SELECTIVE SIGNAL TRANSMISSION TO INLAID MICROCOILS BY INDUCTIVE COUPLING

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ABSTRACT

Inductive links are widely used to power medical implants, but are not suited to transmitting low frequency waveforms. Here, using amplitude modulation, such waveforms are transmitted via an integrated receiving coil and a passive demodulator. An "inlaid electroplating" process is used to fabricate microcoils of small resistance to obtain enhanced output. As our inductive link resonates at a frequency specific to the secondary microcoil, selective signal transmission is demonstrated with frequency-multiplexing input. Further, combining two coils with switching transistors, biphasic pulses are produced at the load, which has potential applications for biomedical electrical stimulations.

INTRODUCTION

An inductive link, when used in implanted biomedical instruments, is also known as a transcutaneous transformer, since its primary and secondary coils are physically separated and communicate through the electromagnetic field linking the two coils. Extensive research has been performed to analyze and optimize the operation of such inductive links [1], [2], and there exist many applications of inductive links to deliver power and data into implants [3], [4].

Recently, microfabricated coils have been adopted as receivers in inductive links in order to reduce the size of implants [5], [6], [7], which also offers the possibility of future integration with other micro-devices to establish a system-on-a-chip. On the other hand, integrated coils have non-negligible internal resistance, which dissipates power within the coil and damps the signal deliverable to the load. In order to increase metal thickness, we developed an inlaid electroplating process that can build microcoils of centimeter side length and hundred micron height, and at the same time preserve the integratability of microcoils.

Our ultimate goal is to transmit biphasic waveforms across the scalp to implanted electrodes for purposes of deep brain stimulation, so waveform control of relatively long biphasic pulses, on the order of 0.1 to 10 ms, is important [8]. Generally, transcutaneous transformers have limited time constants, since the inductance of the primary coil is minimized so as to produce usable large output voltages for integrated secondaries of few turns. Small time constants pose difficulties when transmitting low frequency signals of certain waveforms. In the biomedical field, this problem has been approached by implanting and inductively powering a signal generator inside the body [9]. Here, signal generators pre-store a few waveforms that can be selected through an inductive data link [10], [11], [12]. To some extent, this approach loses flexibility in programming waveforms, which is of importance when seeking optimum stimulation conditions.

Amplitude modulation has been used to transfer analog waveforms [13], [14], and enhanced output was observed at the link resonant frequencies. The link resonant frequency is set within the series tank circuit by the link leakage inductance and microcoil capacitance. This feature has been utilized to selectively drive microactuators [15]. We demonstrate that selective transmission of signals can be realized as well using these resonance phenomena.

Amplitude modulation can transmit signals of only one polarity. However, combining outputs from two separate receiving coils with analog switching transistors can be used to generate biphasic waveforms on-chip. Hochmair [16] used an RC circuit and one coil to generate simple biphasic waveforms. Our technique is more flexible for implant circuit design and any generated waveform. Experimental results on biphasic waveforms will be presented in this paper.

FABRICATION OF INLAID MICROCOILS

Microcoils are well known for their non-negligible internal resistance, which limits their applications in integrated systems. Electroplating can increase metal thickness, such as in copper damascene and LIGA (a German acronym of Lithographie, Galvanoformung und Abformung, meaning lithography, electroplating, and molding) [17]. However, damascene typically produces metal only a few microns thick [18], [19], [20], and LIGA does not allow for lithography steps after electroplating, which may be undesirable for some system integration. Our microcoils are designed to have
side length up to 14 mm, so the damascene process cannot sufficiently reduce the coil resistance. Also, polymer or polyimide molding as used in LIGA is hard to realize for the geometries of 14 mm length, 100 microns height and only 10 microns width, such as the structures demonstrated below. At these dimensions, the polymer molds have a tendency to de-adhere and deform [21].

We have developed an inlaid electroplating procedure to build copper coils of large dimensions within silicon substrates. The fabrication procedure is shown in Fig. 1. Trenches into the Si substrate by deep reactive ion etching (DRIE) were used as the electroplating mold, thus circumventing those molding difficulties in microelectronic electroplating mentioned above. Working copper coils were obtained after electroplating to a height below the surface of the silicon (Fig. 2a), however incorporating a polishing step after overplating (Fig. 2b) offers the potential for integration with circuits. In addition, the sidewalls between copper coils can be controllably removed (Fig. 2c) in order to fine-tune the coil capacitance. Polymer molding processes do not allow the adjustment of coil capacitance for the sake of tuning resonant frequencies. Note that removal of the silicon sidewalls allows inspection of the copper plating, and that no voids are visible.

**SIGNAL TRANSMISSION**

The amplitude modulation scheme for pulse transmission is shown in Fig. 3. The receiving coil is an integrated inductor, so a lumped-element model is used for its parasitic effects, where \( R_2 \) accounts for its internal resistance and \( C_2 \) for its parasitic capacitance. \( C_2 \) comes into series resonance with the leakage inductance of the inductive link, resulting in an enhanced output level at link resonant frequencies. However, large \( R_2 \) typically associated with integrated coils will reduce the output peak, which is alleviated by the inlaid plating process discussed above. Passive demodulation is used after the receiving microcoil. This method is simple, reliable and can accommodate variations in load resistance.

For comparison, signal transmission experiments were performed with both thin film and electroplated copper receiving coils. Our inductive link had a coupling coefficient of 0.42. The coils were 20 turns and exhibited the following impedance properties: thin film: \( R_2=275.8 \) \( \Omega \), \( L_2=23.6 \) \( \mu \)H, \( C_2=114.8 \) pF; electroplated: \( R_2=5.95 \) \( \Omega \), \( L_2=23.6 \) \( \mu \)H, \( C_2=243.6 \) pF, corresponding to the elements shown in Fig. 3. A 1N4148 diode and an 11 nF capacitor were used for demodulation with the load resistance varying as 200 \( \Omega \), 700 \( \Omega \) and 1200 \( \Omega \). The obtained output pulse amplitudes across the load are shown in Fig. 4.
The output amplitudes for the electroplated coils are noticeably higher than those of thin film coils, as predicted. With the same circuit components, the system resonant frequencies depend primarily on the coil properties. In fact, it can be observed that the output reaches maximum amplitude at lower frequencies for electroplated coils than for thin film coils, because electroplated coils have more parasitic capacitance than thin film coils.

The variation in resonant frequencies of two coils can be utilized to selectively transmit signals, as shown in Fig. 5. The input signal has two carrier frequencies: 335 kHz with 250 μs duration and 7.2 MHz with 125 μs duration, which are the system resonant frequencies with two different receiving coils (40 turns and smaller 10 turns). The pulse trains of different carrier frequencies were retrieved separately from loads connected to the two

receiving coils. In principle, selective driving of microdevices, e.g. implanted MEMS actuators, could be realized by powering with different microcoils.

**BIPHASIC WAVEFORM GENERATION**

By amplitude modulation, waveforms of only one polarity are obtained. However, in many implanted biomedical applications, waveforms of both polarities or so-called biphasic waveforms are desired. A promising application area is electrical stimulation of the nervous system, which requires charge-balanced biphasic pulses [22], [23], [24].

In the previous section, two pulse trains were separated from the frequency-multiplexed input with coils of different resonant frequencies. As in Fig. 6, combining the two outputs with two switching transistors, which are also driven by the coil outputs, pulses of different polarities can be isolated from the other coil and coupled to the load. Experiments on biphasic waveform generation were performed using discrete circuit elements and fabricated inlaid microcoils. The obtained biphasic waveform at the load is shown in Fig. 7.

Therefore, it is possible to reduce an implanted pulse generator to the size of two receiving coils, as the circuit

![Fig. 4. Demodulated pulse amplitude for two types of coils.](image1)

![Fig. 5. Selective signal transmission. Top is input modulated at two frequencies, middle is output of 335 kHz resonant coil (40 turns), and bottom is output of 7.2 MHz resonant coil (smaller 10 turns).](image2)

![Fig. 6. Biphasic waveform generation circuit](image3)

![Fig. 7. Biphasic waveform obtained at the load. Top is input at two frequencies and bottom is output from a single load resistor (1 kΩ).](image4)
elements can be integrated on-chip. In addition, as the output waveform at the load is determined by the input signal, which can be modified easily outside the body, this scheme has more flexibility in producing waveforms. This can be better adjusted to suit the medical conditions of individual patients compared with inductive powering of pre-programmed signal generators.

CONCLUSIONS

Based on the above results, pulse generators might thus be substituted by an external unit in coordination with implanted receiving coils and circuit components. Since the coil fabrication process is CMOS-compatible, the implant can be further miniaturized by integrating the dual-coil with circuit components.

REFERENCES