Clinical Neurophysiology xxx (2010) xxx–xxx

Contents lists available at ScienceDirect

Clinical Neurophysiology

journal homepage: www.elsevier.com/locate/clinph

Operation of a P300-based brain–computer interface by individuals with cervical spinal cord injury

Shiro Ikegami [a,b], Kouji Takano [a], Naokatsu Saeki [b], Kenji Kansaku [a,*]

[a] *Cognitive Functions Section, Department of Rehabilitation for Sensory Functions, Research Institute of National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Saitama 359-8555, Japan*

[b] *Department of Neurological Surgery, Chiba University Graduate School of Medicine, Chiba 260-8670, Japan*

ARTICLE INFO

Article history:
Accepted 23 August 2010
Available online xxxx

Keywords:
BCI
BMI
P300
Chromatic change
Cervical spinal cord injury

ABSTRACT

Objective: This study evaluates the efficacy of a P300-based brain–computer interface (BCI) with green/blue flicker matrices for individuals with cervical spinal cord injury (SCI).
Methods: Ten individuals with cervical SCI (age 26–53, all male) and 10 age- and sex-matched able-bodied controls (age 27–52, all male) with no prior BCI experience were asked to input hiragana (Japanese alphabet) characters using the P300 BCI with two distinct types of visual stimuli, white/gray and green/blue, in an 8×10 flicker matrix. Both online and offline performance were evaluated.
Results: The mean online accuracy of the SCI subjects was 88.0% for the white/gray and 90.7% for the green/blue flicker matrices. The accuracy of the control subjects was 77.3% and 86.0% for the white/gray and green/blue, respectively. There was a significant difference in online accuracy between the two types of flicker matrix. SCI subjects performed with greater accuracy than controls, but the main effect was not significant.
Conclusions: Individuals with cervical SCI successfully controlled the P300 BCI, and the green/blue flicker matrices were associated with significantly higher accuracy than the white/gray matrices.
Significance: The P300 BCI with the green/blue flicker matrices is effective for use not only in able-bodied subjects, but also in individuals with cervical SCI.

1. Introduction

Brain–computer interfaces (BCI) or Brain–machine interfaces (BMI) are devices that use neurophysiological brain signals to control external computers or machines (Wolpaw et al., 2002; Daly and Wolpaw, 2008; Kansaku, in press). One approach to using these devices, invasive BCI, relies on electrical signals recorded directly from the cortical surface (electrocorticograph; ECoG) or a single neuron (unit recording) (Kennedy et al., 2000; Leuthardt et al., 2004; Hochberg et al., 2006). The other approach, non-invasive BCI, uses electrical signals from the brain in the absence of surgery. The primary approach for non-invasive BCI is electroencephalography (EEG), where neurophysiological signals are recorded from an array of scalp electrodes.

Several types of electrical brain activity have been proposed for controlling EEG-based BCI. These include sensorimotor rhythm, slow cortical potential, steady state visual evoked potential, and P300 event-related potential. If the chosen communication system results in more than 70% correct responses, it has potential for practical use as a BCI system in people with disabilities (Sellers et al., 2006; Kübler and Birbaumer, 2008; Nijboer et al., 2008). Some BCI systems have already reached this level. Thus, BCI systems based on P300 signals were tested in patients with amyotrophic lateral sclerosis (ALS) and other diseases either in a laboratory setting (Piccione et al., 2006; Sellers and Donchin, 2006; Hoffmann et al., 2008) or the patient's home (Nijboer et al., 2008). The majority of subjects used in these studies were patients with ALS, and no studies have examined age- and sex-matched controls for comparison.

Our research group recently developed a BCI system for environmental control and communication (Komatsu et al., 2008; Kansaku et al., 2010), in which we applied several flicker panels that were modified from the "P300 speller" (Donchin et al., 2000), which uses P300-like evoked signals. We previously reported that a male volunteer quadriplegic SCI (C3/C4) patient successfully controlled our device without significant training (Komatsu et al., 2008). However, we sought to develop better visual stimuli because the white/gray flicker stimuli used could

* Corresponding author. Address: Cognitive Functions Section, Department of Rehabilitation for Sensory Functions, Research Institute of National Rehabilitation Center for Persons with Disabilities, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan. Tel.: +81 4 2995 3100x2573; fax: +81 4 2995 3132.
E-mail address: kansaku-kenji@rehab.go.jp (K. Kansaku).

1388-2457/$36.00 © 2010 International Federation of Clinical Neurophysiology. Published by Elsevier Ireland Ltd. All rights reserved.
doi:10.1016/j.clinph.2010.08.021

possibly induce discomfort or seizures, particularly in subjects with a history of epilepsy. Parra et al. evaluated the safety of chromatic combinations for those with photosensitive epilepsy (Parra et al., 2007). Five single-color stimuli (white, blue, red, yellow, and green) and four alternating-color stimuli (blue/red, red/green, green/blue, and blue/yellow with equal luminance) of four frequencies (10, 15, 20, and 30 Hz) were used as the visual stimuli. Under white stimulation, flickering stimuli with higher frequencies, especially those greater than 20 Hz, have been found to be potentially provocative. Under the alternating-color stimulation condition, as suggested by the Pokemon incidence, the 15-Hz blue/red flicker was the most provocative. It is noteworthy that the green/blue chromatic flicker emerged as the safest and evoked the lowest rates of EEG spikes. Accordingly, we used the green/blue chromatic combination for the visual stimuli used to elicit visually evoked responses (Takano et al., 2009b). We prepared a white/gray flicker matrix for the luminance flicker, a green/blue isoluminance flicker matrix for the chromatic flicker, and a green/blue luminance flicker for the luminance and chromatic flicker. We applied the experiments to the able-bodied subjects, and showed that accuracy rates were significantly higher in response to the luminance chromatic flicker condition than in response to the luminance or chromatic flicker condition. We also found that the green/blue luminance flicker matrices significantly improved subjective feelings of comfort in able-bodied subjects compared to the white/gray flicker matrices (Takano et al., 2009a).

This study focuses on the efficacy of the P300 BCI in individuals with chronic cervical SCI, who are potential users of the system (Daly and Wolpaw, 2008). We compared individuals with chronic cervical SCI and age- and sex-matched able-bodied controls. Subjects had no prior experience with the P300 BCI and were required to input hiragana (Japanese alphabet) characters using our P300 BCI system using either white/gray or green/blue luminance flicker matrices. We show that the P300 BCI with the green/blue flicker matrix is effective not only for use in able-bodied subjects, but also in individuals with cervical SCI.

2. Materials and methods

2.1. Subjects

Ten individuals with chronic cervical SCI (age 26–53, mean 41.9, all male) with no prior experience with BCI devices were recruited as participants. The mean time since SCI was 18.2 years (range, 5.5–29.2 years). Five individuals were diagnosed with complete tetraplegia according to the American Spinal Injury Association (ASIA) impairment scale (Maynard et al., 1997) (summarized in Table 1). All of the SCI patients had severe upper extremity dysfunction and needed the help of caregivers to use appliances for emailing and other tasks, and most needed Alternative Augmentative Communication (AAC) devices (e.g., mouth stick). All SCI subjects were outpatients and visited the laboratory in wheelchairs. In addition, 10 age- and sex-matched able-bodied controls (age 27–52, mean 42.1, all male) with no prior experience with BCI devices were recruited. This study was approved by the Institutional Review Board, and all subjects provided written informed consent according to institutional guidelines.

3. Experimental procedure

We modified the so-called P300 speller (Farwell and Donchin, 1988). The P300 speller uses the P300 paradigm and involves the presentation of a selection of icons arranged in a matrix. According to this protocol, the participant focuses on one icon in the matrix as the target, and each row/column, or a single icon of the matrix is then intensified in a random sequence. The target stimuli are presented as rare stimuli (i.e., the oddball paradigm). We elicited P300 responses to the target stimuli and then extracted and classified these responses with respect to the target.

All subjects sat approximately 100 cm away from a liquid crystal display that displayed a flicker matrix and input window (Fig. 1). SCI subjects used their own wheel chair, and control subjects sat in a desk chair. We prepared an 8×10 hiragana matrix for the P300 speller, modified from a 6×6 matrix using the English alphabet (Takano et al., 2009b). We used two types of intensification/rest flicker conditions, white/gray and green/blue. Luminance was measured using a chromatic meter (CS-200, Konica Minolta Sensing Inc., Osaka, Japan), and was 20 cd/cm(white)/6.5 cd/cm(gray), and 20 cd/cm(green)/6.5 cd/cm(blue), for each condition. The duration of intensification (green or white) was 100 ms, and that of rest (blue or gray) was 75 ms (Blankertz et al., 2006; Sellers et al., 2006). Each row and column of the matrix was intensified once per sequence in random order and, according to the P300 paradigm, the target stimuli were presented as rare stimuli (i.e., the oddball paradigm). One complete cycle of eight row and ten column intensifications constituted a sequence. Online performance was evaluated, and each letter was selected in a series of 10 sequences (180 intensifications for each hiragana character).

We first collected EEG data to derive feature vectors for the subsequent test session. All subjects were instructed to attend to six successive letters of the matrix under each condition (training session). In the test session, using the feature vectors, all subjects were required to input 15 letters from the 8×10 hiragana matrix under each flicker condition. The order of the experimental conditions (white/gray or green/blue flicker matrix) was counterbalanced between subjects.

3.1. EEG recording and analysis

Eight-channel (Fz, Cz, Pz, P3, P4, Oz, PO7, and PO8) EEG data were recorded with a g-Tec cap and g.USBamp acquisition system (Guger Technologies OEG, Graz, Austria) (Krusienski et al., 2008;

Table 1
Summary of spinal cord injury subjects.

	Age	Sex	Level of SCI at injury	Time since injury (years)	ASIA impairment Scale
SCI	37	M	C3/4	16.3	Incomplete
	45	M	C2/3	5.5	Complete
	43	M	C5/6	25.3	Complete
	40	M	C4/5	15.9	Incomplete
	42	M	C4/5	10.4	Complete
	37	M	C3/4	20.5	Incomplete
	48	M	C4/5	27.1	Complete
	48	M	C5/6	21.8	Incomplete
	26	M	C4/5	9.9	Complete
	53	M	C5/6	29.2	Incomplete
Mean	41.9			18.2	

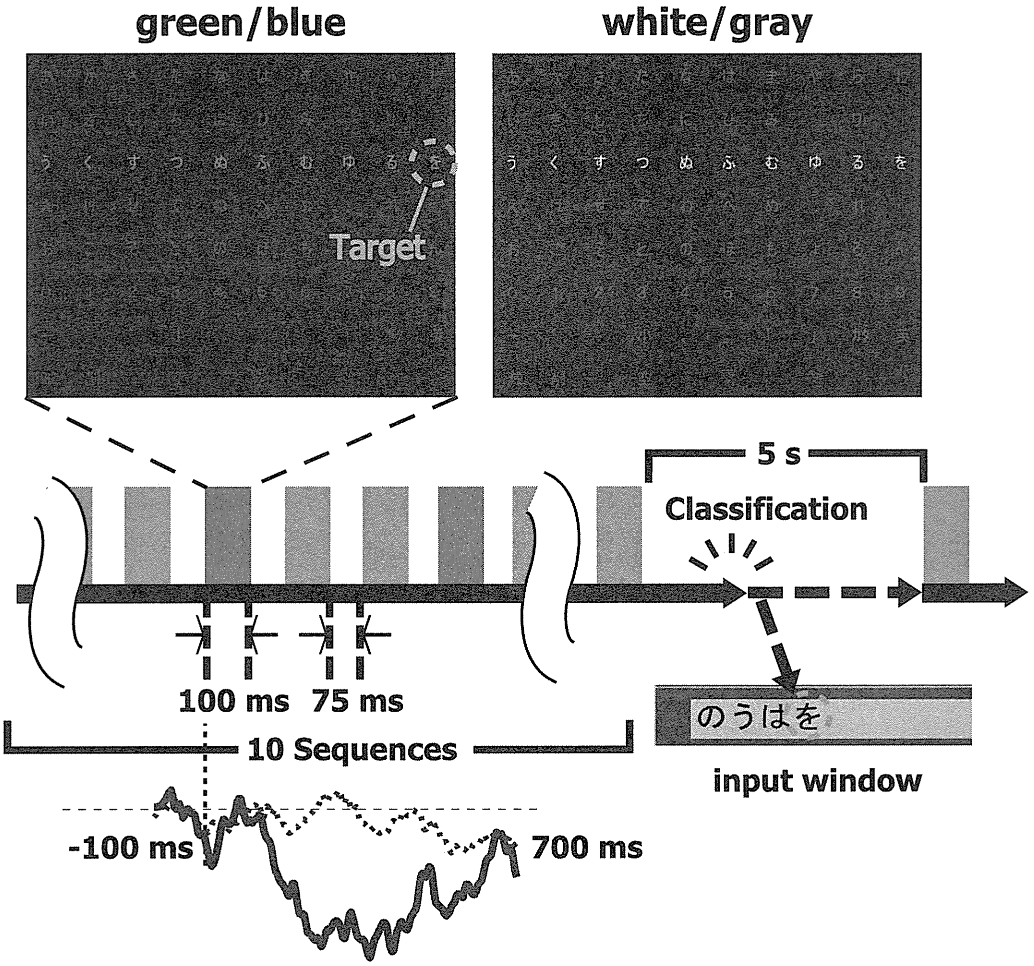

Fig. 1. Task timing for hiragana spelling. Two types of matrix were presented (white/gray and green/blue). The stimulus onset asynchrony was 175 ms, consisting of 100 ms intensification and 75 ms rest. EEG data were collected and used for classification over 10 sequences (180 intensifications). Averaged ERP data (Pz) of the SCI group for the green/blue condition are shown in red (target: thick line, non-target: dotted line).

Takano et al., 2009b). EEG signals were band-pass filtered (0.1–50 Hz), digitized at 256 Hz, and stored. All channels were referenced to Fpz and grounded to AFz. Recorded EEG data were down-sampled to 21 Hz for analysis. A total of 800 ms of EEG data were segmented according to the timing of flash onset. The first 100 ms, occurring just prior to flash onset, was used for baseline correction, and the remaining 700 ms was used for classification. In the training session, feature vectors were derived for each condition (white/gray and green/blue). During the test session, using these feature vectors, target and non-target characters were discriminated using Fisher's linear discriminant analysis. The result of this classification, as the maximum of the summed scores for the each row and column, was used to determine the icon to which the subjects were attending. The intersection of the calculated row and column was regarded as the target.

During online performance, the percentage of characters entered correctly was defined as the classification accuracy and was also translated into bit rate (Wolpaw et al., 2002). Correlations between SCI subjects' accuracy and their demographic characteristics (age, time since injury, ASIA impairment scale score) were evaluated using Spearman's rank correlation coefficient. The effects of patient group (SCI vs. control) and type of flicker matrix (white/gray vs. green/blue) on online accuracy were examined using two-way repeated-measure analysis of valiance (ANOVA).

For offline analysis, the accuracy for each sequence was calculated. The effects of subject group (SCI vs. control), type of flicker matrix (white/gray vs. green/blue), and sequence on accuracy in each sequence were evaluated by three-way repeated-measure ANOVA followed by post hoc paired t-tests with Bonferroni correction.

4. Results

4.1. Online performance

All subjects completed the 15-letter spelling task in both the white/gray and green/blue conditions. The mean online accuracy of all subjects was 82.7% for the white/gray condition and 88.3% for the green/blue condition. Under the white/gray condition, the mean accuracy was 77.3% for the control group and 88.0% for the SCI group. Under the green/blue condition, the mean accuracy was 86.0% and 90.7% for control and SCI groups, respectively (Fig. 2). For the SCI group, the mean bit rates (Wolpaw et al., 2002) were 9.8 bit/min and 10.2 bit/min under the white/gray and green/blue conditions, respectively (Table 2). Note that the time interval between character selections was not included for the bit rate calculation. The mean bit rates for controls were 8.4 bit/min and 9.6 bit/min for the white/gray and green/blue conditions, respectively. No significant correlations were observed between the accuracy or bit rate of SCI subjects and demographic characteristics (age, time since injury, ASIA impairment scale score; Spearman's rank correlation coefficient, $p > 0.05$).

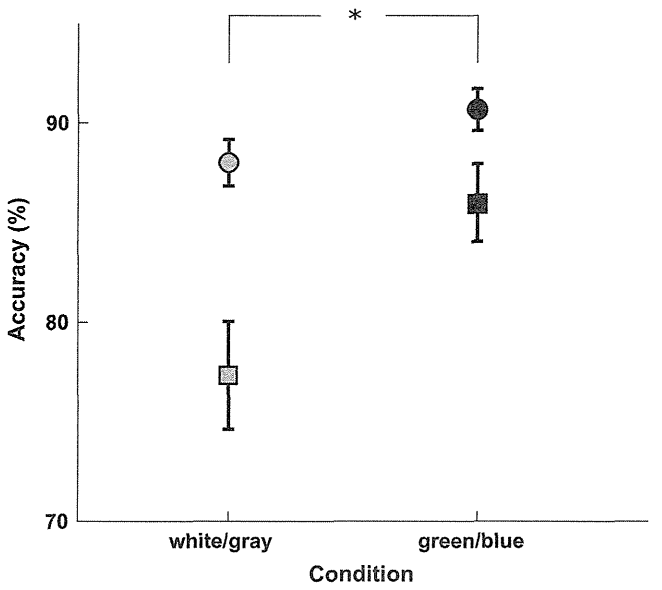

Fig. 2. Online accuracy of subject groups (SCI and controls) for each condition (white/gray and green/blue). Squares, control group; circles, SCI group. Error bars indicate S.E.M.

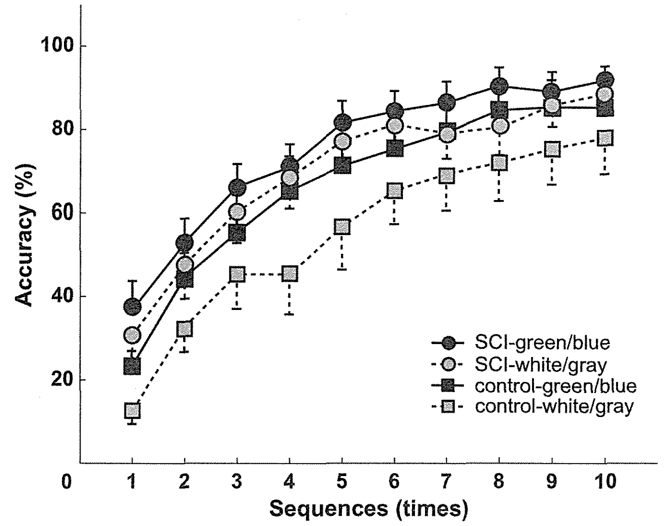

Fig. 3. Offline evaluation for each sequence. Mean accuracy of the control group and SCI group are plotted using squares and circles; the dotted line indicates white/gray, and the solid line indicate green/blue conditions. Accuracy in the first through seventh sequences was significantly lower than that in the tenth sequence, as revealed by post hoc testing ($p < 0.05$, Bonferroni correction). Error bars indicate S.E.M.

Table 2
Offline accuracy, bit rate and letter/min during the tenth, eighth, and fifth sequences in SCI subjects.

Sequence (times)	White/gray		Green/blue	
	Accuracy (%)	bit/min	Accuracy (%)	bit/min
10	88.0	9.8	90.7	10.2
8	80.4	10.9	90.4	12.8
5	77.2	16.2	81.7	17.5

We used a two-way repeated-measure ANOVA to examine the effects of group (SCI vs. control) and of condition (white/gray vs. green/blue) on online accuracy. ANOVA revealed a main effect of flicker matrix condition ($F(1,9) = 5.2$, $p < 0.05$). A trend toward greater accuracy in the SCI group compared to controls was observed; however, no main effect of group ($F(1,9) = 1.2$, $p = 0.30$) and no significant interaction ($F(1,9) = 0.61$, $p = 0.45$) was found. These results did not basically change if bit rate was substituted for accuracy.

4.2. Offline evaluation

Fig. 3 shows the results of the offline analysis of subject groups for each condition. We conducted a three-way repeated-measure ANOVA with group (SCI vs. controls), condition (white/gray vs. green/blue), and sequence number (1–10) as factors. Main effects of condition ($F(1,9) = 9.4$, $p < 0.05$) and sequence ($F(9,81) = 93.2$, $p < 0.001$) were significant, but no main effect of group ($F(1,9) = 1.9$, $p = 0.20$) and no significant interaction ($F(9,81) = 0.89$, $p = 0.54$) was found. Thus, the P300 BCI with the green/blue flicker matrix is effective not only in able-bodied subjects but also in individuals with cervical SCI. Accuracy in the first through seventh sequences was significantly lower than that in the tenth sequence, as revealed by post hoc testing ($p < 0.05$, Bonferroni correction).

In the SCI group, the mean online bit rate was 9.8 bit/min and 10.2 bit/min for the white/gray and green/blue conditions, respectively, as calculated from the tenth sequence accuracy (Table 2). In the fifth sequence, the mean accuracy of the SCI group exceeded 70% under both conditions (77.2% for white/gray, 81.7% for green/blue), and the mean bit rate was 16.2 bit/min and 17.5 bit/min for the white/gray and green/blue conditions, respectively (Table 2). In the fifth sequence, the bit rate was significantly higher than in the tenth sequence, but accuracy was significantly lower under both conditions (paired t-test, $p < 0.05$). By contrast, in the eighth sequence, accuracy was not significantly different from that in the tenth sequence (80.4% for white/gray, 90.4% for green/blue), and the bit rate was 10.9 bit/min and 12.8 bit/min for white/gray and green/blue, respectively. This bit rate was significantly greater than that in the tenth sequence for green/blue (paired t-test, $p < 0.001$), but not for white/gray (paired t-test, $p > 0.05$). Thus, the green/blue flicker matrix was more effective than the white/gray flicker matrix by the eighth sequence.

5. Discussion

We investigated the accuracy of P300-based BCI performance in individuals with chronic cervical SCI using white/gray and green/blue flicker matrices. SCI patients successfully controlled our BCI system without significant training, and the green/blue flicker matrix provided higher accuracy than the white/gray matrix.

5.1. Effect of the color combination in P300 BCI

A number of studies have attempted to increase P300 BCI performance accuracy, primarily by examining classification methods (Donchin et al., 2000; Kaper et al., 2004; Krusienski et al., 2006; Bashashati et al., 2007; Hoffmann et al., 2008). Other studies have examined modifying matrix size and inter-stimulus intervals (Sellers et al., 2006), type of flash (Guger et al., 2009; Townsend et al., 2010) and background colors (Salvaris and Sepulveda, 2009). We recently reported that a green/blue luminance and chromatic flicker matrix provided higher accuracy than a white/gray luminance flicker matrix and a green/blue isoluminance flicker matrix (Takano et al., 2009b). In the present study, the mean accuracy among both able-bodied and cervical SCI subjects was significantly higher under the green/blue condition than under the white/gray condition. No accuracy difference between SCI and able-bodied groups was found.

Online performance of the SCI group reached 90% accuracy and a bit rate of 10.2 bit/min (1.9 letter/min) under the green/blue condition, comparable to previous reports studying disabled subjects (Piccione et al., 2006; Sellers and Donchin, 2006; Hoffmann et al., 2008; Nijboer et al., 2008). This performance is thought to be sufficient for satisfactory use of a BCI, which requires greater than 70% accuracy (Sellers et al., 2006; Kübler and Birbaumer, 2008; Nijboer et al., 2008). Offline analysis showed that number of sequences can be reduced from 10 to 8 while preserving accuracy and significantly increasing the bit rate. This effect was not apparent under the white/gray condition. Thus, the green/blue flicker matrix was more effective for fast communication.

5.2. BCI performance in SCI subjects

P300-based BCI has been examined in SCI subjects in two previous reports of one cervical SCI patient each (Piccione et al., 2006; Hoffmann et al., 2008). One patient controlled a four-choice P300 BCI with an online accuracy of 75.7% (Piccione et al., 2006), and the other controlled a six-choice P300 BCI with an offline accuracy of 100% (Hoffmann et al., 2008). The main BCI method used with SCI subjects is sensorimotor rhythm (SMR) for binary choice (Pfurtscheller et al., 2000; Krausz et al., 2003; McFarland et al., 2005; Kauhanen et al., 2007; Kübler and Birbaumer, 2008). Kauhanen et al. (2007) reported that the mean online accuracy for binary-choice SMR BCI with five cervical SCI subjects was 48%.

Although the brain remains intact in SCI subjects, the deafferentation of sensory input that occurs after SCI can result in brain reorganization and altered scalp EEG activity compared with able-bodied controls (Green et al., 1998; Tran et al., 2004; Herbert et al., 2007). Accordingly, SMR BCI, which uses beta or mu waves from sensory motor areas, would be more affected by this reorganization. Indeed, Kauhanen et al. (2007) reported that the binary-choice SMR BCI performance of five cervical SCI subjects was worse than that of able-bodied subjects (not matched for age and sex). In the present study, individuals with cervical SCI controlled the P300 BCI with similar accuracy to able-bodied individuals. Although the data are limited, the P300 BCI may be easier for SCI subjects to use.

5.3. Toward clinical applications

For practical use of the P300 BCI, the system has to be accurate, fast, and reliable. We used 10 sequences for EEG data acquisition for online analyses, but offline analyses showed that the green/blue flicker matrix was more effective than the white/gray flicker matrix by the eighth sequence. Further reducing the number of sequences to five still provided greater than 70% accuracy with a higher bit rate. The mean accuracy at the fifth sequence became lower than that at the tenth sequence, so if the users needed to complete their sentences by correcting misspelled characters, it would take a longer time (Townsend et al., 2010). The sequence times may be determined by individual user preference, as some prefer to control devices quickly with lower fidelity, whereas others prefer to communicate precisely and more slowly (Sellers and Donchin, 2006).

The severity of the patient impairment may also have implications for practical BMI use. Kübler and Birbaumer (2008) reviewed a number of BCI studies using P300, SMR, and slow cortical potential (SCP) and reported a relationship between physical impairment [subdivided into minor, moderate, major, locked-in state and complete locked-in state (CLIS)] and BCI performance. When they included CLIS patients, they found a strong correlation between impairment and BCI performance; however, after removing the CLIS patients, the correlation disappeared. Nijboer investigated the efficacy of a P300 BCI in eight advanced ALS patients (Nijboer et al., 2008) and showed that online BCI performance was not correlated with the degree of disability according to the ALS Functional Rating Scale (Cedarbaum and Stambler, 1997). Thus, for patients with ALS, it is suggested that BCI be applied before the onset of CLIS (Birbaumer, 2006; Kübler and Birbaumer, 2008). In the present study, we found no correlation between performance and ASIA impairment scale score (complete or incomplete) in SCI patients, nor did we observe a correlation between performance and time since injury. We previously reported that the BMI performance of subacute SCI subjects, whose time since injury was less than a year, was worse than that of chronic SCI subjects (Ikegami et al., 2009). Further investigation is required to determine the optimal time for applying BCI to individuals with SCI.

In conclusion, the P300 BCI system for environmental control and communication with a green/blue flicker matrix provided better accuracy than that with a white/gray flicker in individuals with cervical SCI, and future studies may aid the development of practical BCI for these individuals to expand their range of activity and communication.

Acknowledgments

This study was partially supported by a Grant-in-Aid from the Ministry of Health, Labour and Welfare of Japan. We thank Dr. T. Komatsu and Dr. T. Shimotomai for their help, and Dr. Y. Nakajima for his continuous encouragement.

References

Bashashati A, Fatourechi M, Ward RK, Birch GE. A survey of signal processing algorithms in brain–computer interfaces based on electrical brain signals. J Neural Eng 2007;4:R32–57.

Birbaumer N. Breaking the silence: brain–computer interfaces (BCI) for communication and motor control. Psychophysiology 2006;43:517–32.

Blankertz B, Müller KR, Krusienski DJ, Schalk G, Wolpaw JR, Schlögl A, et al. The BCI competition III: validating alternative approaches to actual BCI problems. IEEE Trans Neural Syst Rehabil Eng 2006;14:153–9.

Cedarbaum JM, Stambler N. Performance of the amyotrophic lateral sclerosis functional rating scale (ALSFRS) in multicenter clinical trials. J Neurol Sci 1997;152(Suppl. 1):S1–9.

Daly JJ, Wolpaw JR. Brain–computer interfaces in neurological rehabilitation. Lancet Neurol 2008;7:1032–43.

Donchin E, Spencer KM, Wijesinghe R. The mental prosthesis: assessing the speed of a P300-based brain–computer interface. IEEE Trans Rehabil Eng 2000;8:174–9.

Farwell LA, Donchin E. Talking off the top of your head: toward a mental prosthesis utilizing event-related brain potentials. Electroencephalogr Clin Neurophysiol 1988;70:510–23.

Green JB, Sora E, Bialy Y, Ricamato A, Thatcher RW. Cortical sensorimotor reorganization after spinal cord injury: an electroencephalographic study. Neurology 1998;50:1115–21.

Guger C, Daban S, Sellers E, Holzner C, Krausz G, Carabalona R, Gramatica F, Edlinger G. How many people are able to control a P300-based brain-computer interface (BCI)? Neurosci Lett 2009;462:94–8.

Herbert D, Tran Y, Craig A, Boord P, Middleton J, Siddall P. Altered brain wave activity in persons with chronic spinal cord injury. Int J Neurosci 2007;117:1731–46.

Hochberg LR, Serruya MD, Friehs GM, Mukand JA, Saleh M, Caplan AH, et al. Neuronal ensemble control of prosthetic devices by a human with tetraplegia. Nature 2006;442:164–71.

Hoffmann U, Vesin JM, Ebrahimi T, Diserens K. An efficient P300-based brain–computer interface for disabled subjects. J Neurosci Methods 2008;167:115–25.

Ikegami S, Takano K, Komatsu T, Saeki N, Kansaku K. Operation of a BMI based environmental control system by patients with cervical spinal cord injury. Program No. 664.16. 2009 Abstract Viewer/Itinerary Planner. Chicago: Society for Neuroscience, 2009. [Online].

Kansaku K. The intelligent environment: brain–machine interfaces for environmental control. In: Ferguson-Pell M, Stefanov D, editors. Smart houses: advanced technology for living independently. Springer Verlag (in press).

Kansaku K, Hata N, Takano K. My thoughts through a robot's eyes: an augmented reality-brain–machine interface. Neurosci Res 2010;66:219–22.

Kaper M, Meinicke P, Grossekathoefer U, Lingner T, Ritter H. BCI competition 2003–data set IIb: support vector machines for the P300 speller paradigm. IEEE Trans Biomed Eng 2004;51:1073–6.

Kauhanen L, Jylänki P, Lehtonen J, Rantanen P, Alaranta H, Sams M. EEG-based brain–computer interface for tetraplegics. Comput Intell Neurosci 2007:23864.

Kennedy PR, Bakay RAE, Moore MM, Adams K, Goldwaithe J. Direct control of a computer from the human central. IEEE Trans Rehabil Eng 2000;8: 198–202.

Komatsu T, Hata N, Nakajima Y, Kansaku K. A non-training EEG-based BMI system for environmental control. Neurosci Res 2008;61(Suppl. 1):S251.

Krausz G, Scherer R, Korisek G, Pfurtscheller G. Critical decision-speed and information transfer in the "Graz Brain-Computer Interface". Appl Psychophysiol Biofeedback 2003;28:233–40.

Krusienski DJ, Sellers EW, Cabestaing F, Bayoudh S, McFarland DJ, Vaughan TM, Wolpaw JR. A comparison of classification techniques for the P300 Speller. J Neural Eng 2006;3:299–305.

Krusienski DJ, Sellers EW, McFarland DJ, Vaughan TM, Wolpaw JR. Toward enhanced P300 speller performance. J Neurosci Methods 2008;167:15–21.

Kübler A, Birbaumer N. Brain–computer interfaces and communication in paralysis: extinction of goal directed thinking in completely paralysed patients? Clin Neurophysiol 2008;119:2658–66.

Leuthardt EC, Schalk G, Wolpaw JR, Ojemann JG, Moran DW. A brain–computer interface using electrocorticographic signals in humans. J Neural Eng 2004;1:63–71.

Maynard Jr FM, Bracken MB, Creasey G, Ditunno Jr JF, Donovan WH, Ducker TB, et al. International standards for neurological and functional classification of spinal cord injury. American Spinal Injury Association. Spinal Cord 1997;35:266–74.

McFarland DJ, Sarnacki WA, Vaughan TM, Wolpaw JR. Brain–computer interface (BCI) operation: signal and noise during early training sessions. Clin Neurophysiol 2005;116:56–62.

Nijboer F, Sellers EW, Mellinger J, Jordan MA, Matuz T, Furdea A, et al. A P300-based brain–computer interface for people with amyotrophic lateral sclerosis. Clin Neurophysiol 2008;119:1909–16.

Parra J, Lopes da Silva FH, Stroink H, Kalitzin S. Is colour modulation an independent factor in human visual photosensitivity? Brain 2007;130:1679–89.

Pfurtscheller G, Guger C, Müller G, Krausz G, Neuper C. Brain oscillations control hand orthosis in a tetraplegic. Neurosci Lett 2000;292:211–4.

Piccione F, Giorgi F, Tonin P, Priftis K, Giove S, Silvoni S, et al. P300-based brain–computer interface. reliability and performance in healthy and paralysed participants. Clin Neurophysiol 2006;117:531–7.

Salvaris M, Sepulveda F. Visual modifications on the P300 speller BCI paradigm. J Neural Eng 2009;6:046011.

Sellers EW, Donchin E. A P300-based brain–computer interface. initial tests by ALS patients. Clin Neurophysiol 2006;117:538–48.

Sellers EW, Krusienski DJ, McFarland DJ, Vaughan TM, Wolpaw JR. A P300 event-related potential brain–computer interface (BCI): the effects of matrix size and inter-stimulus interval on performance. Biol Psychol 2006;73:242–52.

Takano K, Ikegami S, Komatsu T, Kansaku K. Green/blue flicker matrices for the P300 BCI improve the subjective feeling of comfort. Neurosci Res 2009;65(Suppl. 1):S182.

Takano K, Komatsu T, Hata N, Nakajima Y, Kansaku K. Visual stimuli for the P300 brain–computer interface. a comparison of white/gray and green/blue flicker matrices. Clin Neurophysiol 2009b;120:1562–6.

Townsend G, Lapallo BK, Boulay CB, Krusienski DJ, Frye GE, Hauser CK, et al. A novel P300-based brain–computer interface stimulus presentation paradigm: moving beyond rows and columns. Clin Neurophysiol 2010;121:1109–20.

Tran Y, Boord P, Middleton J, Craig A. Levels of brain wave activity (8–13 Hz) in persons with spinal cord injury. Spinal cord 2004;42:73–9.

Wolpaw JR, Birbaumer N, McFarland DJ, Pfurtscheller G, Vaughan TM. Brain–computer interfaces for communication and control. Clin Neurophysiol 2002;113:767–91.

Please cite this article in press as: Ikegami S et al. Operation of a P300-based brain–computer interface by individuals with cervical spinal cord injury. Clin Neurophysiol (2010), doi:10.1016/j.clinph.2010.08.021

ORIGINAL RESEARCH ARTICLE
published: 18 July 2012
doi: 10.3389/fneur.2012.00114

A non-adhesive solid-gel electrode for a non-invasive brain–machine interface

Shigeru Toyama[1]*, Kouji Takano[2] and Kenji Kansaku[2]*

[1] Biotechnological Rehabilitation Section, Department of Rehabilitation Engineering, Research Institute of National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Japan

[2] Systems Neuroscience Section, Department of Rehabilitation for Brain Functions, Research Institute of National Rehabilitation Center for Persons with Disabilities, Tokorozawa, Japan

Edited by:
Emanuel Donchin, University of South Florida, USA

Reviewed by:
Eric W. Sellers, East Tennessee State University, USA
Christopher Leroy Frewin, University of South Florida, USA

****Correspondence:***
Shigeru Toyama, Biotechnological Rehabilitation Section, Department of Rehabilitation Engineering, Research Institute of National Rehabilitation Center for Persons with Disabilities, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan.
e-mail: toyama-shigeru@rehab.go.jp;
Kenji Kansaku, Systems Neuroscience Section, Department of Rehabilitation for Brain Functions, Research Institute of National Rehabilitation Center for Persons with Disabilities, 4-1 Namiki, Tokorozawa, Saitama 359-8555, Japan.
e-mail: kansaku-kenji@rehab.go.jp

A non-invasive brain–machine interface (BMI) or brain–computer interface is a technology for helping individuals with disabilities and utilizes neurophysiological signals from the brain to control external machines or computers without requiring surgery. However, when applying electroencephalography (EEG) methodology, users must place EEG electrodes on the scalp each time, and the development of easy-to-use electrodes for clinical use is required. In this study, we developed a conductive non-adhesive solid-gel electrode for practical non-invasive BMIs. We performed basic material testing, including examining the volume resistivity, viscoelasticity, and moisture-retention properties of the solid-gel. Then, we compared the performance of the solid-gel, a conventional paste, and an in-house metal-pin-based electrode using impedance measurements and P300-BMI testing. The solid-gel was observed to be conductive (volume resistivity 13.2 Ωcm) and soft (complex modulus 105.4 kPa), and it remained wet for a prolonged period (>10 h) in a dry environment. Impedance measurements revealed that the impedance of the solid-gel-based and conventional paste-based electrodes was superior to that of the pin-based electrode. The EEG measurement suggested that the signals obtained with the solid-gel electrode were comparable to those with the conventional paste-based electrode. Moreover, the P300-BMI study suggested that systems using the solid-gel or pin-based electrodes were effective. One of the advantages of the solid-gel is that it does not require cleaning after use, whereas the conventional paste adheres to the hair, which requires washing. Furthermore, the solid-gel electrode was not painful compared with a metal-pin electrode. Taken together, the results suggest that the solid-gel electrode worked well for practical BMIs and could be useful for bedridden patients such as those with amyotrophic lateral sclerosis.

Keywords: EEG, BMI, BCI, non-adhesive conductive solid-gel

INTRODUCTION

The brain–machine interface (BMI) or brain–computer interface (BCI) is a state-of-the art technology that utilizes neurophysiological signals from the brain to control external machines or computers (Wolpaw and McFarland, 2004; Pfurtscheller et al., 2006; Birbaumer and Cohen, 2007). Much effort has been made to help individuals with physical disabilities such as amyotrophic lateral sclerosis (ALS), stroke, or upper cervical spinal cord injury. Non-invasive BMI does not require surgery. Some non-invasive BMI systems make use of hemodynamic signals using functional magnetic resonance imaging (fMRI; Sitaram et al., 2011) or near-infrared spectroscopy (NIRS; Sitaram et al., 2007; Cui et al., 2010), but the majority of papers report methods using electroencephalography (EEG) signals. The P300 speller, a popular BMI system, uses elicited P300 responses to target stimuli placed among row and column flashes (Farwell and Donchin, 1988). We also used EEG signals in a BMI system that enables environmental control and communication using the P300 paradigm (Takano et al., 2009, 2011; Kansaku et al., 2010). Recent studies have evaluated the use of systems relying on these sensory evoked signals of patients with ALS and other diseases (Piccione et al., 2006; Sellers and Donchin, 2006; Ikegami et al., 2011).

Based on BMI studies, it is possible to build an intelligent house in which home electronics or communication tools such as e-mail can be operated with brain waves (Kansaku, 2011). Therefore, we have conducted measurement tests in patient's homes and in hospitals. However, we felt that the conventional paste-based electrode systems are somewhat awkward to use in practical circumstances (here we use "paste" to mean the conventional paste or gel used for an EEG electrode).

Typical EEG-based BMI electrodes use sticky conductive paste to reduce the impedance between the scalp and electrodes. To use electrodes with paste requires elaborate work not only for preparation but also for removal of the paste. In the preparation stage, paste is first placed on cup electrodes, and then the electrodes are fixed on the head. After using a BMI system, the patient's head must often be washed to remove the dried paste, which adheres tightly to hair. Similar comments are sometimes found in the literature, and some authors have proposed the use of dry electrodes for BMI to avoid tedious preparation and after treatment (Popescu et al., 2007; Liao et al., 2011; Zander et al., 2011). Popescu et al. (2007) prepared a cap system equipped only with six dry electrodes and a dry reference electrode. The information-transmission rate of their new system was only 30.8% slower than that of their previous experiments using caps with 64 wet electrodes on the same participants. They stated that the advantages of their system were the ease of preparation for EEG measurements and the long-term monitoring capability.

Various types of dry electrodes for EEG measurements have been developed, and one type is a bundle of microneedles (Griss et al., 2001; Chiou et al., 2006; Ruffini et al., 2006; Ng et al., 2009). In this case, microneedles pierce the outer skin layer (*Stratum corneum*), which has high impedance characteristics, and contact the *S. germinativum* layer where living cells exist and are electrically conductive. Therefore, using these electrodes does not require skin preparation. Griss et al. developed such an electrode with a spiked silicon electrode array using a microfabrication technique. Furthermore, Ruffini et al. developed an electrode with an array of multi-walled carbon nanotubes. Although they reported that their electrode was useful, the microneedles must be designed carefully to prevent cell damage.

Conductive textile-based electrodes have been developed for ECG monitoring (Hoffmann and Ruff, 2007; Beckmann et al., 2010). They are comfortable and also durable for long-term use because of their softness. However, these textile-based electrodes are hard to use on hairy sites. Above all, Lin et al. (2011) developed an EEG electrode made of urethane foam covered with a conductive polymer textile. They reported that their electrode can be used on a hairy site.

A popular dry electrode is the pin-based type, which has a metal-pin that contacts the skin. Zander et al. (2011) used a head cap with three pin-based electrodes, a reference electrode, and a ground electrode for their BMI experiments. They concluded that their dry electrode system showed no degradation in EEG and BMI performance in most cases. Liao et al. (2011) also used a metal-pin electrode system. They obtained similar brain waves from both dry and wet electrodes in simultaneous measurements. Sellers et al. (2009) compared pin-based and paste-based electrodes by simultaneously mounting these electrodes on a specially developed headpiece. The EEG signals of these two types of electrodes were almost identical, and the BMI classification accuracy was also almost identical while performing a copy-spelling task. Although these dry electrodes have been reported to be useful, we are concerned that they may cause pain or may injure the surface of the patient's head when hard solid electrodes are used, particularly when the patient is lying on a bed with his/her head resting on the pin electrodes.

We have developed several types of electrode for BMI (Toyama et al., 2011a,b). In this study, we prepared a specially designed conductive solid-gel as an electrode material for BMI. The solid-gel retains moisture and is not sticky like paste (here we use "gel" as a chemically defined term). We selected the ingredients carefully to produce a moisture-retaining solid-gel. The solid-gel contained a liquid comprising carboxymethylcellulose (CMC), $CaCl_2$, glycerol, and pure water. The superior wettability of our solid-gel was attributed to the nature of the ingredients. $CaCl_2$ and glycerol are water-absorbing materials. The weight of these materials increases when they are exposed to a normal room environment. Due to an electrochemical reaction, a combination of Ag/AgCl electrodes and KCl-containing liquid or solid-gel is usually recommended as an electrode material to measure biopotentials. The potential between the electrode and solution in contact with that electrode is constant for a long period. By contrast, the electrode signal is apt to drift in the long-term when we use non-recommended materials, such as $CaCl_2$. However, this problem has essentially been alleviated with modern circuit technology. Today, we have extremely high-precision analog-to-digital converters (ADCs) for medical research and clinical measurement systems (Aksenov et al., 2001). For example, when a 24-bit ADC is used to capture the voltage signal, a signal with $0.12\,\mu V$ precision and $\pm 1\,V$ full range can be obtained, which is sufficient to cover the drift due to the electrochemical reaction at the electrode surface. Therefore, operators of the measurement system can use digital filters to subsequently obtain brain waves.

Here, we examined the usefulness of a new system with the in-house conductive solid-gel-based electrode by comparing it with in-house metal-pin-based electrodes and conventional paste-based electrodes.

MATERIALS AND METHODS

SOLID-GEL ELECTRODE

The solid-gel chip (**Figure 1**) was made of CMC sodium salt (MW, 700 kDa; Sigma Chemical, St. Louis, MO, USA), calcium chloride dihydrate (Wako Pure Chemical Industries, Ltd., Osaka, Japan), glycerol (Nakalai Chemicals, Kyoto, Japan), and pure water. The solid-gel contained 10.9, 38.0, 7.6, and 43.4% of these compounds by weight, respectively.

METAL-PIN ELECTRODE

The structure of the metal-pin electrode is shown in **Figure 2**. Seven metal-pins extended from one side of a cylindrical main body, which was made of epoxy resin. Each metal-pin was composed of a center rod, an outer sheath, and a spring. The rod could be forced inward under pressure, so it was harmless to the scalp. The effective area of the rod tip was enlarged by sandblasting to decrease the contact impedance between the rod and the skin. Additionally, a wire was fixed at the end of the rod instead of at the outer sheath to minimize the friction noise caused by sliding between the center rod and the outer sheath.

COMMERCIALIZED PASTE

In this experiment, we also used a conventional paste (ABRALYT 2000; abrasive electrolyte-gel; EASY CAP, Munich, Germany) as a control material to compare its usefulness.

HEAD-MOUNTED CAP

We developed an original head-mounted cap with 12 electrode sockets (**Figure 1**) for use with solid-gel electrodes, metal-pin electrodes, and commercial paste. Each electrode socket had a silicone-based grommet equipped with a through-hole where the solid-gel-based, metal-pin, and paste electrodes were inserted. During EEG measurements, one electrode was used as a reference. The diameter and depth of the through-hole were 15 and 8.5 mm, respectively. When using a solid-gel chip, the chip was inserted

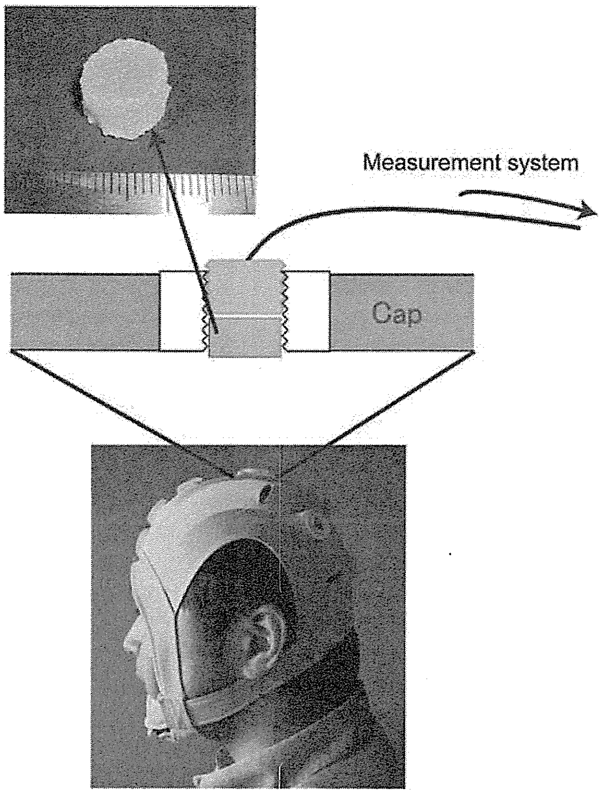

FIGURE 1 | Photograph of the originally developed head-mounted cap, which has 12 electrode sockets. One of the electrode sockets on the cap is enlarged to explain how the solid-gel is used. The solid-gel chip is loaded in the through-hole; the hole is then plugged with an electrode plate. The diameter of the chip was approximately 12 mm, and the height was approximately 7 mm.

into the through-hole of the grommet, which was plugged with an electrode plate. The body of the electrode plate was made of epoxy resin and was cylindrical in shape, and its outer wall had concentric grooves. The inner side of the through-holes of the cap had concentric grooves with the same pitch as that of the plate, so that the plate was held tightly in the through-hole. A flat Ag/AgCl plate was attached to the bottom of the plate, and an electric wire protruded from the rim of the top. The electric wire was solder-mounted on the Ag/AgCl plate at the inside of the plate and penetrated it. Wire terminals were connected to the brain-wave-measurement system.

MEASUREMENT OF VOLUME RESISTANCE: SOLID-GEL VS. PASTE

The volume resistance of the solid-gel and paste was evaluated using a rectangular measurement cell in which the facing sides were a pair of plain electrodes. The cell was filled with sample material, and its impedance was evaluated with a frequency response analyzer (S-5720B, NF Corp., Kanagawa, Japan).

EVALUATION OF VISCOELASTICITY: SOLID-GEL

The dynamic viscoelasticity of the solid-gel was evaluated using a rheometer (NDS-1000, Taisei, Saitama, Japan). The complex modulus of a sample was obtained from the dynamic displacement response on applying sinusoidal pressure (3 Hz). The dynamic viscoelasticity of a silicone rubber plug (E-02, Taiyo Kogyo, Tokyo, Japan) was also measured as a reference sample.

EVALUATION OF MOISTURE-RETENTION PROPERTY: SOLID-GEL VS. PASTE

The moisture-retention property of the samples was evaluated by measuring their weight under a controlled atmosphere. Each sample filled a small cylindrical plastic cup (diameter: 19 mm; depth: 22.5 mm) to the top. Then, the cups were put into a thermo-hydrostat (TPAV-48-20, ISUZU, Tokyo, Japan) without a cover, and the cups were weighed every 2 h. Temperature and relative humidity were constant at 23°C and 40%, respectively. Two kinds of liquid sample were prepared to contrast the moisture-retention property of the solid-gel and the conventional paste. One liquid contained 42.7, 8.5, and 48.7% of $CaCl_2 \cdot 2H_2O$, glycerol, and H_2O by weight, respectively. The other liquid was a 3 M KCl/H_2O solution.

IMPEDANCE EVALUATION: SOLID-GEL VS. METAL-PIN VS. PASTE

The participants in the electrode impedance evaluation were 11 healthy adults (six males, five females; age range 20–39 years; average 28.2 years). The experiment was approved by the Institutional Review Board, and all participants provided written informed consent according to institutional guidelines. The impedances of the solid-gel, metal-pin, and conventional paste electrodes were measured simultaneously by positioning these electrodes at P3, Pz, and P4, respectively.

BMI MEASUREMENTS: SOLID-GEL VS. METAL-PIN

For the P300-BMI tests, four healthy males (age 26–39 years) were recruited as participants. The experiment was approved by the Institutional Review Board, and all participants provided written informed consent according to institutional guidelines. The participants were required to input 15 hiragana characters from the Japanese alphabet using a row and column flicker panel with an 8 × 10 matrix in each condition. For this purpose, we modified the P300 speller (Farwell and Donchin, 1988), which uses the P300 paradigm, which presents a selection of icons arranged in a matrix. The participant focuses attention on one of the icons in the green/blue flicker matrix as a target, and each row/column of the matrix is intensified in a random sequence. The target stimuli are presented as the rare stimuli (oddball paradigm). P300 responses to the target stimuli were elicited, and the extraction and classification of these responses can be used to get the target. One complete cycle of eight row intensifications and 10 column intensifications constitutes a sequence. Online performance was evaluated, and each letter was selected in a series of 10 sequences. Eight-channel (Fz, Cz, Pz, P3, P4, Oz, PO7, PO8) EEG data were recorded using an in-house cap and an amplifier (g.USB amp, Guger Technologies, Graz, Austria). All channels were referenced to the Fpz, and grounded to the AFz electrode. Fisher's linear discriminant analysis was used for classification purposes. The details of experimental setting were same as in our previous study (Takano et al., 2009).

RESULTS

We first evaluated the moisture-retention property. **Figure 3** shows the time-dependent change in the weight of our solid-gel. The decrease in the weight of the liquid, which was the electrolyte of

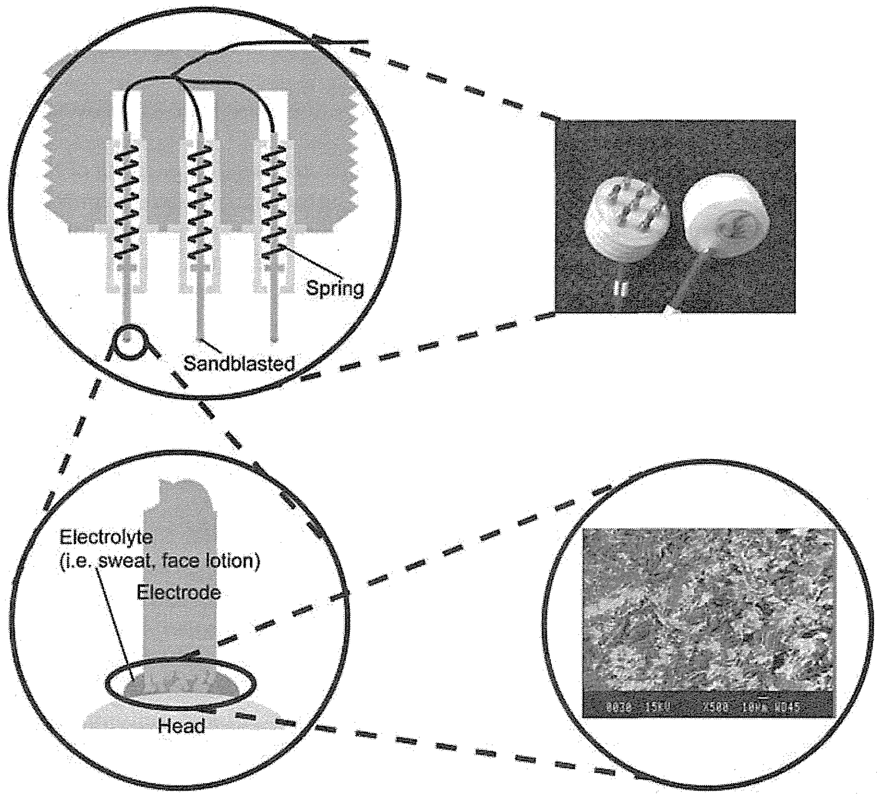

FIGURE 2 | Photograph of the metal-pin electrodes and illustration of the electrode structure. The electrode has seven retractable pins. The tip of each pin was sandblasted to enlarge the effective surface area (see inset scanning electron micrograph).

the solid-gel, was less than that of the KCl solution that is normally used for conventional electrodes. The decrease in the weight of the solid-gel material was also smaller than that of conventional conductive paste, suggesting that the solid-gel remained wet for more than 10 h in the dry environment.

The fabricated solid-gel chips were not sticky, but elastic. The complex modulus was 105.4 kPa, whereas that of the silicone rubber plug was 2088 kPa. Moreover, the volume resistivity of the solid-gel was about 13.2 Ωcm, which is sufficiently low to obtain brain waves. For comparison, the volume resistivity of the conventional paste was measured as 64.8 Ωcm. Before inserting the solid-gel chips into the through-holes of the head-mounting cap, we pushed aside the hair appearing at the bottom of the through-holes. However, this process was not laborious when we became accustomed to it, and it took 30–60 s for each chip setting, even in the worst case. The solid-gel chip was deformed when it was inserted into the through-hole and exposed to pressure from pushing the electrode plate with a finger, thereby penetrating the hair mesh and attaching to the scalp skin surface. The total setting time required before the measurement was 5–10 min.

An impedance evaluation test was carried out to evaluate electrode attachment. **Figure 4** shows the impedance time course between the reference and the brain-wave collecting electrodes including the newly developed electrodes. Although there were 11 participants, some data were omitted. The data from two females were omitted because the impedance of the paste-based electrode

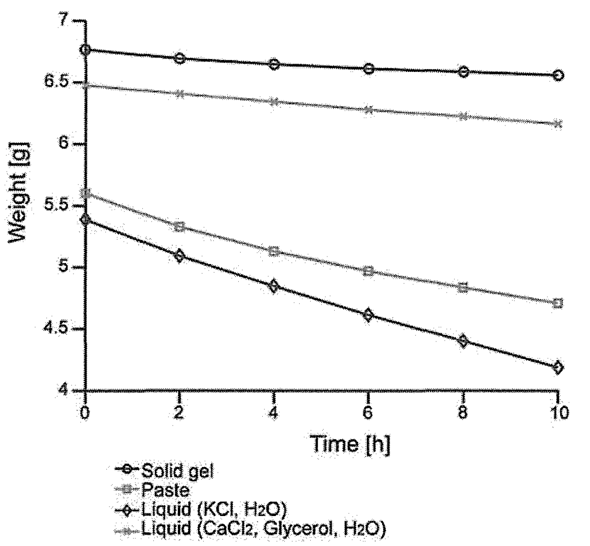

FIGURE 3 | Time-dependent changes in the weight of the solid-gel and other materials. Data of four materials were taken simultaneously.

abnormally increased during the measurement. The impedance for one person was 92 kΩ at 3 h and 116 kΩ at 4 h, and that of the other person was 468 kΩ at 3 h and 5654 kΩ at 4 h. In these cases, we noted that the paste had dried. Data for the metal-pin electrodes in six individuals were also omitted due to irregular impedance values from the start; the main reason was likely the thick hairs between the electrode and scalp, which would cause poor contact between them. Therefore, data of nine participants (six males, three females) for the paste- and gel-based electrodes and five participants (four males, one female) for the metal-pin electrodes remained.

The initial impedance of the solid-gel-based electrode was slightly higher than that of the paste-based electrode, but it decreased gradually and was sufficiently low within 1 h. The impedance ranged from 3 to 25 kΩ (typically 10 kΩ). Moreover, it was noticeable that only the impedance of the solid-gel-based electrode decreased continuously. In two cases, we continued impedance measurements with the solid-gel-based electrode and discovered that the impedance was almost constant for at least 9 h in both cases (data not shown). Furthermore, from the data for the five participants in whom we evaluated the impedance with all three electrode types, ANOVA and the *post hoc* Tukey–Kramer test revealed that the impedance (at 4 h) of the metal-pin electrode was significantly higher than those of the other electrodes [$F(2,8) = 12.01, p = 0.0039$]. A similar tendency in the impedance behavior was observed with a slightly different solid-gel containing $MgCl_2$ instead of $CaCl_2$ (data not shown).

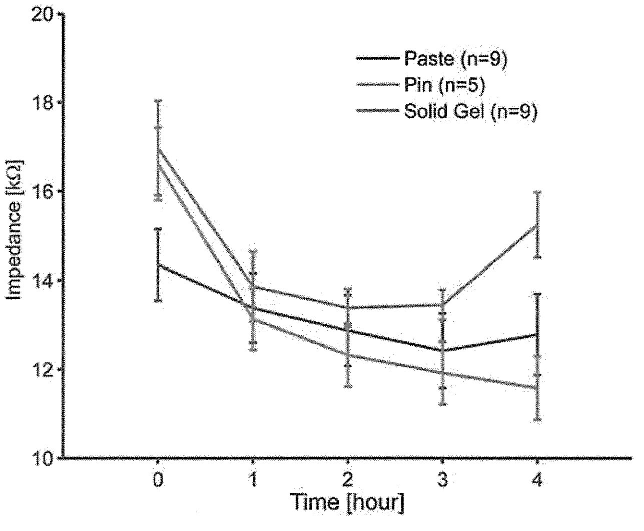

FIGURE 4 | Time course of the electrode impedance of the solid-gel, metal-pin, and conventional paste-based electrodes mounted on the head. Nine participants participated in these experiments at most. However, data for two participants using to the paste-based electrode was omitted, because the values were abnormally high due to drying.

After checking electrode impedance, we successfully obtained brain waves with the solid-gel-based electrodes. As shown in **Figure 5**, we observed similar α-waves and the spike-like waves corresponding to eye blinking during the measurements using the solid-gel- and paste-based electrodes.

Furthermore, we examined the use of the electrodes for operating a P300-BMI to input hiragana characters (**Figure 6**). With the metal-pin electrode, the mean accuracy was 85% ($n = 4$, 73.3–92.3%), whereas with the solid-gel-based electrode, the mean accuracy was 86.7% ($n = 4$, 80–92.3%). These results showed the potential of these electrodes for practical use as a BMI system in individuals with disabilities.

The conventional paste was apt to dry, especially peripherally, where it was exposed to the atmosphere; it was also difficult to remove without washing the hair after use. By contrast, the solid-gel-based electrodes had superior wettability throughout the extended measurements. The solid-gel remained wet for more than 9 h after the measurement; consequently, it was easy to remove, and almost no residual solid-gel was seen after use. Moreover, we observed no skin problems after using the solid-gel-based electrodes in our experiments.

Our solid-gel chips can be stored for more than 3 months at room temperature without any significant change when packed in a sealed container, suggesting the practical usefulness of our solid-gel chip.

DISCUSSION

We prepared a conductive solid-gel-based electrode and a metal-pin-based electrode. An examination of the weight change in

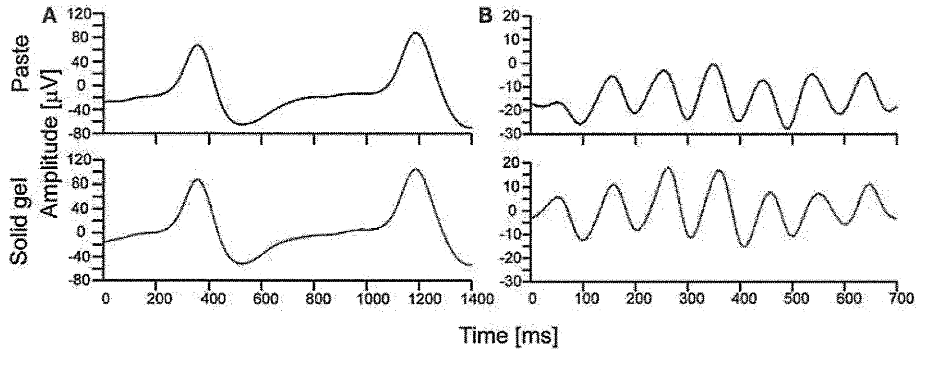

FIGURE 5 | Typical brain waves observed during simultaneous measurement using the solid-gel- and paste-based electrodes. **(A)** Brain waves observed during intermittent eye blinking. **(B)** Alpha waves. The upper and lower waves in each graph correspond to the signals obtained with the paste-based and solid-gel-based electrodes, respectively. The signals were collected with an in-house EEG amplifier (24 bit, 1024 Hz). The represented data were obtained through an eighth-order bandpass filter (1–15 Hz).

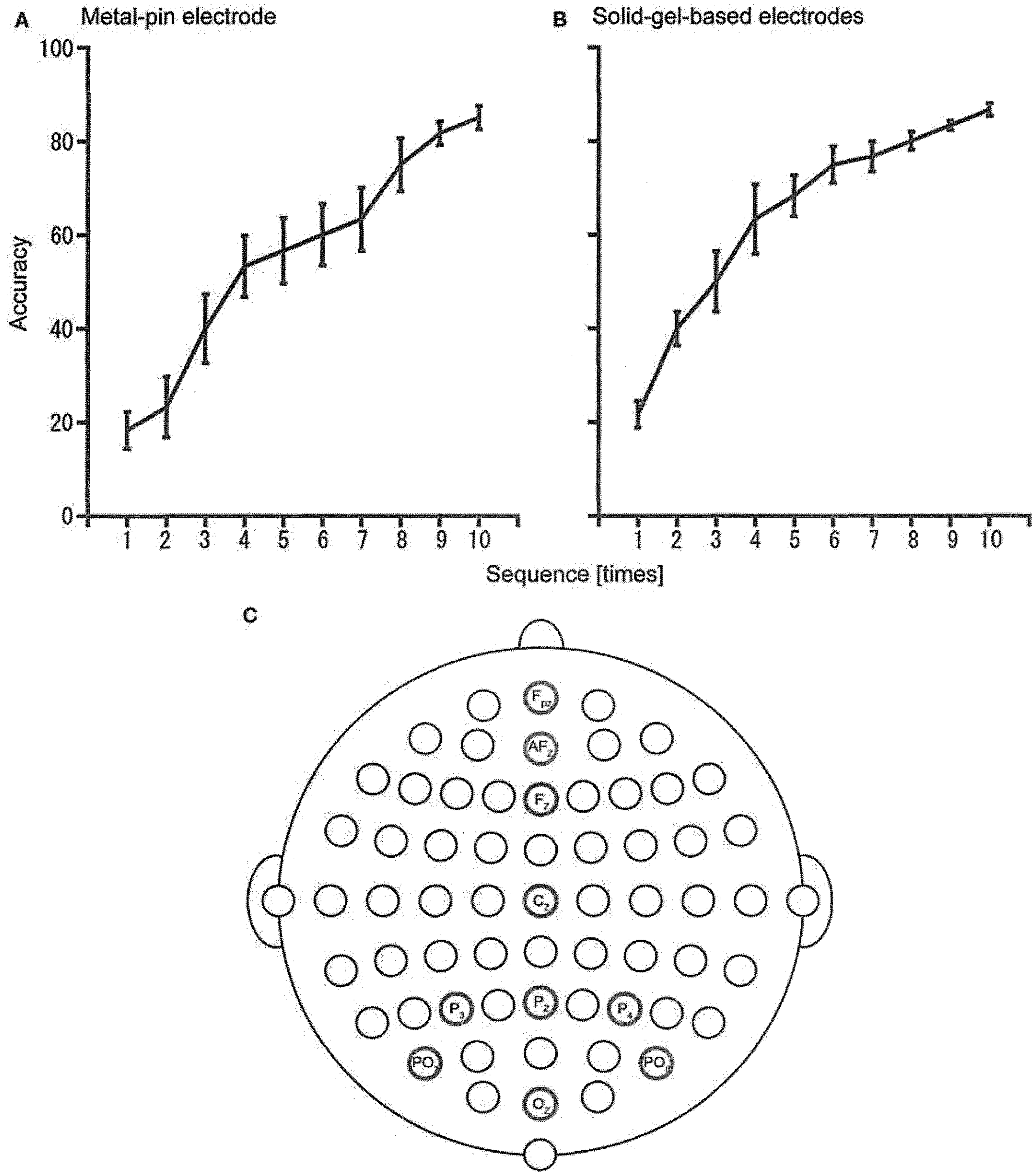

FIGURE 6 | Results of the P300-brain–machine interface (BMI) experiments using (A) metal-pin electrodes and (B) solid-gel-based electrodes. The configuration of electrodes is represented in **(C)**. Data were obtained from the same participants ($n = 4$).

the solid-gel material under a controlled atmospheric environment revealed superior retention of wettability by our solid-gel compared to that of conventional paste. The electrode impedance measurements were comparable between the solid-gel-based electrode and the conventional paste-based electrode, whereas the impedance of the metal-pin electrode was higher than that of the newly developed electrodes. Furthermore, we obtained almost equivalent signals with the solid-gel-based and paste-based electrodes during simultaneous brain-wave measurements. Finally, we successfully performed P300-BMI using both the metal-pin and solid-gel-based electrodes.

The solid-gel is elastic and is soft enough for use on human skin. Boyer et al. (2009) reported that the complex modulus of the skin on the arm is 10.7 ± 2.64, 8.09 ± 1.84, and 7.17 ± 2.06 kPa for participants 18–30, 31–50, and 51–70 years old, respectively. The complex modulus of the solid-gel was 105.4 kPa, whereas that of silicone rubber was 2088 kPa. Although the scalp is slightly harder than the arm skin, the silicone rubber was much harder than either of these.

All ingredients used in our solid-gel are food additives. CMC is used as a thickener in ice cream (Regand and Goff, 2002), and glycerol is used as a sweetener or moisturizer. Calcium chloride is used to prepare a coating material on fruit (Oms-Oliu et al., 2010).

In East Asian countries such as Japan, calcium chloride is also used to coagulate soy solution to produce tofu (bean curd; Prabhakaran et al., 2006). Furthermore, one of the main reasons for adopting calcium chloride was that some commercial EEG pastes already contain calcium (for example, Weaver and Co., Ten20 Conductive Paste; Nihon Kohden, Elefix). Although calcinosis has been reported after using a calcium-containing electrode paste (Wiley and Eaglstein, 1979; Mancuso et al., 1990; Puig et al., 1998), we have not seen unfavorable cases in our experiments. However, we prefer magnesium chloride to calcium chloride for future work because we obtained similar results from the electrodes using either of them in preliminary testing.

In a textbook on EEG (Niedermeyer and Silva, 1999), the recommended contact impedance between the electrode and skin is $<5\,k\Omega$, whereas that of our solid-gel was about $10\,k\Omega$. Although the impedance of our solid-gel-based electrode seems slightly higher than the recommended value, it is acceptable because the impedance depends on geometry and electrode size. In our case, the paste-based electrode with the same geometry and size as those of the solid-gel electrode had similar value. Compared with the dry electrodes in the literature, the contact impedance of the solid-gel-based electrode was comparable or better. The reported impedance is $4–26\,k\Omega$ with a polymer form electrode (Lin et al., 2011), $<20\,k\Omega$ with a bundle of pin electrodes (Zander et al., 2011), and $7–25\,k\Omega$ with arrayed spike electrodes (Ng et al., 2009). In our experiments, the impedance of the metal-pin electrode was about twice as high as that of the solid-gel-based electrode.

We successfully obtained brain waves with the BMI system employing a solid-gel-based electrode. The signals were almost equivalent to those observed with conventional paste-based electrodes. Moreover, the electrode does not require elaborate preparatory work or removal of paste adhered to hair and skin after measurement.

We have already revealed the usefulness of our BMI system using conventional paste-based electrodes (Takano et al., 2009; Ikegami et al., 2011). Carrying out a similar test of P300-BMI, we showed the usefulness of our newly developed metal-pin and conductive solid-gel-based electrodes here. The solid-gel-based electrode is soft and less harmful to scalp skin, whereas pain was sometimes felt with the metal-pin electrodes. Most users will be bedridden individuals such as patients with ALS; therefore, the soft solid-gel-based electrode is a good first choice for that purpose. Furthermore, our developed electrodes can be used for other long-term EEG investigations such as sleep recordings, anesthesia monitoring, intensive-care monitoring, and long-term monitoring of patients with epilepsy (Niedermeyer and Silva, 1999).

ACKNOWLEDGMENTS

This study was supported in part by a grant from the Ministry of Health, Labour, and Welfare to K. Kansaku. We thank T. Komatsu and S. Ikegami for their help and Y. Nakajima and S. Kato for their continuous encouragement.

REFERENCES

Aksenov, E., Ljashenko, Y. M., Plotnikov, A., Prilutskiy, D., Selishchev, S., and Vetvetskiy, E. (2001). "Biomedical data acquisition systems based on sigma-delta analogue-to-digital converters," in *23rd Annual International Conference of the IEEE Engineering in Medicine and Biology* (Istanbul: IEEE), 3336–3337.

Beckmann, L., Neuhaus, C., Medrano, G., Jungbecker, N., Walter, M., Gries, T., and Leonhardt, S. (2010). Characterization of textile electrodes and conductors using standardized measurement setups. *Physiol. Meas.* 31, 233–247.

Birbaumer, N., and Cohen, L. G. (2007). Brain-computer interfaces: communication and restoration of movement in paralysis. *J. Physiol. (Lond.)* 579, 621–636.

Boyer, G., Laquièze, L., Le Bot, A., Laquièze, S., and Zahouani, H. (2009). Dynamic indentation on human skin in vivo: ageing effects. *Skin Res. Technol.* 15, 55–67.

Chiou, J. C., Ko, L. W., Lin, C. T., Hong, C. T., Jung, T. P., Liang, S. F., and Jeng, J. L. (2006). "Using novel MEMS EEG sensors in detecting drowsiness application," in *Biomedical Circuits and Systems Conference, BioCAS 2006* (London: IEEE), 33–36.

Cui, X., Bray, S., and Reiss, A. L. (2010). Speeded near infrared spectroscopy (NIRS) response detection. *PLoS ONE* 5, e15474. doi:10.1371/journal.pone.0015474

Farwell, L. A., and Donchin, E. (1988). Talking off the top of your head: toward a mental prosthesis utilizing event-related brain potentials. *Electroencephalogr. Clin. Neurophysiol.* 70, 510–523.

Griss, P., Enoksson, P., Tolvanen-Laakso, H. K., Merilainen, P., Ollmar, S., and Stemme, G. (2001). Micromachined electrodes for biopotential measurements. *J. Microelectromech. Syst.* 10, 10–16.

Hoffmann, K. P., and Ruff, R. (2007). "Flexible dry surface-electrodes for ECG long-term monitoring," in *29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society* (Lyon: IEEE), 5739–5742.

Ikegami, S., Takano, K., Saeki, N., and Kansaku, K. (2011). Operation of a P300-based brain-computer interface by individuals with cervical spinal cord injury. *Clin. Neurophysiol.* 122, 991–996.

Kansaku, K. (2011). "Brain–machine interfaces for persons with disabilities," in *Systems Neuroscience and Rehabilitation*, eds K. Kansaku and L. Cohen (Tokyo: Springer), 19–33.

Kansaku, K., Hata, N., and Takano, K. (2010). My thoughts through a robot's eyes: an augmented reality-brain-machine interface. *Neurosci. Res.* 66, 219–222.

Liao, L. D., Wang, I. J., Chen, S. F., Chang, J. Y., and Lin, C. T. (2011). Design, fabrication and experimental validation of a novel dry-contact sensor for measuring electroencephalography signals without skin preparation. *Sensors (Basel)* 11, 5819–5834.

Lin, C. T., Liao, L. D., Liu, Y. H., Wang, I. J., Lin, B. S., and Chang, J. Y. (2011). Novel dry polymer foam electrodes for long-term EEG measurement. *IEEE Trans. Biomed. Eng.* 58, 1200–1206.

Mancuso, G., Tosti, A., Fanti, P., Berdondini, R., Mongiorgi, R., and Morandi, A. (1990). Cutaneous necrosis and calcinosis following electroencephalography. *Dermatology (Basel)* 181, 324–326.

Ng, W. C., Seet, H. L., Lee, K. S., Ning, N., Tai, W. X., Sutedja, M., Fuh, J. Y. H., and Li, X. P. (2009). Micro-spike EEG electrode and the vacuum-casting technology for mass production. *J. Mater. Process. Technol.* 209, 4434–4438.

Niedermeyer, E., and Silva, F. H. L. (1999). *Electroencephalography: Basic Principles, Clinical Applications, and Related Fields.* Philadelphia: Williams & Wilkins.

Oms-Oliu, G., Rojas-Graü, M. A., González, L. A., Varela, P., Soliva-Fortuny, R., Hernando, M. I. H., Munuera, I. P., Fiszman, S., and Martín-Belloso, O. (2010). Recent approaches using chemical treatments to preserve quality of fresh-cut fruit: a review. *Postharvest Biol. Technol.* 57, 139–148.

Pfurtscheller, G., Brunner, C., Schlogl, A., and Lopes da Silva, F. H. (2006). Mu rhythm (de)synchronization and EEG single-trial classification of different motor imagery tasks. *Neuroimage* 31, 153–159.

Piccione, F., Giorgi, F., Tonin, P., Priftis, K., Giove, S., Silvoni, S., Palmas, G., and Beverina, F. (2006). P300-based brain computer interface: reliability and performance in healthy and paralysed participants. *Clin. Neurophysiol.* 117, 531–537.

Popescu, F., Fazli, S., Badower, Y., Blankertz, B., and Müller, K. R. (2007). Single trial classification of motor imagination using 6 dry EEG electrodes. *PLoS ONE* 2, e637. doi:10.1371/journal.pone.0000637

Prabhakaran, M. P., Perera, C. O., and Valiyaveettil, S. (2006). Effect of different coagulants on the isoflavone levels and physical properties of prepared firm tofu. *Food Chem.* 99, 492–499.

Puig, L., Rocamora, V., Romani, J., Saavedra, M., and Alomar, A. (1998). Calcinosis cutis following calcium chloride electrode paste application for auditory–brainstem evoked potentials recording. *Pediatr. Dermatol.* 15, 27–30.

Regand, A., and Goff, H. (2002). Effect of biopolymers on structure and ice recrystallization in dynamically frozen ice cream model systems. *J. Dairy Sci.* 85, 2722–2732.

Ruffini, G., Dunne, S., Farrés, E., Marco-Pallarés, J., Ray, C., Mendoza, E., Silva, R., and Grau, C. (2006). A dry electrophysiology electrode using CNT arrays. *Sens. Actuators A Phys.* 132, 34–41.

Sellers, E., Turner, P., Sarnacki, W., Mcmanus, T., Vaughan, T., and Matthews, R. (2009). "A novel dry electrode for brain-computer interface," in *Human-Computer Interaction. Novel Interaction Methods and Techniques,* ed. J. A. Jacko (Berlin: Springer), 623–631.

Sellers, E. W., and Donchin, E. (2006). A P300-based brain-computer interface: initial tests by ALS patients. *Clin. Neurophysiol.* 117, 538–548.

Sitaram, R., Lee, S., Ruiz, S., Rana, M., Veit, R., and Birbaumer, N. (2011). Real-time support vector classification and feedback of multiple emotional brain states. *Neuroimage* 56, 753–765.

Sitaram, R., Zhang, H., Guan, C., Thulasidas, M., Hoshi, Y., Ishikawa, A., Shimizu, K., and Birbaumer, N. (2007). Temporal classification of multichannel near-infrared spectroscopy signals of motor imagery for developing a brain-computer interface. *Neuroimage* 34, 1416–1427.

Takano, K., Hata, N., and Kansaku, K. (2011). Towards intelligent environments: an augmented reality-brain-machine interface operated with a see-through head-mount display. *Front. Neurosci.* 5:60. doi:10.3389/fnins.2011.00060

Takano, K., Komatsu, T., Hata, N., Nakajima, Y., and Kansaku, K. (2009). Visual stimuli for the P300 brain–computer interface: a comparison of white/gray and green/blue flicker matrices. *Clin. Neurophysiol.* 120, 1562–1566.

Toyama, S., Kansaku, K., and Takano, K. (2011a). Japan patent application 2011–262032. 2011-11-30.

Toyama, S., Kansaku, K., Takano, K., and Ikegami, S. (2011b). Japan patent publication 2011-120866. 2011-06-23.

Wiley, H. E., and Eaglstein, W. E. (1979). Calcinosis cutis in children following electroencephalography. *JAMA* 242, 455.

Wolpaw, J. R., and McFarland, D. J. (2004). Control of a two-dimensional movement signal by a noninvasive brain–computer interface in humans. *Proc. Natl. Acad. Sci. U.S.A.* 101, 17849–17854.

Zander, T. O., Lehne, M., Ihme, K., Jatzev, S., Correia, J., Kothe, C., Picht, B., and Nijboer, F. (2011). A dry EEG-system for scientific research and brain-computer interfaces. *Front. Neurosci.* 5:53. doi:10.3389/fnins.2011.00053

Conflict of Interest Statement: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Received: 06 April 2012; accepted: 30 June 2012; published online: 18 July 2012.

Citation: Toyama S, Takano K and Kansaku K (2012) A non-adhesive solid-gel electrode for a non-invasive brain–machine interface. Front. Neur. ***3****:114. doi: 10.3389/fneur.2012.00114*

This article was submitted to Frontiers in Neuroprosthetics, a specialty of Frontiers in Neurology.

平成22~24年度厚生労働科学研究費補助金（障害者対策総合研究事業（身体・知的等障害分野））
「ブレインーマシン・インターフェイス（BMI）による障害者自立支援機器の開発」

総合研究報告書

発行者　　中島　八十一（研究代表者：国立障害者リハビリテーションセンター）
〒359-8555　埼玉県所沢市並木4-1