Available online at www.sciencedirect.com
ScienceDirect
Sensors and Actuators A 136 (2007) 467–474
SENSORS AND ACTUATORS A PHYSICAL
www.elsevier.com/locate/sna

Switchable polymer-based thin film coils as a power module for wireless neural interfaces

S. Kim [a,*], K. Zoschke [b], M. Klein [b], D. Black [a], K. Buschick [b], M. Toepper [a,b], P. Tathireddy [a], R. Harrison [a], H. Oppermann [b], F. Solzbacher [a]

[a] *Department of Electrical and Computer Engineering, University of Utah, Salt Lake City, UT, USA*
[b] *Fraunhofer Institute for Reliability and Microintegration (IZM), Berlin, Germany*

Received 7 June 2006; received in revised form 5 October 2006; accepted 22 October 2006
Available online 27 November 2006

Abstract

Reliable chronic operation of implantable medical devices such as the Utah Electrode Array (UEA) for neural interface requires elimination of transcutaneous wire connections for signal processing, powering and communication of the device. A wireless power source that allows integration with the UEA is therefore necessary. While (rechargeable) micro-batteries as well as biological micro-fuel cells are yet far from meeting the power density and lifetime requirements of an implantable neural interface device, inductive coupling between two coils is a promising approach to power such a device with highly restricted dimensions. The power receiving coils presented in this paper were designed to maximize the inductance and quality factor of the coils and microfabricated using polymer-based thin film technologies. A flexible configuration of stacked thin film coils allows parallel and serial switching, thereby allowing to tune the coil's resonance frequency. The electrical properties of the fabricated coils were characterized and their power transmission performance was investigated in laboratory condition.

Keywords: Utah Electrode Array (UEA); Neural interface; Thin film coil; Inductive powering; Micromachining

1. Introduction

Recently, efforts have been devoted to develop a fully integrated, wireless neural interface based on the conventional Utah Electrode Array [1]. The Utah Electrode Array (UEA) [2] is a silicon-based structure consisting of a 10×10 array of tapered needle-type electrodes with a base width of 80 μm and a length of 1.8 mm (see Fig. 1). The wireless characteristic of the UEAs will free the patient from the risk of infection associated with wired connections to extracorporeal devices and allow distribution of a network of interface nodes through the central and peripheral nervous system. Towards this wireless neural interface, the UEAs need to have their own power source and electronic circuitry to control the device, process detected neural signals, and transmit them to an extracorporeal data analysis/storage system.

As a power source of such neural interfaces, batteries are not suitable due to their limited lifespan, power density, and the geometric constraints of the envisioned neural interface device. Wireless power delivery through inductive coupling can be a solution to provide power to implanted devices [3–6]. It utilizes electromagnetic induction between two magnetically coupled coils and provides the device a virtually infinite lifetime. An efficient receiver coil needs to be designed to inductively power the device over a certain distance. The technical requirements for the power coils of the neural interface device envisioned [1] are:

- Maximum diameter: 5 mm.
- Minimum transmission distance: 5 mm.
- Transmission frequency: 2.64 MHz.
- Minimum voltage delivery: 4.5 V_{peak}.
- Minimum power delivery: 10 mW.

Besides the specifications listed above, the factors to be considered during the design and fabrication of such power coils are high inductance and quality factor of the coils, biocompatibility, and interconnection with other functional sub-components of the neural interface device.

In this study, coils to serve as a power source of the integrated neural interface were designed, fabricated, characterized, and

* Corresponding author. Tel.: +1 801 581 8013; fax: +1 801 581 6096.
E-mail address: soheek@eng.utah.edu (S. Kim).

0924-4247/$ – see front matter © 2006 Elsevier B.V. All rights reserved.
doi:10.1016/j.sna.2006.10.048

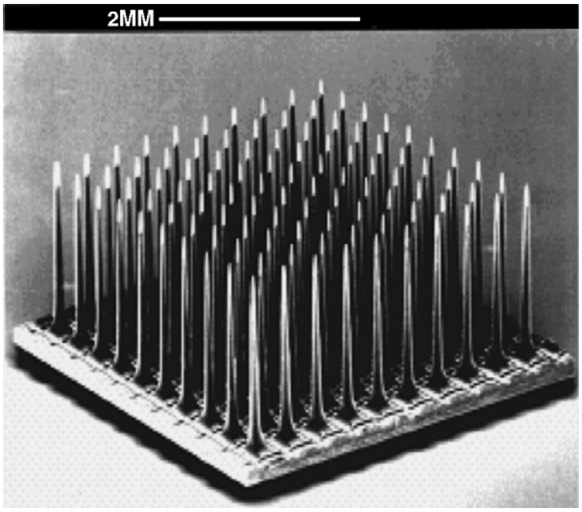

Fig. 1. SEM photograph of the Utah electrode array.

finally its power transmission performance was tested in laboratory condition.

2. Design of power coils

The coils were designed to maximize the inductance and quality factor (Q) and to minimize the parasitic losses. When a coil has a high inductance, it needs only a low capacitance to resonate at a frequency, allowing the use of small-sized SMD (surface mounted device) capacitors or the integration of the capacitor into an IC (integrated circuit). The higher the Q-factor of a coil, the higher the power it can receive. It is important to keep the parasitic losses as low as possible, to ensure the functionality of inductive power transmission in a certain range of operating frequency.

The dimensions of the UEA allow the maximum diameter of power coils to be 5 mm. Since a coil will have a maximum Q-factor when the opening in its center is 20–25% of the coil diameter [7], the inner diameter of power coils was determined to be 1.25 mm. Other geometrical parameters such as the width, spacing, and height of coil turns were determined to achieve high number of windings within the technical limitations (e.g. minimum structure size, maximum aspect ratio achievable with thin film technologies) that still guarantee good fabrication quality. A trace width and spacing of 15 μm were considered as a feasible dimension for coil turns, with which a reasonable production yield was possible for coil thicknesses in excess of 10 μm. A 200 μm thick ferrite platelet was used to increase the Q-factor of coils as well as to protect the underneath electronics against electromagnetic interferences (see Fig. 2). Several different types of single- and double-layer coils were fabricated and tested for the power transmission.

Stackable coils were proposed to avoid undesirable high inter-winding capacitance that may exist between coil layers in double-layered coils. An interconnection layout was devised to accommodate these stackable coils, allowing to switch between different coil configurations as shown in Fig. 3. The stacked coils can be switched in parallel or series to tune the coil parameters and therefore the resonant frequency of power transmission. With this feature, the layout can provide a certain degree of flexibility in coil configurations. Generally, micromachined MEMS (micro-electro-mechanical system) devices do not allow easy changes in fabrication design, since high cost and time investments are required to change or modify the design of devices once a mask design is determined. For the initial stage of device development, however, a flexible configuration for the coil design was required such that the inductance and Q-factor of the coils, as a consequence, the resonance frequency can be modified. This allows compensating for unknown power signal variance due to the tissue and the placement accuracy of primary (driving) and secondary (receiving) coil.

Fig. 4(a) shows the layout for interconnection of the coil, SMD capacitors, and the IC chip that is flip chip bonded on the backside of the UEA. Fig. 4(b) shows the schematic side view of the integrated neural interface device, in which all sub-components are electrically connected through the layout as shown in Fig. 4(a). Two spacers and a jumper are used to adjust the coils' electrical characteristics to various parasitic capacitances and voltage gains. These spacers and jumper allow the operation of the coils in three different arrangements: (1) a single-layer or a double-layer coil, (2) two stacked coils connected in series, and (3) two stacked coils connected in parallel (see Fig. 3).

3. Coil fabrication

Taking into account numerical simulation results [7], technological considerations, and the device assembly process [8], six different coil designs were manufactured. The coils were fabricated using biocompatible materials such as gold and polyimide. To achieve sufficiently high density of coil turns within the given coil diameter, thin film technologies were used. Polyimide-based Au coils were fabricated on a 4-in. silicon wafer. Each wafer carried 100 coils with different designs. Single- and double-layer coils were manufactured with line width and spacing of 15 or 20 μm, and the thickness of Au coil turns was approximately 10 μm.

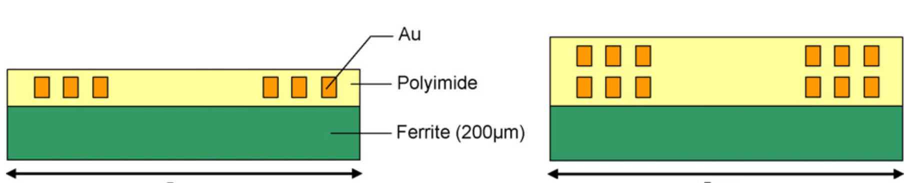

Fig. 2. Schematic of the cross-section of single-layer and double-layer coils with a ferrite platelet underneath.