Chronic, Percutaneous Connector for Electrical Recording and Stimulation with Microelectrode Arrays

Kedar G. Shah, Kye Young Lee, Vanessa Tolosa, Angela Tooker, Sarah Felix, William Benett, Satinderpall Pannu

Abstract— **The translation of advances in neural stimulation and recording research into clinical practice hinges on the ability to perform chronic experiments in awake and behaving animal models. Advances in microelectrode array technology, most notably flexible polymer arrays, have significantly improved reliability of the neural interface. However, electrical connector technology has lagged and is prone to failure from non-biocompatibility, large size, contamination, corrosion, and difficulty of use. We present a novel chronic, percutaneous electrical connector system that is suitable for neural stimulation and recording. This system features biocompatible materials, low connect and disconnect forces, passive alignment, and a protective cap during non-use. We have successfully designed, assembled, and tested *in vitro* both a 16-channel system and a high density 64-channel system. Custom, polyimide, 16-channel, microelectrode arrays were electrically assembled with the connector system and tested using cyclic voltammetry and electrochemical impedance spectroscopy. This connector system is versatile and can be used with a variety of microelectrode array technologies for chronic studies.**

I. Introduction

The ability to translate fundamental developments in electrical stimulation and recording of neural tissue into clinical practice will require the ability to perform chronic studies in awake and behaving animal models [1,2]. In recent years, significant advances have been made to extend the lifetime of neural interface and probe technology [3-6]. An example is the polyimide-based microelectrode arrays developed by Lawrence Livermore National Laboratory for the Department of Energy Artificial Retina program that demonstrate a lifetime of greater than 20 years *in vitro* and have been implanted in humans for more than 7 years [7].

Percutaneous microelectrode arrays stimulate or record from neural tissue by interfacing with external instrumentation using electrical connectors. A variety of electrical connectors have been translated for use in neural interface devices from the electronics industry, such as zero-insertion-force connectors [8], Omnetics connectors [9,10], and Amphenol [11]. While these connectors work well for electronics application, they are not designed for the rigors and constraints of *in vivo* experiments in awake and behaving animals. There is a need for percutaneous electrical connectors for neural interface devices with high-density, biocompatibility, ease-of-use, and robustness.

In clinical practice, devices will be fully implanted and wireless, but such devices are not always practical for use in research and pre-clinical studies due to cost, complexity, and difficulty to reconfigure. There is a need for reliable and chronic percutaneous electrical connectors to bridge the gap between acute experiments and clinical trials [12].

II. Motivation

State-of-the-art percutaneous electrical connectors used in neural interfaces are often adopted from the electronics industry, where it is desired to have high-performance and robust connections that do not easily separate. Furthermore, the choice of materials is largely driven by the cost-reduction and high-volume manufacturing practices in the industry. The neural interface community has adopted some of these connectors for use with microelectrode arrays, such as Omnetics connectors [9] and zero-insertion-force (ZIF) connectors [8] as shown in Fig. 1.

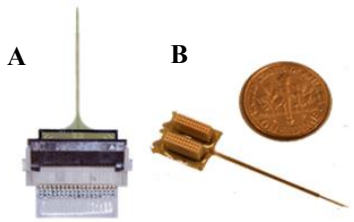

Figure 1. (A) polyimide microelectrode array with zero-insertion-force (ZIF) connector (LLNL); (B) polyimide microelectrode array with dual 16-channel Omnetics connectors (LLNL)

Though these connectors may be adequate for acute studies, there are significant drawbacks that need to be addressed:

A. High Connect-and-Disconnect Forces

High connect-and-disconnect forces required by a majority of current connectors are significant disadvantages for use in awake and behaving animal models. Animal movement during the connect/disconnect cycles can cause stress to the animal which can affect behavioral studies, and in some cases, physical damage/trauma to the animal when significant force is used with small animals. The difficult handling also makes the interface of the device with the connector prone to failure, such as broken or damaged connector pins.

In awake and behaving animals, a safety release mechanism that disengages the tethered connector from the

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implanted connector is desirable. This ensures the device is not damaged and the animal is not harmed during device implantation or during recording and stimulation studies.

B. Low Density and Large Size

Most connectors are fabricated using machining and molding practices that limit miniaturization. The result is that most connectors cannot practically make more than 32 or 64 connections without a large footprint, making them impractical for smaller animals or in cases where multiple connectors and devices are used for high-channel count recording and stimulation studies.

C. Non-biocompatibility

Most microelectronic connectors use materials and processes that permit high-volume manufacturing, enabling significant cost reduction for non-biological applications, for example solders, copper, plastics, and epoxies. As a result, most connectors are not built from chronically biocompatible materials.

III. DESIGN

The design of the chronic percutaneous connector described in this publication is based on detailed discussions with, and feedback from neuroscientists who perform chronic experiments in a variety of animal models and for a range of applications such as auditory prostheses, brain-machine interfaces, cortical implants, deep brain stimulation, and spinal cord stimulation.

A. Components and Materials of the Connector

The connector consists of three main components: (1) percutaneous connector, (2) interconnect sheet, and (3) external connector, as shown in Fig. 2. These components are designed to feature high-density device channels, chronic biocompatibility, reliable electrical contact, and ease-of-use.

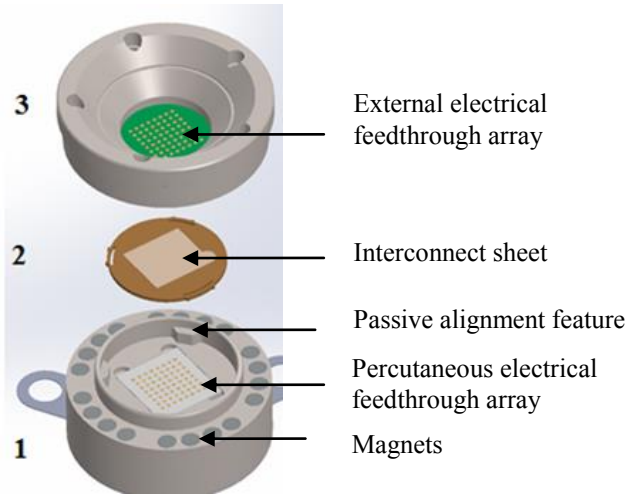

Figure 2. Schematic of connector system showing the three main components: (1) percutaneous connector, (2) interconnect sheet, and (3) external connector with magnetic clamp.

1. Percutaneous Connector

The percutaneous connector, which is exposed to tissue, consists of a medical grade titanium housing that provides high strength and corrosion resistance and is suitable for long-term *in vivo* experiments on the order of years. The perimeter of the connector contains magnets that engage the clamp mechanism. It also has a screw plate with holes allowing it to be attached to the animal using metal screws and/or cement. The housing holds an alumina electrical feedthrough array (EFA); a flexible thin-film neural device is attached to the backside of EFA by rivet bonding [13].

2. Interconnect sheet

The interconnect sheet is a biocompatible polymer sheet with gold-plated wires embedded in. When this dense forest of wires is compressed between the opposing percutaneous and external connectors, electrical contact is made from the neural device to the external connector (Fig. 2(3)).

In this chronic connector system, the magnitude of compressive force is fixed by the magnetic clamping mechanism. This interconnect sheet is inexpensive and replaceable, ensuring the long term use of this chronic connector.

3. External Connector

The external connector's housing consists of titanium for its strength and corrosion resistance. The perimeter of the external connector also contains magnets whose polarities match with those of magnets of the percutaneous connector to allow magnetic clamping mechanism. The EFA for the external connector is made of laminates, a typical material used for building printed circuit board; it is held by the external connector housing and can freely rotate inside the magnetic clamp prior to the assembly of the connector system. Both the external and percutaneous EFA have identical numbers and sizes of contact pads; these contact pads must be properly aligned to make reliable electrical contacts.

B. Magnetic Clamping Mechanism

The novel magnetic clamping mechanism utilized by our connector system significantly improves the ease-of-use. The magnetic clamp applies the necessary force for making reliable electrical contact, without the need for screws or clips.As shown in Fig. 3A, the magnets in alternating polarity were arranged around the perimeter of both the percutaneous and external connectors. The two connectors are aligned using the passive alignment mechanical features (Fig. 2) in the connectors with the interconnect sheet between them. The magnetic clamp of the external connector is brought into contact with the percutaneous connector and rotated until the opposing magnetic polarities align to clamp the two connectors together. For intentional disconnection, the external connector is rotated partially, which aligns magnets of equal polarity together, and the opposing force disconnects the connectors. In accidental disconnection through unexpected movement of the animal, the mated magnetic force is overcome, which disconnects the system, thus preventing injury to the animal and damage to the

device. In this case, the force required to separate the external and percutaneous connector depends on the angle at which the force is applied. A force applied at an angle can induce rotational motion between the external and percutaneous connector, causing the equal polarity to align and thus making the separation easier. The extreme case of an axial pull of the external connector from the percutaneous connector approximately requires 4.9N for complete separation.

The external connector is only attached to the animal during the time of electrical stimulation and/or recording. During non-use, a magnetic mating cap with an O-ring seal keeps moisture and contamination out, as shown in Fig. 3B.

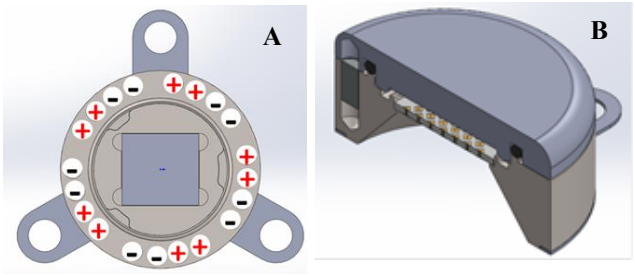

Figure 3. (A) schematic showing alternating polarity of magnet arrangement on percutaneous connector side, and (B) cross section of the connector system assembly with the protective cap during non-use.

C. *Design Factors for Channel Density*

In our design of EFAs, each contact pad is square and equidistant to each other. The channel density was determined by performing tests of a range of bond pad dimensions, separated with a range of gaps between them. The minimum size of a contact pad and minimum gap among contact pads (Table 1) were determined by testing daisy-chained chips (Fig. 4).

To determine the minimum size of a contact pad, a pair of chips (Fig. 4A and 4B) for a given contact pad dimension is designed to form a daisy chain (Fig. 4C) when in electrical contact with each other with an interconnect sheet between the contact pads. Good electrical contact is made if embedded wires of the interconnect sheet connect the contact pad on top with the corresponding contact pad on the bottom (Fig. 5A), indicated by a resistance measurement at the probe pads. If the contact pads are too small, then the wires of the interconnect sheet fail to connect the corresponding top and bottom contact pads, resulting in an open circuit (Fig. 5B). To test the minimum gap between contact pads, a pair of identically designed chips (Fig. 4D) for a given gap were brought in contact to each other with an interconnect sheet between the contact pads. A sufficient gap allowed the interconnect sheet to connect only the matching top and bottom contact pads, keeping the circuit open. However, if the gap was too small, it created an undesirable short circuit (Fig. 5C). By testing various sizes and gaps, we determined a value of 300 µm as the minimum contact pad side length and gap between them. We exceeded these minimum values when designing the EFAs to ensure reliable electrical contact, as shown in Table 1.

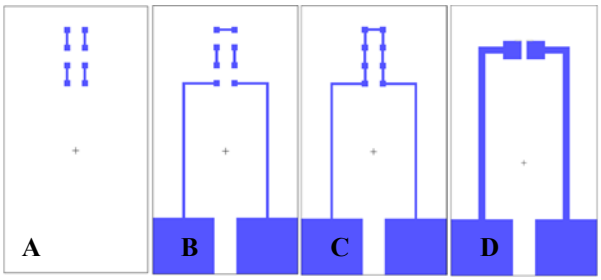

Figure 4. Chip schematics to test design factors for channel density; the probe pads are large squares located at the low ends of the chips.

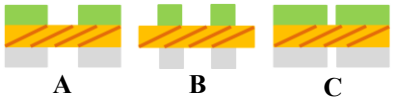

Figure 5. Cross sections of contact pads (green, gray) and interconnect sheet consisting of polymer (gold) and slanted metal wires (red)

TABLE I. DESIGN FACTORS OF EFAS

	16-Channel EFA	64-Channel EFA
Side Length	838 µm	305 µm
Edge-to-edge Gap	432 µm	330 µm

IV. TEST

We tested the functionality of the chronic connector system *in vitro* by assembling the connectors with a custom-fabricated neural device of 16 channels and 100 µm diameter electrodes, connecting the assemblage to a potentiostat (PAR VersaSTAT 4, Oak Ridge, TN), and performing electrochemical characterization in a three-electrode cell. All electrochemical data was acquired at room temperature in non-purged phosphate-buffered saline (PBS) with a silver-silver chloride (Ag/AgCl) electrode used as a reference electrode and a platinum wire used as a counter electrode. Electrochemical impedance spectroscopy (EIS) data was collected at frequencies from 300 kHz to 1 Hz with a 10 mV amplitude sinusoidal signal. Cyclic Voltammograms (CVs) were collected at a 100 mV/s scan rate over the potential range of -600 mV to +800 mV.

The lifetime of the chronic connector system and its electrical contact reliability were studied by connect-disconnect tests of the percutaneous and external connectors with an interconnect sheet reassembled between each test. To simplify the process, the contact pads on the percutaneous and external EFAs were daisy-chained so that a single electrical path from a contact pad at one end to another end is formed. Then, resistance at these end contact pads was measured to ascertain electrical contact of the entire connector system. Prior to resistance measurement, the external connector was made sure to make necessary contact by gently pressing down on the connector. Assuming a practical chronic connector usage condition, 1 connect-disconnect per week for 2 years, we conducted 104 tests on both the 16-channel system and 64-channel systems.

V. Results and Discussion

Fig. 6 shows the fully assembled connector system and its individual components.

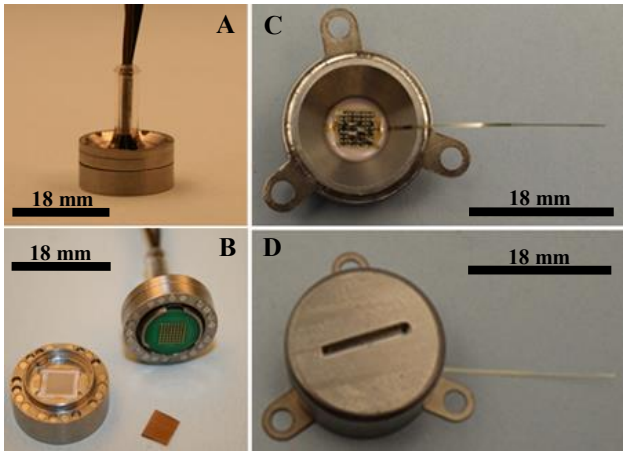

Figure 6. Fully assembled connector: (A) side profile of connector attached to wires leading to external electronics (B) image of three main components of the connector showing details of magnets, EFAs, and alignment features, (C) bottom view showing polyimide device attached to the percutaneous EFA, (D) top view with protective cap for non-use.

Electrochemical data, shown in Fig 7, verified successful installation of the chronic connector system of a 16 channel device. Each CV and EIS curve corresponds to one of 16 electrodes of the installed device, and the smooth overlaps of both the CVs and EIS data indicates that the chronic system did not impede electrical performance of the device. The impedance of the neural device was measured to be 26.9 kΩ ~ 40.15kΩ at a frequency of 1.2 kHz; this range of impedance is reasonable compared to other similar electrodes of 100 µm diameter, indicating that the chronic connector system has low impedance.

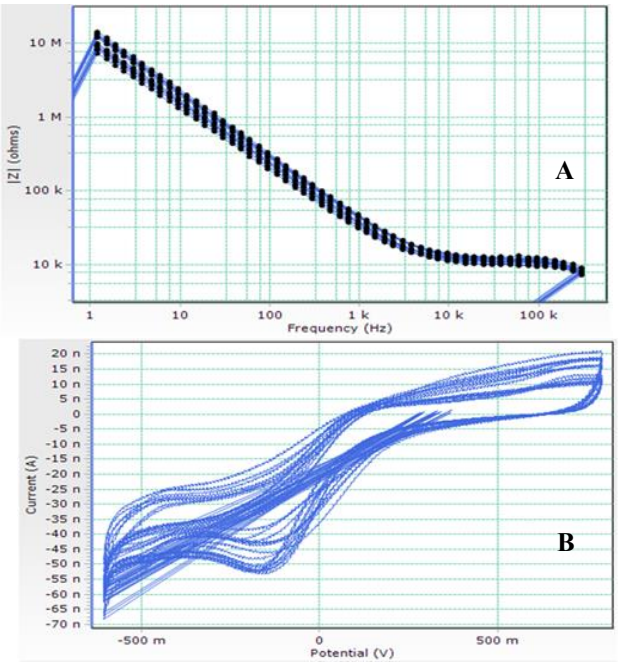

Figure 7. Electrochemical data for the electrode array assembled onto a chronic connector (A) cyclic voltammetry, and (B) electrochemical impedance spectroscopy for all 16 channels.

The results of connect-disconnect tests verified the chronic connector's reliability: good electrical contacts of all contact pads without opens were made 97.12% in 16-channel system and 98.08% in 64-channel system. In the case of successful contact, the average resistance of 16-channel system assembly was 19.1Ω and the average resistance of 64-channel system assembly was 159.9Ω. These resistance values are insignificant compared to the typical kilo-ohm resistance range of microfabricated neural interface devices. Current work is underway to use this connector in a chronic animal study preparation, and the results will be reported in a future publication.

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