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Applications of Cortical Signals to Neuroprosthetic Control: A Critical Review

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***Abstract*—Cortical signals might provide a potential means of interfacing with a neuroprosthesis. Guidelines regarding the necessary control features in terms of both performance characteristics and user requirements are presented, and their implications for the design of a first generation cortical control interface for a neuroprosthesis are discussed.**

***Index Terms*—Cortical interface, electroencephalogram (EEG), neuroprosthesis.**

I. INTRODUCTION

Neuroprosthetic systems provide function by electrical stimulation of paralyzed muscles in a coordinated fashion. The individual using the system can control the stimulation, usually through movement of some nonparalyzed part of the body. For example, in the Case Western Reserve University (CWRU)/VA hand-grasp neuroprosthesis [1], [2], the user controls opening and closing of his or her hand by movement of the contralateral shoulder. This movement is sensed by an external position transducer that is taped to the user's shoulder and chest. The command signal is sent to an external control unit, which converts the signal into the appropriate stimulus level for each muscle. This signal is sent through a radio frequency (RF) link to the implanted stimulator unit, which in turn generates a stimulus pulse train of the appropriate magnitude to each electrode placed on the different muscles of the forearm and hand. By coordinating the activation of each muscle, a functional grasp pattern is achieved.

The user controls the degree of opening and closing of the hand by movement of the contralateral shoulder—moving the shoulder forward (protraction) results in hand closing, moving the shoulder back (retraction) results in hand opening. The control is proportional, allowing the user to modulate the grasp force for the desired task by adjusting shoulder position. The neuroprosthesis also uses a state control input (typically a chest mounted switch) that enables the user to select different pre-programmed grasp patterns, to turn the device on and off, and to lock and unlock the hand.

Ongoing research and clinical experience has defined limitations that are inherent in the shoulder generated command signal and its hardware implementation. First, shoulder control is restricted to the contralateral arm, thus restricting bilateral implementation of the neuroprosthesis. Second, the external mounting is cumbersome, necessitates external wiring, and performance varies somewhat with mounting differences. Current research is designed to overcome these deficiencies. In particular, an implantable transducer that senses the position of the ipsilateral wrist has been designed and clinically implemented [3]. Also, myoelectric control has been assessed as an alternative command signal, using the EMG signal from retained muscles [4], [5] as the control source. Nevertheless, all of these signal sources are somewhat unnatural and require the user to learn to relate an artificial command with the intended movement. It is in this dimension of natural control that a cortical interface provides the greatest potential.

II. CHARACTERISTICS OF THE COMMAND CONTROL INPUT

The characteristics of the command signal for a hand neuroprosthesis should enable the user to utilize a natural method to select a grasp pattern, regulate hand opening/closing and the grasp strength, and maintain grasp. The principal feature of the command signal is proportional, single degree of freedom information under user volitional control delivered at a sufficient accuracy and speed to provide appropriate control of the hand. Acceptable factors to achieve this type of control can be separated into performance and user criteria. A partial review of these criteria has been compiled elsewhere [6], [7]. These criteria are shown in the following sections.

A. Performance Criteria

1) One Degree of Freedom: A single degree of freedom input signal will enable control of both hand opening/closing and grasp strength. Control of some other upper extremity movement (e.g., elbow extension, forearm pronation) can be linked to synergistic movements [8], [9], thus reducing the demands of the controller. Other movements (e.g., shoulder) will require a separate control input. For the operation of the contralateral hand, a second control input will also be required since sequential control of the hand is not clinically acceptable.

2) Stability over Time: Stability is required to enable day-to-day consistency of the command input. This is addressed further in the discussion section as applicable to cortical signals.

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3) Minimal Delay: Delays between intention and action degrade performance. A nominal, acceptable value to this has been placed at 200 ms [10].

4) Number of Discrete Levels of Activation: The command input should at least match the performance capabilities of the hand neuroprosthesis. Although approximately 100 to 200 stimulus levels are available, only five to eight levels are used in practice by the user [6]. This number of levels can be achieved by either using the unaltered signal or through additional processing. For example, a signal with only two discrete levels can be used to gate a command up and down at a predetermined rate, and thus achieve graded control [20]. However, the deficit in this method is the introduction of a command delay. This trade-off between the number of available command levels and command delay is an essential performance criterion.

5) Selectivity: The command signal should not be affected by other intentional voluntary movements, movements generated by electrical stimulation, or the stimulus artifact (see Discussion).

B. User Criteria

User criteria are more difficult to describe quantitatively, but are critical for user acceptance of any control method. The control must be "natural" for the user, i.e., it must not require extensive special attention. The control must also be convenient to access, easy to learn and utilize, and cosmetically acceptable. Clinically, it is also desirable that the interface is easy to implement and safe from electrical hazard and biological compatibility perspectives.

It is clear that the performance and user criteria are in conflict with one another. For example, a surface cortical potential system utilizing external electrodes has performance restrictions and is not likely to be cosmetic, but it will be safe. Alternatively, an intracortical recording system will likely have considerably better performance and be cosmetic, but will be more difficult to deploy clinically and have considerably greater concerns of safety based purely on being implanted. Thus, judgement will be required in determining the balance between these factors for the specific intended clinical application.

III. Cortical Interface Development for Neuroprosthetic Control

The development of a cortical interface to the hand neuroprosthesis can take one of two forms, either through the use of an intracortical recording array [11]–[14] or though the use of surface cortical potentials [15]–[18]. Research undertaken at CWRU and the Cleveland VA Medical Center (MC) has focused on assessing the feasibility of using surface cortical signals to operate a hand grasp system.

The interface is shown in Fig. 1. The controller is a combination of the BCI system and the neuroprosthesis developed in the laboratory at CWRU and the Cleveland VAMC. The EEG signal, recorded from electrodes on the scalp, is amplified and filtered using a Laplacian spatial filter. The signal is then converted into a voltage by performing a spectral analysis. Further noise reduction is achieved using an adaptive step-size [4] filter, and the signal is converted into a command to control grasp using a gated control algorithm. This algorithm only allows for dynamic hand function, converting high voltages into a state command to ramp open the hand and low voltages into a state command to ramp closed the hand. Features such as holding the object for long periods of time are not possible with this algorithm. However, even this minimal amount of function, should enable a subject to grasp and manipulate objects.

Our initial work has identified a number of important aspects of the cortical signal that must be considered in order for it to be used for neuroprosthetic control. First, it is uncertain that the surface EEG signal will have the information content of some of the already established

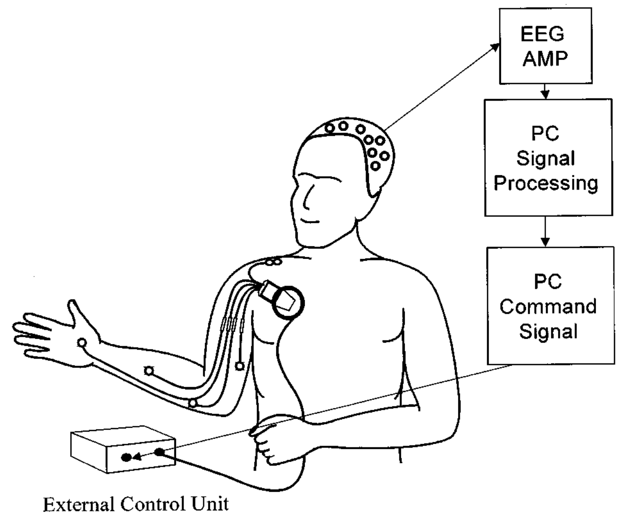

Fig. 1. Schematic of the EEG-based controller for the Hand Grasp Neuroprosthesis.

methods of control, such as shoulder position. Most studies to date have focused on a binary "high–low" signal using the EEG, although McFarland *et al.* [20] has shown that there may be potential to achieve as many as four discrete states with further subject training and improved signal processing. However, it may be that the EEG signal is better suited to provide a state signal for functions, such as switching between grasp patterns or turning stimulation on and off. It may also be possible to generate a "proportion in time" signal from this information similar to that used for myoelectric control, in which the signal gates the command to increase or decrease at a known fixed rate [21]. Thus, if the subject is able to hold and maintain a high or low level for a discrete period of time (with a resolution of 0.5 s), then it is possible to convert the state information into proportional information, allowing for greater control.

The information outflow rate of the EEG system might be rather low because of the signal processing involved and the nature of the EEG signal. Appropriate spectral analyses usually require a 200-ms sample size. To accomplish this and still allow for a 10-Hz update rate for the interface, a 100-ms data backlogging function is often used which introduces a delay. Another delay occurs in the generation of an appropriate change in the EEG signal. Studies of changes in the frontal rhythms due to cognitive processing suggest that this delay will be about 300 ms [22]–[24] which may also apply to the μ and central β rhythms. These delays would have the effect of introducing a noticeable delay to the user between the initiation of the command and the response of the neuroprosthesis. It was stated in the Introduction that during the accomplishment of any task, a delay of greater than 0.2 s will degrade performance. This will be the limiting factor to the use of EEG for direct proportional control of grasp, but may not be especially limiting for a state signal.

The intracortical signal, in comparison, appears to have a large information content and a quick response time. The signal may also be easier to use than the EEG signal if, as expected, less training and attention is required. However, there are two major issues regarding the use of intracortical recordings. The first of these is the stability of the signal over the course of days and years. In this case, the requirements for neuroprosthetic control may be more lenient than the requirements currently placed on this type of recording for neurophysiological studies. Stability has typically been defined as recording from the same neuron or population of neurons over time, and much effort has gone into demonstrating this type of stability. However, for neuroprosthetic control, the requirement is that the user is able to consistently generate the control signal reliably, without the need for frequent re-tuning. This may not require constant recording from the same group of neurons, but rather requires that the user can always generate a reliable signal from the neurons within the recording range of the electrode array. It may also be that slow changes in the relative sensitivity of an electrode between neurons can be naturally accommodated for by the user, without the need for specific retraining. A simple recalibration that could occur intermittently or during in-patient visits at perhaps six-month intervals would seem to be acceptable in this regard.

A second issue regarding the use of intracortical recordings is the unresolved issue of cortical plasticity following spinal cord injury. Currently, research is focused on recording neural activity from the hand and arm area of the cortex, and then using this signal to control a robotic arm. This is the first step toward controlling a neuroprosthetic system. However, to restore hand function there is a reliance upon recording hand movement from an intact cortical representation in the somatomotor area. The question arises, however, as to whether or not these neuron populations will be present and functionally active in the motor cortex of an individual with a midcervical level spinal cord injury. It has been hypothesized that the somatomotor cortex undergoes reorganization after a spinal cord injury [25], but the degree is unknown. This is an area that requires further investigation if cortical control of a neuroprosthesis is to be achieved. If the arm and hand areas are no longer present, or if the neuron population representing these areas are sufficiently reduced so that recording from these neurons is no longer possible, then research must also be directed into finding alternative sites in the motor control system which may be used to acquire this information.

Another issue that must be dealt with if cortical signals are to be used to control a neuroprosthesis is the stimulus artifact from stimulation electrodes in the prosthesis. We have verified that even stimulation applied as far away as the forearm produces an artifact on the scalp. Typical stimulation rates for the hand grasp neuroprosthesis range from 12 to 16 Hz. This has been determined to be the optimal frequency for electrical stimulation [26]. Frequencies below this rate result in an unfused muscle contraction and an unusable hand grasp, while frequencies above this result in greater muscle fatigue and a decrease in the amount of time the system can be used. In EMG studies, the stimulus artifact was removed by blanking the recorded signal during the time period that the stimulus pulses are generated (typically a period of 20–30 ms out of every 60–80 ms) [4]. It is not certain whether users can achieve adequate control of the μ or β rhythms, which have a frequency of 8–10 Hz. A higher β rhythm might be a better control signal. This impact of the stimulation artifact is one that will also have to be addressed with the use of intracortical recordings, although the skull may provide substantial artifact suppression.

Another aspect of the EEG signal that is important to consider is the identification of the optimum cortical area for signal acquisition. Most of the reported studies that use EEG signals for the control of external devices record the signal generated over the somatomotor cortex. In these studies, the application is intended for severely disabled individuals with no muscle movement at all [18], or able-bodied users who are resting their extremities [15]. However, in individuals who have sustained midcervical level spinal cord injury, for which cortical control would be an important command signal source for neuroprosthetic use, voluntary movement is retained in the shoulder and upper arm. The stimulated hand muscles of the neuroprosthesis then augment these voluntary movements. Any control source must not interfere with the user's ability to make maximum use of their voluntary musculature. However, arm movement can have a pronounced effect upon the β and μ rhythms recorded from the somatomotor cortex, may interfere with control signals developed in these areas. It is not clear whether users can achieve adequate control of the μ or β rhythm while moving their arms in a simulated task. Therefpre, we have explored recording over the frontal cortex where the effects of arm movement might be less. On the other hand, signals recorded over the frontal cortex can be contaminated by electromyographic (EMG) or electro-oculographic (EOG) activity. Indeed, early data [19] that initially reported to show user control of a frontal β rhythm were found to contain EMG contamination.

The use of the EEG signal for the operation of a practical neuroprosthesis for daily use will also require further advancements in recording electrode technology. Presently, recording an EEG signal requires the placement of individual electrodes or a cap upon the scalp, with the locations based roughly upon physiological landmarks. Because these electrodes must be donned and doffed, there are variances in the recorded signals during each session that require a recalibration during each use. These problems can be overcome with the use of a subdural recording array [27] or an implanted array [28], [29] that would provide stable and repeatable signals. It is our expectation that an implanted recording electrode will be needed for users to find this type of system cosmetically acceptable for daily use in social situations. Nevertheless, it will be critical for investigations to demonstrate that this interface is safe as well as effective before introduction in a neuroprosthesis can be anticipated.

IV. Conclusion

The use of the cortical signal for the operation of a hand grasp neuroprosthesis is particularly attractive because it provides a means of restoring the link between thought and hand movement which was lost at the time of injury. Guidelines for both performance characteristics and user requirements have been established. However, there is still much which needs to be explored, not only with the technology, but also with the underlying neurophysiology of spinal cord injury and with the conversion of the signal into neuroprosthetic control.

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A Natural Basis for Efficient Brain-Actuated Control

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***Abstract*—The prospect of noninvasive brain-actuated control of computerized screen displays or locomotive devices is of interest to many and of crucial importance to a few 'locked-in' subjects who experience near total motor paralysis while retaining sensory and mental faculties. Currently several groups are attempting to achieve brain-actuated control of screen displays using operant conditioning of particular features of the spontaneous scalp electroencephalogram (EEG) including central μ-rhythms (9–12 Hz). A new EEG decomposition technique, independent component analysis (ICA), appears to be a foundation for new research in the design of systems for detection and operant control of endogenous EEG rhythms to achieve flexible EEG-based communication. ICA separates multichannel EEG data into spatially static and temporally independent components including separate components accounting for posterior alpha rhythms and central μ activities. We demonstrate using data from a visual selective attention task that ICA-derived μ-components can show much stronger spectral reactivity to motor events than activity measures for single scalp channels. ICA decompositions of spontaneous EEG would thus appear to form a natural basis for operant conditioning to achieve efficient and multidimensional brain-actuated control in motor-limited and locked-in subjects.**

I. INTRODUCTION

Recent work in several laboratories has demonstrated that noninvasively recorded electric brain activity can be used to voluntarily control switches and communication channels, allowing a few so-called locked-in near-totally paralyzed subjects the ability to communicate, however slowly, with their families and aides ([4]; [14]; [2]). Communication rates achieved to date are in the range of several bits a minute, far from rates that would allow locked-in persons access to normal social interaction. This communication briefly describes a technique for blind decomposition of electroencephalogram (EEG) data into temporally and often functionally independent components that would appear to provide a natural basis for optimizing brain-actuated control ([7]; [9]). An example is given of a decomposition of spontaneous EEG in one subject into four components accounting for spatially distinguishable though widely overlapping posterior alpha and central μ-rhythmic activities. Learned control of the amplitude of motor-related central μ-rhythms in the alpha frequency range (8–12 Hz) ([5]) is being used for brain-actuated control by at least two groups ([13]; [15]). We demonstrate that the motor-response related spectral perturbations demonstrated by the independent component analysis (ICA)-defined

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