Phase and magnitude spatiotemporal dynamics of β oscillation in electrocorticography (ECoG) in the monkey motor cortex at the onset of 3D reaching movements

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***Abstract*—β oscillations in local field potentials, electrocorticography (ECoG), and electroencephalograms (EEG) are ubiquitous in the motor cortex of monkeys and humans. However due to their lack of contributions, compared to other frequency ranges, to decode effector kinematics especially in ECoG signals, spatiotemporal dynamics of ECoG β oscillations has not been examined despite the larger areas that ECoG arrays can cover than standard intracortical multielectrode arrays. Here, we used ECoG grids to cover large areas of motor cortex and some somatosensory cortex in monkeys while they performed an unconstrained reaching and a lever pulling task at two force levels in three dimensional space. We showed that under the pulling task β power increased around movement onset. However, the β phases were locked around the movement onsets and their peak timings were spatially aligned in the motor cortex. These results may indicate that spatiotemporal dynamics of β oscillation conveys task relevant information and that ECoG arrays will be useful to study larger spatiotemporal patterns in the motor cortex, or any cortical areas in general, than intracortical multielectrode arrays.**

***Index Terms*—Motor cortex, Electrocorticography (ECoG), β oscillations, spatiotemporal dynamics, non-human primates, kinetics**

I. Introduction

Electrocorticography (ECoG) has been widely used both in humans [1], [2] and non-human primates [3], [4] due to its stable recording and less invasive surgical procedures than intracortical multichannel electrode arrays. In particular, ECoG arrays implanted in motor cortical areas have been used as a means to record signals that were relevant to decode kinematics of the effectors [1], [4]. However, despite its larger spatial coverage over the cerebral cortex than that of intracortical multichannel electrode arrays, detailed study of spatiotemporal dynamics of ECoG signals has hardly been performed. In particular, since the most prominent oscillation frequency in the motor cortex, a broadly defined β range, $15 \sim 30$ Hz, has been show to contribute little to decoding performance [5], dynamics of β oscillations in ECoG signals has received little attention.

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Beta oscillation in local field potentials (LFPs) and electroencephalograms (EEG) in the motor cortex has been prevailed in mammalian motor cortex including monkeys and human [6], [7], [8], [9], [10], [11]. Power of β oscillation is enhanced around the external cue onset and is attenuated around the movement onset [12] for point-to-point discrete movements. However, temporal dynamics of β oscillation in the motor cortex when subjects need to exert forces at different levels has not been studied much [13]; rather corticospinal coherence over γ range is modulated based on force levels [14]. Thus, in our current study, we examined the temporal and spatial aspects of β oscillations in ECoG signals over the motor cortex using two tasks: unconstrained 3D reaching task and level pull task with two different force levels.

II. Materials and methods

All experimental procedures were performed in accordance with the Guidelines for Proper Conduct of Animal Experiments of the Science Council of Japan and approved by the Committee for Animal Experiment at the National Institutes of Natural Sciences (Approval No.: 11A157). The data presented for all experimental sessions were obtained from two Japanese monkeys (*Macaca fuscata*).

A. Behavioral Task

Two monkeys (Monkey A: female, 4.7 kg; Monkey B: male, 8.9 kg) were trained to perform a uni-manual reach-to-grasp task (Fig. 1(A)). The monkeys kept pushing the home button for 1 sec (monkey A) or 2 sec (monkey B) till a go cue. Then monkeys reached for and grasped an object (monkey A) or pulled a lever at two different levels of forces (monkey B), and retrieved their hands at the home position again. We defined reaching onset as the time when the monkeys hand left from the home button and an individual trial period was defined as the period between 250 ms before and 250 ms after the reaching onset. Monkey A performed a total of 297 successful trials, and monkey B performed 76 trials in each force level in this study.

B. Implant surgery for an ECoG array

Both monkeys underwent surgery on different days to implant an ECoG array under anesthesia after they completed behavioral training. The monkeys were anesthetized with ketamine (1.0 mg/kg) and xylazine (0.5 mg/kg). The

978-1-4244-7929-0/14/$26.00

inhalation of $1 \sim 2\%$ isoflurane maintained anesthesia during the surgeries.

A platinum ECoG array (a diameter of circular electrode: 1 mm, an inter-electrode distance: 3 mm; Unique Medical Corporation, Tokyo, Japan) was chronically implanted over the left primary motor cortex (M1), which had 15 (monkey A: 5×3 grid) and 32 (monkey B: 4×8 grid) channel electrodes (Fig. 1(B) and (C) respectively). Electrode locations were identified from anatomical views during surgery. Four silver wires as reference and ground electrodes were inserted to the subdural space. The wires were shunted (single-end mode) via connectors.

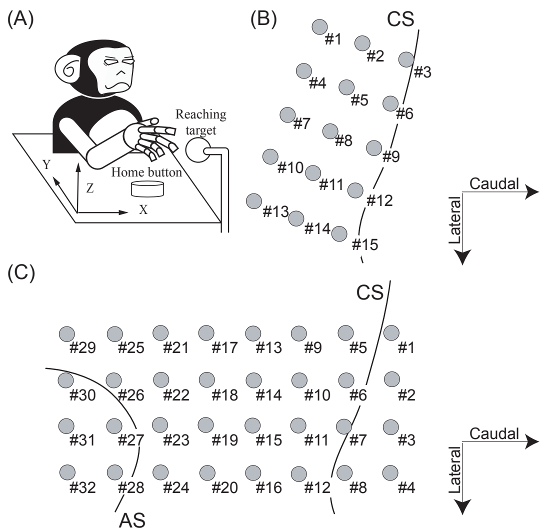

Fig. 1: Experimental setup. (A) Behavioral configuration. (B) Channel configuration and location of the 15-channel ECoG grid in monkey B. (C) Channel location of the 32-channel ECoG grid in monkey A. CS: Central sulcus, AS:Arcuate Sulcus

C. Recorded kinematics of monkey arm movements

The 3D position of the monkey arm was recorded using reflective markers tracked by an optical motion capture system (Eagle digital system; Motion Analysis Corporation, Santa Rosa, CA). The system used 12 infrared cameras operating at 200 frames/s to track the positions of multiple reflective markers (4-mm-diameter spheroids) with submillimeter accuracy. A total of 15 markers were attached to the forelimb of the monkey from the shoulder to the fingers. A kinematic model of the arm and hand was matched with the observed marker data using the systems software package (EVaRT 5.0.4) before analyzing the movement data. Sequential data of the maker position in the orthogonal coordinate system were linearly interpolated and then filtered (low pass with cutoff edge at 10Hz) in each orthogonal coordinate axis. In addition to the optical data, the motion capture system also recorded the analog signals from the external stimulator (SEN-8203; Nihon Kohden Corporation, Tokyo, Japan) as the synchronized time-stamped data with neural and muscle recordings. The motion data were then up-sampled to 500 samples per second to match the neural and muscle data. Wrist speed profiles for the three conditions (normal reach to grasp, low force reach to grasp, and high force reach to grasp) examined in this studies show comparable trajectories.

D. Recording of ECoG signals

ECoG signals were sampled at 4 kHz (Plexon MAP system; Plexon, Inc., Dallas, US). ECoG signals were filtered with band-pass filters through amplifiers (monkey A: 1.5 Hz high-pass and 1 kHz low-pass analog filters, MEG-6116, Nihon Kohden Corporation, Tokyo, Japan; monkey B: 0.7 Hz high-pass and 8 kHz low-pass analog filters, Plexon, Inc., Dallas, USA). The data of the ECoG contained synchronized signals for *post hoc* matching of the time-lines to the marker-position data from the motion capture system. The data was first bidirectionally bandpass filtered over $1-250$ Hz and then down-sampled to 500 samples per second to match the marker-position data.

E. Analysis of ECoG signals

All ECoG signals spanning ± 250 ms around the movement onsets were analyzed in this study. Multitaper spectrum methods were used with Chronux Matlab library[15] with the time-bandwidth product of 3 and 5 tapers were used. The β frequency range computed over all valid trials for each condition for this animal was found to be $15 \sim 21$ Hz across most of the channels. For the amplitude and the phase of β oscillation, first raw ECoG signals were bidirectionally bandpass filtered over $[15,21]$ Hz cutoff with a 3rd order Butterworth filter. Then, Hilbert transform was performed to the filtered signal to compute the amplitude and phase of the transformed signals. For the amplitude, an average of the amplitude of β oscillation over all the valid trials for each channel for each condition was computed. In order to assess the locking of β oscillation recorded at each channel to the movement onset, a percentage of phase locking (PPL, as in [12], [16]) was computed at time t as

$$\begin{aligned} PPL(chan,t) &= 100(1-H(\phi(chan,t))/H_{max}), \\ H(\phi(chan,t)) &= -\sum_{k=1}^{N} p_k log_2(p_k), \\ H_{max} &= log_2(N), \end{aligned}$$

where $\phi(chan,t)$ is an instantaneous phase at time t of the β band signal recorded by ECoG electrode *chan*, N is the number of trials, and $H(\cdot)$ is the Shannon entropy where p_k is the fraction of the values of $\phi(chan,t)$ for *chan* at time t that lie within the kth in out of12 bins used in the current study.

III. Results

A. Spectral properties of β oscillations in ECoG Signals

During normal reach to grasp task, power in β oscillation attenuates in all the channels in the motor cortex [12] as well

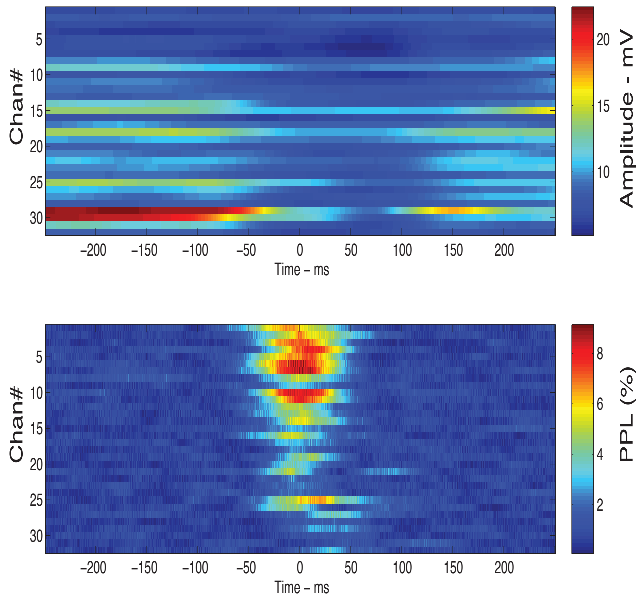

Fig. 2: Amplitude and phase characterization of β oscillations around the movement onset for the 32-channel ECoG grid. Top panel shows the average amplitude in mV of β oscillations and bottom panel shows the percentage of phase locking of β oscillations ranging from 0 to 9 %.

as the channels placed over somatosensory cortex (Fig. 2 (A)) and strong phase locking happens among most channels that are placed rostral to CS (Fig. 2 (B)).

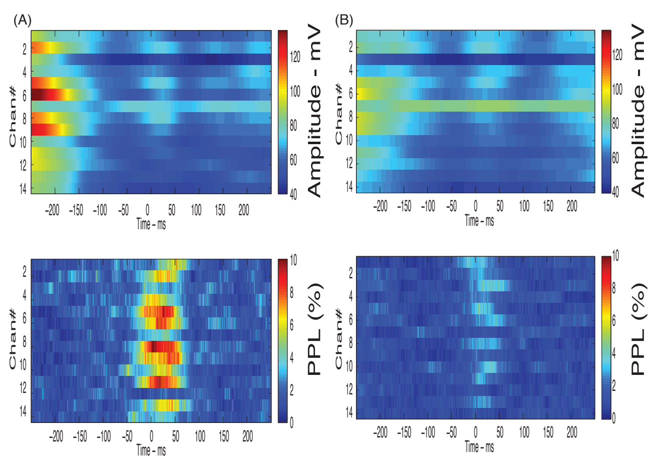

Fig. 3: Amplitude and phase characterization of β oscillations around the movement onset (time = 0 ms) for the 15-channel ECoG grid for unconstrained reach to grasp task. (A) Large force condition (B) Small force condition. Top panel in each column shows the average amplitude of β oscillations in mV and bottom panel shows the percentage of phase locking of β oscillations, respectively.

During the pull lever task, β power increased slightly, especially on the medial side of the array, around the movement onset (Fig. 3 top panels of (A) and (B)), and slightly more increased for the low force condition than the high force condition. However, the PPL for the large force condition across most channels are comparable to those of the normal reach to grasp (Fig. 3 bottom panel of (A)) while much smaller PPL across almost all the channels were observed for the small force condition (Fig. 3 bottom panel of (B)).

B. Spatial property of β oscillations in ECoG signals

As shown in Fig. 4, there are spatial variations in the timings of maximum PPL. Especially among the channels that were placed over the motor cortex, there appears to be a gradient in the timing along caudorostral direction, while channels over somatosensory cortex achieve their PPL peaks after those for channels in pre-central sulcus. Among the pre-central sulcus channels there are spatial gradient in the timings of maximum PPL, especially among the channels that were placed over the motor cortex.

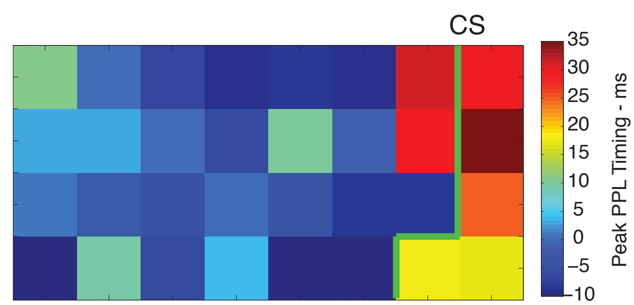

Fig. 4: Spatial distribution of peak percentage phase locking timings of β oscillation for the 32-channel ECoG array around movement onset. The timings in ms are in relation to movement onset. Green line indicates the central sulcus as depicted in Fig.1 (D).

Fig. 5 shows the flow field of the gradient of the peak PPL timing relative to the movement onset in the 32 ECoG grid. The three lower gradient vectors pointing caudal to the central sulcus in the primary motor cortex are all pointing to the central sulcus and resemble the β wave propagation directions axis shown in [12].

IV. Conclusion

We have shown that β oscillations recorded by ECoG arrays exhibited amplitude attenuation around the movement onset as reported previously [12] during normal reach to grasp task while slight amplitude increase when the subject needed to exert forces at the target location. For the normal reach to grasp task, phase locking peaks in the somatosensory cortex occurred later than those in the motor cortex, and within the motor cortex, there appears to be a PPL gradient resembling the propagation axis of β waves observed in monkeys [12]. For the lever pull task, the β power increased briefly around the movement onset (e.g., [13]) especially when the force level was low while the PPL was higher for the high force condition. It is not clear the origin of

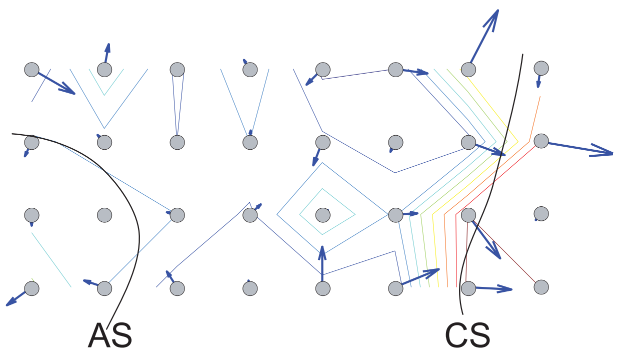

Fig. 5: Spatial gradient flow of peak PPL timings depicted in Fig. 4. The magnitude of each arrow represents the time difference between two adjacent peak PPL values. Each gray circle denotes a location of one of the 32-ECoG channels. The color lines denote the contour plots of peak PPL timings. CS: Central sulcus, AS:Arcuate Sulcus.

this asymmetry across the two conditions, but it is possible that the force information is encoded as a mixture of the magnitude and phase of β oscillation.

The encoded properties in the signals from MI, particularly on kinematics and kinetic variables remain unclear at single neuron and their population level [17], [18]. However, linking cellular level activities and aggregate signals such as ECoG over large spatial coverage may give us better insight to understand encoding properties in MI. Especially in brain machine interface applications, considerable amount of efforts has been made to decode kinematics and kinetic variables using motor cortical unit spiking ensemble activities [19], [20], but not much has been successful to particularly decode kinetic parameters such as end point forces unless inherent effector dynamics was accounted for [19]. Thus, in order for upper limb motor brain machine interface to be practical, our finding focusing on spatiotemporal properties of ECoG recording may give us a further insight as to design a better hinematics and and kinetic hybrid decoder for brain machine interface applications.

V. Acknowledgement

The authors would like to thank M. Togawa, Y. Yamanishi, and N. Takahashi at NIPS and T. Umeda currently at department of neuroanatomy, Yokohama City University for surgery and training of monkeys.

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