Proceedings of the 25th Annual International Conference of the IEEE EMBS
Cancun, Mexico • September 17-21, 2003

The Benefits of Modular Brain-Machine Interface System Design

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Abstract-**Construction of a brain-machine interface system for neuroprosthetic purposes is at the forefront of many current neural engineering thrusts. Due to recent breakthroughs in device technology and implantation techniques, a basic framework is now sufficiently developed to allow design of systems level interface strategies producing robust, scalable BMIs that adapt quickly to optimize information transfer at the interface. It is useful to develop brain-machine interface systems in a modular fashion, enabling individual component research and development. This study investigates cortical microstimulation as a mode of operation for a sensory encoding component of a brain-machine interface system. It has previously been shown that cortical stimulation of sensory cortical areas produces sensations. In this report we compare behavior induced by either natural auditory cues, or cortical microstimulation of the primary auditory cortex. Five rats were implanted with multi-channel microwire arrays in auditory cortex and required to discriminate cortical microstimulation separated by 1.75 mm. The behavior was compared to auditory discrimination of tones separated by four octaves. The microstimulation resulted in 17% faster response times across the five rats.**

Keywords **- neuroprosthetics; microelectrode arrays; brain-machine interface; intracortical microstimulation; auditory cortex; operant conditioning**

I. INTRODUCTION

Brain-machine interface systems have the potential to revolutionize medicine and the quality of life for many users. Brain-machine interface systems have been suggested to have application potential in the treatment of: Parkinson's Disease, Amyotrophic Lateral Sclerosis (ALS), spinal cord injury and many more. Thus, optimal brain-machine interface system design strategies are the focus of much ongoing research in the field of neural engineering.

Modularity, or designing a system composed of individual components, has been shown to be a historically beneficial engineering system design strategy. One of the benefits of modular system design is the availability of independent component optimization. This allows different design teams to improve the performance of a single component regardless of design successes and failures of other groups working on the same system. A second major benefit of modular system design is scalability. Individual components can be duplicated and augmented in order to divide the workload of that component of the system.

Brain-machine interface systems reported in the literature thus far have largely been described in terms of the entire system [1-4]; however, upon inspection, brain-machine interface system design in a modular fashion exhibits a strong potential [5]. The first step in modular system design is casting the system process in terms of a system flow diagram. Fig. 1 shows such a brain-machine interface system flow diagram. In this diagram, each processing stage indicates a discrete component that can be optimized in an individual fashion. The chart is organized such that the brain/organism processing steps are represented by the three components on the left side of the figure, enclosed by a dotted line. The effector, whether that be a robot or a virtual effect such as surfing the internet, is shown as the far right component, "action", indicating some form of work. The components in between the brain and the effector comprise the interface components that handle the communication and information processing necessary for effective system performance. This modular system layout provides opportunity for individual component research and development to improve overall system function.

In this study we investigated sensory communication strategies for sensory cue and feedback transmission in brain-machine interface systems. In Fig. 1 this component is labeled as the "sensory encoder". In this role, the sensory encoder directly passes information about the system and the external environment to the brain.

There are several different modes under which the sensory encoder component can operate to transfer sensory information. An obvious solution is to use the natural sensory transducers of the brain to encode the information. A potentially more favorable mode of operation for sensory transmission incorporates electrical microstimulation of the brain. Electrical microstimulation possesses several valuable traits that suggest it is worthy of investigation. Some natural sensory transducers, such as the chemical receptors of the olfactory system, are difficult to interface with on a long-term, automated basis. The chemical reservoirs deplete over time, and the release of the odorants into the nasal canal is difficult to control with respect to the spread and ability to purge the chemicals. Conversely, electrical microstimulation through a chronically implanted electrode in a sensory location of the brain results in very controlled sensations with respect to specificity and the temporal aspect of the sensation. There are other difficulties associated with interfacing with the other sensory systems within the body. This is another advantage of electrical microstimulation, since it utilizes the same physical encoding parameter to interface with every system, electrical excitation of neurons. This multi-modal aspect of electrical microstimulation, and many others, validates further investigation of its potential role as a mode of operation for the sensory encoder shown in Fig. 1.

This work was supported by grants from the NIH P41-EB00230 and DARPA (MDA972-00-1-0027 and N66001-02-C-8059).

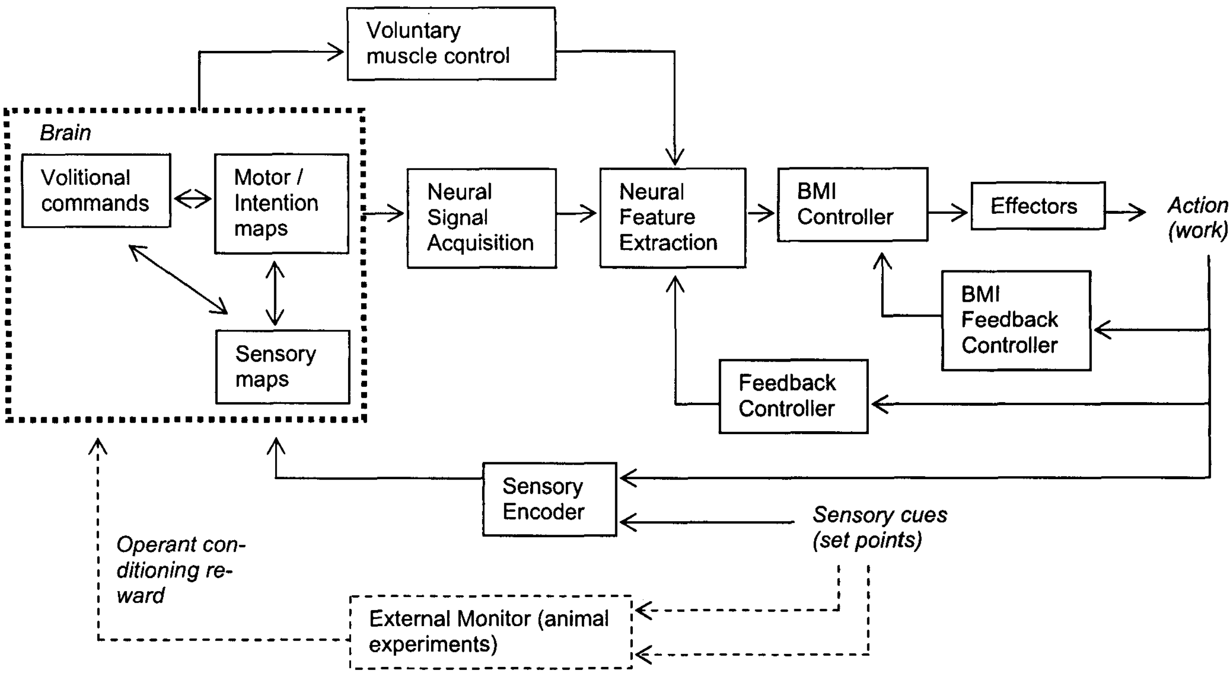

Fig. 1. Brain-machine interface system flow diagram. Each box represents an individual component that can optimized independently and off-line so that the overall system performance is improved. The modular nature of this brain-machine interface is apparent in this system-based design approach.

The objective of this study was to evaluate sensory information transmission through an electrode array chronically implanted in the primary auditory cortex of a behaving rat. In order to compare and contrast the behavior induced by cortical microstimulation, the animals were first trained on an auditory pitch discrimination task. The results were then compared against the results of discrimination of cortical microstimulation location within the auditory cortex.

II. METHODOLOGY

A. Behavioral Task – Auditory Discrimination

Initially, five male Sprague-Dawley rats (250g - 300g) were food deprived to 80% of their free-feeding weight. They were subsequently trained in a two-choice, go/no-go discrimination task. Subjects responded in standard operant conditioning behavioral boxes (Med Associates, St. Albans, VT) located within an anechoic chamber. Subjects were positively reinforced for correct responses via single food pellets (P.J. Noyes, 45 mg rodent diet I, Lancaster, NH) delivered in a 5 cm by 5 cm tray located at the base of one wall of the cage. A 28 V house light at the rear of the box was used for cage illumination. The behavioral apparatus was controlled and monitored by in-house software running on a PC interfaced with digital input-output hardware (System III, Tucker-Davis Technologies, Gainesville, FL). This equipment was also used to generate all auditory stimuli used in the experiment. The auditory stimuli were delivered via a speaker (Yamaha NS-10M Studio, Yamaha Corporation, Buena Park, CA) located 1 m directly above the test box. The system delivered a near-flat frequency response between 500 Hz and 32 kHz. The system was calibrated to a position at the food delivery tray; although calibration measurements indicated that the test box approximated a free field.

Each experimental session continued until the animal received 200 food pellets. This resulted in hundreds of individual discrimination trials for each session. The subjects were signaled to start a single trial by the extension of the center lever. Trials were subject-initiated by two recorded presses of this center response lever. Subsequently, the center lever was retracted, and a train (250 ms on, 250 ms off) of five pure-tone bursts was delivered. Auditory training stimuli trains at either 1 kHz, or 16 kHz were delivered at 70 dB SPL. Upon completion of the auditory stimulus presentation, the two outer levers were extended. A fixed-ratio (FR4) response paradigm was utilized, and subjects were reinforced after four responses on a given lever within 7 s of outer lever presentation. Responses were designated correct and positively reinforced for a left lever response to the 1 kHz stimulus or a right lever response to the 16 kHz stimulus. Left responses to the 16 kHz stimulus or right responses to the 1 kHz stimulus were designated incorrect and negative reinforcement was given in the form of a 30 s dark timeout. A response was considered null, and negatively reinforced, if the subject did not respond within the 7 s response window. Null response trials in all of the training or testing were rare (zero for > 95 % of the sessions) and were not used in the behavioral data analysis.

One measure of performance is assessed by quantifying the response times, and it has been shown that a decreasing response time indicates better performance [6]. In this experiment the response latency is the time between the end of the stimulus presentation and the response. For each session, the average response time was calculated and the response times for the auditory and microstimulation sessions were statistically evaluated.

B. Device and Surgical Implantation

Upon successful auditory discrimination training, each rat was chronically implanted with arrays of 16-channel microwires. Microelectrode array and surgical details are described elsewhere [7]. Briefly, electrode geometry consisted of sixteen microwires, 50 μm in diameter separated by 250 μm. The microwires are arranged into two rows of eight electrodes each. The rows are also separated by 250 μm. The electrodes were implanted into the primary auditory cortex of the left hemisphere, determined by the stereotaxic coordinates of AP: -4.0 mm, ML: 7.0 mm, as described by Sally and Kelly [8]. Upon implantation the crainiotomy was closed with dental acrylic (Co-Oral-Ite Dental Mfg. Co.), and the animal was allowed ~10 days to recover from surgery. All procedures complied with the United States Department of Agriculture guidelines for the care and use of laboratory animals and were approved by the Arizona State University Animal Care and Use Committee.

C. Behavioral Task – Microstimulation Discrimination

After surgical recovery, the rats were then reinforced for successful discrimination of microstimulation delivered through electrodes spaced 1.75 mm apart. Microstimulation pulse trains consisted of cathodic first, charge-balanced, biphasic square-wave pulses (250 μs pulse width) delivered at 200 Hz and 68 μA. This stimulus intensity was chosen based on a calculated estimation of current spread based on parameters, reported in the literature, that led to a minimal effective stimulation radii (100 μm) between neighboring electrodes at 68 μA [9]. A waveform generator (WaveTek, Everett, WA) was used to generate the pulse train, which was delivered through an optical stimulus isolator (A-M Systems, Carlsborg, WA) in constant-current stimulation mode. The cortical microstimulation stimulus intensity was confirmed using a 1 kΩ resistor circuit prior to testing. The cranial stainless-steel screws served as the stimulation return pathway. The temporal parameters of the microstimulation were chosen to mimic the temporal envelope of the auditory stimuli. Microstimulation pulse trains were delivered in five bursts (250 ms on, 250 ms off). The behavioral apparatus software recorded responses to both the task stimuli and the probe stimuli.

III. Results

A. Behavioral Data

Auditory and microstimulation discrimination response time results from the five rats are shown in Fig. 2. The response time values for auditory discrimination range from 1.31 s to 1.95 s and reflect the peak performance after 4-5 months of daily training.

Post surgical recovery, each subject performed between 5 and 7 microstimulation discrimination sessions. The response time values for microstimulation range from 0.98 s to 1.86 s and include all microstimulation sessions, including the first day of microstimulation discrimination.

In four of the five rats the response latency of the auditory trials was significantly longer than the response latency of the microstimulation trials (t-test, $p<0.05$). The average auditory response latency was 1.7 s. There was a 16.7% drop in average response latency in the microstimulation trials to 1.4 s.

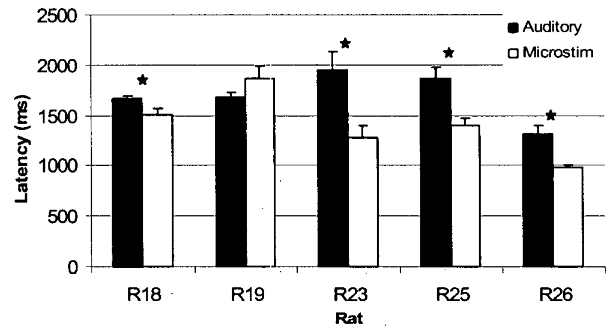

Fig. 2. Cumulative response latency results of tone and microstimulation discrimination from five rats. Response latency is defined as the time required for the subject to respond after the stimulus presentation. In four of the five rats the auditory response latency is significantly greater than the microstimulation response latency (t-test, $p<0.05$).

IV. DISCUSSION

A modular brain-machine interface system design strategy allows for independent component research and development. In this study we investigated the mode of operation of the sensory encoder component of a brain-machine interface system. Recent results have shown that closed-loop feedback is important to increase performance in brain-machine interface systems [2-4]. From these results, it follows that enhanced sensory communication from machine to brain will result in even better performance of the system as a whole. The results of this study indicate not only that cortical microstimulation is an effective communication method, but that, in the relative paradigms presented here, microstimulation is more effective than natural auditory stimulation as a method to provide a behaving animal with sensory cues.

B.F. Skinner's law of latency indicates that the response latency is inversely proportional to the strength of the stimulus [6]. The results of this experiment indicate, for the relative stimulation parameters of the auditory and microstimulation stimuli used in this study, that cortical microstimulation is a stronger stimuli.

The results provided here also have implications for sensory neuroprosthetic applications. Several recent studies have indicated that cortical microstimulation may be useful for sensory replacement in blind and deaf subjects [10, 11]. This study supports the findings of these studies and adds relevance to the idea of cortical microstimulation for the deaf and blind. In fact, the increased strength of the microstimulation stimulus has important implications for the information content of single channel cortical microstimulation, as well as the number of channels that will be required for adequate sensation.

ACKNOWLEDGMENT

The authors would like to thank Chris Visser for electrode construction and Roger Hurtado for animal training assistance. We would also like to thank Drs. Mark Reilly and Peter Killeen for operant conditioning assistance. This work was supported by grants from the NIH P41-EB00230 and DARPA (MDA972-00-1-0027 and N66001-02-C-8059).

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