34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

A Microwave Powered Injectable Neural Stimulator

Bruce C. Towe, *Member, IEEE,* Patrick J. Larson, and Daniel W. Gulick

Abstract— **An unexpectedly simple implantable device that can achieve wireless neurostimulation consists of a short 1 cm long dipole platinum wire antenna, a Schottky diode, and a pulsed microwave transmitter. Fabricated into a 1 cm long by polyimide tubing, the implant can have a sub-millimeter diameter form factor suited to introduction into tissue by injection. Experiments that chronically implant the device next to a rat sciatic nerve show that a 915 MHz microwave transmitter emitting an average power of 0.5 watts has an ability to stimulate motor events when spaced up to 7 cm from the body surface. Tissue models consisting of saline filled tanks show the possibility of delivering milliampere pulsed current to neurosimulators though 5 centimeters or more of tissue. Such a neurostimulation system driven by microwave energy is limited in functional tissue depth by microwave SAR exposure. This report discusses some of the advantages and limitations of such a neurostimulation approach.**

I. Introduction

Shrinking the size of neurostimulation devices is desirable to reduce the trauma of surgery and the discomfort of a large implant. If neurostimulators could be implanted through a syringe needle, it would be a tool for neuroscience research and perhaps for medical therapeutic applications. Powering such small devices however is generally a limiting problem since the usual approach with magnetic induction requires relatively large loop coils in order to adequately couple to devices at multi-centimeter order depths in tissue.

In this work we present the concept of power transfer to very small implanted stimulators at microwave frequencies. The approach uses short linear dipolar antennas within the implant and rectification of the microwave pulses to produce neurostimulation events.

II. Background

Power transfer to implanted bioelectronic devices typically employs magnetic coupling that is dependent on the mutual inductance between an implanted coil and an external coil. Characterization of this coupling process dates back to Heetderks [1] who described resonant inductive coupling by coaxial coils through tissue at frequencies up to 30 MHz.

Perhaps the most well known application of inductive coupling principles to small implanted neurostimulators is the BION [2]. A problem is that efficient magnetic coupling to deeply implanted devices need relatively large implanted coil loop areas whose size requires open surgery to implant.

Manuscript received March 29, 2012. Authors are with the Harrington Biomedical Engineering, School of Biological and Health Systems Engineering, Arizona State University. (email:bruce.towe@asu.edu). Project supported in part by NIH grant 5R21NS059815-02 and Arizona State University.

There have also been reported other energy transfer design approaches that have included ultrasound coupling [3], floating light-activated devices [4] and an RF microstimulator array [5].

The use of microwave energy has been considered by Poon [6-8] for power transfer in biomedical applications. Reported optimal energy transfer for various tissues occurs in the GHz microwave region. At elevated frequencies, coupling by way of the electric field rather than magnetic becomes important and the rules that govern capacitive coupling and antennas become important.

A. Microwave Powered Neurostimulator

In this work we investigate an approach to the design of very small implantable neurostimulators. The basic design strategy is to capture microwave energy applied to the body surface through a short dipole antenna within the implant, rectify it with a diode, and then apply it to platinum electrodes. Thus rather than using the microwave energy to power a circuit that internally forms neurostimulation pulses, the implant is a type of RF probe that is dependent on the applied microwave energy pulse generation circuitry for neural stimulation parameters of amplitude and pulse characteristics.

B. SAR limits

For non-ionizing radiation, the maximum safe electric field in tissue is limited by the specific absorption rate (SAR). Under US regulations, exposure to the head and torso is limited to 1.6 W/kg, averaged over a gram of tissue over six minutes [9].

The SAR is dependent on the applied RF power and the duty cycle. For neurostimulation applications the duty cycle is on the order of 1-2%. For example a nerve neurostimulation protocol might have 200 μs pulses at 50 Hz, so having a 1% duty cycle. Thus in principle, 50 watt peak pulses at up to a 1% duty cycle can be used to help overcome path losses and still be comparable or less than 1 watt cell phone average power levels.

In addition to safety limits, RF transmission is also subject to FCC regulation in the US and most countries have similar regulations. Part 18 (47 CFR 18) allows for unlicensed use of the ISM (industrial, scientific, and medical) bands, including 902 to 928 MHz in the U.S. and 2.4 to 2.5 GHz internationally.

C. Implant Circuitry Design

A Schottky microwave diode bisecting a short length of platinum wire forms a dipole antenna that rectifies the incoming RF and produces a monophasic DC potential across its junctions. Its DC output over a range is linearly proportional to microwave intensity however diodes have conduction thresholds and so low threshold voltage Schottky diodes are desirable to reduce losses and are more efficient.

We find that application of rectified RF will stimulate nerve even without a smoothing capacitor to rid the GHz pulsation. Tests show that the nerve responds to this signal as it does to a similarly shaped DC pulse waveform having an amplitude equal to the DC average of the rectified GHz wave.

III. METHODS

The neurostimulator device shown in Fig. 1a was made of 0.8 mm (outer diameter) polyimide tubing, encapsulating a 100 micron platinum wire 1 cm long dipole antenna with a bare 300 µm square Schottky diode die (Skyworks CDC-7621) connected in the center. The antenna wire tips terminate in roughly 1mm diameter platinum balls that extend beyond the insulation and perform the function of stimulating electrodes.

The platinum balls were made first by melting 100 micron diameter platinum wire into a ball using an electrical arc and leaving the formed ball suspended from the wire. The wires were then bonded to the diode die using silver conductive epoxy and the internal tubing lumen filled with EpoTek #301 medical epoxy and baked according to manufacturer recommendations.

Two RF exciters were used, one at 915 MHz and a second at 2.45 GHz. Both were obtained surplus as part of cell phone base station RF driver amplifier boards. The amplifier circuit boards were removed and driven by an HP8657d RF pulse generator tuned to their respective frequencies and by a ZHL-42 (Minicircuits Inc.) preamplifier. Their power output was adjusted to produce the same level from both boards as measured by a spectrum analyzer.

Each exciter system produced pulses up to 50 W at 100-750 µs duration with an adjustable repetition rate. At 10 Hz, their average power output under any combination of conditions was kept to less than one watt. Figure 1b shows a portable microwave generator box and transmitting dipole antenna.

A. *Saline Tank*

Tests assessed the performance of the system to generate stimulator current in a tissue-mimicking salt water tank as a function of separation distance from the exciter antenna. We used an aquarium of saltwater (Fig. 2) of 8 mS/cm conductivity to approximate an average of body tissue and muscle. The microwave antenna was placed against the tank wall and tuned in length over the range of 13-15 cm for maximum output.

Determining the current generated in the solution was accomplished by placing two recording electrodes near the device and measuring the amplitude of the resulting voltage.

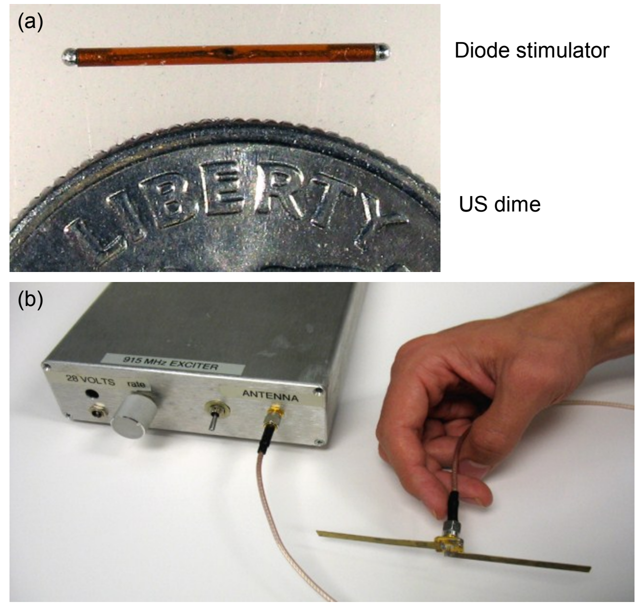

Fig. 1. (a) Implantable micro-stimulator showing the polyimide outer tube and platinum-iridium ball electrodes on either end. (b) Microwave driver at 915 MHz and antenna.

The system was then calibrated by replacing the device with a similar platinum electrode pair having the same spacing and position. These electrodes were driven by an isolated constant current stimulator (A-M Systems Inc. 2100) to produce known currents in solution. The corresponding voltage output of the recording electrodes then allowed calibration of solution voltage to device current.

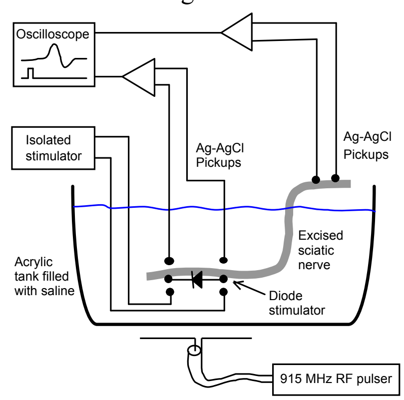

Fig. 2. Apparatus for testing current output of the wireless stimulator used also for stimulating amphibian sciatic nerve.

Recordings were taken as a function of distance of the device from the antenna over the range of 4 to 15 cm and at the two different ISM frequencies. Closer approach than about 4 cm risked damaging the Schottky diode by overcurrent.

B. *Animal Experiments*

An in-vivo experimental protocol was approved by an institutional IRB. A length of excised frog (*Rana Castebeiana*) sciatic nerve was used for testing. Part of the nerve was fully immersed in the saline tank setup and the neurostimulator device placed in contact with its surface (Fig 2.) Pulses of microwave energy at 250 µS duration emitted at the tank wall antenna produced evoked CAP events as monitored by silver chloride electrodes placed on part of the nerve emerging out of the tank.

A 250 gram Sprague Dawley rat was surgically implanted with a 0.8mm x 15 mm long device next to the sciatic nerve in the lower hind limb and nestled between the folds of the muscles. The device was not anchored to the nerve and the rat recovered.

RF pulses at 915 MHz were applied to the hind limb by a dipole antenna held up to 7 cm away in air from the implant site (Fig. 3). Data was recorded from EMG electrodes attached to the hind limb as well as from a strain gauge force transducer attached by a thread to the rat foot.

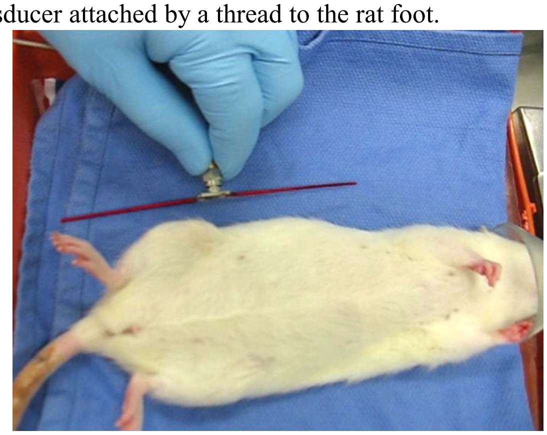

Fig. 3. Testing the implanted stimulator *in vivo*. This is a frame from a video showing the leg of the rat twitching as the microwave antenna is placed in proximity to the hind leg. This test was performed 13 weeks after implantation.

IV. Results

Fig. 4 shows a compound action potential (CAP) from the frog sciatic nerve evoked by the neurostimulator driven at 915 MHz. The upper trace shows a stimulus artifact (left) and the evoked CAP (right). The lower trace is a microwave RF probe output used for timing reference.

At 50 watt pulse powers we find that the nerve would respond to 750 µsec 915 MHz microwave pulses though solution path lengths up to 12 cm. The primary variables are the dipole length and solution conductivity. Lower solution conductivities (not shown) representing fat for example, give greater ranges than this.

Fig. 5 shows the result of the tank study with the current flow generated by the diode as a function of device distance from the antenna at the tank wall. It can be seen in this test that mA-order currents can be generated by the system at 4-8 cm distances. The greater attenuation at 2.45 GHz is apparent (Fig. 5) but the slope of the curve is faster rising than at 915 MHz suggesting advantages to the use of the higher frequency for nearer surface applications.

Fig 6. Shows the EMG and strain gauge evidence of the hindlimb forceful contraction in response to the microwave pulse.

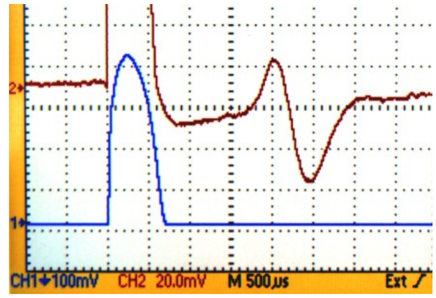

Fig. 4. Oscilloscope trace of a wireless stimulator inducing excitation of a sciatic nerve. The lower trace shows the stimulus monitor, and the upper trace shows the recording from the distal end of the sciatic nerve.

The choice of 915 MHz as the exciter frequency for the animal experiments was determined mostly due the reduced path length losses relative to 2.45 GHz.

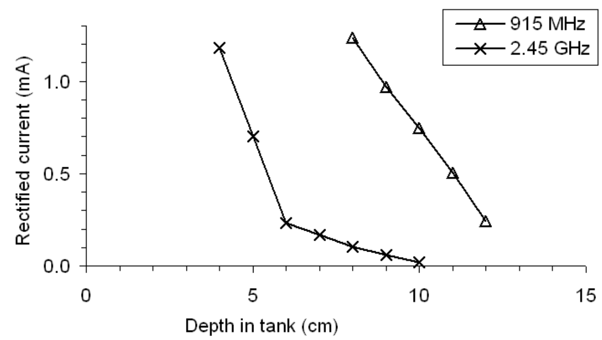

Fig. 5. Maximum current from stimulator versus depth in a salt water tank.

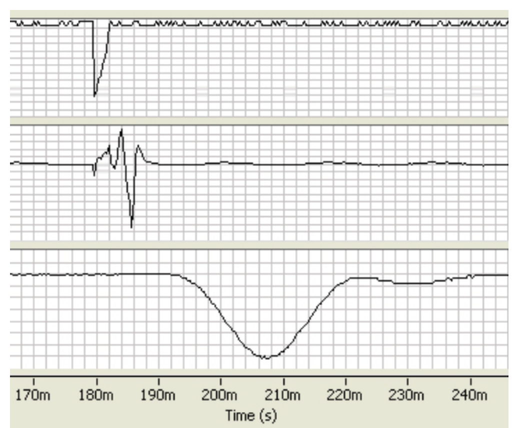

Fig. 6. Rat hindlimb force generated by stimulation. (Top) timing event showing the onset of the RF pulse. (Middle) EMG waveform. (Bottom) muscle force, as recorded by a strain gage.

Although no definitive SAR studies were conducted as part of the tests at 915 MHz, no detectable temperature rise to the nearest 0.1 C was measureable by a thermocouple placed on skin at peak pulse power levels of 50 watts at 100-750 µsec at 10 Hz.

The use of a polyimide and medical epoxy as an encapsulant is not optimal but in combination worked successfully over a 12 month implantation time in one rat. No corrosion was noticed around the anode of the diode system.

Although the device was not anchored after implantation within the hind leg such as through the use of a nerve cuff [3], and the rat was ambulatory for a year, the migration of the device amounted to about 2 mm that still left it functional for the full duration. We suspect that the device small size and lack of lead wires tending to pull on the device is the reason for its relative immobility.

V. Discussion

In this work we investigate the use a microwave diode connected to two short platinum wires to form a simple and compact neurostimulator. Despite large inefficiencies in microwave coupling due to a non-resonance of the implant at both 915 MHz and 2.45 GHz, the delivered current levels were observed to reach several milliamperes at depths on the order of 7 cm centimeters. These current levels are comparable or greater than that needed for many neurostimulation applications.

Such a system might be useful for example in research investigations with small animals where low repetition rate monophasic DC pulses can be tolerated. A charge recovery capacitor on the electrode output or a charge balancing circuit however is clearly desired for most applications but was not evaluated as part of this work. In application, the device size for example suggests that it might be wholly integrated into a closed rat skull. The simplicity of the neurostimulation system would not be outside the resources of many university research labs.

This work is consistent with the possibility that useful levels of stimulation current could be delivered at depths adequate for some neurostimulation applications on humans. SAR would be the primary limiting factor for any specific combination of implant depth, pulse width, and pulse repetition rate.

There are multiple variables in the neurostimulator device design with a primary one being its overall length. In general longer implant dipoles than the 1-1.5 cm used in these experiments and ranging up to 3 cm length offer 50% or more higher current for other conditions held constant. No frequency resonance peaking was noticed at any implant antenna length tested. We suspect that the mutual coupling between the external and implanted antennas is mostly capacitive in nature but this requires a more rigorous study.

Multichannel systems however are not easily accomplished and would depend on the spatial separation of the devices and the use of multiple external antennas if there was not the addition of implanted digital logic address circuitry.

As pertinent to the simple system described here, basic limitations in a practical sense result from the lack of implant current control. This kind of design strategy involving very few implanted components sacrifices local control over stimulation current parameters to achieve simplicity and miniature size. Variations in antenna coupling positioning for example cause changes in implant current. The feedback that regulates the stimulation current is the operator increasing the exciter power level to achieve the desired physiologic effect. Although perhaps adequate for some non-demanding applications, methods for more general application need to be developed to remotely monitor and control the current output of the implant.

The choice between the microwave-range ISM bands (915 MHz in the US and 2.45 GHz internationally), depends on the application. Low frequencies are less attenuated by tissue, and so are better suited for neurostimulator placement more than a few centimeters deep. Higher frequencies have higher coupling efficiency in the near field and allow for efficient use of short antennas.

VI. Conclusion

Implantable neurostimulators of small form factor can be powered by microwave energy using electric field coupling. The tested implant employs a short dipolar antenna that doubles as a neurostimulator electrode system. The system achieved rat sciatic nerve stimulation with driving antennas spaced 7 cm away in air. Milliampere-order neurostimulation currents were achievable at depths up to 9 cm in tissue phantoms. This approach appears attractive for a number of research and medical therapeutic neurostimulator applications. The further development of such microwave powering approaches may allow new designs of neuroengineering tools and bioelectrical implants.

References

[1] W.J. Heetderks, "RF powering of millimeter- and submillimeter-sized neural prosthetic implants," *Biomedical Engineering, IEEE Transactions on* , vol.35, no.5, pp.323-327, May 1988.

[2] G.E. Loeb, R.A. Peck , W.H. Moore, K Hood , BION(TM) system for distributed neural prosthetic interfaces, Medical Engineering & Physics, Volume 23, Issue 1, pp. 9-18. January 2001.

[3] P. Larson, B.C. Towe, "Miniature Ultrasonically Powered Wireless Nerve Cuff Stimulator", Proceedings of the 5th International IEEE EMBS Conference on Neural Engineering, Cancun, Mexico, April 27 - May 1, 2011

[4] A. Abdo, M. Sahin, "Feasibility of Neural Stimulation with Floating-Light-Activated Microelectrical Stimulators", IEEE Trans. Biomed. Circuits Systems, Vol. 5. No. 2., April 2011.

[5] G. Suaning and N. Lovell, "CMOS Neurostimulation ASIC with 100 Channels, Scaleable Output, and Bidirectional Radio-Frequency Telemetry", Biomedical Engineering, IEEE Transactions on, vol. 48, no. 2,pp. 248 –260, 2001.

[6] A.S.Y. Poon, S. O'Driscoll, and T. H. Meng, " Optimal Frequency for Wireless Power Transmission into Dispersive Tissue" , IEEE Trans. Antennas and Propagation, vol. 58, no. 5, pp. 1739–1749, May 2010.

[7] A.S.Y. Poon, A mm-Sized Implantable Wireless Power Receiver" Proc. IEEE Engineering in Medicine and Biology Society Annual International Conference (EMBC), Minneapolis, Mn, Sept. 2009 (invited).

[8] A.S.Y. Poon, "Miniaturization of implantable wireless power receiver," *Engineering in Medicine and Biology Society, 2009. EMBC 2009. Annual International Conference of the IEEE* , vol., no., pp.3217-3220, 3-6 Sept. 2009.

[9] IEEE Std C95.1™-2005 "Standard for Safety Levels with Respect to Human Exposure to Radio Frequency Electromagnetic Fields, 3 kHz to 300 GHz".