Proceedings – 23rd Annual Conference – IEEE/EMBS Oct.25-28, 2001, Istanbul, TURKEY

DEVELOPMENT OF A WIRELESS BRAIN IMPLANT: THE TELEMETRIC ELECTRODE ARRAY SYSTEM (TEAS) PROJECT

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Abstract- **The Telemetric Electrode Array System (TEAS) project aims at developing and embedding entirely into the head, a three-dimensional intracortical electrode array with all electronics required for signal acquisition, processing, and wireless communication. A general description of the system, the main design issues, and its capabilities are briefly described.**

***Keywords* – Intracortical electrode array, telemetry, wireless, neuronal recording, cortex, TEAS**

I. INTRODUCTION

The ability to monitor the activities of ensembles of single neurons is critically important in understanding the principles of information processing in the brain that underlie perception, cognition, and action. Multiple micro-electrode recording using appropriate neuronal implants provides this ability and, moreover, may lead to many possible applications such as the development of prostheses for restoration of sensation such as vision as well as control of limb movement, to name a few.

Although understanding the human brain with its several billion neurons is a formidable task, recent breakthroughs are leading the way to the realization of such “brain-machine interfaces”. For instance, in latest experiments performed at Duke University, it was demonstrated that it is possible to detect a monkey’s intention of making a specific movement a few tenths of a second before it actually happens by tapping into multiple neurons in different parts of the cerebral cortex [1]. This demonstration among others suggests that it is worth pursuing the development of neurosurgical implants designed to restore motor functions that could eventually be used to allow human brains to control artificial devices such as a prosthetic arm or even regain control over their own limbs.

For such applications, it will be highly desirable to eliminate percutaneous connectors and cables. By encapsulating an entire brain implant into the skull and using wireless communication through the skull would eliminate pathways through the skin that could risk infections and also eliminate electrical artifacts due to cable movement. Many researchers have recognized the potential advantages of a wireless brain implant and it is not surprising that Duke’s biomedical engineering department has begun to develop a telemetry chip that would collect and transmit data through the skull.

Our development strategy in the Telemetric Electrode Array System (TEAS) project differs from the one pursued by the group at Duke University. Although we believe that their approach to design at the chip-level has many advantages, we initially chose a system-level approach where we prioritize the selection of existing off-the-shelf technologies as much as possible while concentrating our development efforts on system integration, and only design and fabricate parts that do not exist. The chip-level approach has the advantages of potentially developing smaller systems and to allow more control over the system’s specifications. On the other hand, the increasingly variety of off-the-shelf components with proven specifications available in increasingly higher density in smaller packages, makes the system-level approach a viable and powerful alternative with the opportunity to take advantage of many new technologies suited for wireless brain implants.

II. PROPOSED SYSTEM

A “total implant” where the goal is to embed as much functionality as possible might be considered as a step backward in the development of future brain implants. Besides the fact that implementing much hardware and software outside the skull instead of being part of the implant would reduce the size of the implant itself, decrease its power consumption and heat generation, while decreasing the risk of permanent defects, it would allow easier services and upgrades, an important concept considering the fast rate at which the electronics and computer technologies improve. As such, the TEAS project aims at developing the minimum hardware and software required for an external computer to interface effectively with the brain, by adopting the “brain-machine interface” concept over the “total implant” concept. Nonetheless, such a wireless implant based on the “brain-machine interface” concept is still a relatively complex system requiring many components.

The first version of the proposed brain implant consists initially of a 64 needle-type electrodes in a 8×8 array connected to an electronic system through a special polyimide flexible cable. The neuronal signals recorded by the electrode array at 1 mm deep in the cortex are routed to front-end amplifiers through the special flexible attachment. The amplified signals are digitized at 37.5 kHz per electrode using 12-bit converters prior to being sent to a custom triggering circuit in the implant. The triggering circuit allows the detection of extracellular action potential waveforms (i.e. spikes) in real-time in order to remove the inter-spike data and hence, to optimize the wireless communication bandwidth as well as minimize the size of data storage required by the implant. As the initial version of the system is developed as a research instrument and not solely as a prosthesis device, the need to temporarily store pre- and post-triggered data, so as to record the entire spikes for further analysis, complicates to a large extend the entire implementation through additional digital control circuitries and buffering. The triggered data are fed to a wireless communication interface to be transmitted through the skull

0-7803-7211-5/01$10.00©2001 IEEE

Report Documentation Page

Report Date	Report Type	Dates Covered (from... to)
25 Oct 2001	N/A	-

Title and Subtitle Development of a Wireless Brain Implant: The Telemetric Electrode Array System (TEAS) Project	**Contract Number**
	Grant Number
	Program Element Number
Author(s)	**Project Number**
	Task Number
	Work Unit Number
Performing Organization Name(s) and Address(es) BioInstrumentation Laboratory Massachusetts Inst of Technology Cambridge, MA	**Performing Organization Report Number**
Sponsoring/Monitoring Agency Name(s) and Address(es) US Army Research, Development & Standardization Group (UK) PSC 802 Box 15 FPO AE 09499-1500	**Sponsor/Monitor's Acronym(s)**
	Sponsor/Monitor's Report Number(s)
Distribution/Availability Statement Approved for public release, distribution unlimited	
Supplementary Notes Papers from 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, October 25-28, 2001, held in Istanbul, Turkey. See also ADM001351 for entire conference on cd-rom., The original document contains color images.	
Abstract	
Subject Terms	
Report Classification unclassified	**Classification of this page** unclassified
Classification of Abstract unclassified	**Limitation of Abstract** UU
Number of Pages 4	

to an external computer for further processing and storage. The wireless communication interface is based on BlueTooth technology with a maximum transmission rate of 700 kb/s over several meters. The initial version of the implant is designed to sustain a peak neuronal firing rate of 375 Hz simultaneously on 64 recording channels while transmitting 900 spikes per second with 1.65 ms of recorded data per spike. The bi-directional wireless communication link allows the pre-selection of a sub-group of electrodes for real-time transmission among 64 possible channels prior to recording.

III. Three-Dimensional Electrode Array

A. Types of Multi-Electrode Arrays

Multi-electrode systems for single unit recording can be grouped into two broad classes: the ones that include microdrive mechanisms and those that have fixed electrode arrays. Systems with microdrive mechanisms allow for the vertical positioning of electrodes in the tissue being examined. In its most elaborate configuration, it allows the independent manipulation of a few tens of electrodes that can be driven through the dura and into the brain. These systems have the advantages that they can actively search for neurons of interest and position the tip of the electrode near the soma of the neuron to improve the signal-to-noise ratio (SNR). On the other hand, their disadvantages are two-fold. First, they generally involve daily insertion and removal of the electrodes so that they cannot record from the same neuron over long periods of time. The ability to record from the same neuron over a lengthy period is important for many applications such as the observation of neuronal plasticity that accompanies behavioral learning or memory. The other disadvantage is that these systems do not allow for recording from more than several dozen electrodes because of space limitations and the time required to independently position, so many electrodes.

Fixed electrode arrays, on the other hand, are placed in the brain during an initial surgical procedure and are not re-positioned thereafter. One of the simplest technologies is recording from multiple micro-wires or hatpin like electrodes that are inserted individually into the brain. Despite its simplicity, it is extremely time consuming and therefore, has not seen widespread use. More recently, wire bundles have been developed which are inserted into the cortex as a unit [1]. The University of Michigan Center for Neural Communication Technology has developed a variety of multi-electrode devices using thin film silicon substrates [2, 3].

A very interesting approach is the silicon-based electrode array developed at the University of Utah [4, 5]. It consists of 100 electrodes arranged in a 10×10 matrix with recording sites at their tips and separated from each other by 400 µm. This particular approach is interesting for us because the entire three-dimensional electrode array consists of a single component. A lower number of components often lead to better repeatability in the final specifications, a more robust implant with a lower risk of defects in long-term applications, and faster implantation during surgery.

The major disadvantage of all these fixed array systems is that they do not offer the ability to actively hunt for neurons and to place the electrode tips near the soma of the neurons. For this reason, the input impedances of the electrodes are generally lowered to enhance their ability to record distant signals at the cost of lowering the SNR as well. Nonetheless, it is of paramount importance in the field of systems neuroscience to simultaneously record from very large numbers of neurons in order to better understand distributed neural representations in the brain and as such, the electrode array of the TEAS project is based on a fixed array for multiple neuron recording somewhat similar to the Utah array.

B. TEAS Multi-Electrode Array

The structure of the TEAS array is done using wire electro-discharge-machining (EDM) techniques. This numerically controlled approach allows us to precisely machine more complex microstructures than would be impossible to do with a diamond saw as is used to create the Utah array. Furthermore, this approach allows us to use commercially available computer-aided mechanical drawing software and to see the result before the array is machined. Although any conductive materials could be used including doped silicon, tungsten carbide was initially used because of its fine grain and hardness.

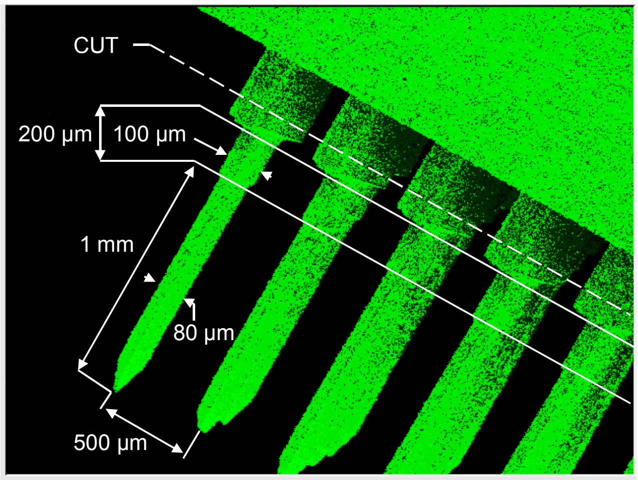

Fig. 1. Image of the structure of the TEAS initial electrode array

Fig. 1 shows an electron-microscope image of the array. The entire array or just the tips of the needle electrodes are coated with platinum (for enhancing the charge transfer capabilities) using electron-beam deposition prior to applying an insulation coating of glass using electron-beam deposition or a biocompatible epoxy through a dipping process. In the case of dipping, the surface tension is often too high and heating the array above the annealing temperature of tungsten carbide prior to the dipping process to reduce the surface tension may also be required.

IV. Interconnections and Insulation

A. Flip Chip Process

Providing independent electrical access to each electrode in the array is a non-trivial task. Our first approach to interconnect the TEAS array was to use flip chip mounting based on stud bumping. With this high-density mounting method, an additional coating of aluminum or gold is applied as for the wire bonding techniques over tungsten carbide, platinum or iridium at the interconnection surface of each electrode to allow ultrasonic wire bonding. The process of stud-bump formation is the same as that of wire bonding IC die to lead frames. First, as in conventional wire bonding, a gold wire is passed through the hole in a capillary tube and subjected to an electrical discharge to form a gold ball, which is bonded to the base of the electrode in the array by thermo-compression and ultrasonic energy. Then, the capillary tube clamping the wire is withdrawn and the wire is cut. The resulting stud bump has non-uniform bump height due to the relatively low Young's modulus of conventional bonding wire. Even with a higher modulus bonding wire, stud bump leveling or coining process where a flat surface is pressed at approximately 30 grams per bump is often required to optimize bump contact especially when conductive adhesive is used in the connection process. A simultaneous coining process on all stud bumps is desirable for better co-planarity.

One method to fix the stud bumps on the electrode array itself or the flexible attachment (because of possible space constraints on the wire bonder), is to dip the stud bumps in an isotropic conductive adhesive prior to being aligned and cured to provide the final connection. As the dimensions of the array and/or stud bumps decrease, the dipping process may become difficult and as such, another approach is to use anistropic conductive adhesive (ACA) or film (ACF). The last approach typically requires an aligner-bonder with a work stage that can be heated to a minimum of approximately 200^{o}C for curing purpose while applying a force with a load of approximately 100 grams per bump to squeeze the conductive particles within the adhesive to provide a low resistance electrical conduction path between the stud bumps and the bond pads. Past experiments have shown that the most suitable amount of conductive particle content is within the range of 2.5 to 12.5-vol%. The insulation resistance of the neighboring interconnections measured in the condition of 12.5% content of conductive particles at 50 Volts is said to be more than 10^{12} Ω. For both approaches, the tips of the electrodes must typically be encapsulated into a temporary plate or block (possibly wax) to create proper air suction during the pick-and-place process. Furthermore, in the case where ACA or ACF is being used, the temporary block (made of wax or similar material) must protect the needles when the compression force is exerted.

B. Through Hole Process

With the flip chip process, an additional layer for electrical insulation between the needle electrodes is required. Even though such a layer can be made relatively easily, by melting and cooling glass powder or curing biocompatible epoxy, it was desirable to simplify the process and reduce the number of components by using the flexible connection attachment itself not only to provide electrical pathways between the needle electrodes and the electronics block but also to hold the electrodes together while providing the insulation layer.

As a first step, the needle electrodes are inserted through an aligned matrix of un-plated holes on the flexible attachment until the top of the attachment reaches the end of the 100 μm width section (see Fig. 1). The holes are un-plated to minimize input capacitance. Each hole has a minimum diameter of 7 mils (~175 μm) to allow the 100 μm × 100 μm square section of each needle electrode to be inserted while compensating for a maximum positional tolerance of ± 25 μm during the drilling process. A 100 μm and 200 μm sections are also added on top of the 80 μm section to form a pyramid-like shape that helps achieving a higher aspect ratio of the needle electrodes and/or make them more robust. Once completely inserted, the 200 μm width section is cut (see Fig. 1) with an EDM wire to form square pads, which are then soldered on top of the larger pads used as electrical contacts on the flexible attachment. Once soldered, the whole interconnection region on top of the array is covered with a biocompatible epoxy.

B. Flexible Attachment

The flexible attachment provides the electrical connections between the needle electrodes and the electronic block. We designed the attachment using computer-aided design (CAD) software such as schematic entry and printed circuit board (PCB) layout tools. The initial attachment consists of a single conductive layer of a polyimide-based 200 μm maximum thickness flexible substrate with 64 (¼ once) 50 μm wide conducting traces with 50 μm inter-trace spacing running in parallel on the top layer between the array and the electronic block. These specifications were chosen to ensure small dimensions while providing a good yield during manufacturing.

A stiffener is inserted in the flexible attachment on the top section of the electrode array to prevent excessive bending of the array. On the other hand, flexibility between the array and the electronic block is of prime importance due to brain motion. As the brain moves relative to the skull, it is important to minimize tethering forces that would result in movement of the electrode array relative to the cortex. The TEAS array has been designed to initially sustain brain shifts of up to 2 mm in the skull due to cardiac and respiratory rhythms and other mechanical perturbations. Because a horizontally flat flexible attachment where the width is much higher than its thickness only provides low stiffness for up-down brain shifts, cuts using a laser are made between the parallel conductors with just sufficient lengths to ensure minimum stiffness within the maximum expected range of motion in all directions.

V. ELECTRONIC BLOCK

Unlike the rest of the system, the electronic block is fixed and attached to the skull. It consists of the following main sections: the front-end amplification, the analog-to-digital (A/D) conversion and multiplexing, the triggering, control, and buffering subsystem, the wireless communication interface, and the power section.

Although an analog multiplexer at the front-end could have reduce the number of analog electronic components in term of amplifiers, it would have also significantly degraded the input characteristics while increasing noise through additional loading errors through a reduction of the input impedance, additional charge coupling effects, offset, drift, leakage current, capacitance, and settling errors. The amplifiers are arranged to provide high input impedance and sufficient gain to maximize SNR. The amplification block uses quad-amplifier TSSOP14 packages mounted on both sides of the board.

Eight 8:1 analog multiplexers integrated with the 12-bit A/D converters provide the digitized signals to the triggering circuit at a rate of 37.5 kHz per channel using 62 samples (1.65 ms) per spike with a lock-out time of 1 ms. A programmable 12-bit threshold on each channel is provided on the triggering block. Although a maximum of three pre-selected recording channels can be used simultaneously when detailed waveforms of all the spikes must be transmitted through the wireless communication path when a maximum spike firing rate of approximately 300 Hz per channel is expected, other recording modes are also available. For instance, up to 60 recording channels can be sampled when a maximum neuronal firing rate of 15 Hz per channel is expected. Bursts of spikes can be captured by an onboard 2MB buffer, which can hold up to 20,000 spikes, or one second of 300 Hz activity on all 64 channels. The information in the buffer can be transmitted in 25 seconds over the wireless link. Finally, if the waveform of each spike does not need to be transmitted but only the time stamp values of the triggered neurospikes, the implant will support real-time transmission of all 64 channels at a maximum neuronal spiking rate of 300 Hz per channel. Power is provided from batteries charged using induction through the skin.

VI. CONCLUSION

A brief overview of the conception of a wireless brain implant has been provided. The whole system to be embedded into the skull consists of a three-dimensional electrode array interconnected to an electronic block through a special flexible interconnection cable. The electrode array is fabricated using electro-discharge machining techniques. Two attachment processes based on flip chip and through hole techniques have been described. Although through hole assembly has been chosen for the first version of the implant to minimize the number of parts and simplify the assembly process, it is anticipated that the flip chip will be used as the feature sizes decrease. As the number of electrodes within the same array increases, the diameter of the needle electrodes will need to be decreased to minimize the insertion force and damage to the brain. As such, other fabrication techniques will be required as well as improvement in wireless communication.

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