# TOWARDS A WIRELESS AND FULLY-IMPLANTABLE ECOG SYSTEM

*E. Tolstosheeva[1], J. Hoeffmann[2], J. Pistor[2], D. Rotermund[3], T. Schellenberg[4], D. Boll[1], T. Hertzberg[1], V. Gordillo-Gonzalez[5], S. Mandon[5], D. Peters-Drolshagen[2], M. Schneider[4], K. Pawelzik[3], A. Kreiter[5], S. Paul[2], and W. Lang[1]*

[1]Institute for Microsensors, -actuators and -systems, University of Bremen, Germany
[2]Institute of Electrodynamics and Microelectronics, University of Bremen, Germany
[3]Institute for Theoretical Physics, University of Bremen, Germany
[4]RF & Microwave Engineering Laboratory, University of Bremen, Germany
[5]Brain Research Institute, University of Bremen, Germany

## ABSTRACT

This paper presents a wireless system, designed for electrocortical (ECoG) neural recordings, consisting of an implantable flex-rigid ECoG array and a wireless electronic platform. The array is designed for implantation on top of the visual cortex of a *Rhesus monkey*. The electronic platform contains pre-amplifiers and multiplexer stages, analog-digital converters (ADCs), an in-house developed and reconfigurable ASIC, an RF data transceiver and an inductive energy link. Our ASIC provides user-defined selection of channels (≤128), adjustable channel resolution (1-16bit), sample rate (39S/s-10kS/s), and configurable analog filters. The functionality of the whole system is successfully demonstrated by an in vitro test.

## KEYWORDS

Wireless ECoG system, local field potentials, flex-rigid neural array, configurable digital ASIC, wireless RF datalink

## INTRODUCTION

In order to understand how the brain processes information and based on this information to create neuronal medical devices, it is essential to observe the neuronal activity with a high temporal resolution at as many locations as possible. For this task high density electrocortical (ECoG) electrode arrays [1,2], placed on top of the cortex surface are very promising tools. They offer higher quality signals than EEG, and are less invasive than single-neuron recordings. Clinical ECoG arrays are used to localize seizure areas prior to epilepsy surgery. Wireless and fully-implantable measurement systems provide the basis for the realization of functional neuroprosthesis for medical applications. For such an ECoG implant to be applicable, high spatial and temporal resolution are required. Moreover, placement of the entire system inside the skull will abolish the risk of infection along cable connections. In order to fulfill the above listed requirements, the wireless system has to record many electrophysiological signals in parallel and it needs to be highly configurable for adapting its data processing to the available signal and recording quality. Furthermore, it needs to transmit the collected data wirelessly with high data rate to the outside of the body. The system should also be powered wirelessly without any implanted batteries. Last but not least, the system design is highly restricted by the size of the implant and the power dissipation.

Miranda *et al.* has tackled the challenge of high data rate transmission by placing their system in a skull-mounted housing [3]. Song *et al.* has divided their implant in two blocks, connected them by cable, placed one block directly on the cortex, and the power-demanding block between the skull and the skin [4]. Harrison *et al.* has developed a fully-implantable single-chip, with a high-degree of integration [5].

Our approach to designing a wireless, fully-implantable neural wireless system combines key advantages of the measurement systems listed in [3-5]: significant flexibility in terms of system architecture and therefore functionality, the aim of a fully implantation and the high degree of integration. Our passive ECoG electrode array, presented in [6], has been optimized in terms of connector assembly and its impedance was measured in saline. In parallel, our work, presented in [7], has been extended to a wireless system. In this work the functionality of the wireless ECoG system (the passive ECoG array connected by cable to the wireless system) is successfully demonstrated by an in-vitro test in saline.

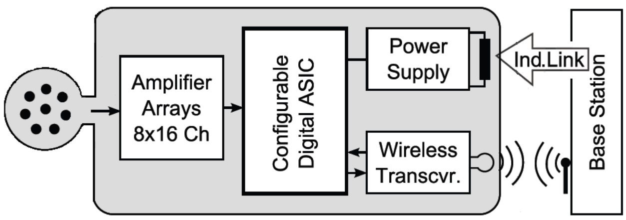


*Figure 1: Block schematic diagram of the wireless ECoG system.*

The structure of the wireless ECoG system is depicted in Fig.1. A 128-electrode array is our direct interface to the visual cortex. It feeds them into an amplifier array, containing 8x16 channels. The signals are digitized and then processed in a user-defined way by an in-house developed digital ASIC. The data are transferred to a wireless medical implantable RF transceiver, which sends them in the MICS-band (401-406MHz) to an external base station. The data are analyzed and visualized by our software. Energy is fed wirelessly into the system via an inductive link.

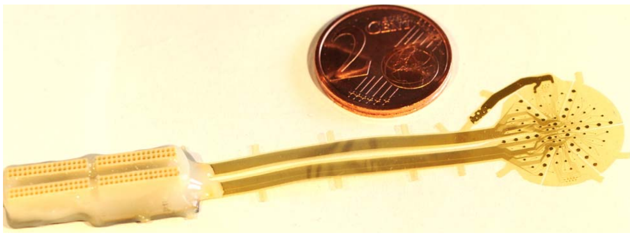

*Figure 2: 128-electrode flex-rigid ECoG array.*

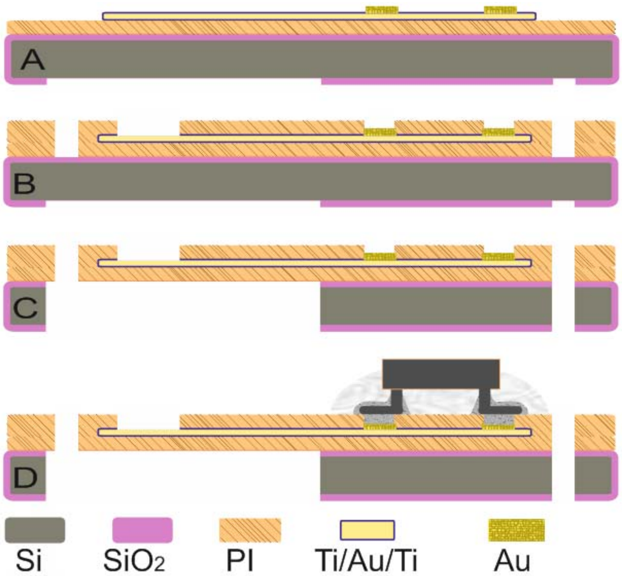


*Figure 3: Fabrication steps of the flex-rigid array: (A) On a Si/SiO$_2$ wafer polyimide film is applied, Ti/Au/Ti is sputtered and Au is electroplated. (B) Polyimide (PI) is applied and etched. (C) The Si and SiO$_2$ on the wafer backside is etched. (D) OMNETICS connector is soldered onto the array.*

The first part of the system, represented by the passive ECoG electrode array, is depicted in Fig.2. It consists of 128 electrodes, situated on the edges of concentric hexagons. The reference electrode of the array is situated beside the largest hexagon. A cable links the electrodes to OMNETICS connectors. The whole system is made from a thin flexible foil, with an extra rigid platform, situated underneath the connector area.

## ARRAY MICROFABRICATION, ASSEMBLY & TEST

Several groups have addressed the challenge of assembling electrical connectors onto thin flexible neural arrays. Rubehn et al. [1] designed a custom-made holder, which was used during soldering as an external rigid platform, Baek et al. [8] used an external through-hole PCB board. However, the requirements for long-term implantable arrays shrink the size of the connector area significantly and limit the types of the connectors that can be used. Therefore, we have fabricated an array with a Si rigid platform, which is an inherent part of the system, as presented in [6], whereas conductive glue was used for the electrical connections. In this paper, aiming at higher mechanical stability, we present an optimized fabrication, allowing soldering of connectors onto the array and

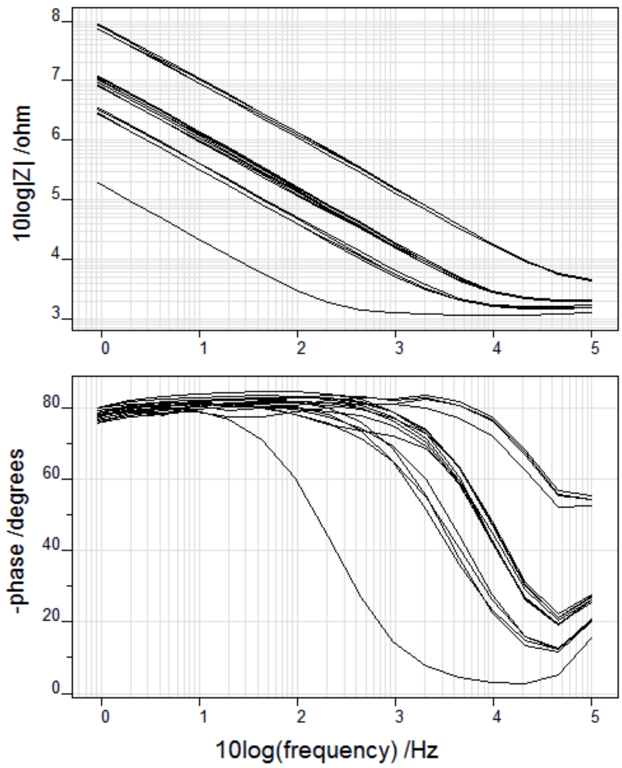


*Figure 4: Impedance plots (top: magnitude, bottom: phase) of the ECoG electrodes of a 32-pin OMNETICS connector: three different sized electrodes (100, 300 and 500µm in diameter) and an ECoG reference electrode.*

providing better adhesion between the flexible material and the metal. To allow soldering, pads of least thickness 1-2µm are required; otherwise too thin metal layers will be dissolved into the solder material and later detach from the array. For this purpose an electroplating step was incorporated in the array's fabrication steps. Furthermore, to increase the adhesion between the flexible material and the sensing metal, Ti films were applied.

An overview of the microfabrication steps of the new array is depicted in Fig.3. On a 380µm/800nm-thick Si/SiO$_2$ substrate a 5µm-thick polyimide (PI) film was spun and cured, followed by a 20/300/20nm-thick Ti/Au/Ti sputtered stack. A 2µm-thick Au film was electroplated on the pad areas, predefined by a photoresist mask. Next, the Ti/Au/Ti stack was wet etched (Fig.3.A) into electrodes, lines and pads. Then a second PI film was applied and structured in two steps (to diminish the undercut in polyimide) in O$_2$+CF$_4$ plasma to open the electrode and pad areas (Fig.3.B). The Si on the wafer backside was then dry etched by a DRIE process in two steps (during the second step a dummy wafer was used for better mechanical stability) to deliver a flex-rigid structure (Fig.3.C), which is held to its carrier wafer by PI straps (not shown).

On wafer level, a Sn/Bi solder paste was dispensed on the array pads. An OMNETICS connector was aligned and soldered onto the rigid platform by a FINEPLACER tool. The space under and around the connector was filled with epoxy to ensure higher mechanical stability (Fig.3.D). The array was then released from its carrier wafer by cutting through its polyimide straps.

An electrochemical impedance spectroscopy test was then performed. The electrode area of the array was inserted in saline. A Pt electrode of $2cm^2$ surface area served as counter electrode and an Ag/AgCl as an external reference electrode. A 10mV sinus was applied at the Pt-electrode, whereas its frequency was swept from $10^5$ to 1Hz. Thereby the impedance was delivered by a COMPACTSTAT Impedance analyzer. The switching between the electrode channels was performed by a NI-PXI 2530 module. An in-house developed LabView program provided the communication between the PXI and the COMPACTSTAT.

The impedance plots indicate four different impedance groups, which match the different pre-designed ECoG electrode diameters of 100, 300 and 500µm and its reference electrode (Fig.4). At 1kHz, the impedance ranges according to groups as follows: 4-7kΩ, 16-20kΩ and 120-160kΩ, for the 500, 300 and 100µm diameters, respectively. Despite their relatively high-impedance, such electrodes can successfully acquire ECoG signals, as demonstrated below.

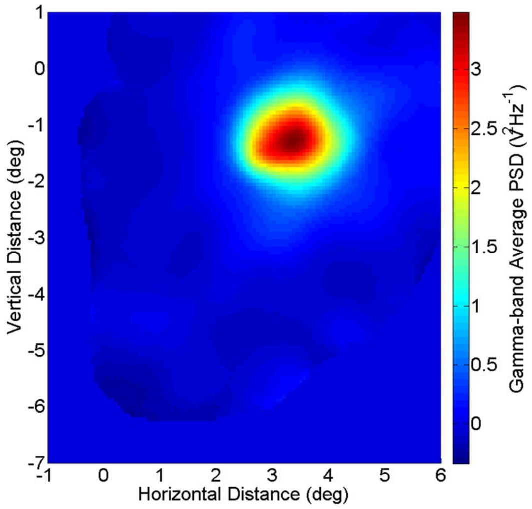


*Figure 5: Receptive field map for one of the electrodes (500µm in diameter), recorded epidurally from an awake, fixating Rhesus monkey's primary visual cortex.*

In this work, we present field potentials from the primary visual cortex of an awake, behaving Rhesus monkey (*Macaca mulatta*). Fig. 5 shows one visual receptive field for one of the 128 electrodes, indicating the good spatial resolution of in-vivo recordings.

# WIRELESS ELECTRONIC PLATFORM

The wireless electronic platform is presented in Fig.6. The thin flexible ECoG array is connected by OMNETICS cables (not shown) to the bio-signal amplifiers and ADCs, realized by eight RHA2116 from intan Technologies. Within the amplifier array the signals pass a DC-decoupling stage, then a LNA stage with adjustable bandwidth, to be next multiplexed and digitized by a low power 16-bit ADC. Then the signals are fed into a configurable in-house developed digital ASIC, containing a set of pre-defined instructions. These allow the user to reconfigure the data processing. For example one can reduce the data transmission rate by changing the number of the measured channels, by adjusting the resolution (1-16bit) and/or reducing the sample rate (10kS-40S/s/channel). Furthermore, analog filters can be configured from 0.02Hz to 1kHz for the lower-end and 10Hz up to 20kHz for the upper-end of the band. Afterwards, the user- defined SPI-coded data run into a wireless medical implantable RF transceiver (Zarlink ZL70102 from Microsemi), which sends them in the MICS-band to an external base station. The data are analyzed and visualized by our software. Energy is fed wirelessly into the system using an inductive link based on the bqTESLA (bq500210EVM-689-System from Texas Instruments). For the implant side of the inductive energy link, we designed a small implantable coil. The power consumption of the system was 84mW.

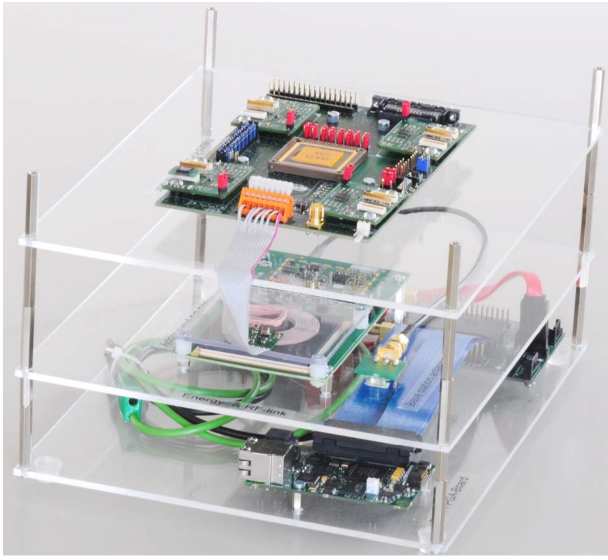

*Figure 6: The electronic platform.*

Our ASIC was designed with the capability to dynamically control the power routing to the RHA ICs, allowing to save energy if not all of these ICs are temporarily not needed.

# IN VITRO TEST

The ECoG wireless system was tested in vitro. An overview of the in-vitro test set-up is shown in Fig.7. Our ECoG array was connected by an OMNETICS cable to the electronic platform. The ECoG array and a reference electrode were immersed in saline. A sinus of 1Vpp, 5Hz was applied between a $2cm^2$-Pt-electrode and a ground electrode, both electrodes being immersed in saline. The recorded signal was wirelessly transmitted to the base station. The noise level of the wireless system, was measured (Fig.7). Its root-mean-square value is 7µV at a sample frequency of 250Hz. A signal, acquired from the ECoG array, is depicted in Fig.9 and its power spectral density is shown in Fig.10. These measurements demonstrate the functionality of the full chain of signal and data processing from the electrodes to the external base station.

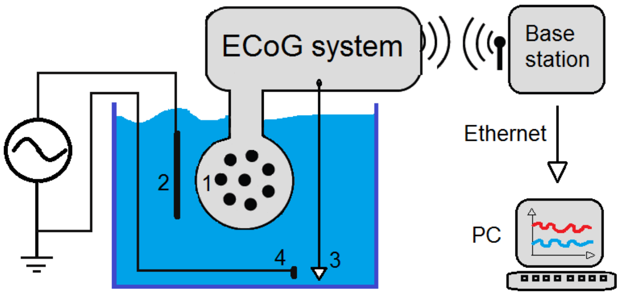


*Figure 7: In-vitro test set-up of the ECoG wireless system: (1) ECoG array, (2) Counter electrode, (3) Reference electrode, (4) Ground.*

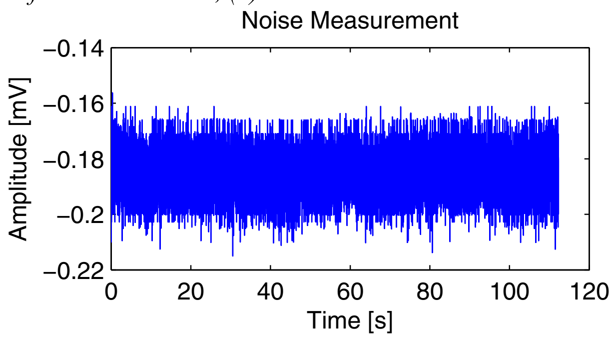


*Figure 8: Noise level of the wireless system, including the ECoG array in vitro, with Vrms=7µV.*

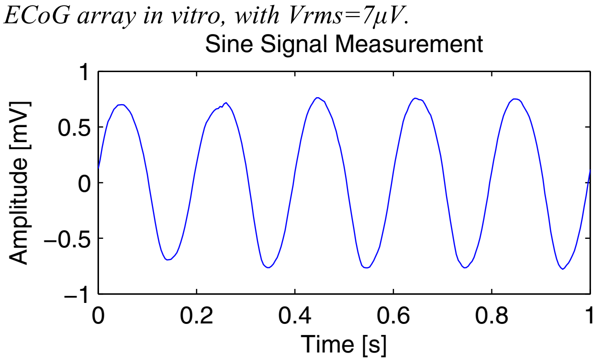


*Figure 9: Output sinus signal of the wireless data transmission chain.*

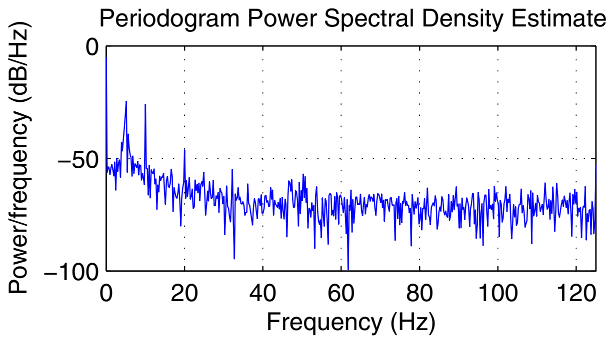


*Figure 10: Power spectral density of the output signal.*

## CONCLUSION

We present our ECoG passive array, which is planned to be used as an assembly platform of analog and digital components, required by the wireless implant. At the moment, the wireless system is further miniaturized and tested with the electrode array in saline solution. Next steps will be the integration of the electronics on the electrode array and test of the system implanted in a Rhesus monkey.


## ACKNOWLEDGEMENTS

The authors would like to acknowledge the financial support of Bundesministerium fuer Bildung und Forschung (BMBF), Grant 01 EZ 0867 (Innovationswettbewerb Medizintechnik), the company Brain Products GmbH and the hospital of Bonn, (Department of Epilepsy).


## REFERENCES


[1] B. Rubehn, C. Bosman, R. Oostenveld, P. Fries, T. Stieglitz, "A MEMS-based flexible multichannel ECoG-electrode array", in *Journal of Neural Engineering*, vol. 6, no.3, 2009.

[2] J. Viventi, D. Kim, L. Vigeland, E. S. Frechette, J. Blanco, Y. Kim, A. Avrin, V. Tiruvadi, S. Hwang, A. Vanleer, D. Wulsin, K. Davis, C. Gelber, L. Palmer, J. Van der Spiegel, J. Wu, J. Xiao, Y. Huang, D. Contreras, J. Rogers, B. Litt, "Flexible, foldable, actively multiplexed, high-density electrode array for mapping brain activity in vivo", in *Nat Neurosci*, vol. 14, no.12, pp.1599-1605, 2011.

[3] H. Miranda, V. Gilja, C. Chestek, K. Shenoy and T. Meng, "HermesD: A High-Rate Long-Range Wireless Transmission System for Simultaneous Multichannel Neural Recording Applications," in *IEEE Trans. On Biomedical Circuits and Systems*, vol. 4, no. 3, June 2010, pp. 181-191.

[4] Y. Song, D. Borton, S. Park, W. Patterson, C. Bull, F. Laiwalla, "Active Microelectronic Neurosensor Arrays for Implantable Brain Communication Interfaces," in *IEEE Trans. On Neural Systems and Rehabilitation Engineering*, vol. 17, no. 4, pp. 339-345, 2009.

[5] R. Harrison, R. Kier, C. Chestek, V. Gilja, P. Nuyujukian, S. Ryu, B. Greger, F. Solzbacher, K. Shenoy, "Wireless Neural Recording with Single Low-Power Integrated Circuit," in *IEEE Trans. On Neural Systems and Rehab. Eng.*, vol. 17, no. 4, pp. 322-329, 2009.

[6] E. Tolstosheeva, V. Gordillo-González, T. Hertzberg, L. Kempen, I. Michels, A. Kreiter, W. Lang, "A novel flex-rigid and soft-release ECoG array," in *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, Boston, August 30-Sept.3, 2011, pp.2973-2976.

[7] J. Pistor, J. Hoeffmann, D. Peters-Drolshagen, S. Paul, "A programmable neural measurement system for spikes and local field potentials," in *Design, Test, Integration and Packaging of MEMS/MOEMS (DTIP), 2011 Symposium on*, 11-13 May 2011, pp.200-205.

[8] D. Baek, C. Han, H. Jung, S. Kim, C. Im, H. Oh, J. Pak and S. Lee, "Soldering-based easy packaging of thin polyimide multichannel electrodes for neuro-signal recording", in *J. Micromech. Microeng.*, vol. 22, no. 11, 2012.


## CONTACT


*E.Tolstosheeva, +49 421 218 62612, et@imsas.uni-bremen.de