Stereoelectroencephalography for continuous two-dimensional cursor control in a brain-machine interface

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Stereoelectroencephalography (SEEG) is becoming more prevalent as a planning tool for surgical treatment of intractable epilepsy. Stereoelectroencephalography uses long, thin, cylindrical “depth” electrodes containing multiple recording contacts along each electrode’s length. Each lead is inserted into the brain percutaneously. The advantage of SEEG is that the electrodes can easily target deeper brain structures that are inaccessible with subdural grid electrodes, and SEEG does not require a craniotomy. Brain-machine interface (BMI) research is also becoming more common in the Epilepsy Monitoring Unit. A brain-machine interface decodes a person’s desired movement or action from the recorded brain activity and then uses the decoded brain activity to control an assistive device in real time. Although BMIs are primarily being developed for use by severely paralyzed individuals, epilepsy patients undergoing invasive brain monitoring provide an opportunity to test the effectiveness of different invasive recording electrodes for use in BMI systems. This study investigated the ability to use SEEG electrodes for control of 2D cursor velocity in a BMI. Two patients who were undergoing SEEG for intractable epilepsy participated in this study. Participants were instructed to wiggle or rest the hand contralateral to their SEEG electrodes to control the horizontal velocity of a cursor on a screen. Simultaneously they were instructed to wiggle or rest their feet to control the vertical component of cursor velocity. The BMI system was designed to detect power spectral changes associated with hand and foot activity and translate those spectral changes into horizontal and vertical cursor movements in real time. During testing, participants used their decoded SEEG signals to move the brain-controlled cursor to radial targets that appeared on the screen. Although power spectral information from 28 to 32 electrode contacts were used for cursor control during the experiment, post hoc analysis indicated that better control may have been possible using only a single SEEG depth electrode containing multiple recording contacts in both hand and foot cortical areas. These results suggest that the advantages of using SEEG for epilepsy monitoring may also apply to using SEEG electrodes in BMI systems. Specifically, SEEG electrodes can target deeper brain structures, such as foot motor cortex, and both hand and foot areas can be targeted with a single SEEG electrode implanted percutaneously. Therefore, SEEG electrodes may be an attractive option for simple BMI systems that use power spectral modulation in hand and foot cortex for independent control of 2 degrees of freedom.

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KEY WORDS • stereoelectroencephalography • depth electrode •
brain-machine interface • brain-computer interface • epilepsy monitoring unit

In a subset of patients with medically intractable focal epilepsy, invasive brain monitoring is recommended to best identify the location of the epileptogenic zone. These patients are admitted to an Epilepsy Monitoring Unit (usually between 3 and 15 days) to await seizure occurrence to better localize the epileptic focus.³³ Subdural grids are thin flexible sheets or strips of electrodes and are the most commonly used electrodes for invasive epilepsy monitoring in the US. Alternatively, SEEG uses “depth electrodes” consisting of very thin (approximately 1–2

mm), flexible cylindrical leads with multiple concentric metal electrodes along the length of the lead. Subdural grids require implantation via craniotomy, whereas each SEEG lead only requires a small bur hole placed percutaneously. A stiff wire insertion tool is used to insert the flexible SEEG electrode into the brain to a stereotactically defined depth and angle.

Subdural grid electrodes have the benefit of sampling a large cortical surface area. However, identifying activity in deeper structures, such as the hippocampus, is more difficult with subdural grids. Stereoelectroencephalography electrodes, on the other hand, have the benefit of sampling deeper structures but are limited in terms of cortical

Abbreviations used in this paper: BMI = brain-machine interface; EEG = electroencephalography; SEEG = stereo-EEG.

mapping.³² Whereas SEEG was originally described by Bancaud et al. in 1965,³ it has only become increasingly used in the US since 2009.^{14,32}

Brain-machine interfacing is an area of research that is increasingly taking place in tandem with invasive brain monitoring for epilepsy,^{1,2,6–8,10,11,18–30,35} even though BMI systems are being developed primarily for paralyzed individuals. Brain-machine interface systems “decode” some aspect of brain processing in real time and use that decoded information to control an assistive device. Motor-system BMIs decode one’s attempted or desired movement from the recorded brain signals. The decoded movement information can then be used in real time to control the movement of a device such as a computer cursor, a wheelchair-mounted robotic arm, or even one’s own paralyzed arm reanimated via implanted peripheral nerve stimulators that can activate the paralyzed muscles.³⁶ Brain-machine interfaces are also being developed to enable fully paralyzed or “locked-in” individuals to communicate via brain-driven typing programs. These communication BMIs often use nonmotor signals, such as changes in brain activity that occur when the letter the person wishes to type flashes on the screen.

A limited number of paralyzed individuals have now been implanted with invasive brain recording electrodes specifically for use in prototype BMI systems.^{4,5,12,13,15–17,31,34} Patients in the Epilepsy Monitoring Unit provide a unique opportunity to speed up BMI development by providing a pool of research participants that already have invasive brain recording electrodes implanted. Current literature on BMI studies performed in the Epilepsy Monitoring Unit reflects the prevalence of subdural grid electrodes in brain monitoring.^{1,2,6–8,10,11,19–29,35} To date, there have been very few BMI studies conducted using SEEG,^{18,30} and none, to our knowledge, have investigated the feasibility of using SEEG for continuous control of 2D cursor movements. In this paper we present our initial experiences conducting motor BMI testing with 2 patients who underwent SEEG electrode implantation for localization of ictal onset. Our results show that both foot and hand movements could be detected on a single SEEG electrode lead when recording contacts were located both in the hand knob area and in the foot area. Our findings point to some potential advantages of SEEG electrodes over subdural grids in certain types of BMI systems.

Methods

Epilepsy Workup and Patient Selection

Prior to invasive electrode implantation, both patients were discussed in a multidisciplinary epilepsy management conference that included neurosurgeons, epileptologists, radiologists, and neuropsychologists. Standard preoperative workup for epilepsy surgery was performed for both patients, which included high-resolution MRI, magnetoencephalography, SPECT, and PET testing. Patient 1 also underwent preoperative neuropsychological testing based on the location of the lesion. The diagnosis of medically refractory focal epilepsy was made using video-EEG.

After discussion in the epilepsy management confer-

ence, both patients were deemed candidates for SEEG electrode implantation because of the deep location of their lesions. The epileptogenic focus in Patient 1 was hypothesized to be located along the left lateral ventricle, where small foci of subependymal gray matter heterotopias were identified. A total of 13 SEEG electrodes with 10–12 contacts each were implanted in the left hemisphere and an additional 2 SEEG electrodes (10 contacts each) were implanted on the right side. Magnetic resonance imaging in Patient 2 suggested a deep lesion in the dorsal surface of the right posterior cingulate, suggestive of cortical dysplasia. A total of 11 SEEG electrodes were implanted in the right hemisphere (10–12 contacts each), and 2 electrodes (10 contacts each) were implanted in the left hemisphere. Electrodes and the insertion stylus were manufactured by Ad-Tech Medical Instrument Corporation.

Decisions on the number of SEEG electrodes to implant and their locations were based solely on clinical diagnostic requirements. The BMI study was not a factor in determining the electrode implantation locations. Patient 1 participated in 2 testing sessions on different days. Patient 2 was only available for 1 testing session. All research-related activities were approved by the Cleveland Clinic Institutional Review Board, and signed consent forms were obtained from both participants.

Methods for 2D Cursor Control

The Y-shaped splitter connectors were inserted into the SEEG electrode leads, allowing for a copy of the voltage traces to go to our BMI testing equipment while clinical monitoring continued uninterrupted. For this study, 32 SEEG signals were connected to our BMI system via the Y-shaped splitter connectors. However, with Patient 2, 4 of the connectors experienced technical difficulties. Therefore, Patient 1 used all 32 SEEG electrodes for the BMI experiments, but Patient 2 used only 28 SEEG signals. The subset of electrode contacts was chosen based on their proximity to areas of the brain that might convey useful movement information. Figure 1A and B show the approximate SEEG lead locations overlaid on each patient’s brain, reconstructed from imaging data; circles indicate the electrode contacts used in the BMI study.

This study used the same testing protocol we have used in previous BMI studies, in which brain signals recorded via EEG electrodes were used to control 2D movements of an object on a screen.⁹ The experimental paradigm and signal processing methods used were similar to those we published in our previous EEG study and are briefly summarized below.

In this study, we were not trying to decode precise limb motion from the associated brain signals. Instead, we used more robust power spectral changes associated with moving or resting 2 different body parts to control the horizontal and vertical components of cursor velocity in real time. Both participants received individual SEEG leads with recording contacts in both the hand knob area and in the foot motor cortex. Patient 2 also had an SEEG lead spanning hand and foot sensory cortex. Therefore, hand and foot movements were chosen for use in this task. Specifically, SEEG power spectral changes reflecting the degree of wiggling compared with resting the hand were

Fig. 1. Stereoelectroencephalography electrode locations in the 2 patients. **A and B:** Electrode locations for Patients 1 and 2, respectively. Circles indicate contacts that were used for the BMI study. Blue = contacts not significantly modulated with movement; pink = contacts modulated with hand movement; yellow = contacts modulated with foot movement; black = contacts excluded due to noise contamination; star = contact outside of the brain. **C and D:** Coronal section diagrams illustrate the electrode leads traversing hand and foot cortex for Patients 1 and 2, respectively. Circle size indicates the relative accuracy of the hand (pink) or foot (yellow) movement information decoded from each individual contact as measured by the correlation coefficient between the predicted and actual hand or foot movement state. Scaling ranges from $r = 0.56$ (largest circle) to $r = 0.27$ (smallest circle excluding dots).

assigned to control the horizontal component of the cursor velocity. Power spectral changes reflecting the degree of wiggling compared with resting the foot were assigned to control the vertical component of the cursor velocity in real time.

To generate a decoding function to translate each participant's brain activity into 2D cursor motion in real time, we first collected 3 minutes of baseline data in which the participants wiggled/rested their hands and feet on cue. To cue the participants, 4 targets were presented on the screen, 1 at a time. The participants were instructed to wiggle or rest their hands or feet in the 4 different combinations that would be required to move the cursor to each target during the upcoming brain-control task. This process also served to acclimate the participants to performing the different combinations of hand and foot movements. Figure 2 illustrates the 4 movement combinations as well as how moving and/or resting the hand and foot by different degrees would be used to move the brain-controlled cursor during the closed-loop target acquisition task.

The collected baseline data were used to generate a decoding function by first using a common spatial pattern algorithm to generate 8 linear combinations of the raw signals that maximized the difference in magnitude between the moving and resting states for the hand and foot (algorithm described in Foldes and Taylor⁹). Power

spectral data from the original raw SEEG signals along with the 8 new combined signals were then calculated. X and Y movement decoding functions were generated by regressing the power values against the X and Y components of the 4 target locations used to cue the different hand and foot movements. Power values from only 4 frequency bands were used on Patient 1's first day of testing (0–0.75 Hz, 0.75–2.25 Hz, 2.25–9 Hz, and 9–30 Hz), but power values in 7 frequency bands were used for Patient 1's second testing session and for Patient 2's testing session (4.5–7.5 Hz, 7.5–10.5 Hz, 10.5–13.5 Hz, 13.5–18 Hz, 18–22.5 Hz, 22.5–36 Hz, and 36–100.5 Hz). The regression coefficients became the decoding function that would map each participant's power values to an X and Y cursor velocity at each time step in the subsequent cursor control task.

Once each patient's decoder coefficients were generated, the task was switched to the 2D target acquisition task in which participants had to use their decoded brain signals to move a cursor to the displayed targets in real time. At the beginning of each trial, the cursor would appear in the center of the screen along with 1 of the 4 radial targets. Participants had up to 5 seconds to use their brain signals to steer the cursor to the target before the cursor would be recentered, and a new target would appear. The participants were instructed to wiggle their hands when they needed to steer the cursor right and to rest their hands

FIG. 2. Illustration of the hand and foot actions used for 2D cursor control. **Left:** During baseline data collection, participants performed the 4 listed combinations of hand and foot movements when each target appeared. **Right:** During the brain-controlled cursor task, participants had to vary how strongly they wiggled or rested their hand or foot to steer the cursor from the center of the workspace to the displayed target. In the example shown, the participant would have wiggled his hand and relaxed his foot to cause the cursor to move to the target.

when they needed to steer the cursor left; similarly they were instructed to wiggle their feet when they needed to steer the cursor down and rest their feet when they needed to steer the cursor up (Fig. 2). At the start of each session, the targets were very large so the participants did not have to move the cursor very far to hit the target. However, as performance improved, the task difficulty was increased by automatically reducing the target size.

Additional Post Hoc Analysis

Additional analyses were performed after the testing sessions were over to determine which electrode contacts were most useful for movement control and to determine if cursor control could have been improved had we eliminated nonuseful electrode contacts from the decoder. We also estimated how accurately 2D movement could have been controlled had we used 1 or 2 electrode leads passing through both the hand and foot cortical areas.

The usefulness of each individual electrode contact was determined by calculating how well the signal from that contact alone could be decoded into the movements needed to reach the target. Data from the brain-control task were used for building new decoding functions offline for each individual electrode. Individual electrode decoding functions were generated in a manner similar to the original full decoding function except that only power bands from 1 electrode contact at a time were included in the regression. Because the cursor was moving during the brain-control task, the regression was performed between the power from each individual contact and the movement needed at each time step to move the cursor to the target from its current position.

To ensure accurate assessment of the decoding performance of each individual electrode contact, a 10-fold cross-validation process was used in which 90% of the time points from the brain-control task formed the training data set that was used to build the individual decoding functions. The remaining 10% of the data (testing set) was used to determine the accuracy of the decoders by calculating the correlation between the actual movement needed at each time step and the movement predicted by

each contact's decoding function. This process was then repeated 10 times, each time using a different 10%/90% of the data for testing/training, respectively.

Not surprisingly, the only electrode contacts within the brain that were significantly modulated with hand and foot movements were those contacts directly in hand or foot motor and/or sensory cortex (pink and yellow circles in Fig. 1). Figure 1C and D illustrate these electrodes from a coronal view in which the circle size indicates how well each individual contact could decode hand (pink) or foot (yellow) movements.

The entire data set from the brain-control task was then used to build the new decoding functions that tested how well 2D movements could be controlled if only the 1 or 2 SEEG leads spanning the hand and foot areas were used for control. Those new refined decoders were then applied to the original baseline data set during which participants just wiggled or rested their hands and feet on cue. We chose to build our new decoding functions on the brain-control data set and apply it to the baseline data set for two reasons. First, the brain control data set was much larger, which should, in itself, improve the accuracy of the decoding functions built from it. Second, the baseline data set consisted of uniform segments of data in which participants were consistently performing the combinations of hand and foot wiggles and resting needed to move directly to each of the 4 targets. Therefore, perfectly decoded movement trajectories from the baseline data set should point directly toward the intended targets, thus making it easy to visualize and quantify the offline movement accuracy.

Results

Figure 3 shows the cursor trajectories generated when the patients used their decoded brain signals to control the cursor in real time (plots labeled "closed-loop") as well as additional theoretical trajectories calculated offline using the signals recorded on only 1 or 2 SEEG leads spanning the hand/foot cortex (plots labeled "open-loop"). The number of electrode contacts used for decoding in each plot is listed below the "closed-loop" and "open-loop" labels. Because it is difficult to clearly see all the overlapping trajectories, trajectories are also replotted in an exploded view underneath each primary plot. These exploded views separate out trajectories to each target, making it easier to see the spread and consistency of movements to each target. The black lines in the exploded views span the center starting position and the center of the outer target positions. Similarly the thicker, straight colored lines go from the center start position to the target centers in the normal trajectory plots. Therefore, perfectly decoded trajectories should be in line with each black line in the exploded view and in line with each thick, straight colored line in the standard view.

Although neither participant performed particularly well during closed-loop control, the exploded view plots show that trajectories to most targets at least went toward the correct half of the workspace and were therefore not random. Notice how Patient 1's trajectories were most accurate in the X direction on the first day and most accurate in the Y direction on the second day (Fig. 3). However, the

FIG. 3. Two-dimensional cursor trajectories generated during the brain-control task (closed-loop) and calculated offline from the baseline data using only 1 or 2 SEEG leads spanning hand and foot cortex (open-loop; 13 contacts for 2 leads and 6 or 7 contacts for 1 lead). Trajectories are color coded by their intended targets. *Black dots* in the closed-loop plots indicate when the correct target was hit. The upper row of plots for each participant shows the trajectories plotted normally. Those same trajectories are replotted directly underneath with the trajectories for each target separated for easier viewing. The rX and rY numbers listed below each plot indicate the correlation coefficients between the actual X and Y movements needed and the movement decoded from the neural signals at each time step. Elect. = electrode; M&S = motor and sensory.

offline trajectories generated from the 1 SEEG lead spanning the hand and foot regions were fairly good and fairly similar between days. These results suggest the inclusion of nonuseful electrode contacts during closed-loop control likely degraded the decoded movement command with unrelated noise. The decoders built on different days handled this unrelated noise differently, resulting in different problems with closed-loop control on different days.

Note that the exploded views of the closed-loop trajectories do not show any trajectories that make it out to the distal ends of the black lines, even though open-loop trajectories do tend to move outward around each black line. During closed-loop control, the movement trial ended as soon as the cursor hit the target. Therefore, closed-loop trajectories that successfully hit the target were trun-

cated at the target boundary. However, trajectories that missed the targets continued to grow until either the trial ended or the person was able to redirect the trajectory back to the target. This truncation of the good trajectories but not the bad trajectories should be kept in mind when assessing the spread of the closed-loop trajectories. For the closed-loop trajectories, the rX and rY values listed (Fig. 3) are the correlations between the movement vectors that were actually generated and the direction vectors needed to move the cursor toward the target at each time step. Because of this truncation of good movement trials, there was a disproportionately large number of time steps from missed trials compared with good trials in these closed-loop correlation calculations. For closed-loop control, these correlation values are not a true measure of

movement accuracy as in the open-loop plots, in which all data were included in the rX and rY calculations and no truncation took place. For the offline data, all correlations were significant at $p < 0.00001$.

From the exploded trajectory plots, one can see that most trajectories calculated offline generally went in the correct direction with some degree of spread. In most cases, the spread stayed within the correct quadrant of the workspace. Because those trajectories were generated offline, no corrective movements could be made to redirect the cursor back toward the hypothetical target. The offline decoding accuracy values observed here with single SEEG electrode leads are consistent with those from our EEG-BMI studies in which participants used 32 scalp EEG electrodes for similar open- and closed-loop 2D cursor control tasks. The closed-loop example given in Foldes and Taylor⁹ suggests that the quality of offline decoding noted in this study can result in reasonably functional 2D cursor movements when the person has visual feedback and can make corrective movements online if a trajectory starts to drift off course.

Patient 2 had 2 leads that spanned hand and foot cortex: 1 lead was in the motor area and 1 was in the sensory area. Trajectories were generated offline using just the single SEEG hand/foot motor lead (7 contacts), using just the single hand/foot sensory lead (6 working contacts), and then using both the motor and sensory leads together (13 contacts). While offline movement accuracy was better when both motor and sensory leads were used, the increase in performance was small over either one alone. For paralyzed individuals, motor areas may be more easily modulated with attempted movements than sensory areas, although both have been shown to be useful in EEG studies.

Discussion

Patients undergoing invasive electrode monitoring for medically intractable focal epilepsy offer a unique opportunity to assess the benefits and drawbacks of a variety of BMI systems. Because the majority of patients in the US undergoing invasive testing for epilepsy are implanted with subdural grid electrodes, there is a great deal of literature and research related to subdural grid-based BMI systems.^{1,2,6–8,10,11,19–29,35} On the other hand, there is a relative dearth of SEEG-related BMI studies.^{18,30}

Previous SEEG BMI studies have demonstrated the possibility of using SEEG electrodes in communication BMIs in which letters were selected by involuntary brain responses.^{18,30} This study, for the first time, demonstrated the use of SEEG electrodes for continuous 2D motion control using voluntary activity in hand and foot cortical areas. Trajectories generated during the real-time cursor control task were more accurate than one would expect by chance. However, further post hoc analysis suggested that better directional control could have been achieved by including only the electrode contacts that were directly in the hand and foot motor areas and excluding contacts conveying unrelated activity.

Most notably, the only electrode contacts within the brain that were significantly modulated with hand/foot

movements were all located on the same 1 or 2 SEEG leads, suggesting that BMI control of 2D movements could be achieved with electrode leads implanted through just 1 or 2 bur holes. Plots from Patient 2 in Fig. 3 showed the relative difference between using 1 versus 2 leads for control. Although there is some improvement with 2 electrodes, 1 lead may be adequate as BMI users learn to modulate their brain areas more effectively with practice.

The circle sizes in Fig. 1C and D show that 3–4 different SEEG contacts within each hand or foot cortical area were all generating similarly useful signals for control. These similarities suggest that the implantation depth within the various cortical layers does not have to be very precise because there is a large range in which electrode contacts can be placed for good BMI signals to be acquired. Furthermore, with BMIs that use scalp surface EEGs, the left and right foot movements cannot be differentiated because the current from both hemispheres blends together to form 1 combined generic “foot” signal at the top of the head. Stereoelectroencephalography electrodes could potentially be placed bilaterally, acquiring separate signals from both left versus right hand and left versus right foot regions. Bilateral implants could theoretically provide 4 relatively independent control signals for more complex BMI applications.

Conclusions

Stereoelectroencephalography electrodes for BMIs have many of the same pros and cons as when using SEEG electrodes in epilepsy monitoring. Stereoelectroencephalography electrodes allow us to reach deep structures, such as the foot motor cortex, which was shown to be useful for cursor control in this study. Stereoelectroencephalography electrodes also allowed us to reach both superficial and deep structures with the same lead and could be implanted percutaneously through small bur holes instead of the larger craniotomies required by subdural grid electrodes. This study demonstrated the feasibility of performing 2D continuous movement control using only a single SEEG electrode lead. These results warrant further exploration of SEEG electrode technology for BMI applications.

Disclosure

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