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A Microfabricated Transduction Coil for Inductive Deep Brain Stimulation

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Abstract: “Inductively Coupled Deep Brain Stimulator” describes a chip/system design to inductively couple arbitrary waveforms to electrodes embedded in the brain for deep brain stimulation or other neurostimulation. This approach moves the conventionally implanted signal generator outside the body and provides flexibility in adjusting waveforms to investigate optimum stimulation waveforms. An “inlaid electroplating” process with through-wafer plating is used to reduce microcoil resistance and integrate microstructures and electronics. Utilizing inductive link resonance specific to microcoils, waveforms are selectively transmitted to microcoils, which further produces biphasic waveforms that are suitable for deep brain stimulation.

Keywords: micro-electro-mechanical system, inductive coupling, electrical stimulation

1. Motivation

Deep brain stimulation (DBS) is used to treat medical conditions such as essential tremor and symptoms of Parkinson’s disease, e.g. dyskinesia, bradykinesia, tremor, and rigidity. It is implemented by providing pulses to electrodes implanted into the thalamus, globus pallidus, or subthalamic nucleus of the brain [1, 2, 3, 4].

DBS has conventionally been approached by implanting and inductively powering a signal generator inside the body [5]. For example, in the Medtronics Activa® system [6], pulse generators (similar to pacemakers) are implanted into a patient's chest wall to provide signals [7]. Leads tunnel under the skin from the chest to the top of the skull, connecting the pulse generator to the implanted electrodes. Signal generators pre-store a few waveforms that can be selected through an inductive data link [8, 9]. To some extent, this approach loses flexibility in programming waveforms, which is important when seeking optimum stimulation conditions. The existing generator implant surgery incurs a high cost both financially and physically, and might cause side effects e.g. system infection, battery/connector problems, lead migration, and hemorrhage [10].

So far, the implantation of pulse generators is indispensable because the pulse trains for DBS have widths from 20 μs to 1.5 ms [11], and such long duration waveforms are difficult to transmit wirelessly by inductive links (transcutaneous transformers) if the generator is moved outside the body. Typically, transcutaneous transformers have limited coil inductance so as to produce usably large output voltages for secondary coils of few turns. However, this causes the coils to have small time constants, causing droop and backswing of waveforms during transmission. Transmitted waveforms are distorted to spikes when waveform durations are much longer than the system time constant.

Our inductively coupled deep brain stimulator (ICDBS) seeks to move outside of the body the waveform generator by inductive coupling while preserving the waveforms with high fidelity. The only parts that remain implanted are the electrodes and a small receiver, i.e. packaged silicon wafer about 1 cm^2 and 1 mm thick that contains the pickup coil and passive electronics. Compared with implanted generators, this new system offers a variety of advantages to the patient:

- Less surgery, less risk of infection and higher reliability
- No need for implanted battery and subsequent replacement
- Easy system upgrade
- Ability of the patient to vary treatment parameters (within limits) without doctor visits
- Considerably lower total cost of system and procedures
- Option for testing DBS in the patient before permanent implantation of Activa generator
- Possible applications to improved DBS research on primates

2. Inductively Coupled Deep Brain Stimulator

The key concepts of the inductively coupled deep brain stimulator (ICDBS) include: 1) Employ amplitude modulation for waveform transmission, thus inductively transmitting signal with a suitable carrier frequency and applying external control to modulating (stimulation) signals, 2) Fabricate monolithic, low-resistance, large-area, silicon-process-compatible microcoils [12, 13] to intercept modulated signals, 3) Utilize self-resonance of MEMS coils due to loose inductive coupling in transcutaneous applications and coil parasitic capacitance, so that selective waveform transmission [14] to a specific MEMS coil can be realized, and 4) Combine outputs from two microcoils with switches that are also controlled by microcoil outputs, so that biphasic waveforms suitable for DBS are reconstituted at the load [15]. (For electrical stimulation of the nervous system, waveforms with both polarities, and "biphasic," are desired to maintain charge balance [16, 17].)

ICDBS is schematically shown in Fig. 1 and 2. External control of waveforms is realized by amplitude modulation of carrier signals. The modulated signals are inductively intercepted by microfabricated receiving coils. *RC* envelope detectors connected to the integrated coils demodulate the output from the coils and deliver the waveforms to the target brain tissue, which here serves as the circuit load. The receiving coils and demodulation circuits can be monolithically integrated to minimize the volume of the implanted components.

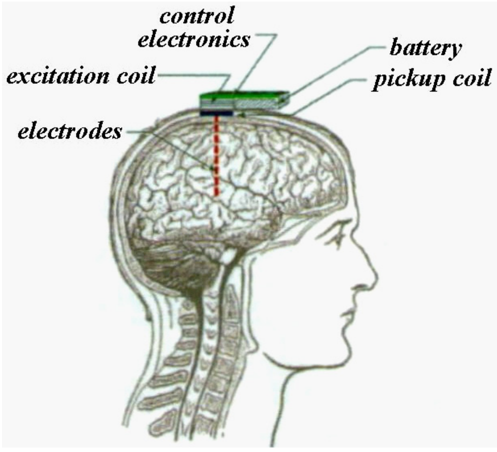


Fig.1. Implementation of inductively coupled deep brain stimulator (ICDBS) on human.

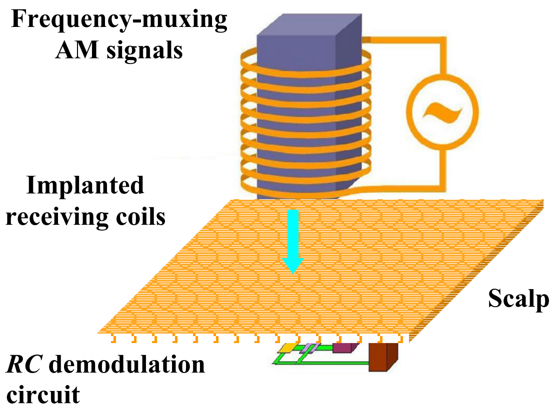


Fig. 2. Electronic composition of inductively coupled deep brain stimulator (ICDBS).

The circuit implementation for transmitting waveforms is shown in Fig. 3. The receiving coils are integrated inductors with parasitic capacitance and resistance. Because the integrated coils operate at frequencies no higher than 10 MHz, a lumped 3-element model as shown in the dashed box is adopted to account for microcoil parasitic effects, consisting of the coil inductance, coil internal resistance and coil parasitic capacitance.

Corresponding to the dual coil design in Fig. 2, two microcoil geometries were fabricated on silicon wafers, as shown in Fig. 4, with in-house developed "inlaid electroplating" process [12, 12]. This process can form microcoils of tens of microns thickness to reduce their internal resistance while preserving their integrability. Both coils have copper lines of 80 μm width on a 100 μm pitch. One coil has ten turns of 10 mm outer-sidelength and 8 mm inner-sidelength, and the other has twenty turns of 14 mm outer-sidelength and 10 mm inner-sidelength. The via-holes are ~ 400 x 400 μm2, etching through ~575 μm-thick <100> Si substrates, connecting microcoils with circuitry on the other side. Coil parameters are: L=2.2 μF, R=2.3 Ω, C=60.5 pF for 10-turn coils, and L=9.9 μF, R=5.9 Ω, C=238.2 pF for 20-turn coils. The coil capacitance can be modified by removing the silicon sidewalls between the copper lines (reducing C) or adding a metal layer over the coils (increasing C), which provides a means of fine tuning the resonance frequencies for signal transmission.

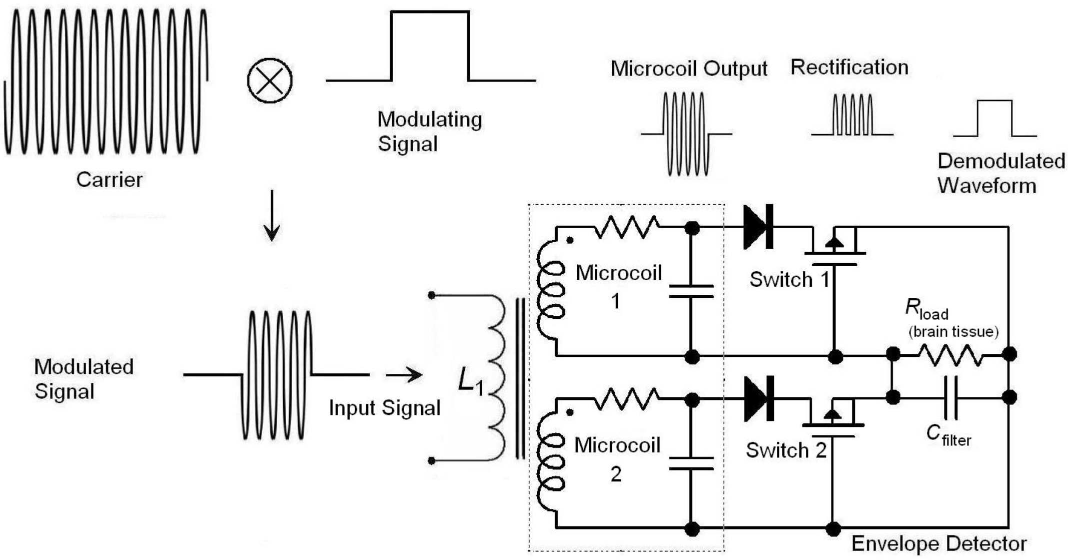
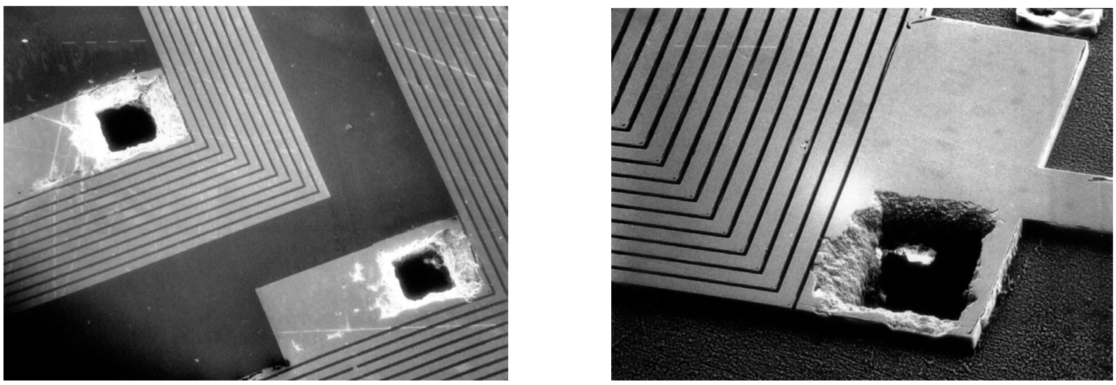


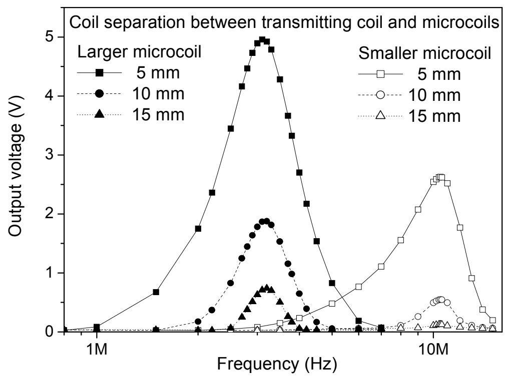
Fig.3. Amplitude modulation scheme for inductive waveform transmission.



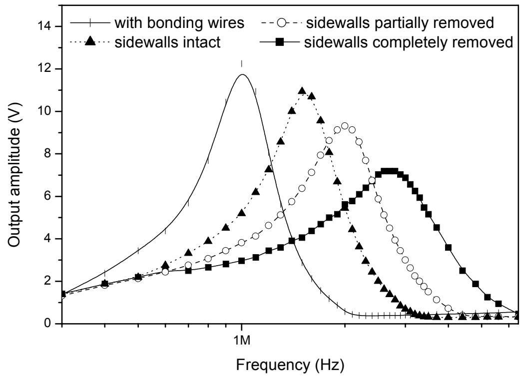
(a) (b)

Fig.4. (a) Dual-microcoil realized by our inlaid electroplating process, and (b) copper coil with Si sidewall removed to adjust coil capacitance.

Due to loose inductive coupling in transcutaneous applications and microcoil parasitic capacitance, the inductive link undergoes resonance at a frequency specific to the microcoil, resulting in enhanced output at the coil resonant frequency. As shown in Fig. 5 (a), two microcoils have two distinct operating frequencies that provide maximum outputs, with little dependency on the separation between the transmitting coil and the receiving coil. If a larger separation in carrier frequency is needed, the resonant frequency of a microcoil can be adjusted by changing coil capacitance, as shown in Fig. 5 (b). Consequently, selective transmission of waveforms to individual microcoils can be realized.



(a)



(b)

Fig.5. (a) Operating frequencies of two microcoils with little dependence on coil separation, and (b) fine-tuning of coil operating frequency by modifying its parasitic capacitance.

While amplitude modulation transmits waveforms of one polarity, electrical stimulation prefers biphasic waveforms for charge balance. Hence, we implement here two analog switches to provide biphasic waveforms from a two-channel modulated input, with separate control of wave shapes in either polarity. The two-channel input has two carrier frequencies corresponding to specific microcoil resonances. Therefore, the microcoils themselves can realize bandpass filtering to extract the modulated signal designated to them. After rectification, the outputs from both coils are connected with opposite polarities through two switching transistors. The two transistors are also driven by the coil outputs. At one coil's resonant frequency, the resonant coil has sufficient voltage to turn on its switching transistor, while the other coil cannot provide enough voltage to drive its transistor, and is thus isolated from the load. As a result, arbitrary biphasic waveforms are reproduced at the load, as shown in Fig. 6.

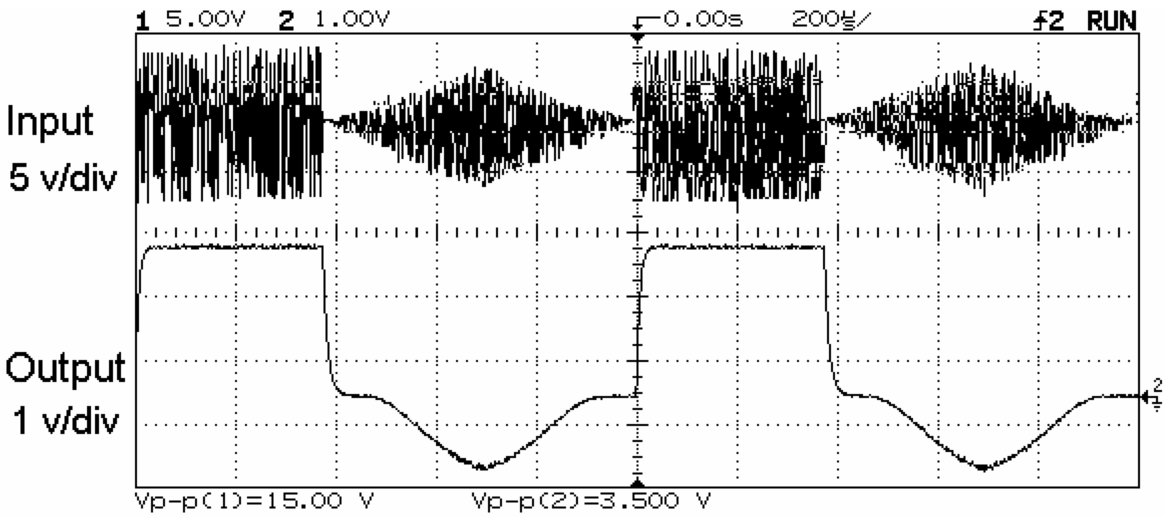


Fig. 6. Generated biphasic waveform at the load with two-channel input.

Besides the proof of concept, the robustness of the system was also examined. 1) The system can tolerate a large variation of load resistance, which is important because brain tissue resistance varies with individuals. The system operation has been verified for load resistance from 200 to 1200 Ω. 2) The system can also tolerate some degree of misalignment, which is expected in transcutaneous applications. The link coupling coefficient as a function of lateral deviation from co-axial is measured at a separation of 10 mm. The coupling coefficient decreases by only ~ 28 % with as much as 8 mm lateral misalignment, resulting in a similar drop in output voltage.

For biomedical applications, interaction between electromagnetic fields and living tissues is always a concern. As our inductive link works at frequencies lower than 10 MHz, no detrimental biological effects are expected, so safety is unlikely an issue. Also, no effects on coupling were observed when a hand was placed between the transmitting and receiving coils, so scalp material is not expected to absorb electromagnetic energy.

3. Conclusions

In conclusion, ICDBS shows promising feasibility and offers many benefits. In North America, about 1 million people suffer from Parkinson's disease, with 50,000 new cases reported each year, and about 4 million total worldwide. The only DBS system available today is the Medtronics Activa, and the cost of the installation procedure is about $ 50,000 ~$ 60,000 per bilateral procedure. The batteries are depleted in 3 to 5 years, after which the stimulator(s) is(are) replaced. As a comparison, other than implantation of the stimulation electrodes (which will be identical to existing systems) it would be a relatively minor operation to install our system, therefore the DBS treatment will be accessible to many more people, and the flexibility in adjusting waveform parameters makes the system well suited for DBS research.

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