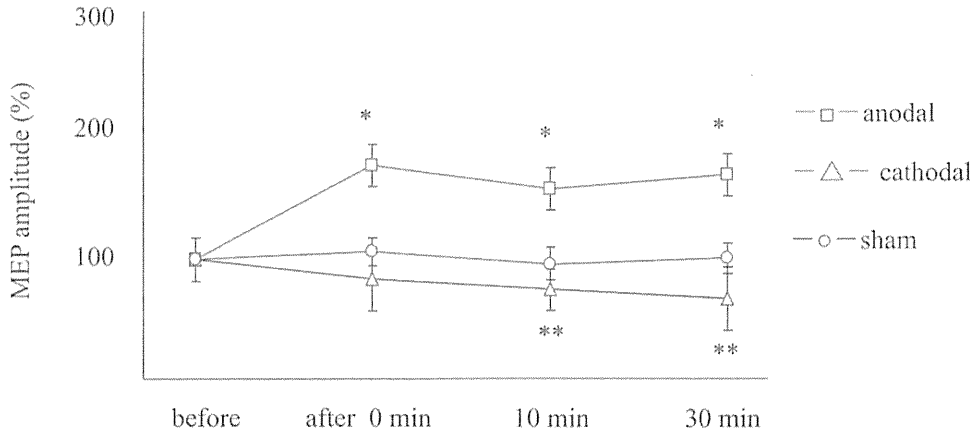

FIGURE 1. Change of the MEP amplitudes before and after (after: 0 min, 10 min, 30 min) the anodal tDCS (open square), cathodal tDCS (open triangle), and sham tDCS (open circle) among healthy subjects. The size of the amplitude is expressed as a percentage of the preconditioning control. Anodal tDCS increased MEPs in 0, 10, and 30 min after stimulation and cathodal tDCS decreased MEPs at 10 and 30 min after stimulation. $^{*}p < 0.001$; $^{**}p < 0.05$.

(before and at intervals after tDCS), stimulation (anodal, cathodal, and sham), and category (healthy subjects and stroke subjects; $F_{6,9} = 5.369$, $p = 0.013$).

In healthy subjects, anodal tDCS increased MEPs at 0 min ($p < 0.001$), 10 min ($p < 0.001$), and 30 min after stimulation ($p < 0.001$). Cathodal tDCS decreased the MEPs at 10 min ($p = 0.023$) and 30nin ($p = 0.04$) after stimulation compared with before stimulation. Sham stimulation did not induce any significant changes of MEPs.

In stroke patients, anodal tDCS increased the MEPs significantly at 0 min ($p = 0.024$) and 10 min ($p = 0.031$) after tDCS. In cathodal tDCS, the MEPs were significantly increased at 0 min after tDCS compared with before tDCS ($p = 0.016$). Sham stimulation did not induce any significant change.

Silent Period and F-Wave Amplitude

Table 2 shows the mean duration of the silent period and mean amplitudes of the F-wave. There was no significant interaction of time, stimulation, and category in the duration of the silent period ($F_{6,66} = 0.816$, $p = 0.562$) and F-wave amplitude ($F_{6,66} = 0.348$, $p = 0.909$).

DISCUSSION

We found that both anodal and cathodal tDCS increased cortical excitability in patients with subcortical stroke. Cathodal tDCS, however, decreased cortical excitability in healthy volunteers. Our results showed that MEP change induced with tDCS has significant interaction

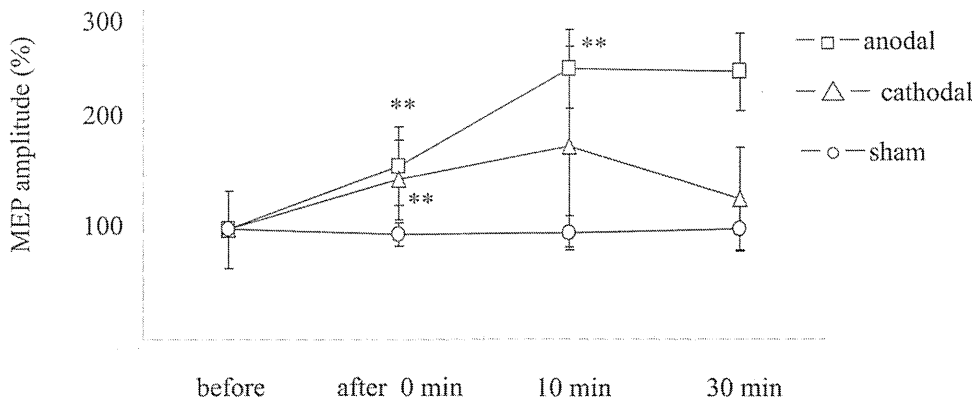

FIGURE 2. Change of the MEP amplitudes before and after (after: 0 min, 10 min, 30 min) the anodal tDCS (open square), cathodal tDCS (open triangle), and sham tDCS (open circle) among patients with stroke. The size of the amplitude is expressed as a percentage of the preconditioning control. Post-hoc paired *t*-test showed significant increased MEPs in 0 and 10 min after anodal and 0 min after cathodal tDCS. $^{**}p < 0.05$.

TABLE 2. The mean values (SD) of silent period and F-wave amplitude

Group	Parameter	Stimulation	Before	After 0 min	After 10 min	After 30 min
Healthy group	Silent period (ms)	Anodal tDCS	98.8 (47.0)	103.0 (32.0)	102.7 (46.7)	109.4 (48.9)
		Cathodal tDCS	130.7 (19.5)	133.2 (22.5)	131.5 (19.7)	135.1 (24.1)
		Sham	124.4 (39.8)	124.3 (43.6)	125.4 (39.5)	125.1 (39.9)
	F-wave amplitude (mV)	Anodal tDCS	0.24 (0.14)	0.18 (0.05)	0.24 (0.10)	0.22 (0.10)
		Cathodal tDCS	0.21 (0.16)	0.21 (0.11)	0.18 (0.04)	0.19 (0.10)
		Sham	0.22 (0.15)	0.22 (0.12)	0.23 (0.11)	0.24 (0.08)
Stroke group	Silent period (ms)	Anodal tDCS	188.1 (78.8)	190.5 (79.1)	192.3 (78.6)	189.5 (80.3)
		Cathodal tDCS	196.1 (71.2)	196.7 (71.9)	203.8 (75.0)	202.8 (76.0)
		sham	178.9 (72.4)	175.8 (72.1)	179.3 (78.3)	174.1 (70.9)
	F-wave amplitude (mV)	Anodal tDCS	0.31 (0.05)	0.33 (0.06)	0.30 (0.08)	0.34 (0.10)
		Cathodal tDCS	0.27 (0.12)	0.23 (0.11)	0.26 (0.09)	0.27 (0.15)
		Sham	0.30 (0.11)	0.32 (0.13)	0.30 (0.11)	0.35 (0.14)

of category (stroke and healthy), stimulation (anodal, cathodal, and sham), and time (before and at intervals after tDCS). It implies that the modulation of motor cortex excitability with tDCS depends on the state of motor cortex.

All patients recruited in this study had a subcortical lesion. Liepert et al. [17] reported that motor cortex excitability of patients with subcortical stroke had been decreased. Neuronal circuits within the basal ganglia facilitate the motor cortex either through antidromic excitation of cortical-basal ganglia fibers or through orthodromic activation of a basal ganglia-thalamocortical pathway. Therefore, a stroke-induced disturbance of basal ganglia may result in change of motor cortex excitability. It was hypothesized that the condition of projection from the basal nucleus to M1 in these patients would be different from conditions in healthy subjects. Therefore, cortical modulation by tDCS would be different in the affected hemisphere.

The differential effects of cathodal tDCS between healthy subjects and patients with stroke underline the point that the effects of brain stimulation depend on the physiological state of neuronal populations at the time the stimulus is applied. This is evident in concepts such as "homeostatic" plasticity [18,19], where the state of neural activity determines the production of LTP/LTD-like synaptic effects. Siebner et al. [18] showed that inhibitory preconditioning with cathodal tDCS resulted in 1 Hz repetitive TMS (rTMS) increasing corticospinal excitability, whereas 1 Hz rTMS alone induced reduction in corticospinal excitability. The baseline corticospinal excitability in patients with stroke should be reduced in the affected hemisphere because the mean RMT value of the affected hemisphere was significantly higher than that of healthy subjects. The after-effect of noninvasive brain stimulation should depend on the state of excitability before stimulation. According to the homeostatic hypothesis, cathodal tDCS should increase corticospinal excitability in a reduced-excitability state affected hemisphere, such as stroke, whereas it should decrease corticospinal excitability in healthy subjects.

When changes that bring about membranal potential excitement or stimuli that would induce changes in synaptic transmission occur at a time when excitability is lowered by the state of the precondition, as with homeostatic plasticity, there is a possibility for cortical excitability to be increased after cathodal tDCS in the direction of homeostatic plasticity.

The mechanism how cathodal tDCS increases the motor cortex excitability may be explained by cathodal tDCS-induced depression of cortical inhibitory interneuron, which is abnormally increased in the affected hemisphere among patients with stroke.

It should be, however, noted that it is not always the case for the opposite phenomena to occur in anodal and cathodal tDCS [20,21].

Age difference between healthy subjects and patients with stroke might be other mechanism, which explains the different after-effect induced cathodal tDCS. Normal aging is associated with relative decrease in the excitability of inhibitory circuits within motor cortex. Peinemann et al. [22] showed that short-interval intracortical inhibition correlated negatively with age, whereas intracortical facilitation (ICF) was preserved in the elderly persons. It is, therefore, supposed that cathodal tDCS induces facilitatory effect more than inhibitory effect in elderly person.

Hummel et al. [8] found decreases of short intracortical inhibition (SICI) with anodal tDCS. Liepert et al. [23] conducted TMS with stroke patients as subjects. Compared with the unaffected side, SICI decreased in a significant manner, which suggests disinhibition among stroke patients. They reported that the cortical silent period was significantly prolonged among stroke patients, which suggests that different inhibition mechanisms are at work in SICI and the silent period.

With our current assessment, no changes were seen in the silent period with anodal or cathodal tDCS. With

1 mA tDCS for 10 min, it is possible that the effects were not large enough to bring about changes in the silent period. It should be noted, however, that there are many unknown elements in the mechanisms of occurrences during the silent period. In the case of stroke patients, effects of excitement and inhibition between hemispheres must be considered. It is necessary, therefore, to carry out in future assessments on SICI and ICF during the silent period and double stimuli (paired pulse TMS) after conducting anodal and cathodal tDCS on both the affected and unaffected areas.

As for the F-wave, results from our study showed no significant changes before or after anodal or cathodal tDCS. Nitsche et al. [5] reported that no changes occurred on the spinal level as no change was seen in H-reflex amplitudes before and after tDCS. Nitsche et al. [24] reported no change in the assessment of F-wave. When combining these reports with our assessment on F-wave of stroke patients, it is considered that MEP changes are caused by changes on the cortex level and not on the spinal level. However, it has been pointed out that effects on the spinal level cannot be completely excluded when examining F-waves [25]. Thus, future studies are warranted.

We found no relationships between the change of corticospinal excitability induced by tDCS and time from onset in this study. We could not find consistent differences between the subjects with putaminal lesions, the subjects with thalamic lesions, and the subjects with subcortical white matter lesions. Patients with hypertensive hemorrhages and ischemic stroke often have small vessel microangiopathic changes on MRI. Extensive small vessel ischemic disease affects cognition and motor function [26]. We did not find extensive microangiopathic changes on MRI. We could not, however, exclude the effect of microangiopathic changes. However, we had a limited number of subjects in this study, so further studies are needed. In particular, we need to study factors that influence the change of corticospinal excitability induced by tDCS, such as lesion size, location, time after events, and impairment level.

Hummel et al. [9] reported that tDCS of the motor cortex improve motor function in the paretic hand of patients with chronic stroke. We did not find whether anodal and cathodal tDCS have beneficial effect on motor and sensory function of the affected hand or not. Future studies are necessary to study the relationship to the behavioral consequences of stimulation in patients with stroke.

CONCLUSION

In this study, we found that both anodal and cathodal tDCS increased the affected MI excitability in patients with stroke. It is thought that the after-effect of tDCS is different in patients with stroke, compared with healthy subjects. In applying tDCS to patients with stroke, we further need to study factors that influence the after-effect induced by tDCS.

Declaration of interest: The authors report no conflicts of interest. The authors alone are responsible for the content and writing of this paper.

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RESEARCH ARTICLE

Modulation of event-related desynchronization during motor imagery with transcranial direct current stimulation (tDCS) in patients with chronic hemiparetic stroke

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Abstract Electroencephalogram-based brain–computer interface (BCI) has been developed as a new neurorehabilitative tool for patients with severe hemiparesis. However, its application has been limited because of difficulty detecting stable brain signals from the affected hemisphere. It has been reported that transcranial direct current stimulation (tDCS) can modulate event-related desynchronization (ERD) in healthy persons. The objective of this study was to test the hypothesis that anodal tDCS could modulate ERD in patients with severe hemiparetic stroke. The participants were six patients with chronic hemiparetic stroke (mean age, 56.8 ± 9.5 years; mean time from the onset, 70.0 ± 19.6 months; Fugl-Meyer Assessment upper extremity motor score, 30.8 ± 16.5). We applied anodal tDCS (10 min, 1 mA) and sham stimulation over the affected primary motor cortex in a random order. ERD of the mu rhythm (mu ERD) with motor imagery of extension of the affected finger was assessed before and after anodal tDCS and sham stimulation. Mu ERD of the affected hemisphere increased significantly after anodal tDCS, whereas it did not change after sham stimulation. Our results show that anodal tDCS can increase mu ERD in patients with hemiparetic stroke, indicating that anodal tDCS could be used as a conditioning tool for BCI in stroke patients.

Keywords Electroencephalography · Cerebrovascular disease · Rehabilitation · Noninvasive brain stimulation

Introduction

The functional recovery of the upper extremity is limited in patients with hemiparetic stroke. Most patients with stroke have difficulty performing activities of daily living (ADL) using their weakened upper extremity. The functional recovery depends on the severity of their motor impairment (Hendricks et al. 2002). Therefore, therapeutic options for patients with a severely hemiparetic upper extremity are limited.

Recently, technological innovations such as the brain–computer interface (BCI) have been developed. Buch et al. (2008) reported the possibility of using the sensorimotor mu rhythm over the affected primary motor cortex (M1) recorded with magnetoencephalography (MEG) for neurorehabilitation. Using this signal, the patients learned to use motor imagery to control the mu rhythm and to operate an orthotic device that opened and closed their paretic hand. The mu rhythm is a spontaneous characteristic feature of the EEG/MEG pattern that has 8- to 13-Hz activity and appears maximally over the central rolandic or sensorimotor area

Yuko Kasashima and Yayoi Matsushika contributed equally to the study.

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during a relaxed state. It is attenuated by movement execution and motor imagery. That phenomenon is referred to as event-related desynchronization (ERD) (Arroyo et al. 1993). The ERD of the mu rhythm, named mu ERD, is interpreted as the desynchronized activities of the activated neurons, and it appears around the motor area during motor execution, preparation and motor imagery (Pfurtscheller and Aranibar 1977; Pfurtscheller and Lopes da Silva 1999). Recently, electroencephalogram (EEG)-based BCI was applied to patients with chronic stroke (Daly et al. 2009; Shindo et al. 2011). These EEG-based BCIs controlled an orthotic device to extend their paretic fingers. The EEG-based BCIs detected mu ERD during motor imagery with EEG and reported that BCI training improved hand motor function in patients with chronic stroke.

However, the application of BCI to patients with severe motor disabilities has been limited, because of the difficulty detecting stable brain signals (Platz et al. 2000; Leocani et al. 2006). If it is possible to potentiate ERD, it would be easier to apply BCI to these patients.

Matsumoto et al. (2010) reported that tDCS could modulate mu ERD in healthy persons. Anodal tDCS (10 min, 1 mA) increased the magnitude of mu ERD in M1. If tDCS could also increase mu ERD in patients with severe hemiparetic stroke, it may be useable as a conditioning tool to facilitate the detection of more stable ERD for BCI application. Therefore, the aim of this study was to test whether anodal tDCS could increase ERD in patients with severe hemiparetic stroke.

Methods

Participants

Six patients with chronic hemiparetic stroke (4 males and 2 females) participated in this study. Inclusion criteria consisted of the following: (1) first unilateral subcortical stroke, not involving sensorimotor cortex as confirmed with brain MRI; (2) time from the stroke onset more than 6 months; (3) moderate to severe hemiparesis (participants could not move their paretic fingers individually); and (4) no motor improvement in the last 1 month before starting the intervention as confirmed by physicians and patients' testimonies. Exclusion criteria were as follows: (1) history of major psychiatric or previous neurological diseases, including seizure; (2) cognitive impairment precluding informed consent; (3) use of central nervous system-active drugs; and (4) implanted pacemaker or other metallic object. Participants' mean age was 56.8 ± 9.5 years. The mean time from the onset was 70.0 ± 19.6 months. The mean score on the Fugl-Meyer assessment of upper extremity motor score was 30.8 ± 16.5 (Fugl-Meyer et al. 1975), and the median score of the modified Ashworth scale for finger flexors was 1+ (range = 1+ to 2) (Bohannon et al. 1987). All participants were right handed. Clinical details of the participants are shown in Table 1. Additionally, seven age-matched healthy persons were recruited. All were right handed. Their mean age was 54.4 ± 6.1 years. We found no significant difference in the age between the stroke and age-matched healthy participants (unpaired t test, $P = 0.593$). The purpose and procedures of the study were explained to the participants, and written informed consent was obtained. The study was approved by the institutional ethics review board and performed in accordance with the Declaration of Helsinki.

Measurement of event-related desynchronization (ERD)

We assessed mu ERD during imagery of extension of the affected fingers just before and after anodal and sham tDCS over the motor area of the affected hemisphere in the stroke participants. The order of the stimulations was randomized,

Table 1 Clinical details of participants

Participant (sex)	Age	Dx	Lesion	Paretic side	Time from onset (months)	FM U/E	Modified Ashworth scale
1 (M)	67	CH	L thalamus	R	96	35	1+
2 (M)	44	CH	L putamen	R	64	50	1+
3 (M)	63	CH	R thalamus	L	49	12	2
4 (M)	46	CI	R corona radiata	L	48	49	1+
5 (F)	61	CI	R putamen	L	85	24	1+
6 (F)	60	CI	L putamen	R	78	15	2
Mean	56.8				70	30.8	1+[a]
SD	9.5				19.6	16.5	

CI cerebral infarction, *CH* cerebral hemorrhage, *FM U/E* Fugl-Meyer assessment score upper extremity motor score, *L* left, *R* right

[a] median value

and the interval between the stimulations was more than 2 days. In the healthy participants, we assessed mu ERD during imagery of right finger extension before and after anodal tDCS over the left motor area.

EEG signals were recorded with 15 Ag/AgCl disk electrodes (1 cm in diameter) with binaural references according to the international 10-20 system of electrode placement (FC3, FC1, FCz, FC2, FC4, C3, C1, Cz, C2, C4, CP3, CP1, CPz, CP2, CP4) with the average of bilateral earlobe references. Impedance for all channels was maintained below 10 kΩ throughout the experiment. Electromyograms (EMGs) were simultaneously recorded from the bilateral extensor digitorum communis muscles (EDC) with surface Ag/AgCl disk electrodes (1 cm in diameter) to monitor EMG activities during the imagery task to avoid unexpected muscle contraction. EEG and EMG were amplified, digitized with sampling frequency of 1,000 Hz and band-pass filtered (EEG 0.53–100 Hz, EMG 20-1 kHz) using a commercially available biosignal recorder (Neurofax EEG-9100, Nihon Kohden Corporation, Japan).

The participants sat in an upright position in an armchair with their eyes open facing the computer monitor showing the task. The monitor was placed approximately 0.5 m in front of the subjects at eye level. One trial started with an 8-s period of relaxation during which the word "Rest" was shown on the monitor. After that, the word "Ready" was shown for 2 s, then the word "Start" was presented for 5 s, and the participants were asked to imagine extension of their affected fingers. The trial ended when the word "Rest" reappeared, and the next trial began. They were given no feedback regarding EEG changes to avoid a learning effect. One session consisted of 20 trials. Before and after tDCS, three sessions were conducted with approximately 5 min of rest between each session. All three sessions were completed within 30 min (Fig. 1).

Quantification of ERD

Event-related trials lasting 5 s during motor imagery were selected for off-line data processing. All trials were visually assessed. The trials with artifacts resulting from eye movement and the trials with increased EMG activities were excluded. All trials were segmented into successive 1-s windows with 900 overlapping samples, and the Fourier transform with the Hanning window was applied in each segment. The power spectrum densities of each segment were estimated over the trials by Welch's averaged periodogram method (Welch et al. 1967).

The mu ERD was expressed as the percentage of the power decrease in relation to the 1-s reference interval before the direction of "Ready." The ERD at a certain frequency was calculated for each time (resolution = 0.1 s) and frequency (resolution = 0.98) according to Eq. (1).

$$\mathrm{ERD}\ (f,t) = \{[R(f) - A(f,t)]/R(f)\} \times 100\,(\%) \qquad (1)$$

where $A(f, t)$ is the power spectrum density of the EEG at a certain frequency band f[Hz] and time t[s] since the imagery task was started, and $R(f)$ is the power spectrum at the same frequency f[Hz] of the baseline period (a 1-s interval before the direction of "Ready" was displayed). The largest power decrease during motor imagery was selected as the value of mu ERD. The values of mu ERD before tDCS application were compared in all adjacent pairs of bipolar derivations of EEG and determined the electrode pairs showing the strongest value of mu ERD for individuals. The values of mu ERD in two stimulation conditions (anodal and sham stimulation) were calculated from the same bipolar derivation of EEG. All off-line analyses of EEG data were performed using MATLAB (The MathWorks, Inc. USA).

Transcranial direct current stimulation (tDCS)

The tDCS was applied through rectangular saline-soaked sponge electrodes (50 × 70 mm) with a battery-driven stimulator (CX-6650, Rolf Schneider Electronics, Gleichen, Germany). In the stroke participants, the position of M1 of the affected hemisphere was determined as the symmetrically opposite side of M1 of the unaffected

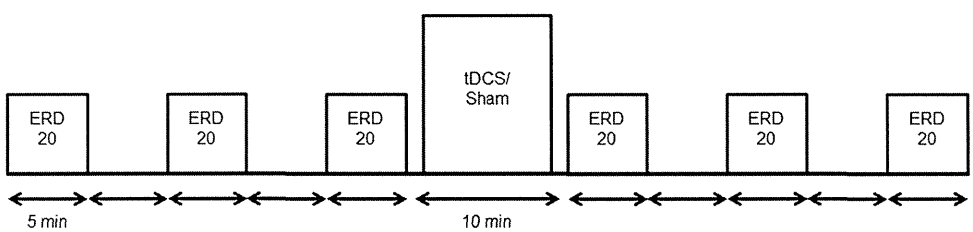

Fig. 1 The paradigm of the experiment. We assessed the ERD during imagery of the affected fingers extension just before and after the anodal and sham tDCS over the motor area of the affected hemisphere. The order of the stimulations was randomized, and the interval between the stimulation was more than 2 days. One ERD assessment session consisted of 20 trials. One trial consisted of an 8-s period of relaxation, a 2-s period of ready state, and a 5-s period of imagery. Before and after tDCS or sham stimulation, three sessions were conducted with approximately 5 min of rest between each session

hemisphere confirmed by the induction of the largest motor-evoked potentials (MEPs) in the unaffected EDC muscle with constant stimulus intensity using TMS with a figure-eight stimulation coil connected to a Magstim 200 magnetic stimulator (Magstim, Whitland, UK). This is because the MEPs could be evoked in the unaffected hemisphere but not in the affected hemisphere in all stroke participants. The anode was placed over M1 of the affected hemisphere, and the cathode was placed over the opposite side in the supraorbital region. In the active condition, tDCS was applied for 10 min with a current intensity of 1 mA. Participants sat awake in a comfortable armchair during the stimulation. In the sham stimulation, the electrodes were arranged similarly to the anodal stimulation and applied stimulation within the first 10 s only to mimic the transient skin sensation at the beginning of actual tDCS without producing any conditioning effects on the brain (Furubayashi et al. 2008). In the healthy participants, the anode was placed over the left M1 determined by TMS, and the cathode was placed over the right supraorbital region. TDCS was applied to them for 10 min with a current intensity of 1 mA. To place the tDCS electrodes on the head, 3 to 4 EEG electrodes over the stimulus site were removed after marking the scalp. After the stimulation, the EEG electrodes were set again on the same position as before. Because it took less than 3 min for electrode replacement, the effect of elapsed time after tDCS on the ERD measurement was limited.

Statistical analysis

To analyze the difference in mu ERD value and baseline EEG power spectrum with stimulation (both anodal tDCS and sham stimulation), Wilcoxon signed-rank test was used. To compare the mu ERD value between the stroke and healthy participants, Mann–Whitney test was used. Statistical analysis was performed with SSPS 18.0 J (SSPS Japan).

Results

None of the participants reported any adverse effects during or after the experiment. All participants showed mu ERD during motor imagery. The changes of ERD with tDCS in the stroke and healthy participants were shown in Fig. 2. Anodal tDCS significantly increased mu ERD in both the stroke ($P = 0.028$) and healthy participants ($P = 0.018$), though we did not find any significant change of mu ERD in the stroke participants with sham stimulation ($P = 0.084$). The mean (SD) values of mu ERD in the stroke participants were 21.2 % (11.7) before anodal tDCS and 21.4 % (15.8) before sham stimulation, and there was no significant difference in the baseline mu ERD values ($P = 0.818$). The mean mu ERD value (SD) before tDCS in the healthy participants was 28.0 % (7.2). The mu ERD values before tDCS in the healthy participants were relatively larger than those in the stroke participants, though the difference was not significant ($P = 0.317$).

We found no significant difference in the baseline EEG power spectrum between before and after tDCS in both the stroke and healthy participants. The mean (SD) value of the baseline EEG power spectrum was 0.46 μV^2 (0.24) before anodal tDCS and 0.47 μV^2 (0.26) after anodal tDCS in the stroke participants ($P = 0.715$), and 0.23 μV^2 (0.12) before anodal tDCS and 0.27 μV^2 (0.15) after anodal tDCS in the healthy participants ($P = 0.398$).

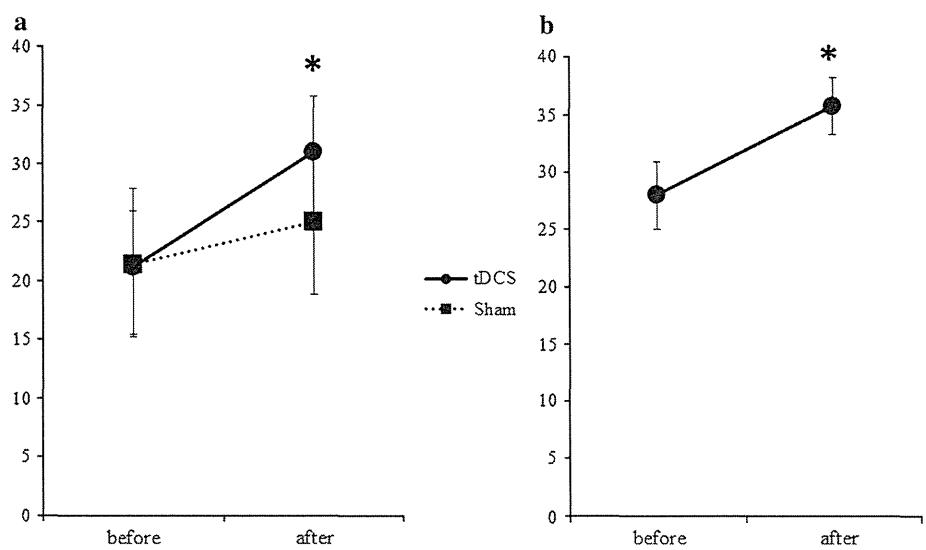

Fig. 2 Changes of mu ERD during the motor imagery with tDCS in the stroke participants (**a**) and the age-matched healthy participants (**b**). The *circle* shows the mean ERD before and after anodal tDCS, and square shows the mean ERD before and after sham stimulation. *Error bars* are standard errors. *Wilcoxon signed-rank test $P < 0.05$

Discussion

We found that anodal tDCS was able to increase ERD during imagery of extension of the affected fingers in patients with chronic severe hemiparetic stroke as same as age-matched healthy persons. This result was similar to younger healthy persons as demonstrated by Matsumoto et al. (2010). It has been reported that anodal tDCS increases cortical excitability (Nitche and Paulus 2000). It was supposed that anodal tDCS increased spontaneous neuronal firing (Bindman et al. 1964; Purpura and McMurtry 1965) and depolarization of the resting membrane potentials (Bindman et al. 1964; Nitche and Paulus 2001; Nitsche et al. 2003). The mechanism of ERD is thought to be a decrease in synchrony of the underlying neuronal population (Pfurtscheller and Lopes de Silva 1999). Therefore, modulation of ERD with tDCS could be explained by changes in the oscillatory behavior of cortical neurons, such as membrane potentials in the primary motor area, and the neurons firing according to input signals in response to motor imagery. An increase in cortical excitability, such as depolarization of the membrane potential of the cortical neurons in the M1, will result in more activated and desynchronized neurons, based on the input signals from motor imagery, which will strengthen ERD.

EEG patterns in patients with stroke are different from those in healthy subjects. Platz et al. (2000) showed that stroke patients with somatosensory deficits had reduced alpha centroparietal ERD during movement preparation and execution. We found that the baseline ERD values of the stroke patients in this study were relatively smaller than the values of the age-matched healthy participants. Anodal tDCS may lead to normalization of the pattern of EEG by increasing ERD of the affected hemisphere in patients with severe hemiparetic stroke. Since ERD was fully detected in every patient before tDCS, it might be interesting to replicate this study in more severe patients in order to fully test for the interest of the present findings.

There are several limitations to be considered in this study. First, we determined the position of M1 of the affected hemisphere using the symmetrical opposite side as a marker, that is, M1 of the unaffected hemisphere. This is not the exact position decided by motor-evoked potential (MEP) of the affected EDC by directly stimulating the affected hemisphere. This is because MEP could not be evoked from the affected EDC. Second, because we used fairly large (5 cm × 7 cm) electrodes for tDCS, we could not exclude the aftereffect of the premotor cortex or sensory motor cortex. Thirdly, there is the possibility that some participants did not imagine well before the stimulation. This is very difficult to assess. It could be supposed that tDCS directly influenced the attention (Kang et al. 2009). Further study of the relationships between modulation of ERD and stimulation site among patients with stroke is needed.

In conclusion, anodal tDCS can increase mu ERD of the affected hemisphere in patients with severe hemiparetic stroke as well as in healthy persons. Therefore, it could be a conditioning tool for EEG-based BCI to make detection of ERD easier.

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