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1pEAb9. A modulator and driver for a high-efficiency ultrasonic link providing power and signal to an implanted hearing aid
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We present results from the design and testing of a high-efficiency amplitude modulation and demodulation circuit for use in an ultrasonic power link for implanted hearing aids. The modulator provides an 8 KHz modulation bandwidth around a carrier frequency of 1.024 MHz and achieves high efficiency from a switched driver design employing a 50 KHz ramp-compare pulse-width modulation scheme. It can deliver 27.2mW to an ultrasonic link with 76% efficiency. The design of an implanted demodulator circuit is also discussed. When coupled to a set of matched high-efficiency ultrasound transmitter/receiver transducers achieving a 45% transmission efficiency, this system holds promise as a compact, lightweight, highly directional means of providing power and signal to implanted devices.

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INTRODUCTION

Implanted hearing aids and cochlear implants have traditionally been powered through magnetic induction coils. Recent advances in acoustic power transmission have opened up the possibility of powering such devices using an ultrasonic power link. Such a link has the advantages of high directivity, a compact form factor and the possibility of eliminating implanted magnetic components that interfere with magnetic resonance imaging. With the advent of newer piezoelectric materials like lead magnesium niobate-lead titanate (PMNPT) which offer a high electromechanical coupling and low dielectric loss, a large power handling capability and high power transmission efficiency can be obtained even through several millimeters of soft tissue. In order to take advantage of this technology for implanted hearing aids, it is desirable to have the ultrasonic link carry not only power but also an acoustic signal with a bandwidth of 8 kHz or more. In the present work we will discuss progress in designing electronic circuitry to achieve efficient signal and power transmission on an acoustic link designed for integration into an implanted hearing aid system.

There has been considerable research into ultrasonic links for delivering power to implants in recent years, most comprehensively by Ozeri et al [1, 2]. A few groups have also developed systems capable of carrying both power and signal. Suzuki et al.[3] outline power and information transfer using lead zirconate titanate (PZT) devices and were successful in delivering power and data to an internal unit at a transmission speed of 9.6kbps. In [4] Lawry et al. designed and implemented a system for simultaneous high power transfer at 50W and uni-directional data transfer at 12.4Mbps. However, an appropriately sized system that provides acoustic signal and power to an implanted hearing aid using ultrasonic transducers, while also maintaining high efficiency, has yet to be considered to the best of our knowledge. The design proposed in this paper is an amplitude modulated system that delivers electrical power and signal with an 8 kHz bandwidth. The efficiency of the electronic driver/modulator was measured to be 76%.

CIRCUIT REQUIREMENTS & DESCRIPTION – MODULATOR

The design of a low-power transmitter that runs off a single CR2450 3.0V lithium coin-cell battery is presented along with an efficient rectifying demodulator that extracts the encoded audio signal. The CR2450 is rated for 620mAh and so can provide 11.25 mA of current for 8 hours a day for 1 week. This is sufficient to power many low powered active middle ear and bone conduction hearing aids that require total current draw from the transmitter of less than 4mA. Commercially available devices in this category include the Medel Vibrant Sound Bridge and the BoneBridge.

The electronic driver and modulator were designed for integration with an ultrasonic power link designed and fabricated in our laboratory. The link consisted of a matched pair of piezoelectric transducers made from PMN-32%PT. The transducers are 5 mm diameter unfocused 3-1 composite transducers[5] with a center frequency of 1.024MHz and an electromechanical coupling coefficient k_t=0.78 and capable of achieving a link transfer efficiency of 45%. The design and fabrication of these transducers is presented elsewhere in these proceedings[6].

In powering a piezoelectric ultrasound transmitter, the key consideration is the achievement of a high-efficiency radio frequency (RF) power oscillator that can be modulated by an acoustic signal. In our design we achieve this by employing a resonant LC bandpass filter coupled to a high-efficiency switched, variable duty cycle digital oscillator. The oscillator is modulated with an acoustic signal using a ramp-compare pulse width modulation (PWM) scheme.

The modulator and transmitter circuit is shown in Figure 1 with sub-circuits illustrated in Figure 2. The design uses a complementary pair of MOSFETs M1 and M2 to switch the input of an LC bandpass filter at frequencies close to the 1.024MHz resonant frequency of the transmit transducer. High-efficiency is achieved with a switching design because the MOSFETs are either in an “on” low resistance state in which case they drop very little voltage or in a “off” high resistance state in which they pass little current. In either state the dissipated power is low. Losses in such a configuration come from dissipation during the transitions, the gate charge required to turn the FETs on and dissipation due to the on-resistance.
The gate signals driving the FETs are generated from a ramp-compare PWM circuit consisting of the ramp generator shown in Figure 2 and the comparator U3. On the left hand side of Figure 1, the audio input is biased at half the DC supply voltage and is compared against a 50 KHz ramp signal. The comparator output provides a PWM digital signal whose baseband component is the audio signal.

**FIGURE 1,** Low power transmitter driver circuitry for delivering power and acoustic signal to an implanted hearing aid.

The PWM digital signal is fed to the input of a SN74LVC1G08 AND gate along with a clock signal at the transmit transducer center frequency to generate a 1.024MHz digital carrier wave with amplitude modulation sidebands at the audio signal frequency. The output from the AND gate drives the complementary pair of FETs. An LC 2\textsuperscript{nd} order type-I Chebyshev bandpass filter with a 16 kHz 3dB bandwidth was used to filter around the carrier frequency and to remove the unwanted modulation sideband due to the 50 kHz triangular wave. A 32 dB isolation is achieved between the carrier and the 50KHz sideband.

A computer-based KLM-type model\cite{7} was used to analyze the input impedance looking into the transmit transducer. Within 16 KHz of resonance the impedance of the transducer in water is real and equivalent to a 150Ω resistance labeled as R11 in Figure 1.

The subcircuits in Figure 2 consist of a micropower 1.024 MHz clock driving low power active op-amp integrator in order to generate a 50 kHz triangle wave. An appropriate $RC$ time constant was chosen such that the triangle wave swings from GND to the positive supply without clipping. A mid-point voltage reference is also provided to the operational amplifier in order to center the triangle wave at mid-supply.

**FIGURE 2,** Transmitter sub-circuits responsible for generation of a 50 kHz triangle wave and 1.024MHz clock.

**MODULATOR – CIRCUIT MEASUREMENTS**

A power budget listing the measured current draw of each component in the unloaded circuit is shown in Table 1.
TABLE 1. Current draw characterization for individual sections of the transmitter

<table>
<thead>
<tr>
<th>Purpose</th>
<th>Parts involved</th>
<th>Measured current draw</th>
</tr>
</thead>
<tbody>
<tr>
<td>50kHz triangle wave</td>
<td>LTC6900, LTC6246</td>
<td>1.09mA</td>
</tr>
<tr>
<td>Carrier wave at 1.024MHz</td>
<td>LTC6900</td>
<td>0.66mA</td>
</tr>
<tr>
<td>PWM</td>
<td>MAX9100</td>
<td>0.32mA</td>
</tr>
<tr>
<td>Carrier wave generation</td>
<td>AND gate</td>
<td>0.35mA</td>
</tr>
<tr>
<td>FET switching</td>
<td>BSS84, 2N7002</td>
<td>0.24mA</td>
</tr>
</tbody>
</table>

The total draw from Table 1 is 2.7mA and so the remaining 8.55 mA of the 11.25mA power budget can be delivered to the transducer.

When connected with the transducer load in the absence of the bandpass filter, the measured input power drawn from the battery and the output power delivered to the transducer was measured to be $P_{in} = 34.2\,\text{mW}$ and $P_{out} = 27.2\,\text{mW}$ which gives an overall power efficiency of 81%. With the output stage filter connected, a drop in efficiency occurs due to the DC series resistance of the inductor (L1 in Figure 1) which is ~8Ω. With the 150Ω load, this corresponds to a 5% drop in efficiency giving an estimate for the circuit efficiency at 76%.

MODULATOR – 50 KHZ SIGNAL PRESENCE & TOTAL HARMONIC DISTORTION

Initial simulations with LTspice and preliminary measurements at the output stage of the FETs both showed the presence of a 50 kHz component requiring the design of a band pass filter that effectively filters out the 50 kHz component. Below, a simulation is shown with the presence of the 50kHz spikes showing up in the modulated output waveform.

![Simulated input audio signal & Output from 2nd order LC filter](image)

**FIGURE 3**, The amplitude modulated output signal (blue), and the original acoustic signal (green). Note that the amplitude modulated signal remains contaminated by some 50 KHz modulation due to the ramp-compare PWM scheme.

The power spectrum of measured output signal generated by the circuit for a 1KHz audio signal input is shown in Figure 4. Harmonics are generally 40dB down relative to the 1KHz sideband.
The total acoustic harmonic distortion was measured by summing the total power in the sideband harmonics when the modulation was a 1.0kHz sine wave. It was found to be to be 2.4% which is acceptable for a hearing aid application.

**MODULATOR – DISCUSSION**

The largest single source of loss in the circuit is the 8Ω DC resistance of the series inductor in the bandpass circuit. By incorporating this circuit element into the packaging of a behind the ear unit containing the microphone and battery it may be possible to achieve the necessary inductance with a lower DC resistance. The main purpose of the bandpass filter is to prevent significant energy from the 50KHz sideband from being transmitted to the transducer. Since it was found that total power at 50KHz was only 5% of the acoustic modulation sideband power, a more efficient circuit could perhaps be achieved by avoiding the bandpass filter altogether and allowing the 50KHz sideband to be transmitted. It should be noted that the 50KHz sideband is too high-frequency to be audible and so should not cause any problem with sound quality.

In order to achieve even smaller form factor, a similar transmitter design should be considered that operates off a single zinc-air cell. Many conventional induction link powered hearing aids, such as the Vibrant Soundbridge, operate off single high energy density zinc-air batteries. For example, a zinc air cell rated for +1.4V has a capacity of 620 mAh and comes in a package size of 11.6mm (diameter) by 5.4mm (height). The circuit given above can be modified for +1.4V operation using the following component replacements presented in Table 2.

<table>
<thead>
<tr>
<th>Current Part</th>
<th>Replacement</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>LTC6900</td>
<td>TLC551</td>
<td>Clk &amp; Triangle Generation</td>
</tr>
<tr>
<td>LTC6246</td>
<td>LMV951</td>
<td>Triangle Generation</td>
</tr>
<tr>
<td>2N7002, BSS84</td>
<td>SiA533EDJ</td>
<td>FETs</td>
</tr>
<tr>
<td>SN74LVC1G08</td>
<td>SN74AUC08</td>
<td>AND Gate</td>
</tr>
</tbody>
</table>

**Demodulation**

In a high-efficiency ultrasonic link the impedance looking into the transmit transducer is strongly affected by the load impedance of the receive transducer. This effect can be readily modeled using the KLM transmission line

![Figure 4](https://example.com/figure4.png)

**FIGURE 4.** Power spectrum across 150Ω load with 1 kHz audio input and 1.024MHz carrier
model of the transducers coupled to a transmission matrix model of the electrical circuits at the source and the load. For our transducers with an electromechanical coefficient $k_t = 0.78$ connected through 5 mm of tissue, the impedance looking into the transmit transducer when the receive transducer is electrically terminated with an open circuit is $6.3\, \Omega$. By comparison, the impedance looking into the transmit transducer when the receive transducer is terminated by a matched load of $450\, \Omega$ is $690\, \Omega$. One can view this effect as being due to the acoustic wave reflecting from the receive transducer and returning to the source.

This fact is highly relevant for AM demodulation circuitry since in AM modulation at least half of the power is located in the carrier signal. Since the carrier must be transmitted along with the signal, but only the signal will ultimately be used to drive the hearing aid, this modulation scheme might seem like wasteful use of transmit power. In an efficient link, though, the impedances reflected back to the transmitter at the carrier and the signal frequencies can be different with impedance seen by the carrier being significantly higher than that for the signal. As a result, standard, simple AM demodulation techniques can be used with reasonably high efficiency, resulting in a relatively compact implanted demodulation circuit.

**Demodulation circuits**

We designed a Schottky diode rectifier-based AM demodulation scheme that involves a diode rectifier bridge followed by a low pass filter as shown in Figure 5. The efficiency of such a rectifier increases with the ratio of the received signal level to the forward bias voltage of the Schottky diodes, typically 0.2-0.3V. This effect is shown in Figure 6. Because of the higher efficiency available with a higher voltage it is advantageous to design the receive transducer with a high source and load impedance so as to create the largest voltage from the impinging acoustic wave for a given power requirement at the load. Unfortunately a high load impedance also requires a large inductor in the low-pass LCR filter which can create design challenges in an implanted device.

![FIGURE 5, Demodulation circuitry using a full wave rectifier and RLC filter stage](image)

![Graph of Measured Efficiency versus Input Voltage](image)
Conclusions

We have built and tested an efficient AM modulator for powering an ultrasonic power and signal link meant to replace inductive links for implanted hearing aid systems. Using piezoelectric transducers, a transmitter/receiver can be designed with a smaller form factor than conventional hearing aids powered through magnetic induction. The measured efficiency for the current transmitter was measured to be 76%, operates off a single +3.0V design, and is small enough to fit into a compact hearing aid package. The measured total harmonic distortion for this system is 2.4% while the power consumption of the transmitter is ~35mW when connected to a 150Ω load.

The design of an efficient rectifying demodulator is currently in progress with a preliminary design that can be optimized in the future.

References