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Portable, Chronic Neural Interface System Design for Sensory Augmentation

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Abstract- **While existing work in neural interfaces is largely geared toward the restoration of lost function in amputees or victims of neurological injuries, similar technology may also facilitate augmentation of healthy subjects. One example is the potential to learn a new, unnatural sense through a neural interface. The use of neural interfaces in healthy subjects would require an even greater level of safety and convenience than in disabled subjects, including reliable, robust bidirectional implants with highly-portable components outside the skin. We present our progress to date in the development of a bidirectional neural interface system intended for completely untethered use. The system consists of a wireless stimulating and recording peripheral nerve implant powered by a rechargeable battery, and a wearable package that communicates wirelessly both with the implant and with a computer or a network of independent sensor nodes. Once validated, such a system could permit the exploration of increasingly realistic use of neural interfaces both for restoration and for augmentation.**

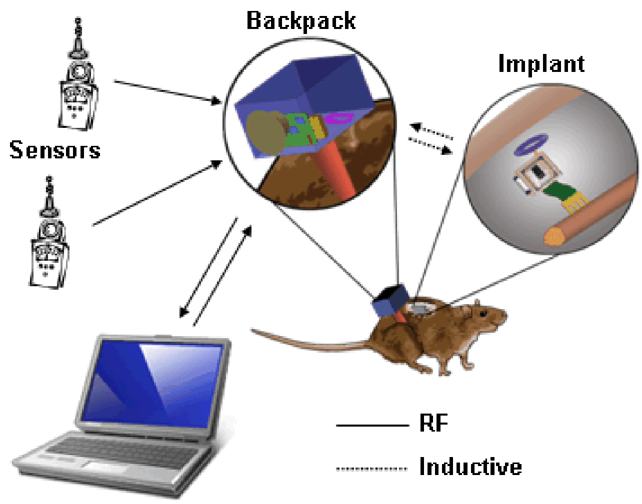

Figure 1. Neural interface system operation. A portable backpack communicates with an implanted microsystem, with sensors and/or a computer.

I. INTRODUCTION

Continuing progress in interfacing to the nervous system through direct electrical connections holds the promise of restoring lost function in amputees and victims of spinal cord injury (e.g. see [1-3]). The development of this technology also raises the tantalizing prospect of someday enhancing human sensorimotor capabilities through direct electrical recording and stimulation of neurons. The possibility of adding new capabilities to individuals, for instance by giving them "thought control" over robots or allowing mind-to-mind communication [4], depends on a central hypothesis that to date is largely untested (though the results of some studies are suggestively supportive): there exists sufficient and accessible excess capacity in the nervous system to permit the performance of significant additional, unnatural functions without substantially interfering with the nervous system's natural functions.

While the idea of human augmentation raises significant and legitimate ethical questions about how and when such technology should be used, if ever, there are undeniable potential benefits. In the military arena alone, conflicts are increasingly fought in confusing, complex environments that decrease the advantages brought by technology due to the unwieldiness of heavy weaponry, the proliferation of small arms and night vision technology, and other effects including the often-overwhelming amount of sensory data to digest. The enhancement of situational awareness for individual soldiers through a set of neurally-tied artificial sensors could provide a new advantage to soldiers operating in these challenging environments and create a new means of protection. Neural interfaces could one day provide a level of intuitive interpretation of sensory data about an environment that would be unattainable through any alternative method of data transmission.

Given the limitations of machine intelligence, such systems may provide an elegant way to efficiently insert the nuances of human intelligence into semi-autonomous systems. Similarly, bidirectional neural interfaces could enable an advanced form of "remote presence," allowing intuitive control with instantaneous response to guide remotely-located vehicles or robots. Such activities could one day be the purview of amputees who choose to return to military duty as "neural interface specialists," who are not only able to control replacement limbs neurally but can also use their neural interfaces to support new, high-performance capabilities.

We have chosen to focus on the question of whether an artificial sense can be introduced through electrical nerve

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stimulation without substantially impairing natural sensorimotor function, and have set as a goal to conduct initial tests of this concept, in partnership with our academic collaborators, in an animal model this summer. With this vision of sensory augmentation accompanying that of restoration, we also approach the need for appropriate neural interface technology from an engineering systems perspective, envisioning a system that could be used beyond the research laboratory in an active, healthy population without assuming unproven technological advances. In this paper we discuss the requirements for a neural interface system that is highly portable and safely implantable in healthy subjects as well as recovering injured subjects. We describe the initiation of our efforts to produce a robust portable, wireless system that can be used regularly for experiments on freely-behaving animals and (in later incarnations) in humans, rather than only in proof-of-concept demonstrations. We provide an overview of the system design, which uses available technology in micro- and meso-scale electronics, neural interface electrodes, and wireless communication, place this system's capabilities in the context of the current state-of-the-art, and discuss progress to date. We also discuss the key technology gaps and critical experimental questions that must be addressed before augmentation can be further pursued.

II. Requirements for a Neural Interface System for Augmentation

At many levels the requirements for a system intended for sensory augmentation are very similar to those required for advanced neurally-controlled prosthetics. However, while safety is a critical concern in any system intended eventually for humans, safety imposes even tighter restrictions when technology is to be used for healthy subjects. For instance, an amputee may be willing to accept some risk of nerve damage in a residual limb that is largely non-functional, while a healthy subject must be certain that there will be no permanent damage. Thus invasiveness is perhaps the paramount safety-related concern for a neural interface system. An ideal system would be completely noninvasive, yet using noninvasive techniques demonstrated to date, the rate and specificity of data flow is insufficient to yield improvements over traditional, non-neural sensory modalities. By contrast, using multiple implanted electrode sites permits a large exchange of information in an extremely small physical space; the greatest challenge lies in extracting and translating this information (e.g. see [5]). With current technology, we believe that implanted electrodes will be required for meaningful and precise neural sensory augmentation.

Assuming an implant will be needed, several choices are available for its placement. Investigators have explored neural interfaces in the cortex [1-3], spinal cord, and several locations in the peripheral nervous system (PNS) including the dorsal root ganglion [6], the sciatic nerve [7], and the auditory and optic nerves. To serve the goal of adding a sense, the preferred location is not obvious. The cortex, when stimulated, may prove more amenable to plastically adapting to the concept of a new sense, as opposed to peripheral nerves which are hard-wired for specific purposes. Experiments in which rats [1], monkeys [2] and humans [3] have gained the ability to control computer cursors or robots with their thoughts demonstrate an ability to gain new, unnatural functions, whether or not they are replacing lost functions. Peripheral nerve signals are subject to substantial pre-processing before reaching the cortex, which could prove advantageous or disadvantageous. Individual neurons are certainly more easily isolated in the PNS. Aside from the (largely unanswered) questions of effectiveness in this application, however, we argue that in practice, implanting healthy individuals in their limbs or torsos will prove much more tractable than cortical implants. Peripheral nerve signals are also more straightforward and deterministic than the complex signals observed in the cortex.

While the overwhelming majority of neural recording and stimulation is currently done using percutaneous wires or connectors, the risk of irritation and infection makes this clearly impractical for chronic human use, particularly in healthy subjects. Thus the implanted portion, apparently essential for high resolution, must be completely implanted, and must communicate wirelessly across the skin. Regardless of where it is implanted, the device must also be small; less than perhaps 5 cm^3 is a reasonable target, especially if it is to be implanted in a limb.

Nearly as important as the unobtrusiveness of the implant is the portability of the portion of the system that communicates with the implant across the skin. While several groups have demonstrated and continue to refine multi-channel wireless neural interface implants [8,9], this work has largely been conducted in research environments to prove the concept, and has not been reduced to practice for use in regular experiments. One significant problem is that most of these systems attempt to harvest power, as well as data, wirelessly. This makes the implementation of a highly-portable, safe and power-efficient wearable system rather difficult. In general, wireless powering requires at least several volts to be induced at the implant. If the implant is small and it is necessarily separated from the powering coil by at least the thickness of the skin and tissue, requirements on the voltage at the powering coil can quickly rise over 100 V. In addition, to achieve high communication bandwidth this voltage is often varied in the hundred kHz to low MHz range. Constantly generating such high-voltage, high-frequency signals with enough current to inductively power an implant in a compact, safe and moderately low-power package can be problematic. A more practical, truly portable system would require a modest (<10 V) driving voltage in order to make the "wearable" portion of the system appropriately compact, lightweight, safe and power-efficient.

Ultimately it is very likely that the system must be bidirectional, capable of both neural recording and stimulation. While a sense could perhaps be added with stimulation only, humans and animals are used to having motor control over their senses by some means, and this may prove to be an important requirement for cognitively integrating a new sense. Much neuroprosthetics work to date has focused on recording of motor activity, and most of the high-profile findings have derived from this effort (e.g. [1-3]). A group at the University of Utah has recently studied stimulation as well as recording in the peripheral nervous system [10], and numerous investigators have recently expressed the inclination to focus more resources on sensory stimulation. This should lead to an increase in experiments with bidirectional interfaces, which have to date been presented only in limited forms (e.g. see [4]). Our system will support these efforts.

Several practical issues of varying difficulty must also be addressed. Important functional safeguards must be in place to prevent stimulating with excessive magnitude, over-heating surrounding tissue, and other hazards. A significantly more challenging problem is that the system should be made ultimately able to be safely removed, or at least able to be shut down and left indefinitely without danger. This may ultimately require the development of entirely new electrode technologies, perhaps using novel materials that demonstrate a new level of biocompatibility. Given existing technology, the options are generally limited to microwires, penetrating microfabricated probes [11,12], cuff electrodes [13], or sieve electrodes [14]. The latter requires transection and regrowth of nerves, an extremely risky and invasive approach for healthy subjects. Cuff electrodes have not demonstrated sufficient resolution. This leaves electrodes that penetrate the nerve, which may not prove adequate over long periods of time [7]. Improved electrode technologies are an important need for practical implementation of neural interfaces, and are being explored throughout the research community.

A final, significant challenge in interfacing sensors to the nervous system concerns the question of how best to handle the flow of information between sensors or computers and the nervous system. For simple, low-dimensional applications this may be easy, but as this area of study evolves, the efficiency of communications between machines and the nervous system is likely to become an increasingly important challenge.

Given the assessment of requirements and the state-of-the-art summarized above, we have chosen to pursue the design of a bidirectional peripheral nerve interface system with a small, wireless implant using existing penetrating electrodes and a portable, small, and lightweight wearable package. The design of the system and its components are discussed in the following section.

III. System Design

A. *System operation and signal flow*

Given the considerations listed above, we have designed a system to function as depicted in Fig. 1. The system consists of an implant and a “backpack” worn by the test animal. The backpack communicates wirelessly with the implant through an inductive link, and also communicates wirelessly through a second RF link to sensor nodes or a computer. Signals can be sent and received by the computer, or they can be sent from

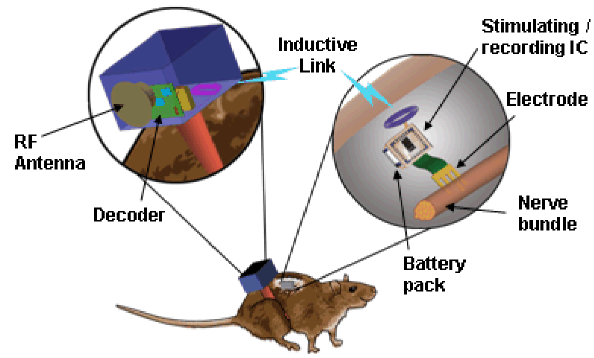

Figure 2. Rat-mounted system components.

individual sensor nodes, each of which has its own processor; the computer is not necessary for system operation. Sensor nodes can be carried by the animal or placed in fixed locations, adding capabilities such as chemical sniffing, infrared or ultrasonic sensing. The computer or sensor node sends the backpack desired neural stimulation commands, which the backpack transmits to the implant along with functional directions. The implant returns status information as well as neural recordings when requested; the backpack relays recordings to the computer. Research studies on freely moving animals would likely use a computer to monitor and control the neural interface; ultimate deployment for everyday human use would likely forego the computer and operate strictly with microcontroller-based sensor nodes along with the neural interface implant and backpack. More detail on the operation of each system element is given below.

B. *Implanted microsystem*

The implanted portion of the system, as sketched in Fig. 2, consists of an application-specific integrated circuit (ASIC), off-chip circuit elements including a battery charging and voltage regulating circuit and three RF chip inductors (Coilcraft) for communication, two 4 V, 3 mAh Li-ion batteries (Quallion) and a custom eight-site silicon penetrating electrode array (NeuroNexus Technologies). The 5 mm^2 ASIC, shown in Fig. 3, forms the crux of the portable neural interface system. The chip was fabricated in the AMI 0.5 µm process and requires a 5 V supply. It reads commands, which are sent digitally in combinations of 1 and 2 MHz pulses, from one of the chip inductors. Each bit consists of two pulses at 1 MHz and two pulses at 2 MHz, where the order of the pulses determines the value of a bit; thus the input bitrate is 333 kbps. The sensitive receiving circuitry can detect pulses with amplitude in the low mV range, and accommodates significant uncertainty in the coupling through the skin. This requires a signal with amplitude of just 5 V at the coupled coil outside the skin for communication. Commands include selecting a site address, stimulating, recording, and communications diagnostics. The ASIC includes a 10-bit analog-to-digital converter (ADC) for recording, which can sample at rates up to 20 ksps. Recorded data and diagnostic information are sent across a second chip inductor to the backpack through a

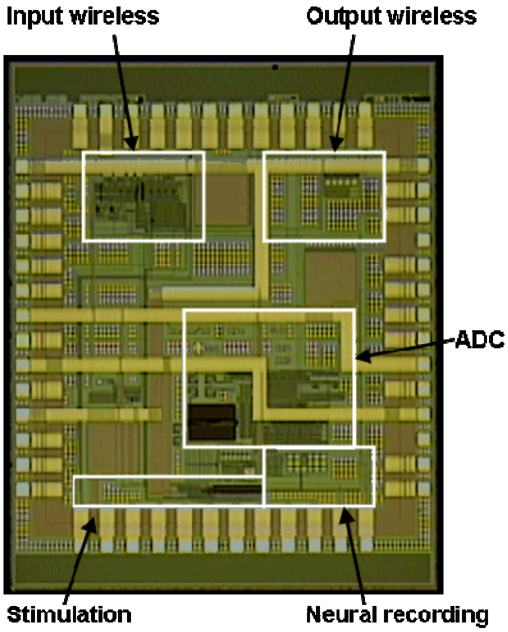

Figure 3. Neural recording and stimulating ASIC.

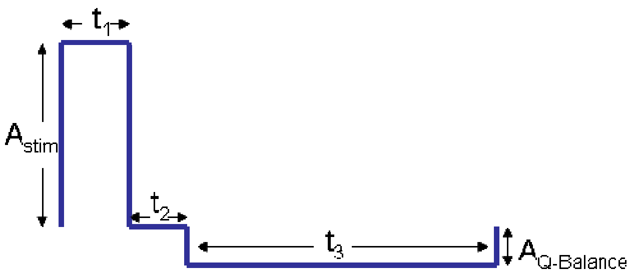

Figure 4. Example of a stimulation waveform.

Manchester-encoded waveform operating at 8.25 and 10.25 MHz. Stimulation waveforms are created by sending successive commands to change the current level, permitting generic waveforms such as the biphasic waveform depicted in Fig. 4. Currents can range from -35 to +35 µA in 2.19 µA increments for sites with impedance less than 50 kΩ. The temporal resolution is 25 µs. Refinements to the ASIC for an upcoming second generation are discussed in section *E*.

Central to the ASIC's ability to accurately read inductive data with low amplitude is the use of battery power. On a single charging cycle, the microsystem can run approximately 8 hours when recording, 15 hours when stimulating, or 20.5 hours when idle. This permits essentially a full day of experiments on a single charge. The third chip inductor is used for wireless battery charging, which requires a sufficient voltage in a coil outside the skin to induce approximately 3 V at the coil (~100 V across driving coil). The battery-charging circuit rectifies and smoothes the input waveform, boosts the voltage with a charge pump, and uses a current mirror to produce a constant current charge (1.5 mA). With the ASIC shut down, batteries can be recharged in approximately 3 hours. Thus the battery can be charged while the subject is in proximity to a high-voltage amplifier (e.g. while the subject sleeps), but such high-voltages are not required for regular

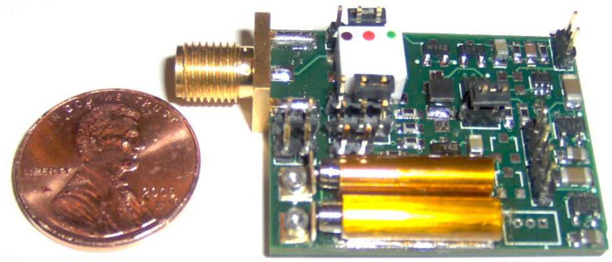
Figure 5. Test board for battery charging and power regulating circuit. Large connector and jumpers are for testing only, not needed for operation.

operation. The circuit also regulates the battery voltage to 5 V and 2.5 V for the ASIC. A test board for the off-chip circuits is shown in Fig. 5.

Arguments against including batteries in wireless neural implants include size and limited lifetime. Despite our use of batteries, the total size of our implant is quite modest; a scale mockup including all necessary components is shown in Fig. 6, and measures just 2.25 cm in diameter and less than 0.8 cm thick. A subsequent generation (described in section *E*) will be significantly smaller still. Because we can run an entire day of testing on a single charge, we expect battery life to be months to years (cycle testing is ongoing). This provides sufficient life for most animal trials. As battery technology continues to evolve, we expect this lifetime to grow to a duration appropriate for long-term human use. Critically, implanted batteries permit a dramatic reduction in voltage levels needed in the backpack (5 V), facilitating true portability. Rechargeable batteries also have a well-established track record for use in implanted devices.

C. Backpack

The backpack, sketched in Fig. 2, includes a microcontroller (Atmel) that communicates serially with an RF chip (HAC) for communication to the sensor nodes or PC. The microcontroller also manages communication with the implant through two dedicated circuits, one for transmitting and one for receiving. The transmit circuitry translates serial data into the necessary sequences of 1 and 2 MHz pulses and drives the inductor circuit. The receive circuitry amplifies the Manchester waveform to a detectable level and identifies the data frequencies. The microcontroller generates stimulation patterns,

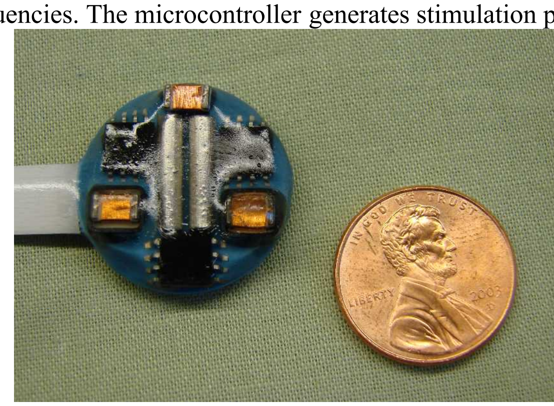
Figure 6. Scale mockup of implant with major components. The three communication coils and two cylindrical batteries are visible on the top side; the ASIC and most of the off-chip circuitry are located on the bottom.

limiting them to a safe, predetermined set of values. A battery pack powers the system. The backpack circuitry can be configured in several different geometries to suit the implant site; most critical is that the coils be aligned with their respective implanted counterparts, and held in place either mechanically or with adhesive. The total volume, which depends on configuration is several in^3. The backpack radio can communicate at a range of up to approximately 500 ft.

D. Sensor nodes and drive PC

Each sensor node consists of a sensor, an RF chip, a microcontroller to read sensor data and drive the RF link, and a power supply. Footprint and weight depend largely on the requirements of individual sensors, as the space and power needed for the radio and microcontroller are modest (comparable to the backpack). To date we have explored ultrasonic range finders (SensComp).

An alternative is to use a PC to direct the neural interface system, either by relaying sensory information or by generating signals directly. A Labview program serves as a user interface, permitting a virtually unlimited array of options for creating signal inputs for various experiments. The user can also monitor recorded activity through this interface.

E. Second generation device

As testing on the system described above is ongoing, we have also designed a second generation ASIC that adds capabilities while shrinking the implant size by an estimated 30%. The next-generation design integrates the battery charger, eliminating one chip inductor and the majority of the off-chip components. It will also permit parameterized stimulation waveforms to be pre-programmed and stored on chip, so that they can be triggered by a single command. Waveforms will parameterize the general format depicted in Fig. 4, with each time parameter stored in 3-bit values and each amplitude parameter stored in 5-bit values. This will permit simultaneous stimulation of up to all eight channels, as well as simultaneous stimulation and recording. System functionality will otherwise be similar.

IV. Ongoing Work and Conclusions

Currently, the ASIC and test versions of the other system components have been fabricated and are being tested. Upon the completion of component testing, the system will be integrated and tested, *in vitro* and then *in vivo*. Although discussion of this is beyond the scope of this paper, we are also directing animal testing of the core hypotheses underlying the idea of sensory augmentation; testing is being conducted by our partners at the University of Michigan and the University of Pittsburgh. When the wireless system is complete, it will be used in these experiments. Initially, experiments will seek to determine whether a freely-moving animal can gain the ability to interpret new sensory information, delivered through a peripheral nerve interface, without substantially losing sensorimotor function. If successful, subsequent studies will explore the details of sensory augmentation in search of an optimal implementation.

Once it has been thoroughly validated, we believe that the system discussed and similar systems will facilitate a new era of studies in more freely-ranging animals, in more natural habitats. We hope to widely distribute such technology to researchers in the neuroscience community, allowing them to repeat key experiments in new environments, and to create new experiments that to date have been impossible. This should provide a critical step toward deploying neural interface technology in human subjects, for restoration of lost function as well as possibly for augmentation.

Also beyond the scope of this paper is a full discussion of the ethical issues associated with the potential use of neural interface technology to enhance healthy subjects. We encourage such discussion among neuroscientists, philosophers, medical personnel, political leaders and citizens, and we believe that exploration and discovery of technical possibilities is an important element of this ongoing conversation.

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