Behavior Research Methods, Instruments, & Computers
1997, 29 (3), 337–345

Design and operation of an EEG-based brain–computer interface with digital signal processing technology

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We are developing an electroencephalographic (EEG)-based brain–computer interface (BCI) system that could provide an alternative communication channel for those who are totally paralyzed or have other severe motor impairments. The essential features of this system are as follows: (1) EEG analysis in real time, (2) real-time conversion of that analysis into device control, and (3) appropriate adaptation to the EEG of each user. Digital signal processing technology provides the speed and flexibility needed to satisfy these requirements. It also supports evaluation of alternative analysis and control algorithms, and thereby facilitates further BCI development.

People with severe movement disorders need alternative means of communication and control. Those who are totally paralyzed cannot use conventional assistive devices because all these devices require some degree of voluntary muscle function. In recent years a variety of studies have addressed the possibility that scalp-recorded electroencephalographic (EEG) activity might be the basis for a brain–computer interface (BCI) that could be a new alternative communication channel for those without any useful voluntary movement (Farwell & Donchin, 1988; McFarland, Neat, Read, & Wolpaw, 1993; Pfurtscheller, Flotzinger, & Kalcher, 1993; Sutter, 1992; Wolpaw & McFarland, 1994; Wolpaw, McFarland, & Cacace, 1986; Wolpaw, McFarland, Neat, & Forneris, 1991).

A BCI system measures particular components or features of EEG activity and uses the results as a control signal. Some systems use evoked potentials, which are EEG components produced by stereotyped sensory stimuli. For example, Farwell and Donchin (1988) and Sutter (1992) used visual-evoked potentials. Other systems, including our own, use EEG components that are spontaneous in the sense that they are not strongly linked to specific sensory inputs. Our system uses the mu rhythm, an 8–12 Hz rhythm recorded from the scalp over somatosensory cortex and/or closely related higher frequency components (McFarland et al., 1993; Wolpaw & McFarland, 1994, 1995; Wolpaw et al., 1986; Wolpaw et al., 1991). Pfurtscheller and his colleagues (1993) have used EEG features defined by neural network analyses.

Requirements for a BCI System

Each BCI system must record EEG activity and convert it into a control signal in real time. To be effective, it must also adjust to the characteristics of each user's EEG and must adapt to short-term and long-term changes in those characteristics. Analog signal processing methods, which have been widely used in EEG studies (e.g., Black, 1971; Tozzo, Elfner, & May, 1988) are generally too rigid and imprecise to satisfy these requirements. Digital signal processing (DSP) technology, incorporating readily available hardware and easily modified software, has the necessary flexibility and precision.

BCI System Overview

Using DSP technology, we have designed a system for developing and testing methodology for EEG-based communication. This laboratory BCI system digitizes 64 EEG channels from the system user (i.e., the subject), performs real-time spatial filtering and spectral analyses, uses the results to control a video display, continually adapts its analysis algorithm so as to convert the user's EEG control as efficiently as possible into display control, provides performance data on-line to the system operator (i.e., the investigator), and stores all data for later off-line analyses. The comprehensive data collection allows off-line evaluation of alternative control algorithms, and the flexibility of the on-line software permits promising alternatives to be tested on-line.

Figure 1 summarizes the on-line operation of this system. As shown in Figure 1A, the user sits facing the screen of a video monitor. EEG activity recorded by scalp electrodes is amplified and digitized. The voltage in specific frequency bands at specific scalp locations is determined and translated into cursor movement on the screen. Over

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We thank Lynn McCane and Theresa M. Vaughan for excellent technical assistance and Gregory W. Neat for his important contributions to software design. This work was supported in part by the National Center for Medical Rehabilitation Research of the National Institute of Child Health and Human Development (Grant HD30146). Correspondence should be addressed to D. J. McFarland, Wadsworth Center for Laboratories and Research, New York State Department of Health, P.O. Box 509, Empire State Plaza, Albany, NY 12201-0509 (e-mail: mcfarlan @wadsworth.org).

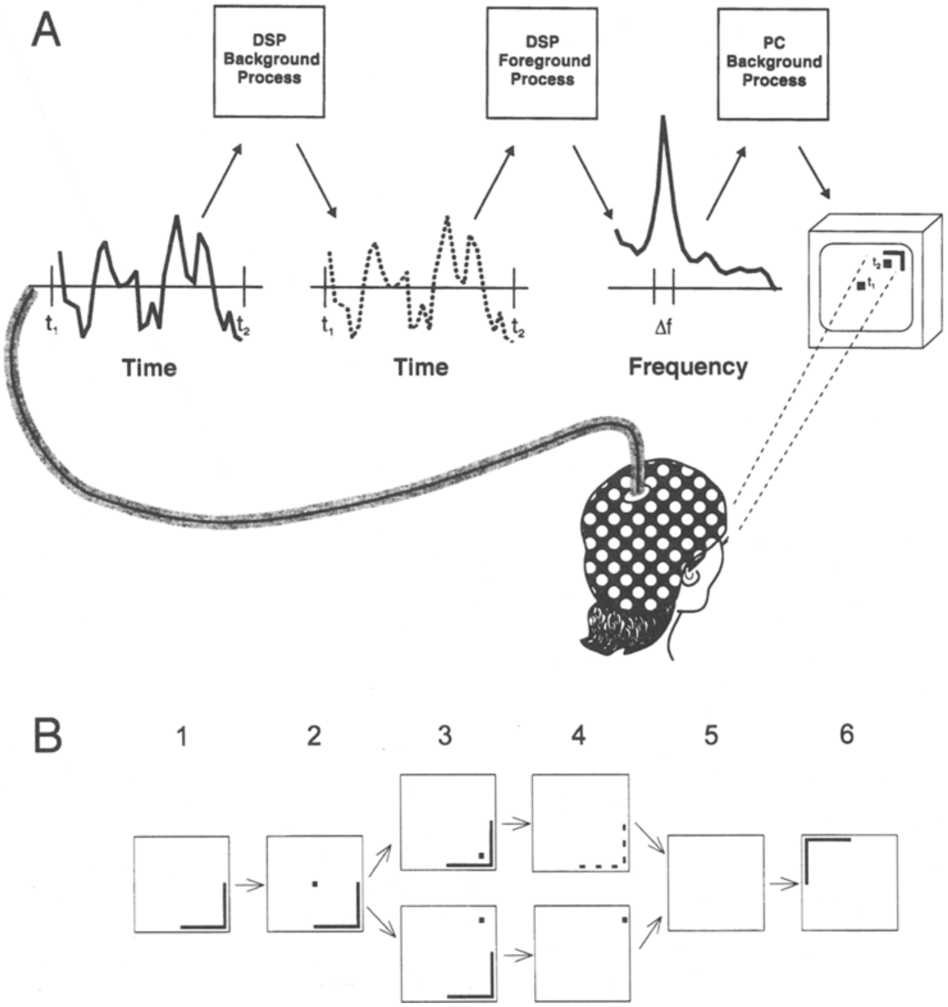

Figure 1. (A) Outline of brain–computer interface (BCI) on-line operation. For simplicity, only one channel of electroencephalographic (EEG) and one-dimensional (i.e., vertical) cursor control are shown. Voltage at the scalp is amplified and digitized and an EEG cursor-control channel is derived and submitted to frequency analysis 5–10 times/sec. The amplitude in a specific frequency band is converted into vertical cursor movement. The upper boxes show the flow of the foreground and background processes on the digital signal processing (DSP) board and the PC. The background DSP process records incoming data and derives the EEG cursor-control channel. At fixed intervals, the foreground DSP process performs a spectral analysis and generates an interrupt request to the PC. The background PC process acquires spectral data from the dual-ported DSP memory and updates the user's screen. The foreground PC process stores data to disk, updates the operator's screen, and controls the sequence of events during each trial and the duration of each 2–3 min run. (B) Sequence of events on the user's video screen during a two-dimensional trial (see text for details).

a series of training sessions, the user learns by trial and error to modulate these EEG components so as to move the cursor toward a target located on the edge of the screen. This laboratory BCI system uses cursor movement as output because it is objective, readily implemented, and serves as a prototype for a wide variety of devices that could be placed under EEG control. For example, achievement of good cursor control could allow the BCI user to access many commercially available mouse-driven programs. The system operator has a second video screen and a keyboard input. She/he sets the control parameters for system operation, initiates that operation, and is provided with continually updated information as to the user's performance.

In the case of one-dimensional (i.e., vertical) movement, the target can be at the top or bottom edge. For two-dimensional movement, the target can be at one of four or more positions anywhere on the periphery of the screen (e.g., one of the four corners or sides). Figure 1B shows the format for a two-dimensional trial. The trial begins when the target appears in one of the four corners of the screen (1). After a 1-sec period that allows the user to perceive the location of the target and initiate the appropriate EEG response, the cursor appears in the center of the screen (2) and moves controlled by the user's EEG (3). Movement continues until the cursor reaches the periphery of the screen (4). If that point is part of the target, the trial is a success and the target flashes for 1 sec. If it is not part of the target, the trial is a miss and the cursor alone remains on the screen for 1 sec. In either case, the screen becomes blank for 1 sec (5) and then the target appears for the next trial (6).

BCI SYSTEM DESCRIPTION

Hardware

The system that supports cursor control consists of a 64-channel EEG amplifier (SA Instruments, San Diego), two 32-channel analog-to-digital (i.e., A/D) converter boards (Spectrum, Inc., Westborough, MA), a TMS320C30-based DSP board (Spectrum, Inc.), and an IBM-compatible PC/80486 equipped with dual monitors. Although only a subset of these 64 channels actually contributes to cursor control on-line, the flexibility they provide allows the on-line algorithm to be adjusted to the unique topographical features of each subject's EEG, and all 64 are stored for later analysis by a second system that is described below under "Data Storage and Off-line analysis."

The DSP board uses the TMS320C30 floating-point microprocessor. DSP boards have instruction sets optimized for the extensive multiply and accumulate operations typical of signal processing algorithms. The TMS320C30 instruction set includes arithmetic instructions that perform these operations in a minimal number of clock cycles. An earlier version of our BCI system used the TMS320C25 fixed-point processor (Neat, McFarland, Forneris, & Wolpaw, 1990). Floating-point DSP processors are better because they do not require specialized algorithms such as integer fast Fourier transform (FFT) routines. Furthermore, they support alternative spectral analysis techniques, such as the autoregressive method (Marple, 1987).

The DSP board is programmable with the TMS-320C30 C compiler. This feature allows use of standard code for FFT and autoregressive spectral analysis routines (see, e.g., Press, Flannery, Teukolsky, & Vetterling, 1988). In addition, Spectrum provides a C-based library of interface routines for DSP–PC communication. Our system uses interrupt-driven communication so that the PC is automatically provided with EEG spectral data at regular intervals.

The PC has a monochrome monitor for the operator and a VGA color monitor for the user. This is one of the several possible PC–video combinations (Wilton, 1988). In other respects, the PC system is standard.

Software

Both the Texas Instruments TMS320C30 processor of the DSP board and the Intel 80486 processor of the PC are programmed to run in a foreground/background configuration. Thus, these two processors run four processes in parallel. All of the software is written in the C programming language. DSP processors are usually programmable either in C or in their native, highly specialized, assembly languages. Use of the C language with both processors allows routines for signal processing to be tested on the PC prior to use on the DSP board. In addition, software can easily be transferred to alternative DSP processors. For example, we have used both the TMS320C25 and TMS320C30 microprocessors, which have very different assembly languages. Finally, the use of C for both the DSP and PC creates consistency across software modules.

The flow of the four parallel processes is summarized in Figure 1A. The background process on the DSP board is initiated by an interrupt request from an A/D board at the end of an A/D conversion. This process acquires the data from all requested channels sequentially and combines them to derive the one or more EEG channels that control cursor movement (see "Data Collection and Processing" below).

The foreground process on the DSP board performs a spectral analysis on the data (i.e., the fourth operation in Table 1; see "Data Collection and Processing" below). When this analysis is completed, the results are moved to dual-ported memory and an interrupt to the PC is generated.

The background process on the PC acquires spectral data from the DSP board, computes cursor movement (see "PC Calculation of Cursor Movement" below), and controls the display on the user's monitor. The PC background process controls the sequence illustrated in Figure 1B: The target appears, the cursor appears and moves, the cursor hits (or misses) the target, the target flashes for a hit (or disappears for a miss), and, after a brief pause, the next target appears. A single variable has a unique value for each of these screen states. This system-state

Table 1
Sequence of Operations That Converts Scalp Electroencephalographic (EEG) Activity Into Cursor Movement

Device	Input	Operation	Output
EEG amplifier	Scalp voltages (μV)	Amplification	Analog voltages (V)
A/D board	Analog voltages	Digitization	Digitized voltages
DSP board	Digitized voltages	Spatial filtering	EEG cursor-control channels
DSP board	EEG cursor-control channels	Spectral analysis	Control signals
PC	Control signals	Transformation	Cursor movements

Note—DSP, digital signal processing; A/D, analog-to-digital.

variable keeps track of the sequence of events controlled by the PC background process and thus allows this process to be initiated at the same point following each DSP-initiated interrupt request. It is also used to index all data stored for later analysis, thereby ensuring accurate matching between data and system state.

The foreground process on the PC provides control parameters for the display on the user's monitor and for the conversion of the user's EEG into cursor movement, records data to disk, and displays information concerning system operation and user performance on the operator's monitor.

In summary, the system has two processors executing four processes. This parallel processing approach has several distinct advantages over the use of a single process. Several procedures must be performed in real time: moving data from the A/D convertor into memory, performing spatial filtering and spectral analysis, translating the results into cursor movement, controlling the video displays, and storing summary data to disk. Since the processing requirements of these individual procedures are relatively independent except for the final results, the parallel design simplifies the program flow. Furthermore, each processor handles those processing requirements for which it is best suited. The TMS320-C30 performs high-speed interrupt-driven data acquisition and signal processing, whereas the 80486 is responsible for multiple input/output operations (e.g., data acquisition from the DSP board, parameter file input, keyboard input, control of two video monitors, and storage of data on disk).

ON-LINE OPERATION OF BCI SYSTEM

Data Collection and Processing

EEG activity is recorded with standard scalp electrodes mounted in a cloth cap (Electro-Cap International, Inc.). Signals from 64 channels, all referred to a right-ear reference electrode, are amplified (20,000×, bandpass 1–60 Hz) and digitized (the first two operations in Table 1). The DSP board derives one or more EEG cursor-control channels from a linear combination of a selected set of the 64 ear-referenced channels provided by the amplifier (the third operation in Table 1). This operation is a spatial filter. Most commonly, each of the EEG cursor-control channels derived is EEG activity at a location over sensorimotor cortex referenced to a common average reference (CAR) composed of the 19 channels of the 10-20 system (Jasper, 1958), which are widely distributed over the scalp. Possible alternatives to the CAR include LaPlacian and bipolar derivations (Pfurtscheller, 1988).

The DSP board then performs a spectral analysis on each EEG cursor-control channel and makes the results available to the PC (the fourth operation in Table 1). Our initial studies used the FFT for spectral analysis. We are now using the maximum entropy method (MEM) of autoregressive spectral estimation (Press et al., 1988), which is preferable for the on-line analysis required in a BCI system. With FFT analysis, resolution in Hz is equal to the reciprocal of the sample duration in seconds (Walter, 1987). Thus, narrow-band (e.g., 1- or 2-Hz) analysis requires relatively long time segments (i.e., 1.0 or 0.5 sec, respectively). This necessitates a potentially deleterious delay in the feedback provided to the user by cursor movement (e.g., 1 movement/sec with 1-Hz resolution) or the use of overlapping time segments (e.g., 4 movements/sec with each based on the previous sec, for 1-Hz resolution). The MEM algorithm supports higher resolution spectral analysis of shorter time segments. Figure 2 shows a comparison of FFT and MEM spectra computed from 1,000-, 250-, and 125-msec time segments. With shorter segments, the frequency resolution of the MEM algorithm is clearly superior to that of the FFT method (Marple, 1987). Thus, it allows a high rate of cursor movement (e.g., 8/sec) with adequate frequency resolution.

The DSP operations of spatial filtering and spectral analysis constitute the process of signal extraction. In the final operation, the PC transforms these signals into actual cursor movements.

PC Calculation of Cursor Movement

At fixed intervals (e.g., 125 msec) the PC uses the voltages at specific frequencies in the one or more EEG

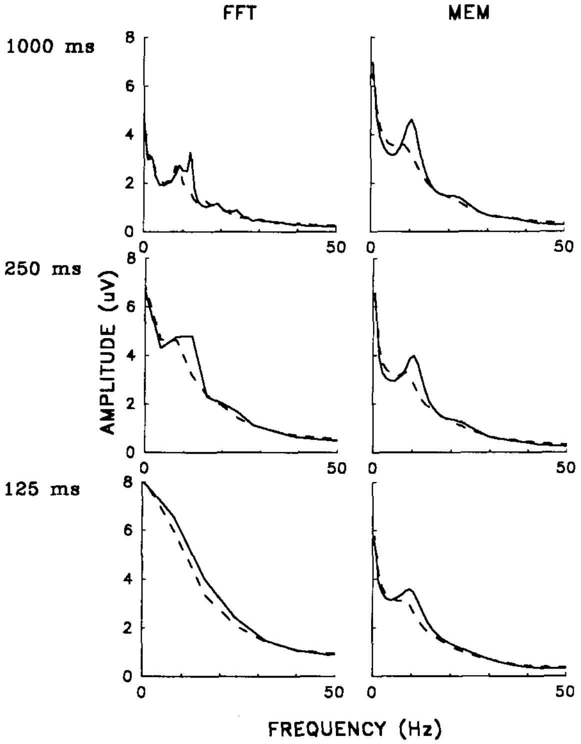

Figure 2. Comparison of electroencephalographic spectra provided by fast Fourier transform (FFT) with those provided by autoregressive (MEM) analysis for 1.0-, 0.250-, and 0.125-sec data segments from a single session in which a well-trained user was moving the cursor to top (solid) or bottom (dashed) targets. Cursor movement was controlled by the voltage at 12 Hz. For shorter time segments, the MEM method provides much better frequency resolution than does the FFT. Thus, it supports a higher rate of cursor movement.

cursor-control channels as the control signals that determine cursor movement (the final operation in Table 1). We have focused on the mu rhythm, which is 8–12 Hz activity recorded over sensorimotor cortex. We have also used related higher frequency (e.g., 20–24 Hz) components recorded from the same locations (Arroyo et al., 1993; Gastaut, 1952; Kozelka & Pedley, 1990; Kuhlman, 1978; Niedermeyer, 1987; Pfurtscheller & Klimesch, 1990). The voltages at these frequencies become the independent variables in the equations that determine cursor movements. To the present, linear equations have been used.

The simplest case is one-dimensional cursor movement to a target located at the top or bottom edge of the screen. If ΔV is vertical cursor movement, S is the voltage at a specific frequency in the EEG cursor-control channel, b is the gain, and a is the mean voltage for the user's previous performance,

$$\Delta V = b(S - a)$$

is the function that determines cursor movement. If the user's mean voltage remains stable, net cursor movement over many trials will be zero and top and bottom targets will be equally accessible. The PC recalculates a at the end of each 2–3 min series of trials or more frequently (i.e., after a fixed number of trials). The value of b determines the rate at which the cursor moves across the screen. Generally, b is increased as a user's control improves. The system can also be set to modify b periodically so as to approach a specific average cursor-movement duration (e.g., 1.5 sec) or rate of movement (e.g., 80 pixels per second).

In the two-dimensional case, one linear function controls horizontal cursor movement and another controls vertical movement. For example, the first successful two-dimensional algorithm used the sum of the mu rhythm voltages over left and right sensorimotor cortices (S_1 and S_2) to control horizontal movement and their difference to control vertical movement (Wolpaw & McFarland, 1994). If ΔV is vertical cursor movement and ΔH is horizontal cursor movement,

$$\Delta V = b_V[(S_1 + S_2) - a_V]$$

and

$$\Delta H = b_H[(S_1 - S_2) - a_H]$$

are the functions that determine cursor movement. Ideally, the values obtained for ΔV and ΔH should be orthogonal, and user success is in part dependent on the degree to which he/she achieves orthogonality.

Proper selection of the intercepts (a_V and a_H) and gains (b_V and b_H) is essential if the user is to move the cursor to the target consistently. For the intercepts, the means of the EEG voltages (i.e., mean of S in the one-dimensional case and means of $[S_1 + S_2]$ and $[S_1 - S_2]$, respectively, in the two-dimensional case) over a substantial number of trials are logical choices, since, as indicated above, they render top and bottom (and right and left) targets equally accessible. At present, the intercept is typically the mean voltage for the most recent 2–3 min run averaged with the previous value of the intercept. This recursive computation dampens variations over time.

Selection of the gains is a more complex problem. An algorithm designed to minimize the least-squared difference between the cursor and the target did not produce stable results in well-trained users. At present, we are using an algorithm that adjusts the gains so that the rate of cursor movement approaches a chosen value. For example, if the desired average vertical cursor movement in pixels/sec is higher than the average vertical movement for the most recent 2–3 min run, vertical gain is increased by a fixed amount (e.g., 10%). As user control improves, the desired movement rate is increased. This algorithm can also be set to maintain a constant relationship between ΔV and ΔH, so that vertical and horizontal movements are commensurate with the dimensions of the screen.

Data Storage and Off-Line Analysis

As noted, the key features of each trial, including the data necessary for the computation of the slopes and intercepts of the control equations, are recorded by the PC background process while the user is moving the cursor. The computation of these constants (e.g., a_V and b_V) is performed by the PC foreground process during the 1-min rest period between 2–3 min runs. During this period, the user's screen is blank and the interrupt that initiates the background process is disabled. The PC background process stores data in buffers as the session progresses, and the foreground process monitors these buffers and transfers data to disk at appropriate intervals.

Three data files are produced. One contains the parameters that control trial timing, translation of EEG into cursor movement, frequency of movement, and target size, as well as the numbers of targets hit and missed. The second contains, for each target location during each 2–3 min run, the frequency spectra for the EEG cursor-control channels. The third contains the control signals (i.e., the voltages at specific frequencies that controlled cursor movement) for each interrupt-driven interval.

Our laboratory BCI system includes a second PC/80486 equipped with two 32-channel A/D boards, a TMS320C25-based DSP board (both from Spectrum, Inc.), and a monitor. This unit simply records all 64 ear-referenced EEG channels along with the system-state variable (transferred via a serial port from the first [i.e., on-line] PC). Like the on-line hardware described above, the DSP and PC processors of this data collection unit are programmed in a foreground/background configuration. The DSP samples up to 64 EEG channels and passes these data to the PC. The PC displays up to 16 selectable EEG channels simultaneously on a monitor visible to the system operator. The data from all 64 EEG channels are stored in extended memory until the end of each 2–3 min run, at which time they are transferred to a hard disk.

OFF-LINE DATA ANALYSIS

These complete data allow comprehensive off-line analyses. The analyses have three objectives: (1) to reveal the major features of the EEG control developed by the user, (2) to detect the interference of non-EEG artifacts, and (3) to define better methods for controlling cursor movement.

Description of EEG Control

Figure 3 shows scalp topographies generated off-line from 64 channels of EEG data collected during performance of one-dimensional cursor control by one well-trained user. The control signal was the sum of the voltages at 10 Hz at two locations centered over right and left sensorimotor cortices, respectively. Figure 3A shows topographies of the mean 10-Hz voltage computed when the target was at the top or bottom edge of the screen. Mean 10-Hz voltages over both sensorimotor cortices are much greater during top targets than during bottom targets. Voltage is highest on the right side. Figure 3B shows the 10-Hz topography of r^2 (the percent of the total variance of the voltages accounted for by target location, Wonnacott & Wonnacott, 1977). This measure can be considered an index of the signal-to-noise ratio at each location. Like the voltage difference evident in A, the value of r^2 is greatest over the sensorimotor cortices. However, in contrast to A, the largest value of r^2 is over the left side. These topographical displays illustrate the typically narrow spatial focus of a well-trained user's EEG control. (Figures 2 and 5 illustrate correspondingly narrow frequency control, which is also typical.)

Figure 4 illustrates, with data from a well-trained user, the timing of the development of EEG control in response to appearance of a top or bottom target. It plots the control signal (i.e., 10-Hz voltage over left sensorimotor cortex) that controlled cursor movement on-line. Average values are shown for every 100-msec time segment from the first appearance of the target until the cursor started moving. (These data were derived by autoregressive spectral analysis, which provides high-frequency resolution with comparatively short time segments, e.g., Figure 2.) It is clear that the user's response to the target develops within the first second after the appearance of the cursor. This information is important for efforts to maximize the rate and accuracy of cursor control by

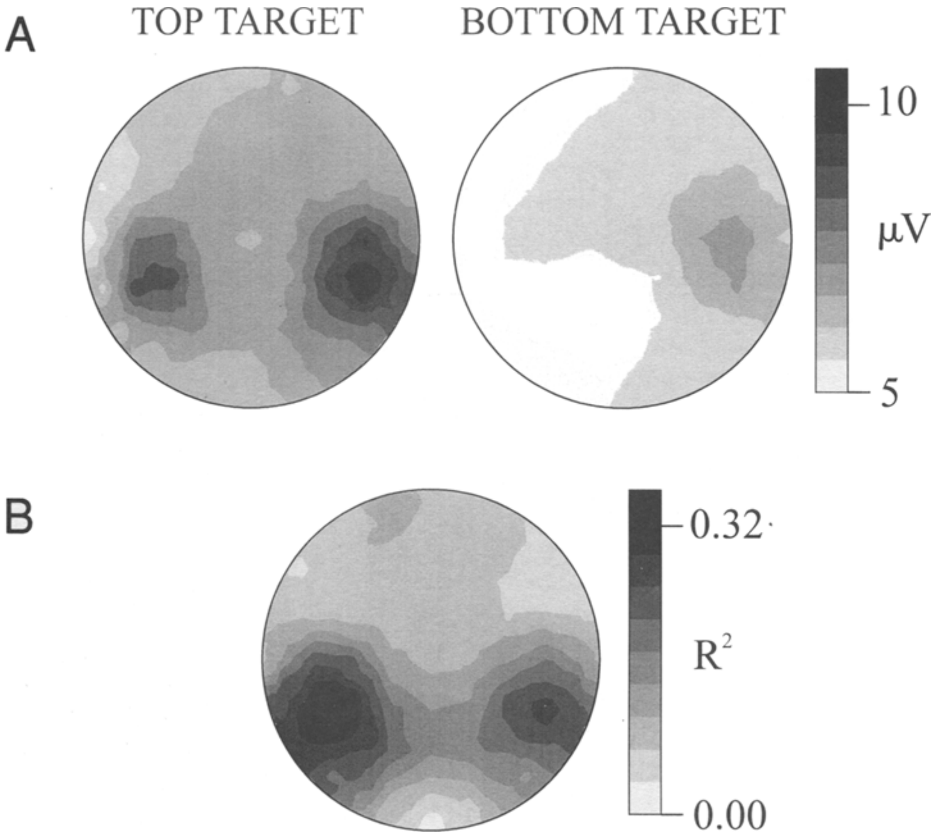

Figure 3. Topographical analysis of four sessions from a well-trained user using the sum of voltages at 10 Hz at locations over right and left sensorimotor cortices (circled) to control vertical cursor movement. (A) Average voltages at 10 Hz for top and bottom targets. (B) Values of r^2 for the top/bottom difference. Electroencephalographic control is focused over the sensorimotor cortex locations that controlled cursor

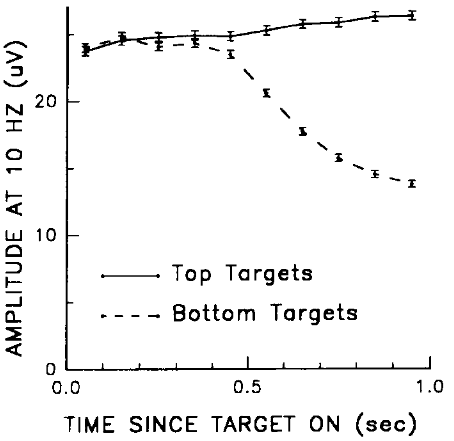

Figure 4. Development of electroencephalographic (EEG) control following appearance of the target at the top (solid) or bottom (dashed) edge of the screen. Voltage at 10 Hz was assessed every 0.1 sec by autoregressive (MEM) spectral estimation. Target position affects EEG within 0.5 sec of target appearance.

modifying trial timing (e.g., shortening the period between target appearance and cursor movement). It is also important because the short latency of EEG control following target appearance rules out the possibility that the mechanism of EEG control is change in respiratory rate or depth (Fried, 1993).

Detection and Elimination of Non-EEG Artifacts

A BCI system, particularly a system intended as a laboratory development tool, must ensure that the electrical activity recorded from the scalp and used for communication is actually EEG. A variety of non-EEG phenomena, including electromyographic (EMG) activity from head and neck muscles, potentials generated by eye movements or blinks, and head movement artifacts can contribute to the electrical activity recorded from the scalp and can masquerade as EEG.

These contaminants can normally be distinguished from true EEG by their spectral characteristics and/or their topographical distributions. For example, true mu rhythm control is centered in a narrow band near 10 Hz and focused over sensorimotor cortex. In contrast, EMG activity is very broad-banded, increases with frequency to a maximum above 100 Hz, and tends to be located near the periphery of the scalp. Eyeblink artifact is mainly low in frequency (i.e., 1–5 Hz) and is concentrated near the forehead. The 64-channel EEG data and the capacity for spatial and spectral analysis provided by our BCI system permit us to detect such artifacts and to prevent them from interfering with BCI performance on-line.

Figure 5 illustrates the spectral distinctions between true EEG control and EMG or eye-movement artifacts. It shows spectral data from a user controlling one-dimensional cursor movement (A) or intentionally producing several non-EEG artifacts (B and C). In A, actual EEG control is seen to be confined to a narrow frequency band. In contrast, in B, tensing of the jaw muscles (i.e., gritting the teeth) produces a broad-banded increase in higher frequency activity, and, in C, rapid eye blinking produces low-frequency activity. We have asked users to generate these and other non-EEG artifacts on numerous occasions and have not found effects comparable in spectral and topographical specificity to true EEG control.

Our BCI system also has the capacity to prevent non-EEG artifacts from contributing to cursor movement and thereby interfering with on-line performance. In the simplest form of artifact detection and rejection, we can specify maximum allowable voltages for specific frequency bands at specific locations (e.g., 5-Hz voltages at frontal electrodes to detect eye blinking). Whenever one of these maxima is exceeded, an artifact is noted and no cursor movement occurs for that time segment. We can also change the color of the cursor when an artifact occurs. This provides additional feedback to the user and thus serves to reduce the frequency of artifacts.

Modification of the On-Line Algorithm

The comprehensive off-line spectral and spatial analysis supported by this DSP-based BCI system permits

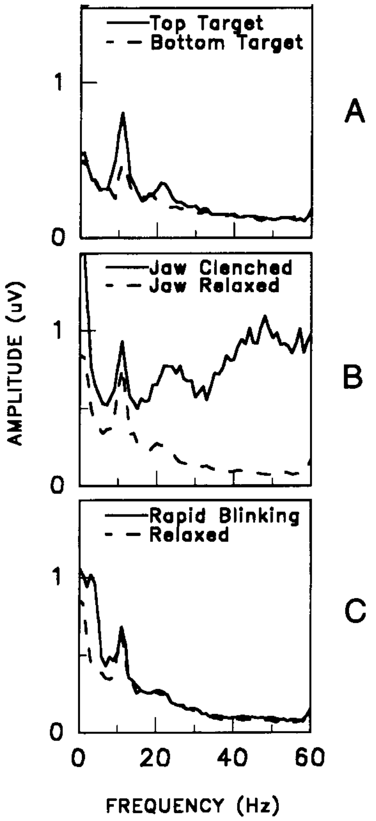

Figure 5. Spectra of activity recorded over sensorimotor cortex from a well-trained user. In A, she is using the voltage at 10 Hz to control cursor movement to top (solid) or bottom (dashed) targets. In B, she is gritting her teeth (solid) or simply sitting quietly (dashed). In C, she is blinking her eyes rapidly (solid) or simply sitting quietly (dashed). The sharply focused electroencephalographic (EEG) control evident in A is easily distinguished from the non-EEG artifacts in B and C.

evaluation of alternative algorithms for translating EEG control into rapid and accurate cursor movement. With analyses like those shown in Figure 3, we can detect and measure control at locations and frequencies other than those that were used to control the cursor on-line.

For example, analysis of EEG stored while users were controlling two-dimensional cursor movement with the sum and difference of mu rhythm amplitudes over sensorimotor cortex (i.e., the original two-dimensional control algorithm, Wolpaw & McFarland, 1994) showed that 12–15 Hz activity over occipital areas was correlated with right/left target location. Current studies are evaluating the use of these posterior channels to control horizontal cursor movement (McCane, McFarland, Vaughan, & Wolpaw, 1995).

Furthermore, we can incorporate alternatives suggested by off-line analysis into the algorithm that controls cursor movement and then apply this new algorithm to the stored data to simulate its on-line performance. Through this analysis, we can predict the effects on the accuracy and speed of cursor movement of using these alternative algorithms on-line. At the same time, it is important to recognize that the on-line results of any modification are likely to differ to some degree from the results predicted by off-line analysis, because the altered performance caused by the modification is likely to affect the user's EEG. Thus, modifications suggested by off-line analysis must eventually be tested on-line to determine whether they actually provide improved cursor control.

POTENTIAL APPLICATIONS

With our present BCI system, users can eventually achieve highly accurate control (i.e., > 90% success rate) of one-dimensional cursor movement. Presently attainable two-dimensional control, although impressive (Wolpaw & McFarland, 1994), is not yet similarly accurate and consistent. In its present state, this system might be of use to individuals who have little or no voluntary movement. People with advanced amyotrophic lateral sclerosis (ALS, or Lou Gehrig's disease), for example, may find it difficult or impossible to operate conventional single-switch control systems. As a result, they may be "locked in" to their bodies, unable to express even "yes" or "no" reliably. Our EEG-based BCI system, which does not depend on voluntary movement, might enable such individuals to control a variety of single-switch assistive communication devices, such as row-column scanning programs and environmental control interfaces. As BCI development proceeds and two-dimensional performance improves, BCI communication could become useful for individuals with less severe disabilities.

A crucial issue for these potential applications is the extent to which those with severe motor disabilities can learn BCI operation. In our limited experience to date (e.g., McFarland, McCane, Vaughan, & Wolpaw, 1994; Miner, McCane, Vaughan, McFarland, & Wolpaw, 1996), disabled users have proved as adept as normal users at mastering EEG-based cursor control. Thus, we are optimistic about the potential usefulness of this new technology as an assistive communication alternative.

CONCLUSIONS

Development and implementation of EEG-based brain–computer communication requires a system that is highly flexible and capable of rapid and complex real-time processing and that provides comprehensive topographical and spectral data. The analog methodologies frequently used in EEG research and applications cannot readily satisfy these requirements. The DSP-based BCI system described here has the requisite flexibility, speed, and processing capacity. It supports high-speed spatial and spectral signal processing, derivation of control signals, control of an output device, and storage of all EEG data and control parameters. These capabilities facilitate comprehensive analysis of the characteristics of EEG control, detection and elimination of non-EEG artifacts, detection and evaluation of alternative control signals, and implementation and evaluation of alternative analysis algorithms. Furthermore, the system can easily incorporate higher capacity hardware components and additional software modifications as they become available or desirable. Finally, although device control is currently limited to cursor movement, the flexibility of the system should permit incorporation of a variety of other outputs, including commercially available mouse-driven programs, row-column scanning devices, and other standard control interfaces.

In its current state of development, our BCI system could serve individuals who are totally paralyzed and thus unable to use conventional assistive communication devices. With further improvements, it could also become useful to the much larger number of individuals with less severe motor disabilities.

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(Manuscript received October 31, 1995;
revision accepted May 8, 1996.)