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# EEG-based communication: presence of an error potential

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**Abstract**

**Background**: EEG-based communication could be a valuable new augmentative communication technology for those with severe motor disabilities. Like all communication methods, it faces the problem of errors in transmission. In the Wadsworth EEG-based brain-computer interface (BCI) system, subjects learn to use mu or beta rhythm amplitude to move a cursor to targets on a computer screen. While cursor movement is highly accurate in trained subjects, it is not perfect.

**Methods**: In an effort to develop a method for detecting errors, this study compared the EEG immediately after correct target selection to that after incorrect selection.

**Results**: The data showed that a mistake is followed by a positive potential centered at the vertex that peaks about 180 ms after the incorrect selection.

**Conclusion**: The results suggest that this error potential might provide a method for detecting and voiding errors that requires no additional time and could thereby improve the speed and accuracy of EEG-based communication. 

*Keywords*: Error potential; Event related potential; Error related negativity; Brain-computer interface; Mu rhythm; Sensorimotor cortex; Electroencephalography; Augmentative communication; Rehabilitation

## 1. Introduction

All presently available augmentative communication systems depend in some measure on voluntary muscle control. Thus, they are useless to those who are totally paralyzed and to some others with severe motor disabilities. EEG-based communication, because it does not depend on voluntary muscle control, could provide a valuable new communication and control option for these individuals. Over the past decade, a number of laboratories have begun developing EEG-based communication as a new augmentative technology for people with motor disabilities (e.g. Wolpaw et al., 1986, 1991; Farwell and Donchin, 1988; Sutter, 1992; Pfurtscheller et al., 1993; Wolpaw and McFarland, 1994; McMillan and Calhoun, 1995; Vaughan et al., 1996; Kalcher et al., 1996; McFarland et al., 1997a; Birbaumer, 1997; Birbaumer et al., 1999).

Like other communication technologies, EEG-based communication faces the problem of errors in transmission; and, in its current early stage of development, errors are frequent. In the Wadsworth EEG-based brain-computer interface (BCI) system, subjects learn to control the amplitude of mu and/or beta rhythms recorded over sensorimotor cortex and use that control to move a cursor to targets on a video screen (McFarland et al., 1997a). Trained subjects, presented with a target in one of two possible locations (e.g. top or bottom of screen), routinely reach the correct target on 80–97% of the trials, but not often on 100%. Recent studies described a response verification (RV) procedure in which errors were greatly reduced by asking the subject to confirm each selection by moving the cursor to a target at the opposite location (Wolpaw et al., 1998; Miner et al., 1998). While this method was effective, it required extra time, and thus it reduced the rate of communication.

The present study set out to explore another option for error detection: the possibility that some feature of the EEG just after the end of a trial reveals whether the trial was a success or an error, that is, whether the outcome was or was not what the subject desired. Such a feature might be used to detect an error and void the outcome, without requiring additional time. This approach to error detection was encouraged by evidence that errors in conventional motor

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performances have detectable effects on the EEG recorded just after the error occurs (Falkenstein et al., 1991, 1995; Bernstein et al., 1995; Gehring et al., 1995). The present results show that errors in EEG-based cursor control are in fact followed by an error potential that might be used to improve the speed and accuracy of EEG-based communication and control.

## 2. Methods

Subjects were 4 adults: one woman (i.e. subject C) and 3 men (i.e. subjects A, B, D), aged 59, 47, 34, and 45 years, respectively. Three had no disabilities, while one had an abnormal gait due to lower limb ankyloses secondary to hemophilia. All gave informed consent for the study, which had been reviewed and approved by the New York State Department of Health Institutional Review Board. After an initial evaluation defined the frequencies and scalp locations of each subject's spontaneous mu and beta rhythm activity, each subject learned EEG-based cursor control over 10 initial 30 min sessions (2–3/week) and then participated for an additional 14–98 sessions. These sessions were devoted to a variety of studies of EEG-based communication (e.g. the present study, McFarland et al., 1998; Miner et al., 1998; Vaughan et al., 1998). Over the course of each subject's participation, off-line data evaluations and concurrent improvements in system hardware and software led to adjustments in the electrode locations, frequency bands, and spatial filter used by the on-line algorithm that controlled cursor movement. The next section summarizes the on-line methodology used in the present study. A detailed description of system configuration and operation is available elsewhere (McFarland et al., 1997a; Ramoser et al., 1997).

### 2.1. Data collection

The subject sat in a reclining chair facing a video screen and was asked to remain motionless during performance. Scalp electrodes recorded 64 channels of EEG (Sharbrough et al., 1991), each referred to an electrode on the right ear (amplification 20 000; bandpass 1.0–60 Hz or 0.1–60 Hz). A subset of channels was digitized at 196 Hz and used to control cursor movement on-line as described below. In addition, all 64 channels were digitized at 128 Hz and stored for later analysis.

In this study, the subjects used mu or beta rhythm amplitude to control vertical cursor movement to a word (i.e. 'YES' or 'NO') located at the top or bottom edge of the video screen. Data were collected from each subject for 5–8 sessions of 30 min each. Each session consisted of 8 3 min runs, separated by 1 min breaks, and each run consisted of 20–32 individual trials. At the beginning of each run, the subject was told to select (i.e. move the cursor to) the word 'YES' or the word 'NO.' 'YES' and 'NO' runs alternated in each session (e.g. for runs 1, 3, 5 the correct selection was 'YES' and for runs 2, 4, 6 the correct selection was 'NO'). Each trial began with a 1-s period during which the screen was blank. Then, 'YES' appeared at the top or bottom edge of the screen and 'NO' appeared at the opposite edge. One second later, the cursor appeared in the center of the screen and began to move vertically 10 times/s controlled by the subject's EEG as described below. The cursor had 188 possible vertical positions. The subject's goal was to move the cursor to the correct word. When the cursor reached a word (usually in 2–3 s), the screen went blank. This event signaled the end of the trial and is defined as time zero for the results presented in this study. Within 80 ms (i.e. the screen redraw time), the selected word, whether right or wrong, appeared in the center of the screen and remained for 100 ms. It appeared on the screen again from 1.0 to 1.1 s and from 2.0 to 2.1 s. Thus, the events on the screen at the end of the trial were the same for hits and misses (e.g. if 'YES' was selected, the screen went blank and then 'YES' flashed 3 times whether or not it was the correct selection). After the third flash, the screen was blank for 0.9 s and then the next trial began. Each word appeared an equal number of times at each location, and its location was randomized in blocks of 8. Thus, accuracy expected in the absence of any EEG control was 50%.

Cursor movement was controlled as follows. Ten times/s, the last 200 ms of digitized EEG from 1–3 channels over sensorimotor cortex was re-referenced to a common average reference or a Laplacian derivation (McFarland et al., 1997b) and then submitted to frequency analysis by an autoregressive algorithm (McFarland et al., 1997a) to determine amplitude (i.e. the square root of power) in a mu and/or beta rhythm frequency band. The amplitudes for the 1–3 channels were combined to give a control signal that was used as the independent variable in a linear equation that controlled cursor movement. During the cursor movement period, eyeblink or other non-EEG artifacts were detected by monitoring specific frequency bands at specific electrodes (e.g. Fpz for eyeblinks); and whenever the amplitude in one of these bands exceeded a defined criterion, cursor movement was aborted. These trained subjects remained relaxed and blinked rarely during cursor movement, so that artifacts were infrequent.

### 2.2. Data analysis

Analysis of the stored 64 ear-referenced EEG channels focused on the period before and after the moment when the cursor reached 'YES' or 'NO' and the screen went blank (i.e. defined here as time zero). For each channel of each subject, we computed the average EEG during this period for all correct trials (i.e. hits) and for all incorrect trials (i.e. misses). We then subtracted the former from the latter to produce the miss-minus-hit difference potential.

Subjects often blinked at or just after the end of a trial, presumably because they had deferred blinking during cursor movement. This behavior was consistent with work

indicating that people tend to defer blinking until the end of performance (Ohira, 1996). As a result, the EEG at the end of the trial (i.e. around time zero) was frequently contaminated by a prominent eyeblink artifact. In response to this problem, we analyzed the data by two different methods and compared the results. In one method, we applied the least-mean square (LMS) adaptive interference canceling (i.e. EOG correction) algorithm (Stearns and David, 1988; Bernardin, 1993) to the individual miss or hit trials prior to averaging. We used channel Fpz as the noise source and parameters $n = 512$, no. of delays $= 20$, $m = 0.025$, $r = 0.2$, and noise-power $= 0.5$. In the other method, we simply eliminated all trials containing large eyeblink artifacts in the period from −220 ms to +580 ms. We did this by eliminating all trials in which EEG amplitude at Fpz exceeded 50 μV at any point during this period (Verleger, 1993). Over all subjects, 38.2% of the hits and 26.5% of the misses remained. As will be seen, for each subject and for all subjects together, the miss-minus-hit difference potentials derived by these two methods were very similar. (As a further verification, we eliminated all trials in which Fpz exceeded 25 μV. While this stringent criterion left only 27.3% of the hits and 16.8% of the misses, the miss-minus-hit difference potentials were still comparable to those produced by the 50-μV criterion and by the EOG correction method.)

## 3. Results

Target accuracies for the 4 subjects, A–D, for the sessions devoted to this study were 94%, 93%, 86%, and 85%, respectively. The miss-minus-hit results were consistent across subjects and across analysis methods. Each subject showed a miss-minus-hit difference potential that consisted of a positive potential that peaked about 180 ms after the cursor reached a word and the screen went blank. This difference, henceforth called the 'error potential,' was centered at the vertex (i.e. Cz). Fig. 1 (EOG-corrected data) and Fig. 2 (eyeblink-free data) show for each subject and all subjects together the error potentials at Cz and the scalp topographies for 40 ms periods near their peaks. The positive peak centered near the vertex is present in each subject and with both methods. The signal-to-noise ratio is clearly higher in Fig. 1, presumably due to the greater number of individual trials comprising each average. The only other noticeable difference between Figs. 1 and 2 is the greater prominence in Fig. 1 than in Fig. 2 of a later negative peak. This difference might reflect an actual difference between all the trials and the subset of eyeblink-free trials, or it might be an artifact of the EOG correction procedure. In any case, the two methods are in close agreement in regard to the major feature of the miss-minus-hit difference, the positive potential.

Fig. 3, derived from the eyeblink-free trials, shows the average EEG traces for hit trials and for miss trials for all subjects together at 9 scalp locations. The positive potential present after misses and absent after hits is evident, and it is largest at Cz.

To assess the signal-to-noise ratio and the statistical significance of the error potential, we analyzed EOG-corrected data from channel Cz. To eliminate low-frequency voltage shifts, we calculated for each trial a linear regression from 0.92 s before the end of the trial to 3.08 s after and subtracted it from the trial's data. We then calculated for each trial the average amplitude for the 40 ms period encompassing the 180 ms peak of the error potential and determined from all the trials of each subject the value of $r^2$, the proportion of the total variance of the amplitudes that was accounted for by whether the trial was a hit or a miss. The results are shown in Table 1. They indicate that, while its signal-to-noise ratio was low, the error potential was clearly statistically significant in every subject.

To assess the possible value of the error potential for EEG-

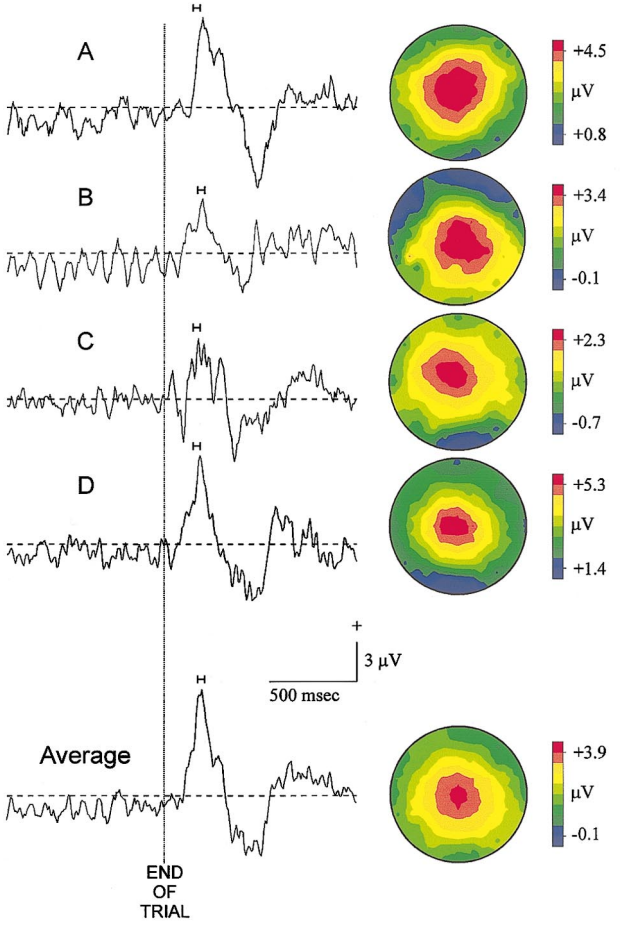


Fig. 1. Average miss-minus-hit EEG traces at Cz from 0.92 s before the end of the trial to 1.08 s afterwards (left), and scalp topographies for 40 ms periods (indicated by bars in the traces) near the positive peak of the error potentials (right), for each subject (A–D) and for all subjects together. All signals were referenced to the right ear and EOG correction was applied to the individual trials prior to averaging (see text). The horizontal dashed lines indicate zero voltage.

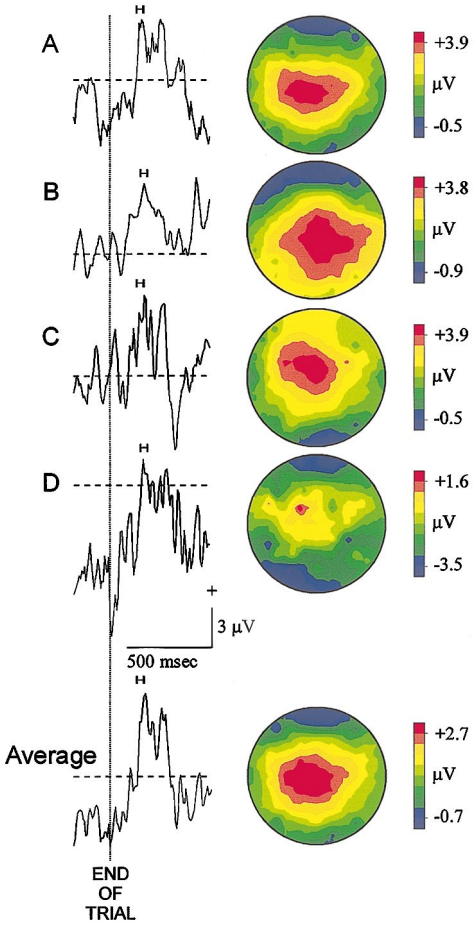


Fig. 2. Average miss-minus-hit EEG traces at Cz from 0.22 s before the end of the trial to 0.58 s afterwards (left), and scalp topographies for 40 ms periods (indicated by bars in the traces) near the positive peak of the error potentials (right), for each subject (A–D) and for all subjects together. Only trials without eyeblink artifacts are included in these averages (see text). All signals were referenced to the right ear. The horizontal dashed lines indicate zero voltage.

based communication, we calculated its expected effects on bits/trial and on accuracy. Information transfer rate, measured here as bits/trial, is a standard method of assessing communication systems (Pierce, 1980). It reflects both speed and accuracy. For example, with two targets and 100% accuracy, bits/trial is 1.00, while with two targets and 90% accuracy, bits/trial is only 0.53. In an off-line analysis of each subject's data, we compared the bits/trial and accuracy obtained without using the error potential to those obtained when we used the error potential to reject trials likely to be errors. Without use of the error potential, accuracy was simply

$$A = \frac{\text{Hits}}{\text{All Trials}} \tag{1}$$

and bits/trial was

$$B = 1 + A\log_2 A + (1 - A)\log_2(1 - A) \tag{2}$$

(Pierce, 1980). When we used the error potential, we rejected trials for which the average amplitude for the 40 ms time period encompassing the error potential exceeded a threshold $t_j$, and varied $t_j$ across the entire range of amplitudes from all trials. The accuracy was

$$A(t_j) = \frac{\text{All Hits} - \text{Rejected Hits}}{\text{All Trials} - \text{Rejected Trials}} \tag{3}$$

and bits/trial was

$$B(t_j) = [1 + A\log_2 A + (1 - A)\log_2(1 - A)]\frac{\text{All Trials} - \text{Rejected Trials}}{\text{All Trials}} \tag{4}$$

We then determined the maximum value of bits/trial. It is important to note that, as Eq. (4) shows, the number of trials used to determine bits/trial was always the total number of trials obtained from the subject, and thus included those trials that had been rejected and therefore conveyed no information (i.e. 0 bits). We also determined accuracy at the value of $t_j$ that gave the maximum value of bits/trial.

Table 1 shows for each subject accuracy and bits/trial without and with use of the error potential. For subjects A, C, and D, the error potential clearly increased both accuracy and bit rate, while for subject B it had only minimal effects. As would be expected, the improvement provided by the error potential correlated with its $r^2$ value, also shown in Table 1.

## 4. Discussion

The results summarized in Figs. 1–3 and in Table 1 indi-

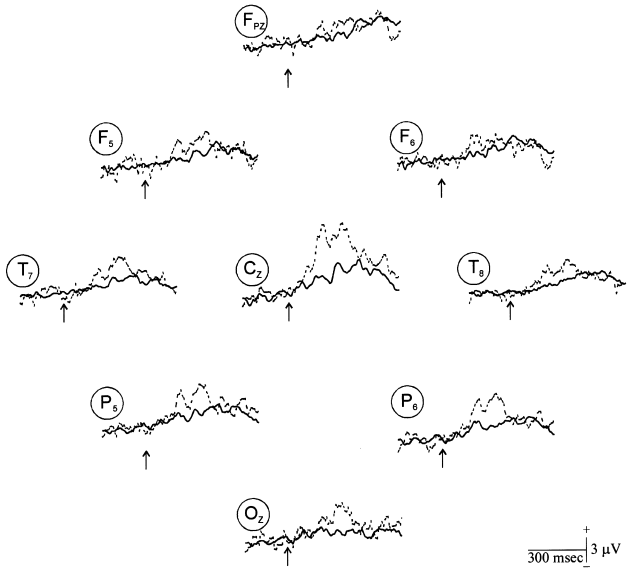


Fig. 3. Average EEG traces from all subjects for hits (solid) and misses (dashed) for different electrode positions from 0.22 s before the end of the trial to 0.58 s afterwards. Only trials without large eyeblink artifacts are included in these averages (see text). The arrows indicate the end of the trial. In order to facilitate observation of the positive error potential that peaks about 180 ms after the end of the trial, hit and miss traces were aligned vertically so as to superimpose their voltage levels prior to the end of the trial. All signals were referenced to the right ear.

Table 1
Signal-to-noise ratio (expressed in $r^2$) of the error potential (EP), accuracy and bits/trial without and with the error potential, and the gain in bits/trial provided by the error potential for each subject[a]

| Subject | $r^2$ | Accuracy without EP (%) | Accuracy with EP (%) | Bits/trial without EP | Bits/trial with EP | Gain in bits/trial (%) |
|---|---|---|---|---|---|---|
| A | 0.064 | 93.9 | 95.3 | 0.670 | 0.691 | 3 |
| B | 0.035 | 93.1 | 93.2 | 0.638 | 0.639 | 0 |
| C | 0.061 | 85.5 | 88.4 | 0.404 | 0.419 | 4 |
| D | 0.124 | 85.4 | 90.6 | 0.399 | 0.481 | 21 |

[a] All $r^2$ values are significant with $P \ll 0.001$.

cate that in these trained subjects errors are associated with a statistically significant error potential centered at the vertex and consisting of a positive potential occurring about 180 ms after the end of the trial. The vertex focus of this potential, combined with the fact that it is visible both in the EOG-corrected data and in the eyeblink free data, indicate that it is error-related EEG activity rather than a non-EEG artifact or EEG activity associated with eyeblinks. Furthermore, the display at the end of the trial was identical for hits and misses, i.e. the selected word flashed whether it was right or wrong; and analysis showed that success probability did not depend on the location of the correct word or on which word was correct. Thus, this error potential cannot be attributed to some aspect of the sensory input occurring at the end of the trial. It can be confidently ascribed to the subject's knowledge that an error has occurred.

While the potential shown in Figs. 1 and 2 clearly depends on the subject's perception of an error, it is not clear what exactly evokes the potential. It might be evoked by the blanking of the screen at the end of a trial and/or by the subsequent flash of the wrong word (though the potential is probably too early to be due to the flash). Alternatively, the process leading to the error potential might begin prior to the end of the trial, as the cursor's approach to the wrong word leads the subject to anticipate an error. If this does occur, the 180 ms peak latency of the error potential in Figs. 1 and 2 is deceptively low and its actual latency is difficult to determine.

While the positive error potential peaking at 180 ms is the most prominent and consistent phenomenon evident in the results, Figs. 1 and 2 prompt several other observations. In subjects A, B, and D at least, the voltage level deviates from zero before the end of the trial. This difference between hit trials and miss trials could reflect a difference in slow cortical potentials related to attention or expectancy (Birbaumer et al., 1990; Rockstroh et al., 1982). In addition, the data of subject B in Fig. 1 show low-amplitude 10 Hz rhythmic activity prior to the end of the trial, and similar activity may be discernible in subjects A and C. This activity might reflect a small difference between hit and miss trials in mu rhythm phase or amplitude or in a cortical potential evoked by the 10/s cursor movements.

### 4.1. Nature of the error potential

Error potentials, both positive and negative in polarity, have been detected with a variety of tasks. They have been noted in choice-reaction tasks (Falkenstein et al., 1991, 1995; Bernstein et al., 1995; Gehring et al., 1995), language processing (Friederici et al., 1993), and time estimation (Miltner et al., 1997). The error potential described here could be a comparable phenomenon. On the other hand, because errors are relatively unusual in these trained subjects, the present error potential could be an example of a P3, or 'oddball' response to an infrequent stimulus, i.e. an error. This interpretation might be tested by studies of new subjects during their first few training sessions, when accuracy is usually low (i.e. 50–70%). If the error potential described here is a P3 response to an unusual event, it should be much less prominent in such subjects, while if it is simply a response to an error it should be equally prominent.

### 4.2. Possible value of the error potential for communication

Whatever the nature of the error potential, the central issue for EEG-based communication is how useful the error potential can be in detecting errors in single trials, and thereby improving accuracy and/or information transfer rate. While its signal-to-noise ratio is low, Table 1 shows that, for 3 of the 4 subjects, the error potential could increase both accuracy and information transfer rate. In the remaining subject, it would probably have only a small effect. (While the improvement provided by the error potential depends on the value of $t_j$, the threshold for trial rejection, which in actual on-line applications would have to be determined by prediction from previous results, experience with prediction of other crucial parameters (e.g. Ramoser et al., 1997) suggest that accurate prediction of $t_j$ should not be difficult.)

Furthermore, more sophisticated EOG correction algorithms and better methods for recognizing and measuring the error potential could substantially improve its signal-to-noise ratio, and thereby increase its impact on accuracy and bits/trial. The linear single-reference LMS EOG correction algorithm used here does not take into account possible phase differences between the noise channel (i.e. Fpz) and the signal channel (i.e. Cz), and it only uses one noise channel, which may not be enough, even for measurements along the midline (Elbert et al., 1985). Other methods could provide superior EOG correction and thus more consistent error potentials (Berg, 1986; Cerrutti et al., 1988; James et

al., 1997). The optimal method for recognizing and measuring the error potential also remains to be defined (e.g. absolute amplitude criterion or pattern matching algorithms). Changes in subject performance could also prove beneficial. The signal-to-noise ratio of the error potential might increase if subjects learned to defer blinking during the period immediately after the trial, as they now do during the trial. In addition, it is possible that, when the error potential is incorporated into the on-line algorithm, subjects will learn to increase its amplitude and consistency, just as they now learn to control mu or beta rhythm amplitude.

With further investigation of these issues, the error potential described in this study could play an important role in EEG-based communication. As explained in Pierce (1980), and originally in Shannon and Weaver (1964), errors greatly reduce information transfer rate. Thus, a procedure that permits many errors to be recognized and voided, and does not require additional time, could substantially improve the rate of EEG-based communication.

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