energy to power a microcontroller that delivers electrical pulses to tissues [11].

II. Designs

Both wireless gastrostimulators are powered by inductive coupling at 1.27 MHz and 4 W. The wearable transmitter coil size is 11.5×11.5 cm^2 made of AWG metal wires. The coil was driven by a class-E amplifier with a biasing voltage of 5 V by a 50% duty-cycle square waveform generated by a microprocessor. Both types of implants consist of coils and regulator to harvest the RF energy and peripheral interface controller (PIC) generating stimulation pulses. The PIC is pre-programmed to generate pulse trains with specific frequencies and duty cycles. The pulse train parameters are defined in Fig. 1 and three different settings based on previous clinical works [3, 12] are shown in Table I.

Figures 2 and 3 show the block diagrams of the systems. The battery based device contains charge pump circuitry and a 3-V 11-mAh rechargeable battery. A magnetic reed switch (KSK-1A80-1015) connects the battery to the charging driver, or to the PIC. The switch is activated by an external magnet between the charging mode and the stimulation mode. In benchtop experiments, it took 30 minutes to recharge the battery from below the working range of 1.8 V back to 2.75 V where required current was generated.

TABLE I
PULSE SPECIFICATIONS

Setting	T_p / T_o (μs/ms)	Pulse frequency f (Hz)	Pulse duty cycle %	T_{on} / T_{off} (s)	Number of pulses per cycle
Low	330/71.4	14	0.46	0.1/5.0	2
Medium	330/35.7	28	0.92	1.0/4.0	28
High	330/18.2	55	1.82	4.0/1.0	216

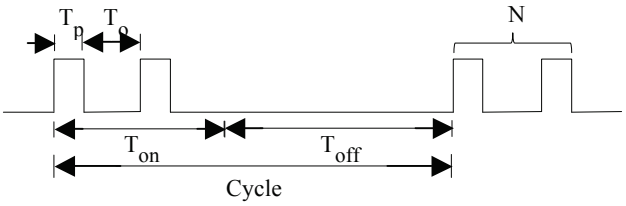

Fig. 1. Pulse train definition.

The batteryless implant harvests the electromagnetic energy and directly generates stimulation pulses. The PIC turns off when the device receives insufficient energy or when the transmitter coil is removed from the body. Both battery-based and batteryless devices were made on miniature printed circuit boards, with AWG-wire coil antennas wrapped around the boards. The boards were then packaged in polydimethylsiloxane coating.

Two wires extending from the implant formed the electrodes. The packaged prototypes have sizes of 12×37×9 mm^3. Figure 4 shows the size comparison of our device with the commercially available Enterra® device.

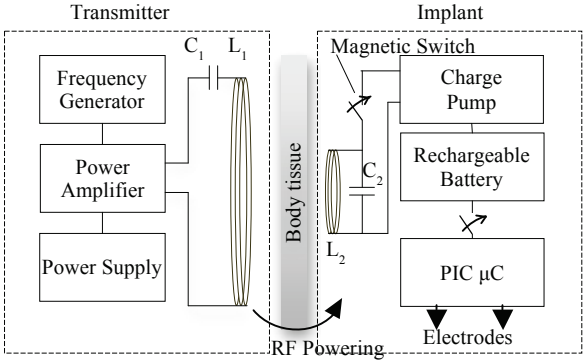

Fig. 2. Configuration of the battery-based wirelessly recharging system.

III. Experiments and Results

Bench-top and animal experiments were performed to demonstrate the feasibility of wireless GES. Animal Control Board of the University of Mississippi Medical Center (UMMC) approved the study. The device was endoscopically implanted in a 100-lb anesthetized pig with two sets of unipolar temporary cardiac pacing leads (model 6414-200, Medtronic). The leads were attached to the gastric mucosa or serosa. The Enterra® device was initially connected to the external leads and recorded before it was replaced by our battery-based and battery-less prototypes separately. The tissue thickness between the external transmitter coil and the implant was about 3 cm. Electrogastrograms (EGG) of stomach were recorded at different stimulation settings.

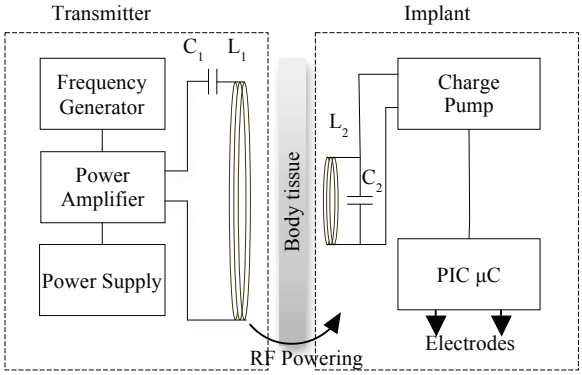

Fig. 3. Configuration of the batteryless wireless system.

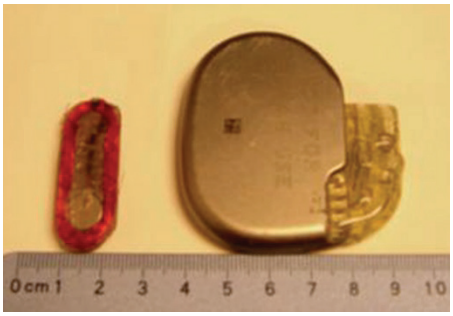

Fig. 4. Size comparison of our wireless gastrostimulator with Enterra® device.

The serosal and mucosal DC impedances were measured as 1179 Ω and 594 Ω, respectively. The voltage outputs of the stimulator and the electrical current delivered into the tissues were measured via wires connected at the electrodes. The EGG recordings were analyzed by signal averaging for mean frequencies and amplitudes, as well as for frequency-to-amplitude ratios (FARs). Figure 5 shows the voltage waveforms recorded in one of the serosal stimulations with battery-based device at low, medium and high settings. The measured electrical currents were 1.93 and 2.26 mA for battery-based and batteryless devices in the serosal stimulations,

respectively, while they were 3.45 and 5 mA in the mucosal stimulations. The currents delivered to the serosal areas were lower than those to the mucosal areas. This was due to the higher impedance in the serosa. The instant powers were 4.39 and 6.02mW for battery-based and batteryless devices, respectively, for the serosal stimulations while they were 7.07 and 14.85 mW for battery-based and batteryless devices, respectively, in the mucosal stimulations. The EGG measurements shown in Table 2 were to confirm stimulation effects on gastric activities such as rhythm frequency ranges, and amplitude ranges of the EGG signals. Control parameters were obtained with the stimulators turned off. Frequencies and mean amplitudes of the signals did vary at different stimulation settings. All rhythms were regular as expected. Frequency-to-amplitude ratios (FARs) were calculated from mean frequencies and mean amplitudes over the 10-minute periods and indicated significant changes in gastric activities during stimulation as compared to the control. It should be noted that the FARs for control varied during the experiments. This was due to the stomach adapting to the stimulation during the short duration of the experiment. The experiments indeed showed that the stomach EGGs could be modulated in different settings.

The average power for each pulse can be calculated as the energy for each pulse divided by the pulse period:

$$P_{av\_p} = \frac{E_p}{T_p + T_0} \qquad (1)$$

where $E_p = P_p \times T_p$ and $P_p = I^2 \times R$. The average power delivered to the tissues during the ON time indicated the intensity of stimulation:

$$P_{av\_on} = \frac{E_c}{T_{on}} \qquad (2)$$

where $E_c = E_p \times N$ is the total energy delivered. The average power for the whole cycle was

$$P_{av\_c} = \frac{E_c}{T_{on} + T_{off}} \qquad (3)$$

In the serosal stimulation, the intensities of stimulation at low, medium and high settings were 20.20, 40.22, and 78.21 μW for battery-based and 27.20, 55.15, and 107.24 μW for batteryless devices, respectively. The average powers were 0.57, 8.12, and 62.61 μW for battery-based and 0.78, 11.13, 85.85 μW for batteryless devices, respectively.

For stimulation in mucosa tissue, the intensities of stimulation at low, medium and high settings were 32.53, 64.76 and 125.91 μW for battery-based and 68.32, 136.01 and 246.46 μW for batteryless devices, respectively. The average powers were 0.91, 13.07 and 100.79 μW for battery-based and 1.92, 27.44 and 211.70 μW for batteryless devices, respectively.

TABLE II
RECORDED EGG SIGNALS

Stimulator	Battery	Setting	Frequency (bpm)	Amplitude (mV)	FAR (bpm/mV)
Control		Off	3.0−4.0	0.15−0.18	22.94
#1	Rechargeable	Low	3.0−4.0	0.20−0.21	17.50
#2	Rechargeable	Medium	3.0−3.3	0.15−0.20	18.10
#3	Rechargeable	High	3.5−4.0	0.10−0.18	26.80
#4	Batteryless	Medium	3.0−3.3	0.05−0.09	44.00
#5	Batteryless	High	3.0−3.0	0.12−0.15	21.43
Control		Off	3.0−3.0	0.08−0.10	33.33
#6	Batteryless	Medium	3.0−3.0	0.06−0.07	50.00
#7	Batteryless	High	3.0−5.0	0.08−0.13	38.30

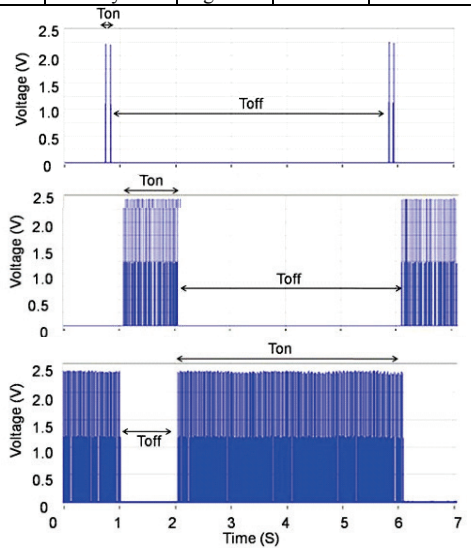

Fig. 5. Stimulator outputs for low (top), medium and high (bottom) settings.

IV. WIRELESS POWER TRANSFER

In the *in vivo* animal experiment, the abdominal wall of the pig was 3cm thick. To compare, the distance between transmitter and implant for the benchtop experiment was kept at 3cm in air. The attenuation of RF energy reduced the stimulation currents to 77% from 2.93 mA to 2.26 mA, and to 76% from 6.56 mA to 5 mA for serosal and mucosal cases, respectively. Nearly 41% (3.8dB) power loss occurred in 3-cm thick tissues for this particular case. The result indicated that, for human applications, the transmitter need to be designed with a variable range of transmission powers to address power losses in different body types.

The effective coverage area was mapped by moving the implant two dimensionally at the plane of interest, which is parallel to the transmitter coil. The receiver coil has a cross section of 10×35 mm^2. Because it is very difficult to conduct such experiments inside the animal body, the tests were conducted in air by assuming a power attenuation of 1.27dB/cm in the tissue. The expected received power

then can be estimated from the coupling loss due to misalignment between transmitter and receiver coils and the tissue loss.

Figure 6 shows the maps of harvested power that can be delivered into tissues, after taking the estimated tissue losses into account, within ±10 cm from the center of coil. The misalignment powers will be sufficient to deliver more than 2-mA currents into mucosal tissues at distances of 2, 3, 4, 5 cm within areas of radii ± 4.35, 4.93, 5.06 and 4.70 cm, respectively. When the current requirement is increased to 4mA, the working areas reduce to radii of ± 3.93, 3.97, 3.61 and 2.11 cm, respectively, for tissue thickness of 2, 3, 4, 5 cm. These estimations assumed that the tissue attenuation is linear with thickness without considering anatomic variations. It is clear that the wireless power transfer efficiencies are low due to the significant size difference in transmitting and receiving coils. However, if the transmitting coil size is reduced, the coverage area will also become smaller adding difficulty in practical settings since the implant is not visible for doctor or patient to align the transmitting coil.

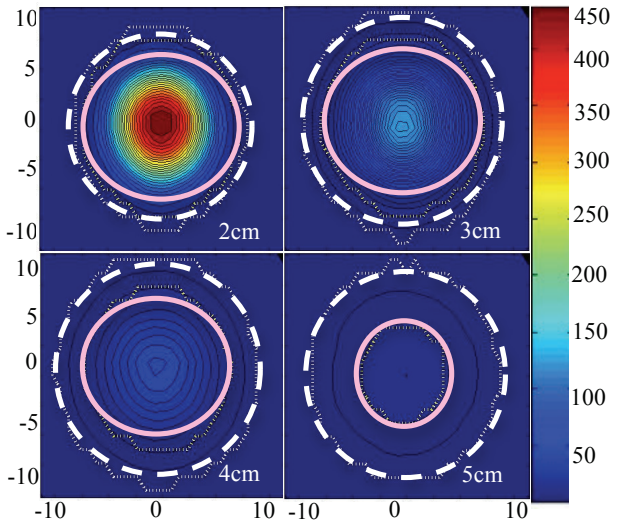

Fig. 6. Received power distributions in the receiver planes at spacing distances of 2, 3, 4, and 5 cm. The delivered currents were greater than 2 mA and 4mA inside the circles with dashed lines and circles with solid lines respectively. The unit is mW.

V. CONCLUSION

In this study we successfully developed and characterized battery-based and battery-less wireless gastric stimulators. These stimulators achieved modulation of myoelectric activities in the animal stomach. Utilization of these miniature devices may eliminate need for repeated surgeries for battery replacement. The small size of the stimulators also makes it feasible to implant them through endoscopic surgeries. Wireless power transfer for charging or powering has been initially investigated. Future works include detailed studies of the coil designs, optimization of the wireless power transfer system and freely-behaving animal experiments.

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