Published in final edited form as:
IEEE Trans Biomed Circuits Syst. 2012 October ; 6(5): 414–423. doi:10.1109/TBCAS.2012.2211872.

Towards A Smart Experimental Arena for Long-Term Electrophysiology Experiments

Uei-Ming Jow, **Mehdi Kiani**, **Xueliang Huo**, and **Maysam Ghovanloo***
GT-Bionics lab, School of Electrical and Computer Engineering at the Georgia Institute of Technology, Atlanta, GA 30308, USA

Abstract

Advancements in wireless power and data transmission have raised the prospects of using a variety of low power microelectronic devices such as biosensors, stimulators, and actuators in long-term uninterrupted electrophysiology experiments on small freely behaving animal subjects in large experimental arenas. This paper presents our progress towards developing such a system, called the EnerCage. This system includes a stationary unit for inductive power transmission through a scalable array of overlapping hexagonal coils, which are optimized using an accurate inductive link model and an iterative optimization process. Furthermore, the EnerCage system is equipped with an array of 3-D magnetic sensors and a smart tracking algorithm for non-line-of-sight positioning of the animal subject. EnerCage also has a small mobile unit, which is either attached to or implanted in the subject's body. This unit includes a magnetic tracer and an efficient power management block, which is capable of closed-loop regulated inductive power delivery to the experimental device(s) of interest. An early prototype of the EnerCage system with six coils has been presented here. The coil array achieves a power efficiency of 17.8% at the worst-case horizontal misalignment of 42 mm (half of the coil radius) at a coupling distance of 70 mm with the mobile unit coil of 20 mm in radius.

I. Introduction

Electrophysiology instruments for long term experiments on awake freely behaving animal subjects often include long cables that deliver power, data, or stimulus current between small and lightweight devices and electrodes attached to or implanted in the animal body and external data acquisition or stimulation instruments [1]. Tethering the animals, however, leads to several limitations due to the weight, sheer stress, and limited length of the cables. Most researchers, particularly those who record from or stimulate the central nervous system through a large number of parallel channels cannot benefit from numerous advantages offered by recently developed wireless data acquisition systems because of the bulky energy sources that should be carried by the animal subjects in long term *in vivo* trials [2], [3]. While using battery-powered wireless neural interfacing technology on nonhuman primates is less problematic, the majority of researchers who use small animal models, such as rodents, should always make a compromise between the duration of the experiment and how much payload the animal can carry on its body before the load severely affects its behavior [4], [5]. One possible solution to above problems could be wireless power delivery to devices on or in the animal body.

*Corresponding author. (mgh@gatech.edu). .

The first step is substituting the battery with a coil that weighs only a fraction of the weight of that battery (see Table I). Unlike transcutaneous wireless power transmission, in which the coils are often aligned with the help of permanent magnets and their relative distance on average remains constant, establishing an inductive power transmission link with a receiver (Rx) coil carried by an awake freely behaving animal is quite challenging. Here, the goal is to maintain the received power across the Rx coil constant despite drastic coupling variations due to animal movements.

Using a single transmitter coil (Tx) encompassing the experimental arena has been successful for ultra low power applications [6], [7]. For higher power applications, however, a single Tx coil limits the size of the arena. Because when the size of the Rx coil is much smaller than the Tx coil, power transfer efficiency becomes non-homogeneous and drops sharply towards the center of the Tx coil. The Tx coil self resonance frequency (SRF) also drops due to increased inductance and parasitic capacitance and may reach the desired carrier frequency, which should be avoided. Therefore, for larger experimental arenas, which are required for behavioral experiments, Tx should be designed in the form of an array of smaller coils.

One approach is using an array of smaller coils that tile the bottom of the cage [8]. This method addresses the single coil issues, but leaves deep valleys in between the coils with low magnetic flux density. An alternate design, which has also been adopted for contactless charging is to arrange the coils geometrically across a multilayer printed circuit board (PCB) such that the peaks of one coil coincide with the troughs of the surrounding coils and create a fairly uniform field over a large surface area [9]. Nonetheless, if all the coils are driven at the same time, a considerable amount of power is consumed for generating the magnetic field outside the area where the animal is located. The extra power can increase the temperature in the experimental arena and affect the results by biasing the animal subject behavior.

In this paper, we present the design considerations, system architecture, and an early proof-of-concept prototype of a smart experimental arena, called the EnerCage system, for long-term electrophysiology experiments on small freely behaving animals. The EnerCage system will be capable of wirelessly powering and communicating with small devices attached to or implanted in the animal body without limiting the size of the arena or increasing the cage temperature beyond a desired range. Geometry of the coil array, shown in Fig. 1a, which tiles the bottom of the arena will be optimized for minimum coupling variations and maximum power transmission efficiency at a nominal coupling distance that depends on the average size of the animal model species (in this case rats). A closed-loop inductive power control mechanism based on radio-frequency identification (RFID) technology will maintain the received power across the Rx coil constant most of the times except when the coil is rotated > 60° or the animal stands on its rear limbs [10]. In these conditions, which often last < 30 s, the power management circuit uses its internal power storage to prevent any possible interruptions. The system will also track the animal position in real time using an array of three-axial magnetic sensor modules (the red dots in Fig. 1a) and a smart non-line-of-sight magnetic tracking algorithm [11] that provides valuable information, which is not only important from behavioral perspective but also necessary in activating the Tx coil that is in the best position to couple onto the Rx coil.

II. EnerCage System Architecture

The EnerCage system, which is conceptually rendered in Fig. 1b, consists of three main units: 1) stationary, 2) mobile, and 3) control. The stationary unit, which simplified block diagram is shown in Fig. 2, consists of an array of overlapping hexagonal planar spiral coils

(PSCs) that cover the bottom of the experimental arena (Fig. 1a). Each PSC is individually driven by an RFID reader chip (TRF7960, Texas Instruments, Dallas, TX) through a class-C power amplifier (PA) at 13.56 MHz. The PA supply voltage, V_{DD}, is adaptively controlled by a DC-DC converter that is part of a closed-loop power control mechanism that adjusts the PA output power based on the rectified (but unregulated) voltage in the mobile unit [10]. Every three RFID readers and their associated PSCs are controlled by a local microcontroller (MCU) (MSP430, Texas Instruments, Dallas, TX), which is tasked with collecting their back telemetry data as well as digitized samples from the 3-axial magnetic sensor modules. Each tile of the stationary unit, shown in Fig. 3, consists of 12 PSCs, 5 of which are complete and 7 of which are partial. The partial PSCs are completed when multiple tiles are joined together. Therefore, each tile has 4 local MCUs.

The main purpose of the mobile unit is to substitute the batteries in the electronic or mechanical (e.g. pumps) devices that need to be attached to or implanted in the animal body and its specifications can be modified accordingly. For instance, in our prototype, the mobile unit is designed to be part of a funnel-shaped headstage with 32 movable tetrodes that will pick up single unit neural activity for a wireless integrated neural recording (WINeR) data acquisition system [12]. The WINeR system wirelessly sends the neural signals to an independent receiver at 915 MHz after conditioning [13]. The mobile unit also has a small permanent magnet, as a tracer, to allow the EnerCage magnetic tracking algorithm to localize the animal subject in real time.

The data collected by the local MCUs is further collected and fused in a central MCU, which is part of the control unit. The collected data is sent to a computer through a universal serial bus (USB) interface. The computer runs the magnetic localization algorithm as well as a graphical user interface (GUI), which informs the user about the EnerCage system status. Once the position of the mobile unit is determined, the computer sends the necessary control signals through the same USB connection to address and consequently activate the PSC that is in the best position to power the mobile unit via the central and local MCUs. Magnetic sensors are sampled at 100 Hz, resulting in ~10 ms response time.

III. Design of The Hexagonal Coil Array

Since the EnerCage system is intended to cover large sized experimental arenas with arbitrary shapes, such as a maze, we chose hexagonal PSCs because of their good coverage [9]. Hexagonal PSCs with an edge length of r need only three metal layers on PCB to cover an area such that there is no point further than $r/2$ from the center of a PSC. In other words, if three identical hexagonal PSC layers are properly aligned, as shown in Fig. 3, the maximum horizontal misalignment of the Rx coil on the mobile unit with respect to the closest Tx coil on the stationary unit will be limited to half of the edge length. The fourth layer of a 4-layer PCB can then be used for the interconnections between PSCs, their driving circuitry, and the sensor modules. Figs. 1a and 3d also show how a rectangular shaped unit tile out of the 4-layer PCB can be selected to easily extend the stationary unit in each direction until it covers the desired experimental arena.

The coil model for the EnerCage stationary PSC array is more complicated than an isolated pair of PSCs, described in [14]. The parasitic mutual inductance, capacitance, and resistance between the metal paths of neighboring PSCs can affect the quality factor (Q) of the activated PSC. Moreover, since the goal is to generate a homogenous magnetic field within the arena, coupling variations should be minimized by considering the worst case coupling conditions in the design of the PSC array as opposed to optimal efficiency in perfect alignment. In other words, designers should maximize the power transfer efficiency (PTE) in the worst case misalignment, which is $\pi/2$ displacement of the Rx coil from the center of a Tx PSC, and worst case overlapping, which occurs in the middle layer.

If we consider the black PSC in the center of Fig. 3d and suppose it to be in the middle layer (Layer-2) of the PCB, it can be seen that this PSC is overlapped by 3 PSCs on the top and 3 PSCs on the bottom layers. It is also surrounded by six other PSCs, by each of which it shares an edge. For the sake of simplicity we can ignore the effects of all other PSCs that are further from the center of this PSC. Fig. 4 shows the equivalent circuit model of the central PSC in Fig. 3d, including the most significant parasitic components. In this model, L is the self inductance, R_S is the series resistance, and C_P and R_P are the parallel parasitic capacitance and resistance between metal trances, respectively. The mathematical formulation for these parameters can be found in [14].

The optimal Tx and Rx coil dimensions are subjected to change with the average size of the animal model species. In this design, the coils are optimized for Long-Evens rats at an average height of about 70 mm from the Rx coil to Tx surface. We followed the same iterative design procedure described in [14] using the model shown in Fig. 4 to achieve overlapping PSC geometries that would maximize the PTE in the worst case misalignment (no angular misalignment was considered in this design). The optimization results were verified using a field solver, HFSS (Ansoft, Pittsburgh, PA). Table I summarizes the geometries of the optimized coils at 13.56 MHz carrier and 500 Ω Rx coil loading. The measured PTE was lower than the calculated and simulated values due to the exclusion of a few additional parasitic effects such as the effect of interconnects on the 4th layer and eddy currents induced in the PSC array, which are ignored in this model.

IV. Measurement Results

To measure the homogeneity of the magnetic field, we assembled a proof-of-concept arena using four EnerCage tiles made of 4-layer FR4 PCB with 32 complete hexagonal PSCs. Fig. 5 shows the measured coupling efficiency of these four tiles, measured using a robotic arm that swept the space above the stationary unit plane at a coupling distance of 70 mm. The efficiency peaks in Fig. 5 are resulted from the coils on the top layer, which have higher efficiency due to slightly shorter coupling distance and less overlapping with other coils, while the deeps are linked to the coils on the 2nd and 3rd layers because of their larger parasitic effects.

We also evaluated the closed-loop power control mechanism and magnetic tracking capability of the system using a setup that included two drivers addressing a total of six PSCs. The mobile unit included a rectifier, regulator, MCU, back telemetry switch, and a 500 Ω load. The 13.56 MHz carrier across the Rx coil was rectified, divided down, and compared with an internal MSP430 reference voltage V_{ref} = 1.5 V. If V_{rec} > 3.3 V, narrow pulses (20 μs) were sent back to the Tx coil by shorting the Rx coil at a rate of 700 Hz to indicate that the received power is more than enough. No switching was done if V_{rec} < 3.3 V [10]. Maximum V_{DD} was 8.5 V (0.5 W including driver power consumption), for a total received power of 20 mW, which is sufficient for our wireless neural recording system with the mobile unit at ~7 cm [13].

Fig. 6 shows the variations of the PA supply voltage, V_{DD}, on the stationary unit as well as the rectifier and regulator outputs on the mobile unit when the mobile unit is moved by 17 cm across 3 adjacent PSCs in 10 s. The peaks and valleys of V_{DD} in Fig. 6 are the opposite of the PTE in Fig. 5. Because the higher the PTE, the less power need to be transmitted to deliver the same amount of energy to the Rx coil. It can also be seen that the closed-loop has maintained V_{rec} and V_{reg} constant regardless of the coupling variation. The magnetic localization has also switched the active coils at the right moments indicated by the vertical dashed lines.

V. Conclusion

EnerCage system is a novel experimental arena for long-term electrophysiology experiments on small, untethered, freely behaving animal subjects in enriched environments. It allows researchers conduct a wide variety of experiments without affecting the outcomes by tethering or weight of the instrumentation attached to the animal subjects.

Acknowledgments

This work was supported in part by the National Institutes of Health, NIBIB, grant 1R21EB009437-01A1, and the National Science Foundation under award ECCS-824199.

REFERENCES

[1]. [Online]. Available: http://www.plexon.com/product/Headstages.html/

[2]. Harrison RR, Kier RJ, Chestek CA, Gilja V, Nuyujukian P, Ryu S, Greger B, Solzbacher F, Shenoy KV. Wireless neural recording with single low-power integrated circuit. IEEE Trans. Neural Syst. Rehab. Eng. Aug.2009 vol. 17(no.4):322–329.

[3]. Azin M, Guggenmos DJ, Barbay S, Nudo RJ, Mohseni P. A battery-powered activity-dependent intracortical microstimulation IC for brain-machine-brain interface. IEEE J. of Solid-State Circuits. Apr.2011 vol. 46(No. 4):731–745.

[4]. Miranda H, Gilja V, Chestek CA, Shenoy KV, Meng TH. HermesD: a high-rate long-range wireless transmission system for simultaneous multichannel neural recording applications. IEEE Trans. Biomed. Circuits and Systems. Jun; 2010 vol. 4(no. 3):181–191.

[5]. Greenwald E, Mollazadeh M, Hu C, Tang W, Culurciello E, Thakor NV. A VLSI neural monitoring system with ultra-wideband telemetry for awake behaving subjects. IEEE Trans. Biomed. Circuits and Systems. Apr.2011 vol. 5(no. 2):112–119.

[6]. Cong P, Chaimanonart N, Ko WH, Young DJ. A Wireless and Batteryless 10-bit Implantable Blood Pressure Sensing Microsystem with Adaptive RF Powering for Real-time Genetically Engineered Mice Monitoring. IEEE J. of Solid-State Circuits. Dec.2009 vol. 44(No. 12):3631–3644.

[7]. Minimitter Animal Monitoring Products. [Online]. Available: http://minimitter.respironics.com/animal_products.cfm

[8]. McCormick D, Hu AP, Nielsen P, Malpas S, Budgett D. Powering implantable telemetry devices from localized magnetic fields. IEEE Eng. Med. Biol. Soc. Conf. Aug.2007 :2331–2335.

[9]. Wireless Power Consortium. [Online]. Available: http://www.wirelesspowerconsortium.com/

[10]. Kiani M, Ghovanloo M. An RFID-based closed loop wireless power transmission system for biomedical applications. IEEE Trans. on Circuits and Systems II. Apr.2010 vol. 57(no. 4):260–264.

[11]. Cheng C, Huo X, Ghovanloo M. Towards a magnetic localization system for 3-D tracking of tongue movements in speech-language therapy. IEEE Eng. in Med. and Biol. Conf. Sept.2009 : 563–566.

[12]. Manns JR, Howard MW, Eichenbaum H. Gradual changes in hippocampal activity support remembering the order of events. Neuron. Nov.2007 vol. 56(no.3):530–540. [PubMed: 17988635]

[13]. Lee SB, Lee HM, Kiani M, Jow U, Ghovanloo M. An inductively powered scalable 32-channel wireless neural recording system-on-a-chip for neuroscience applications. IEEE Trans. on Biomed. Circuits and Systems. Dec.2010 vol. 4(no. 6):360–371.

[14]. Jow U, Ghovanloo M. Design and optimization of printed spiral coils for efficient transcutaneous inductive power transmission. IEEE Trans. on Biomed. Circuits Sys. Sep.2007 vol. 1(no. 3):193–202.

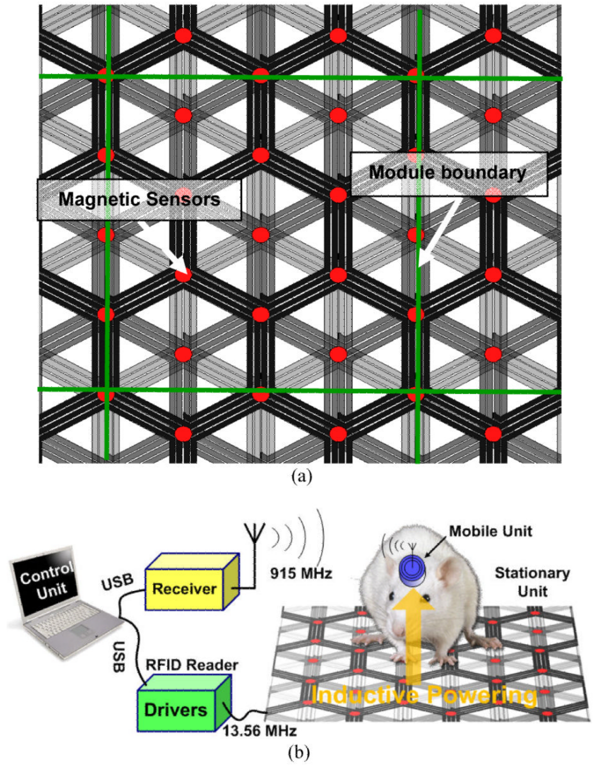

Fig. 1.

(a) Top view of the array of overlapping hexagonal coils and magnetic sensors used at the bottom of the smart experimental arena**.** (b) Conceptual diagram of the EnerCage system, which consists of stationary, mobile, and control units.

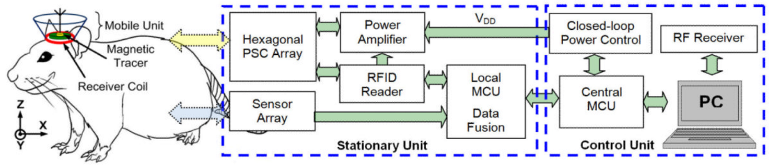

Fig. 2.

Simplified block diagram of the EnerCage system.

NIH-PA Author Manuscript
NIH-PA Author Manuscript

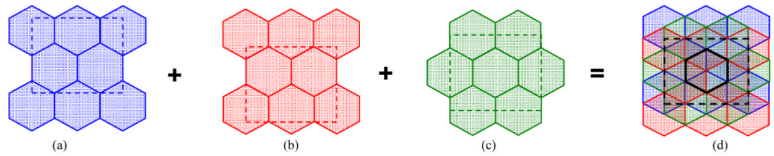

Fig. 3.
Unit repeating tile for the overlapping hexagonal PSCs on (a) Layer-1, (b) Layer-2, and (c) Layer-3. (d) Unit EnerCage rectangular tile when all three layers are properly aligned (also see Fig. 1a).

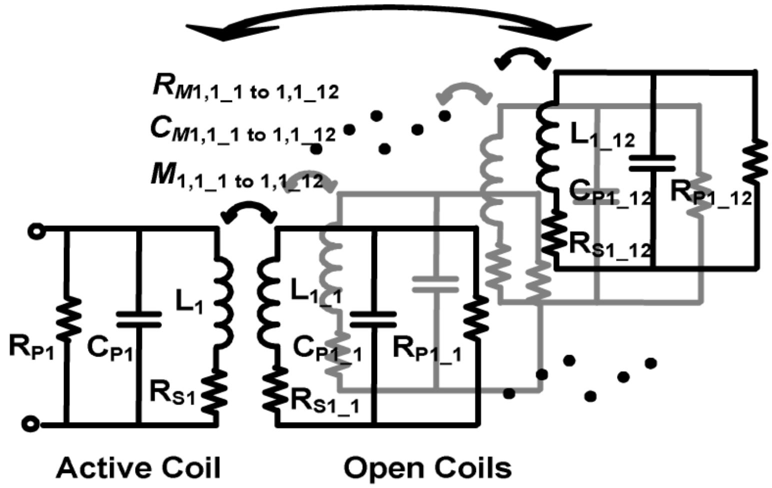

Fig. 4.

The equivalent circuit model of the mutual coupling (M, C_M , and R_M) between overlapping and adjacent PSCs in the stationary unit coil array.

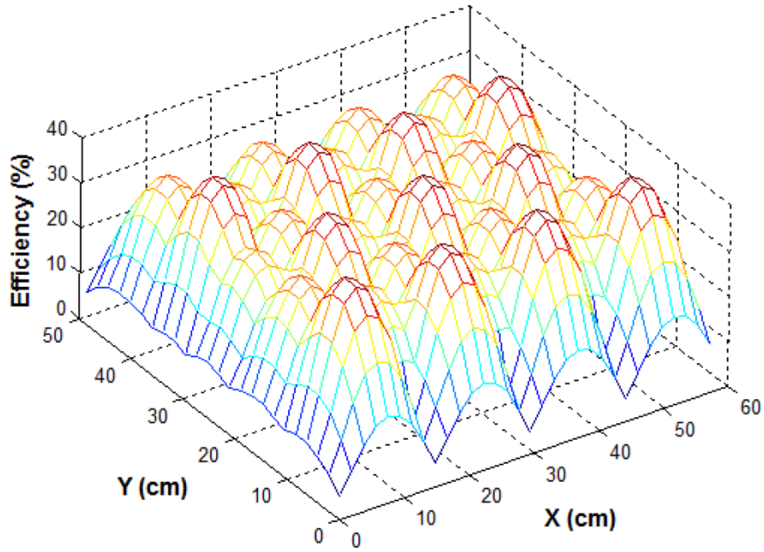

Fig. 5.
Power transfer efficiency mapping of an EnerCage arena consisting of four PCB tiles with 32 complete PSCs at a coupling distance of 70 mm.

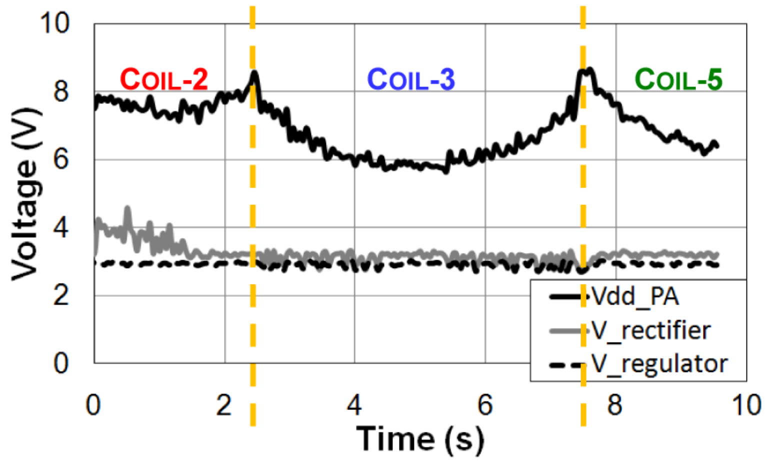

Fig. 6.
Moving the mobile unit by 17 cm over three coils in 10 s at a coupling distance of 70 mm. Vertical dashed lines show when the active coil switches.

Table I

Specifications of the EnerCage Planar Spiral Coil Array*

Parameter	Tx coil (primary)	Rx coil (seconary)
Shape	Hexagonal PSC	Wire-wound
Outer diameter (mm)	168	40
Edge length (mm)	84	20 (Radius)
Number of turns	2	2
Metal trace cross section	35 μm × 10 mm	∅ 0.6 mm
Metal trace spaicng (mm)	3	~0.2
Inductance (μH)	1.28	1.12
Weight of copper (g)	3.6	0.7
Quality factor, Q	168	140
Coupling distance, D (mm)	70	
Mobile unit load (Ω)	500	
Min coupling coefficient (k)	0.021	
Simulated Max/Min PTE (%)	44.8 / 25.8	
Measured Max/Min PTE (%)	42.5 / 17.8	

*From simulation with maximum misalignment = Edge length/2 = 42 mm