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Cortico-muscular Communication during the Generation of Static Shoulder Abduction Torque in Upper Limb Following Stroke

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Abstract— **In this study, we introduced a new index, namely overlap index, to quantify the spatial resolution of cortical activity for muscle coordination based on the measurement of EEG-EMG coherence during a motor task. By applying this index on 4 control and 4 hemisphere chronic stroke subjects we successfully identified that there is a significantly increased overlap between biceps brachii at the elbow and intermediate deltoids at the shoulder when stroke subjects generating a static shoulder abduction torque. Muscles that have increased overlap in cortex are consistent with those that coactivate abnormally in stroke when compared to control subjects. These results not only proof the effectiveness of this index in quantifying the spatial resolution of cortical activity but also point out that the reduced spatial resolution of muscle activity in cortex can be a reason for the abnormal muscle coactivation observed in impaired arms following stroke. Quantification of the cortical overlap index will provide us with new tools to test for the modifiability of the nervous system following clinical interventions. This work will be an important step toward our long-term goal of developing more effective rehabilitation techniques for the treatement of stroke.**

I. Introduction

More than 50% of stroke patients are left with a residual motor deficit, especially reflected as the loss of independent control of joint movement associated with abnormal muscle co-activation patterns in the impaired limb. Quantitative investigation of these stereotypic movement patterns has revealed an abnormal linkage between the activations of shoulder abduction (SABD) and elbow flexion [1-4], and closely parallel the abnormal muscle co-activation patterns reported previously [5]. The abnormal muscle co-activation patterns may reflect brain-reorganization-related changes in descending commands following stroke.

Reorganization of the motor cortices after stroke has been assessed by microelectrode stimulation techniques in animal models and by different functional neuroimaging modalities, such as positron emission tomography (PET), functional magnetic resonance imaging (fMRI), and transcranial magnetic stimulation (TMS) and reconstruction of cortical activity based on electro-encephalography (EEG) and/or magneto- encephalography (MEG) in human during the recent 15 years. However, imaging techniques, such as fMRI, PET, or inverse calculation based on EEG/MEG are not able to identify specific muscle-activation-related cortical activity during a motor task. Therefore, the relationship of brain activations to the abnormal muscle co-activation after stroke remains unclear; hence requires for the EEG/EMG coherence approach proposed in this study.

Functional neuroimaging using coherence between EEG and electromyographic (EMG) is a new and promising method, which can be used to detect the role of cortical reorganization in the abnormal muscle co-activation following stroke. Skilled movements involve the cooperation of many muscles [6]. One central question of motor coordination is how muscles are grouped during different motor tasks. A straightforward answer to the above question is that movements are controlled by intricate communication systems between the sensorimotor cortex in the brain and skeletal muscles. This communication is in part oscillatory [7-10], which is reflected in rhythmic interaction between signals recorded simultaneously from the brain and muscles. It is logical to hypothesize that due to the brain lesion a loss of spatial resolution occurs, which reflected as an increased overlap between coherence distributions of different muscles, and thus leads to obligatory recruitment of muscles demonstrated as abnormal muscle co-activation patterns as observed in chronic stroke. Up to now, there is no validated method to quantify the spatial resolution of muscle-activation-related cortical activity. Therefore, in this study we develop a simple index, namely overlap index, based on EEG-EMG coherence method to quantify such a spatial resolution in cortex.

This study is the first brain imaging study that using a quantitative way to investigate spatial resolution of muscle-activation-related cortical activity. Results of this study provides evidence of an increased overlap in muscle-activation-related cortical activities following stroke, and thus, is able to quantify an expected decrease in cortical spatial resolution for recruiting individual arm muscles for motor task after brain injury.

II. Method

A. Subject

Four able-bodied subjects (all male, 42±13 years old with right hand dominance) and four chronic stroke subjects (see

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Table 1 for stroke subject information) participated in our preliminary studies. All subjects provided written consent prior to participation in the study that was approved by the Institutional Review Board of Northwestern University and in compliance with the principles of the Declaration of Helsinki.

TABLE I
STROKE SUBJECT INFORMATION

Patient	Age	Sex	Affe-cted hand	Domi-nant hand	Site of Lesion	Time to exp.	Fugl-Meyer Score	Sen-sory loss
S1	60	M	L	L	R. Posterior limb of IC	1.5	26/66	No
S2	47	M	L	R	R. IC & R. Put & R. CI	8	32/66	No
S3	57	M	L	R	R. thalamus & IC-G	4	27/66	No
S4	46	M	R	R	L. IC & L. LS & L. SMA	1.5	9/66	No

**S are stroke subjects, M=male, F=female, R=right limb, L=left limb, N/A=not available, IC=Internal Capsule, SMA= supplementary motor area, LS= lateral sulcus, IC-G= internal capsule, Genu, Put= putamen, CI= claustrum*

B. Experimental setup

Participants sat in a Biodex chair that completely supported the trunk. The trunk was restrained to the back of chair with straps crossing the chest and abdomen to prevent trunk and pelvis motion during the experiment. Subjects were casted at the wrist and secured to a six degree of freedom load cell with the shoulder at 75° abduction, 40° flexion and the elbow at a 90° flexion angle.

The motor tasks involved in the study were self-initiated torque generated in the SABD direction from resting to 25% of subject's maximum voluntary torque (MVT). In each of the directions, subjects were required to perform 100-150 trials. At the very beginning of each trial, an auditory signal was given to the subject indicating the start of the task. After that, subjects were required to maintain a resting state for 5-7 seconds, and then, to self-initiate the generation of torque in the required direction to 25%±8.75% of their MVT and to hold it for 2s. Subjects were instructed to avoid eye movements and movements in other parts of their body during the performance of each trial. Prior to the data collection session, subjects went through a training session to make sure that they were able to perform the torque generation task without visual feedback.

During the data collection session, MVTs and maximum EMGs of each subject were recorded in four randomly ordered blocks consisting of the torque generation of shoulder flexion/extension, shoulder abduction/adduction, shoulder external/internal rotation, and elbow flexion/extension. Subjects then performed the required motor tasks without visual feedback and received *post hoc* feedback about their performance at the end of each trial. In an effort to avoid fatigue, subjects completed 100 to 150 trials in several blocks (20-30 trials for one block) with rest periods of 15 seconds between trials, and 20 minutes between blocks.

C. Data collection

We simultaneously collected force/moment signals, 8-channel EMG signals from both arms and 163-channel EEG signals during each data collection session. Forces and torques generated at the wrist were measured using a six degree-of-freedom load cell (JR^3 Inc., Woodland, CA) and then converted online to torques at the elbow (EF/EE) and shoulder (SF/SE, SABD/SADD, and SEXT/SINT) based on a free body analysis of the upper limb [11].

Five surface EMG signals were recorded from the biceps brachii (BIC), brachioradialis (BRD) muscles at the elbow, the anterior (ADL), intermediate (IDL), posterior deltoids (PDL) from the paretic (stroke subject) or dominant (able-bodied) upper limb. In addition, the BIC, triceps brachii lateral head, IDL in the contralateral arm were recorded to test for voluntary muscle activation.

Scalp recordings were made with a 163-channel EEG system using active electrodes (Biosemi, Inc., Active II, Amsterdam, The Netherlands). The electrodes were mounted on a stretchable fabric cap based on a 10/20 system. All data were sampled at 1000 Hz. Collection of the EEG data was synchronized with torque and EMG data using a TTL pulse generated when the primary torque exceeded 0.15 Nm to mark the on-line detected torque onset.

D. Data analysis

1) Torque data: Torque signals were initially visually inspected. Trials with artifacts were eliminated. Remaining shoulder and elbow torques were baseline-corrected and averaged through a 250 ms moving window. The torque responses were then aligned with the off-line adjusted toque onset based on the TTL signal and then ensemble-averaged. Torque responses were then normalized by the MVT yielding in % MVT for each degree of freedom.

2) EMG data for quantifying the muscle coupling patterns: Trials with non-desired EMG activity or with low SNR in the EMG/EEG recordings in the paretic (stroke) or dominant (able-bodied) arm were eliminated. The EMG signals were rectified and baseline corrected (by subtracting the average value in the window from the beginning of the trail to 200 ms before the onset of torque). Movement artifacts were removed (by replacing the values larger than mean+3 times STD inside a window by the mean value in that window) and a low pass filter (zero lag, 6^{th} order Butterworth) with a cutoff frequency at 30 Hz was applied to obtain the envelope of each EMG signal. Off-line detected torque onsets were used to align EMGs from individual trials and compute the ensemble average across trials. Averaged EMGs were normalized to the highest EMG value obtained during the MVT efforts.

3) EMG and EEG data preprocessing for calculating the EEG-EMG coherence: The EEG and EMG signals were first visually inspected for removing the trials with the presence of artifacts. Then, a finite difference Surface Laplacian [12] transformation was applied to each EEG channel as a spatial high-pass filter to reduce the smearing effects caused by head volume conductor and to increase the SNR [13]. This causes the reduction of electrode number from 163 to 131 on the scalp by losing the most outer electrodes. The 'end of trial' was used to align and segment remaining EEGs and EMGs for individual trials. Then segmented signals (0 to -2 s with 0 indicating the end of the trial) was down sampled to 256 Hz and linked together, which resulted in a linked EEG/EMG with more than 2-minute long. The linked EEG and EMG signals were then be band-pass filtered (zero lag, 6th order Butterworth, with a 1 s non-overlapped Hamming window applied) with cutoff frequencies at 5 and 50 Hz.

4) EEG-EMG coherence: EEG and EMG power spectra and the coherence between EEG and EMG were computed using two contiguous 2-s-long sections of EEG and EMG data with a 1-s-long window. This yielded a frequency resolution of 1 Hz. Auto-spectra of *EEG* and *EMG* signals ($P_{EEG,EEG}$, $P_{EMG,EMG}$) and cross-spectra between signals from *EEG* and from *EMG* ($P_{EEG,EMG}$) was calculated using Welch method (i.e., Hamming window was applied when using Fourier Transform). Coherence ($Coh_{EEG,EMG}(f)$) at frequency f Hz was then calculated as:

$$Coh_{EEG_k,EMG_i}(f)=\frac{|P_{EEG_k,EMG_i}(f)|^2}{P_{EEG_k,EEG_k}(f)\times P_{EMG_i,EMG_i}(f)},$$

where k and i are the indices of EEG channels on sensorimotor cortices and EMG channels, respectively.

5) Calculating the overlap index between each pair of the muscles: The overlap index (OI) of the EEG-EMG coherence distribution within the β-band (12 Hz to 30 Hz) between each pair of muscles during the generation of a torque task was calculated in the window of holding phase as:

$$OI=\frac{\sum_{k}^{K}\left(\max_{12\le f\le 30}\left\{Coh_{EEG_k,EMG_i}(f)\right\}\times\max_{12\le f\le 30}\left\{Coh_{EEG_k,EMG_j}(f)\right\}\right)}{\sum_{k}^{K}\max_{12\le f\le 30}\left\{Coh_{EEG_k,EMG_i}(f)\right\}\times\sum_{k}^{K}\max_{12\le f\le 30}\left\{Coh_{EEG_k,EMG_j}(f)\right\}}$$

where K is the total number of electrodes in sensorimotor cortices.

III. Results

Mean and standard error of torques during the holding phase in 4 control and 4 stroke subjects for generating SABD are shown in figure 1. These results demonstrate that there is a significant larger ($p<0.05$) coupling between SABD and EF in the stroke subjects during the generation of SABD torque at the level of 25% of MVT.

The group mean and standard error of normalized EMGs across stroke and able-bodied subjects from muscles during the holding phase of 25% SABD torque generation are shown in figure 2. One-way ANOVA (group) analysis showed that the activities of BIC in stroke subjects during the generation of SABD were significantly greater than that observed in control subjects ($p<0.0001$), which suggests a higher muscle co-activation between BIC and shoulder abductors in stroke subjects than that in control subjects.

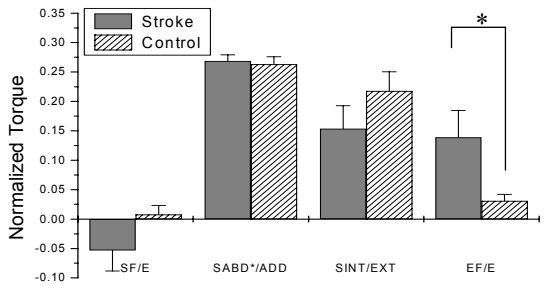

Fig. 1. The mean and standard error (showed by the error bar) of torque during the generation of SABD torque in 4 control and 4 stroke subjects. Significantly larger (*P<0.05) secondary torques in the EF direction were found in stroke subjects.

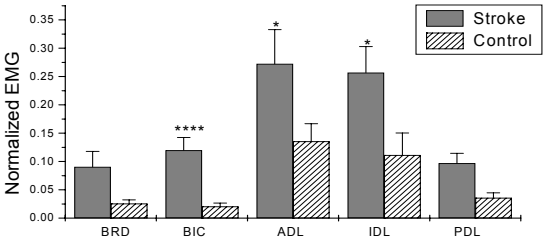

Fig. 2. The group mean and standard error of normalized EMG during the holding phase of SABD and EF torque generation from each of the nine muscles in stroke and control subjects (*P<0.05, **P<0.01, ***P<0.001, ****P<0.0001).

In each of the subjects, we investigated the OI to quantify the overlap of EEG-EMG coherence distribution between different combinations of muscles. A larger OI indicates a larger overlap in the EEG-EMG coherence distribution between two muscles. The analysis of OI focused on coherence in the β-band (12 Hz to 30 Hz) because this band is normally reported to show coherence with muscle activity [14-17]. Results are shown in figure 3. It is obvious that there is a significant overlap between BIC and IDL in stroke subjects when they generating SABD torque. This is in accordance with the results of EMG recordings, which shows significantly larger coactivation between BIC and IDL activities during the generation of SABD or EF torque following stroke (see figure 2).

IV. Discussion

EEG-EMG coherence method was first applied in a patient with intractable seizures [18]; now, is more widely used in studies involving various neural pathologies, including stroke [19, 20]. However, the previous EEG-EMG coherence studies only investigated the coherence between

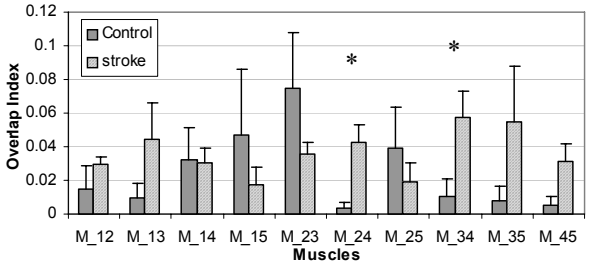

Fig. 3. The group means and standard errors of OI of EEG-EMG peak coherence distribution between different combinations of muscles during the generation of SABD. The indices of muscles represent: 1: BRD; 2: BIC; 3: ADL; 4: IDL; 5: PDL. * P<0.05.

several EEG channels (less than 56) and 1-2 muscles. In this study, we used a high-density EEG system combining a time resolution in the ms range with a spatial resolution of about 1~2 cm. This spatial resolution although not perfect, is enough to show the changes of an individual's muscle-activation-related overlap at the cortex following stroke.

More importantly, we measured overlap between different muscle-activation-related cortical activities, i.e., a loss of spatial resolution at the cortex, for the first time in this study. We found a decrease in cortical spatial resolution for recruiting individual arm muscles for motor task after brain injury. Furthermore, our results showed that muscles with an increased overlap of their EEG-EMG coherence distributions (figure 3) are consistent with those that coactivate abnormally in stroke when compared to control subjects (figure 2). This points out that this overlap in cortical activity can be a reason underlying abnormal muscle coactivation patterns following stroke.

Measurement of overlap in critical activity for different muscles based on coherence between high-density EEGs with multiple EMGs is a new technique to quantify the spatial resolution of cortical activity. This measurement can be used to evaluate the effect of a rehabilitation method on brain reorganization. Additionally, by identifying muscles that have increased overlap in cortical representations and are associated with abnormal muscle co-activations will give us new insights for the design of interventions that seek to overcome abnormal coordination.

REFERENCES

[1] R. F. Beer, J. D. Given, and J. P. A. Dewald, "Task-Dependent weakness at the elbow in patients with hemiparesis," *Archives of Physical Medicine and Rehabilitation*, vol. 80, pp. 766-772, 1999.

[2] D. Bourbonnais, "Abnormal spatial patterns of elbow muscle activation in hemiparetic human subjects.," *Brain*, vol. 112, pp. 85-102, 1989.

[3] J. Dewald and R. Beer, "Evidence for abnormal joint torque patterns in the paretic upper limb of subjects with hemiparesis.," *Muscle & Nerve*, vol. 24, pp. 273-283, 2001a.

[4] J. Dewald, V. Sheshadri, M. Dawson, and R. Beer, "Upper limb discoordination in hemiparetic stroke: Implications for neurorehabilitation.," *Topics in Stroke Rehabilitation*, vol. 8, pp. 1-12, 2001b.

[5] J. P. Dewald, S. P. Pope, J. D. Given, T. S. Buchanan, and W. Z. Rymer, "Abnormal muscle coactivation patterns during isometric torque generation at the elbow and shoulder in hemiparetic subjects," *Brain*, vol. 118, pp. 495-510, 1995.

[6] N. Murayama, Y. Y. Lin, S. Salenius, and R. Hari, "Oscillatory interaction between human motor cortex and trunk muscles during isometric contraction," *Neuroimage*, vol. 14, pp. 1206-13, 2001.

[7] S. N. Baker, J. M. Kilner, E. M. Pinches, and R. N. Lemon, "The role of synchrony and oscillations in the motor output," *Exp Brain Res*, vol. 128, pp. 109-17, 1999.

[8] S. F. Farmer, "Rhythmicity, synchronization and binding in human and primate motor systems," *J Physiol*, vol. 509 (Pt 1), pp. 3-14, 1998.

[9] R. Hari and S. Salenius, "Rhythmical corticomotor communication," *Neuroreport*, vol. 10, pp. R1-10, 1999.

[10] W. Singer and C. M. Gray, "Visual feature integration and the temporal correlation hypothesis," *Annu Rev Neurosci*, vol. 18, pp. 555-86, 1995.

[11] R. F. Beer, E. R. Johnston, and E. Eisenberg, *Vector mechanics for engineers: statistics*. New York: McGraw-Hill, 1995.

[12] R. D. Pascual-Marqui and R. Biscay-Lirio, "Spatial resolution of neuronal generators based on EEG and MEG measurements," *Int J Neurosci*, vol. 68, pp. 93-105, 1993.

[13] B. Hjorth, "An on-line transformation of EEG scalp potentials into orthogonal source derivations," *Electroencephalogr Clin Neurophysiol*, vol. 39, pp. 526-30, 1975.

[14] M. R. Baker and S. N. Baker, "The effect of diazepam on motor cortical oscillations and corticomuscular coherence studied in man," *J Physiol*, vol. 546, pp. 931-42, 2003.

[15] S. N. Baker, E. Olivier, and R. N. Lemon, "Coherent oscillations in monkey motor cortex and hand muscle EMG show task-dependent modulation," *J Physiol*, vol. 501 (Pt 1), pp. 225-41, 1997.

[16] B. A. Conway, D. M. Halliday, S. F. Farmer, U. Shahani, P. Maas, A. I. Weir, and J. R. Rosenberg, "Synchronization between motor cortex and spinal motoneuronal pool during the performance of a maintained motor task in man," *J Physiol*, vol. 489 (Pt 3), pp. 917-24, 1995.

[17] J. M. Kilner, S. N. Baker, S. Salenius, R. Hari, and R. N. Lemon, "Human cortical muscle coherence is directly related to specific motor parameters," *J Neurosci*, vol. 20, pp. 8838-45, 2000.

[18] R. S. McLachlan and L. W. Leung, "A movement-associated fast rolandic rhythm," *Can J Neurol Sci*, vol. 18, pp. 333-6, 1991.

[19] T. Mima and M. Hallett, "Corticomuscular coherence: a review," *J Clin Neurophysiol*, vol. 16, pp. 501-11, 1999.

[20] T. Mima, K. Toma, B. Koshy, and M. Hallett, "Coherence between cortical and muscular activities after subcortical stroke," *Stroke*, vol. 32, pp. 2597-601, 2001.