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Wireless Ultrasound-Powered Biotelemetry for Implants

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***Abstract*— A miniature piezoelectric receiver coupled to a diode is evaluated as a simple device for wireless transmission of bioelectric events to the body surface. The device converts the energy of a surface-applied ultrasound beam to a high frequency carrier current in solution. Bioelectrical currents near the implant modulate the carrier amplitude, and this signal is remotely detected and demodulated to recover the biopotential waveform. This technique achieves millivolt sensitivity in saline tank tests, and further attention to system design is expected to improve sensitivity.**

I. Introduction

WIRELESS biotelemetry is desirable for sensing bioelectrical events in the brain and peripheral nerves. These neural signals could potentially be used to control advanced neuroprosthetics.

Powering wireless implants is a significant problem in the development of neuroprosthetics. Current telemetry approaches use complex on-board electronics powered by induction coils. In addition to the relatively high power requirements of these implanted systems, heat dissipation is also a concern. Alternatively, passive telemetry systems such as RFID tags and card-key entry systems can reduce power needs by employing reflected impedance modulation. A passive approach minimizes power consumption in part because it omits the usual oscillator within the implant. The power received by the device is modulated by the signal of interest and then re-radiated.

Here we look at a passive approach to wireless telemetry of bioelectric events using MHz-range ultrasound energy as a power source for an implantable device. The device design is simplified by using the local bioelectrical signal to directly modulate the MHz carrier, much like an audio signal modulates a carrier in AM radio communications.

This technology has the potential to be extended to multichannel operation and provides key advantages over conventional integrated circuit technology in terms of device simplicity and power requirements.

II. Background

Wireless biotelemetry approaches typically have used radiofrequency (RF) magnetic induction to power integrated circuitry implanted in the body. Using the principle of magnetic induction, RF power is transmitted across the skin to a receiver coil. The AC power from the coil is converted to DC and used to supply amplifiers, multiplexers, and RF oscillators and can perform telemetry over meter-order distances.

Wise et al. reviewed the state of the art in wireless implantable microsystems [1]. Irazoqui-Pastor, Mody, and Judy investigated a miniature EEG recording system to transmit neural signals as weak as 15 microvolts to a remote receiver using a voltage-controlled oscillator [2]. Mosheni et al. reported multichannel unit recording using a digital strategy, detecting simulated 80 microvolt neural spikes [3].

The increasing sophistication of implanted neural telemetry devices requires more wireless channels, and the power demands and heat dissipation in these high density multi-channel systems are recognized as significant and perhaps limiting design issues.

Little attention has been paid to the use of ultrasound energy for transferring power to biomedical implants. Ultrasound can easily be generated by a compact system and transferred through the skin using a small transducer. The mechanical wave energy is received by a piezoelectric element and converted to electrical energy to power the implant. We have observed that pulsed ultrasound is capable of producing milliamp-order currents in piezoelectric devices a few square millimeters in size [4].

We investigate the potential for a small implantable biotelemetry device that is ultrasound-powered and could be introduced into tissue by minimally invasive techniques. In the present implementation, the received ultrasound carrier frequency is amplitude modulated by local bioelectric events. Modulation is accomplished via the nonlinear I-V characteristic of a semiconductor diode. The modulated carrier is conducted through the surrounding tissue, and this signal is detected on the body surface. Subsequent signal processing reproduces the bioelectric waveform.

III. Methods

We developed and tested a small, wireless, potentially implantable ultrasound-powered biotelemetry device. A saline tank was used as a model for the conductivity of human tissue. The experimental setup is shown in Figure 1. A device incorporating the piezoelectric copolymer poly(vinylidene fluoride-trifluoroethylene) P(VDF-TrFE) was submerged into the saline. Electrodes carrying a 100 Hz test signal simulating a bioelectrical current source were placed in the tank, nearly touching the device electrodes. For calibration purposes the test signal was directly monitored as it appeared across the device electrodes. The device was illuminated with tone bursts of 1.7 MHz ultrasound, and the induced 1.7 MHz carrier current emitted

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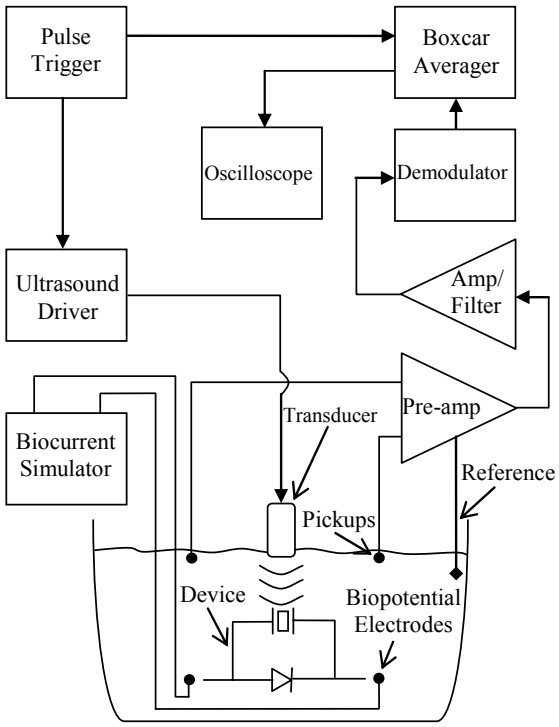

Fig. 1. Diagram of experimental setup.

from the device was picked up by remote electrodes at the surface of the volume conductor. The received signal was then demodulated in order to reconstruct the original bioelectric waveform.

A. Ultrasound Pulse Creation

Ultrasound pulses were generated at a peak intensity of 8 W/cm^2 at 1.7 MHz using a custom 14-mm diameter unfocused piezoelectric transducer placed at the surface of the saline. The emitter was driven at 140 volts peak-to-peak (Vpp). Ultrasound generation was triggered by a pulse stimulator (A-M Systems 2100, Sequim WA) gating a function generator (Wavetek 145) to produce a tone burst feeding into a custom ultrasound driver. The burst was 2 µs long and was delivered at a 9 kHz repetition rate.

B. Device Construction and Energy Conversion

Ultrasound was received at the device by a P(VDF-TrFE) polymer film 90-µm thick (Ktech Inc.). Approximately 15 mm^2 of film was folded into a five-layer stack about 0.5 mm thick, 3 mm long by 1 mm wide. Folding allowed more film to be incorporated into a smaller form factor; greater surface area yields greater current output. A zero-bias diode (Skyworks CDC-7630) was connected in parallel with the piezopolymer by using silver epoxy, and the encapsulant was generic water-proof epoxy. Figure 2 shows a similar prototype that is slightly smaller than the device described and tested here.

The piezoelectric material converts the pressure of the ultrasound wave to an oscillating electrical potential at the ultrasound frequency. Figure 3 shows how an ultrasound

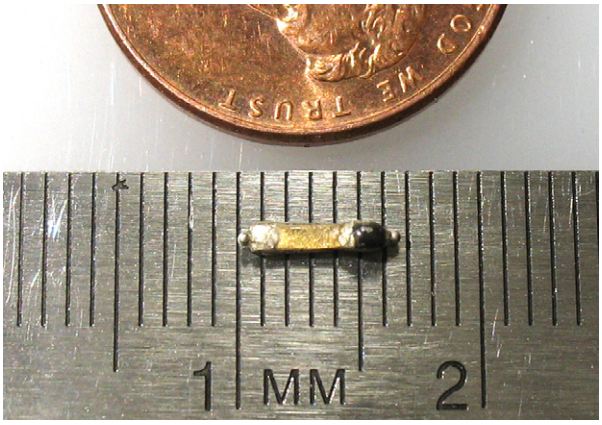

Fig. 2. Photograph of a completely integrated ultrasound powered biotelemetry device with small platinum-iridium ball electrodes on each end.

tone burst with a peak intensity of 8 W/cm^2 produces an *open-circuit* response in the piezopolymer of about 5.4 Vpp.

In the completed device the electrodes were two 1-mm diameter platinum-iridium balls located 1 cm apart that presented a total 140 Ω impedance at 1.7 MHz when loaded by the saline solution. With ultrasound bursts of 8 W/cm^2 (near medical diagnostic ultrasound power levels), at 5 cm immersion depth the device generated current through solution of 3.9 mApp at a voltage of 550 mVpp.

C. Signal Intermodulation at the Device

A function generator (Leader LFG-1300S) connected through an isolation transformer delivered simulated biocurrents to the solution near the device. These biocurrent electrodes were placed immediately beside the device electrodes, nearly touching them. The diode connected in parallel across the P(VDF-TrFE) modulated the local biopotential onto the high-frequency carrier wave in the same way mixer diodes are used in AM radio communications. The ability to do this depends on the diode's nonlinear I-V characteristics. In this sense the diode's forward resistance can be seen as a function of the biopotential amplitude. The diode variably shunts a portion

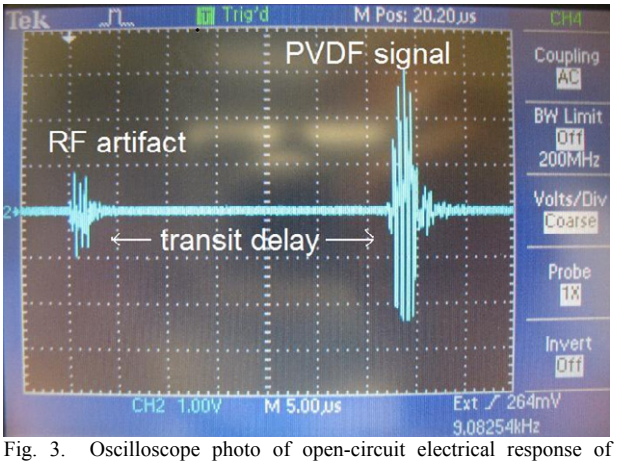

Fig. 3. Oscilloscope photo of open-circuit electrical response of piezopolymer to an ultrasound tone burst.

of the current supplied by the P(VDF-TrFE), so that the remaining current driven through the electrodes and saline is amplitude modulated by the biopotential.

D. Signal Detection at the Surface

Currents driven in solution by the device created electric fields on the surface of the volume conductor that were detected by Ag/AgCl electrodes (IVM, Healdsburg CA). These electrodes were oriented parallel to the dipole axis of the device. The solution was saline at 0.9% NaCl to simulate physiological conductivity. The received signal was preamplified at 8 times gain by a broadband differential amplifier (AD 8251, Analog Devices, Norwood, MA).

E.Demodulation of Surface-Electrode Signals

After preamplification the received pulse was bandpass filtered from 300 kHz to 5 MHz with a gain of 40 dB (Panametrics 5800, Waltham MA). Filtering was crucial to prevent interference from the artificial 100 Hz biocurrent delivered into the solution. The gain allowed a simple diode-capacitor peak-detecting circuit to be used as a demodulator. The time constant of the demodulator was adjusted to ensure a decay to zero prior to the arrival of a subsequent tone burst.

A boxcar averager (EGG 162, Princeton Applied Research, Oak Ridge, IL) was used to sample the demodulated pulse. A single 15-ns sampling aperture was opened for every ultrasound tone-burst. The aperture timing delay was adjusted such that it matched the transit-time of the ultrasound pulse. The propagation time to reach a 5 cm depth was about 35 μs.

IV. Results

The amplitude of the 1.7 MHz carrier signal was directly influenced by the simulated biopotential. The biopotential modulated the ultrasound-driven carrier current that was delivered to the device electrodes, and this carrier was passed through the volume conductor. Our instrumentation successfully detected by the signal via remote surface electrodes and demodulated the carrier to recover the biopotential waveform.

Figure 4 shows a single tone burst and sampling point. The bottom trace is the surface-detected 1.7 MHz carrier waveform after amplification and bandpass filtering. The output of the diode-capacitor demodulator is shown in the middle trace, which is a rectified and lowpass filtered version of the bottom trace. The amplitude was observed to rise and fall in synchrony with the simulated biopotential. The top trace shows the boxcar aperture spike, where the leading edge denotes the moment the aperture was opened to sample the level of the demodulated pulse.

It is notable that the adjustable aperture timing allowed the aperture to sample the pulse or exclude it depending on the adjustment of the delay. If multiple telemetry devices were placed at different depths, this aperture adjustment would allow sampling of a specific device based on the corresponding delay. Aperture-based sampling also excluded RF interference from the ultrasound emitter. (The RF artifact is visible in Figure 2).

Figure 5 shows the output from the entire signal processing chain. The upper trace is the simulated biopotential delivered into the saline solution near the device electrodes. The lower trace is the output of the boxcar averager, which reconstructed the biopotential by sampling the level of each demodulated tone burst. Using 9,000 tone bursts (or samples) per second, the bandwidth was 4.5 kHz. Although a lower sampling rate would have been adequate for the 100 Hz biopotential, the high repetition rate was chosen to illustrate the possibility of reconstructing higher frequency signals such as neuron action potentials.

The smallest simulated biopotential detected using this technique was 9 mVpp. This detection limit was established at a 3:1 ratio of signal amplitude to noise amplitude.

V. Discussion

The electrical simplicity of this potentially implantable device is remarkable in comparison to integrated circuit technology. It lends itself to a very compact design.

The sensitivity of this system however falls short of the microvolt range that we would like to see for cortical

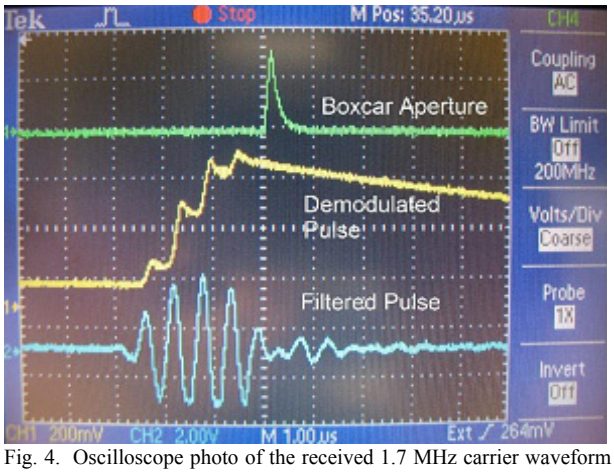

Fig. 4. Oscilloscope photo of the received 1.7 MHz carrier waveform and the result after demodulation. (Time scale = 1 μs/div.)

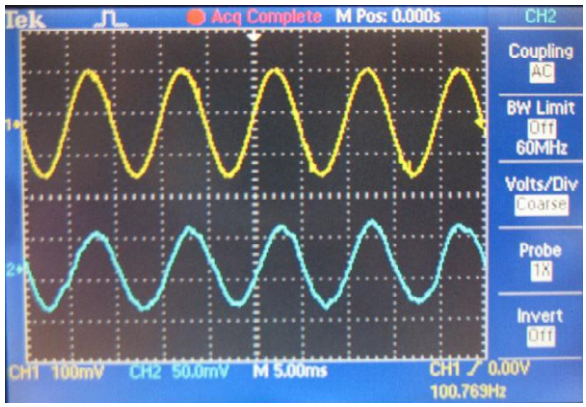

Fig. 5. Simulated biocurrent delivered to the device electrodes (top) and the waveform reconstructed from the demodulated tone bursts (bottom). (Time scale = 5 ms/div.)

neuroprosthetics. The present sensitivity, in the range of a few millivolts, appears to be limited by noise arising from both exciter and demodulator sources. This is encouraging since it does not appear that significant noise arises from the electrodes or the piezoelectric material in the implant. Noise from the RF mixer diode, the only active component, is well known from radio communications to amount to only a few microvolts at the bandwidth and source impedance used here. Thus our assumption is that most of the system noise arises somewhere in the instrumentation and so may be reduced with further attention to system design.

Ultrasound imaging is based on transit-time delays and electronic beam sweeping, and a similar interrogation scheme might be achievable with this biotelemetry system. Range gating and beam focusing could allow techniques to excite specifically located piezoelectric devices within a 3-D volume. Thus, ultrasound has a potentially powerful advantage in terms of addressing multiple small devices implanted in tissue. In principle such a system could address many channels in a neural prosthetic device by time- and space-division multiplexing.

This approach of intermodulation using a diode to encode a bioelectrical signal onto a high frequency carrier wave is an attempt to eliminate high-gain amplifiers and associated electronics that increase the size, complexity, power use, and heat generation of telemetry implants. The intermodulation process is well-known in the field of radio communications, where diodes serve as mixers to convert microvolt level radio signals down to baseband or up to carrier band.

The bioelectric waveforms are reconstructed from a sampling process at the ultrasound pulse repetition rate. The Nyquist bandwidth in the present study is 4.5 kHz, which is sufficient for many bioelectrical recording applications.

Power transmission using ultrasound and piezoelectric elements is not efficient, but because we can generate relatively strong pulses and safely transmit them through the body, the power received can be sufficient for the needs of implanted devices.

Propagation of high-frequency currents through the body is relatively well understood, as described by the volume conductor equation in which the amplitude of surface potentials measured from a current source at some depth in a conductor decline with the square of the depth. At the physiologic conductivity used in our study, several hundred millivolts generated across a 1-cm dipole a few centimeters deep results in several hundred microvolts detected by surface electrodes. Thus, communication is possible from small implanted devices to electronics placed on the surface.

This system then is a method of using a diode to encode a bioelectric event onto a driven carrier current. At the skin surface the electrodes sense the 1.7-MHz carrier that is amplitude modulated with the biopotential seen by the device at the implant location.

The limited dynamic range of the instrumentation system has been a principal constraint on sensitivity. This is problematic because the diode device in this implementation has no ability to provide signal gain, and the modulation index of the biopotential upon the carrier is extremely small. For example, a 10 μV biopotential modulating a 100 mV carrier signal would give at best a modulation index of only 0.01% (assuming ideal modulation). Thus in order to achieve the desired sensitivity the demodulation must be performed with very high precision, which is beyond the capability of our current system.

Future work will also examine the tone-burst repeatability of the ultrasound driver system. The 140 Vpp tone-burst applied to the transducer must be held to much less than 0.01% variability.

Ultrasound-powered biotelemetry systems are similar to passive RF systems in that they use the energy of the incoming waves. However, they have some significant advantages compared to RF technology. Ultrasound powering of small devices is relatively simple, and the current version of the biotelemetry implant has only two electrical components. Scaling the system to multiple channels can be achieved by range gating and ultrasound beam sweeping. Surface detection of volume conducted signals is easily accomplished, and the demodulation process allows easy separation of the signals from ambient bioelectric noise.

VI. Conclusion

We have demonstrated that ultrasound powered wireless telemetry may be a feasible technique for biopotential monitoring. Presently, this approach can detect millivolt-order simulated bioelectric events. The primary limitations on sensitivity seem to be noise in both the excitation and demodulation signal chains, which are amenable to improvement.

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