Quantification of cortical proprioceptive processing through a wireless and miniaturized EEG amplifier

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A. Giangrande, G. L. Cerone, *Member, IEEE,* M. Gazzoni, *Member, IEEE,* A. Botter, *Member,* IEEE, H. Piitulainen

***Abstract*— Corticokinematic coherence (CKC) is computed between limb kinematics and cortical activity (e.g. MEG, EEG), and it can be used to detect, quantify and localize the cortical processing of proprioceptive afference arising from the body. EEG-based studies on CKC have been limited to lab environments due to bulky, non-portable instrumentations. We recently proposed a wireless and miniaturized EEG acquisition system aimed at enabling EEG studies outside the laboratory. The purpose of this work is to compare the EEG-based CKC values obtained with this device with a conventional wired-EEG acquisition system to validate its use in the quantification of cortical proprioceptive processing. Eleven healthy right-handed participants were recruited (six males, four females, age range: 24–40 yr). A pneumatic-movement actuator was used to evoke right index-finger flexion-extension movement at 3 Hz for 4 min. The task was repeated both with the wireless-EEG and wired-EEG devices using the same 30-channel EEG cap preparation. CKC was computed between the EEG and finger acceleration. CKC peaked at the movement frequency and its harmonics, being statistically significant ($p < 0.05$) in 8–10 out of 11 participants. No statistically significant differences ($p < 0.05$) were found in CKC strength between wireless-EEG (range 0.03–0.22) and wired-EEG (0.02–0.33) systems, that showed a good agreement between the recording systems (3 Hz: $r = 0.57$, $p = 0.071$, 6 Hz: $r = 0.82$, $p = 0.003$). As expected, CKC peaked in sensors above the left primary sensorimotor cortex contralateral to the moved right index finger. As the wired-EEG device, the tested wireless-EEG system has proven feasible to quantify CKC, and thus can be used as a tool to study proprioception in the human neocortex. Thanks to its portability, the wireless-EEG used in this study has the potential to enable the examination of cortical proprioception in more naturalistic conditions outside the laboratory environment.**

***Clinical Relevance*— Our study will contribute to provide innovative technological foundations for future unobtrusive EEG recordings in naturalistic conditions to examine human sensorimotor system.**

I. Introduction

The brain may be considered as a predictive organ [1]. Therefore, it needs to continuously integrate the movement intention and the state of the body to achieve efficient and meaningful motor actions through its motor efference. The state of the body is brought to the brain via sensory afferent information. For motor control, the somatosensory, and especially, proprioceptive afference arising directly from "the movement sensors" (i.e. the proprioceptors) of our locomotor system is particularly important for smooth motor actions, since the proprioception monitors the internal state of the body [2]. Quantifying the cortical proprioceptive processing is challenging because of "brain noise" from task/stimulus irrelevant brain activity and low signal to noise ratio of non-invasive on-scalp recordings. The use of non-invasive neuroimaging techniques such as magnetoencephalography (MEG) and electroencephalography (EEG), have allowed investigation of the body-brain interactions during active and passive tasks (i.e. repeated movements of fingers, toes or lower limbs) [3], [4], with clinical relevance [5]. Corticokinematic coherence (CKC) is a robust method to quantify the cortical processing of proprioceptive afference from the body to the sensorimotor cortices during repetitive movements [6]. It has been shown that CKC has the potential to be used as tool to detect and follow proprioceptive impairments throughout lifespan (from newborns to ageing effects), effectivity of rehabilitation and recovery of sensorimotor impairments (stroke, early detection of CP, neuropathy) [4], [7], [8]. CKC quantifies the coupling between the cortical neuronal activity and kinematics (e.g. acceleration, velocity or displacement) in the frequency domain, quantifying their correlation from 0 (no association) to 1 (perfect association) [3], [9]. To the best of our knowledge only a few studies have used EEG to quantify CKC and they were all carried out exclusively in laboratory environments [7]. The conventional EEG amplifiers are mostly wired and bulky, thus non-portable or hardly configurable to other body sensors to readily record different types of signals throughout the body.

To enhance the possibilities to study the brain-body interaction in naturalistic conditions, we have recently designed and developed an innovative wireless EEG acquisition system that is interfaceable with third party devices (e.g. external acquisition unit, feedback systems etc.) with a high degree of synchronization [10].

The aim of this work is to test the feasibility of the wireless-EEG system to examine human cortical proprioception by comparison to a conventional, wired, EEG-amplifier. This study aims to lay the basis for future CKC studies in naturalistic conditions.

II. METHODS

A. Participants

Eleven healthy participants (six males, four females, age range: 24–40 yr) without neurological or motor disorders were recruited for the study. All the participants were right-handed (mean score 92.4 ± 5.5 on a scale from –100 to 100 [11]). The study was approved by the Ethics Committee of the University

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of Jyväskylä (approval number: 369/13.00.04.00/2020). Written informed consent was obtained from all participants in accordance with the Declaration of Helsinki.

B. Protocol

Experimental procedure. Fig. 1 shows the experimental setup aimed at stimulating the proprioceptors of the index finger. The recordings were carried out in a shielded room at the Centre for Interdisciplinary Brain Research (CIBR) of University of Jyväskylä (Jyväskylä, Finland). A custom-made movement actuator using a pneumatic artificial muscle (Aalto NeuroImaging, Aalto University, Espoo, Finland) was used to evoke continuous flexion-extension movement in the metacarpophalangeal joint of the right index finger [12] at 3 Hz for 4 min. The movement range at the tip of the finger was 5 mm in accordance with [13]. Identical proprioceptive stimulation was repeated separately for wireless-EEG [10] and a wired-EEG (NeurOne Tesla, Oulu, Finland) recordings in pseudorandom order. Participants were comfortably seated on a chair with their right-hand relaxed on the hand-support plate of the movement actuator that was placed on a table in front of them. A cardboard was used to block the view to the hand, thereby blocking any visual contamination. The subject was asked to wear shielded earphones (ER-3C, 50 Ohm, Etymotic Research) to attenuate any possible simultaneous although slight acoustic noise emitted by the airflow within the pneumatic muscle. During the experiment, participants were asked to gaze at a fixation cross displayed on a screen in front of them 1.5 m apart.

Acceleration recordings. A three-axis accelerometer (ADXL335 iMEMS Accelerometer, Analog Devices, Norwood, MA) was taped on the right index finger to record its acceleration in real-time. Analog acceleration signals were low-pass filtered at 330 Hz and sampled at 1 kHz with a 16 bit data acquisition unit (Micro1401-4, Cambridge, England, UK).

EEG recordings. A 32 head-mounted electrodes cap (EasyCap GmbH, Gliching, Germany) was used to record scalp signals. The EEG cap was kept in place between consecutive recordings to compare the performances of the two EEG systems. Ag/AgCl electrodes embedded into the cap were positioned in accordance with the international 10-20 system. Each electrode site was gently scrabbed using an abrasive paste (NuPrep, Weaver and Company, Aurora, USA) after having placed the cap on the scalp. Every cavity was filled with a conductive gel (NeurGel, SPES MEDICA, Genova, Italy). Additionally, electrooculograms (EOG) were recorded time-locked with the EEG using two pre-gelled Ag/AgCl electrodes (30 mm × 22 mm Ambu s.r.l., Denmark) placed in the upper-left and lower-right corners of the eye to detect eye movements and blinks. EEG signals were recorded referenced to the FCz electrode of the cap and sampled at 2048 Hz (wireless EEG) or at 2000 Hz (wired EEG). For the wireless-EEG, the wireless synchronization system introduced in [10] was used to achieve a high degree of synchronization among signals from multiple sources (i.e. EEG and acceleration signals).

C. Data Processing

Pre-processing. Data were imported and fully analyzed in Matlab (R2021a, The MathWorks Inc., MA, USA). The Euclidean Norm of the three orthogonal acceleration signals

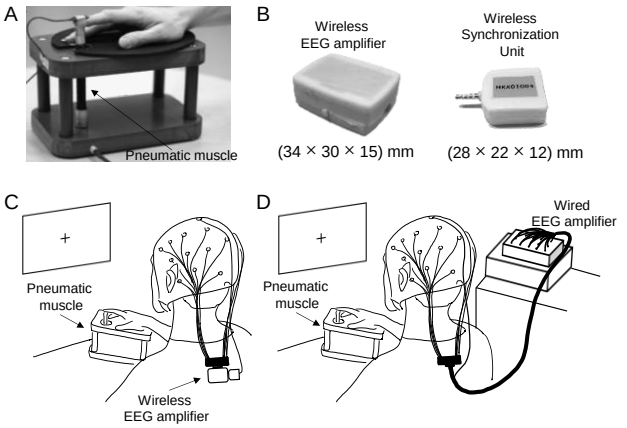

Figure 1. Experimental setup. A: Custom-made movement actuator based on pneumatic artificial muscle with participant's right index finger and the accelerometer attached on it. B: Detail of wireless acquisition and synchronization modules. C: Wireless EEG system setup. D: Wired EEG system setup.

(i.e. the magnitude of the acceleration vector) was then computed. EEG signals were band-pass filtered in the 0.5–95 Hz frequency band (4th-order, zero-phase Butterworth filter). The FieldTrip Matlab toolbox was used for the post-processing of EEG signals. Noisy channels were identified by visual inspection and then replaced with the average of their neighboring channels before further analyses. On average, 2 channels out of 30 were replaced among all the participants. EEG independent components related to eye blink artifact (e.g. those presenting the highest correlation with the time-locked recorded EOG) were identified and removed. Finally, two spatial filters were applied on the 30 monopolar EEG signals: average reference and surface Laplacian [14].

Coherence analysis. The formulation of Halliday et al. [15] was used to compute coherence between EEG and Euclidean norm acceleration of the finger (i.e. CKC), separately for the wireless-EEG and wired-EEG recordings and EEG spatial derivations. A frequency resolution of 0.5 Hz was used for coherence analyses. To this end, continuous data were divided into 2 s epochs with 1 s epoch overlap and coherence spectra between EEG and normalized acceleration epochs were extracted [7], [16]. The magnitude squared coherence was used as coupling measure according to [3], [16]. The strength of CKC was quantified as the peak coherence value among the EEG channels at the movement frequency (F0: 3 Hz) and its first harmonic (F1: 6 Hz), and for each recording and spatial derivation separately. Grand-averaged (i.e. group-level) coherence spectrum was computed by averaging the individual spectra of the participants (n = 11) at their peak channels. Similarly, grand-average topographic scalp distributions of CKC strength at 3 Hz and 6 Hz were obtained.

D. Statistical Analyses

Coherence statistical significance. The statistical significance level (α-level) was set to 0.05/Nc (where Nc = 20 is number of midline and left hemisphere channels), to correct for multiple comparisons of the channel selection. The statistical significance of individual CKC was estimated taking into account the overlapping epochs, according to the hypothesis of linear independence of Fourier coefficients across epochs at each frequency of interest [15].

TABLE I - CKC STRENGTH AND NUMBER OF SUBJECTS SHOWING SIGNIFICANT CKC

EEG recording system [a]	CKC Strength					
	Mean ± SD		*Range*		*#p < 0.05[b]*	
	Average Ref	*Laplacian*	*Average Ref*	*Laplacian*	*Average Ref*	*Laplacian*
Wireless EEG	*F0:* 0.08 ± 0.04 *F1:* 0.12 ± 0.07	*F0:* 0.09 ± 0.07 *F1:* 0.08 ± 0.06	*F0:* 0.03 – 0.16 *F1:* 0.04 – 0.27	*F0:* 0.06 – 0.22 *F1:* 0.05 – 0.17	*F0:* 10 *F1:* 10	*F0: 8* *F1: 8*
Wired EEG	*F0:* 0.07 ± 0.05 *F1:* 0.15 ± 0.11	*F0:* 0.10 ± 0.07 *F1:* 0.14 ± 0.11	*F0:* 0.02 – 0.14 *F1:* 0.02 – 0.36	*F0:* 0.08 – 0.21 *F1:* 0.06 – 0.33	*F0:* 9 *F1:* 10	*F0: 8* *F1: 8*

#p < 0.05: number of participants (out of 11) showing significant coherence values

Comparison of CKC between EEG systems. We used the non-parametric Wilcoxon signed rank test to compare the CKC strength between the two EEG amplifiers. Spearman correlation coefficients on CKC strength at separately F0 and F1 were computed across participants to assess the agreement of CKC strength between the two EEG amplifiers.

III. RESULTS AND DISCUSSION

Qualitatively, the two EEG amplifiers provided similar CKC results at group and individual levels, both in terms of CKC strength and its source location (i.e. scalp topographies).

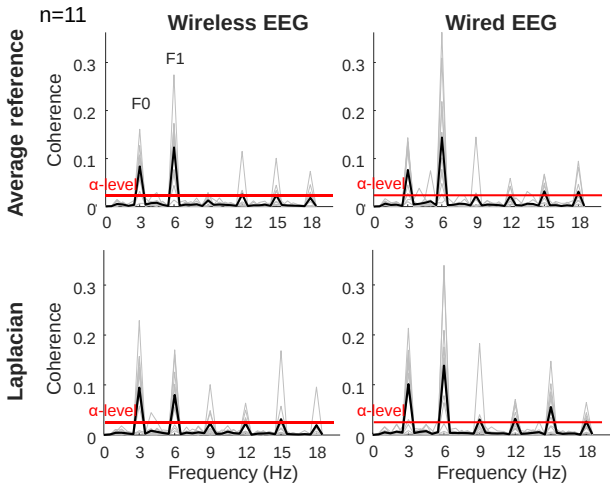

Figure 2. Coherence spectra between finger acceleration magnitude and EEG signals for EEG devices and spatial derivations. Gray solid lines indicate individual coherence spectra (n = 11) of EEG channel in which the CKC peaked. Black solid lines indicate the group-mean coherence spectra. Red horizontal lines indicate the threshold for statistical significant CKC ($p < 0.05$).

Fig. 2 shows coherence spectra for tested spatial EEG filters and EEG systems: grand-average CKC spectra between finger acceleration and filtered EEG signals are superimposed on the individual spectra (n = 11). Overall, 10 or 8 out of 11 participants (respectively for average reference and Laplacian derivation) showed significant CKC ($p < 0.05$) at F0 and F1. Peak CKC strengths are reported in Table I. Similarly to other previous studies using precisely timed proprioceptive stimulation [6], the afferent coupling was stronger at F1 than F0, suggesting that F1 is enhanced by the two afferent proprioceptive volleys arising both from flexion and extension phase of the movement (i.e. occurring twice during the movement cycle), as mentioned in [9].

Fig. 3 shows the grand average topographic scalp distribution of CKC across participants, separately for F0 and F1, spatial derivations and EEG devices. CKC source locations at F0 and F1 were comparable between the EEG devices. As expected, CKC peaked on EEG electrodes above the sensorimotor cortices contralateral to stimulated finger. However, some disparity was observed in the source CKC localization between F0 and F1. The CKC peak at F0 showed two main clusters: one close to the frontal F3 electrode and the other above the left central-parietal area (CP5 and P7 electrodes). Instead, F1 peaked within one cluster above the primary sensorimotor cortex of the stimulated hand (C3 electrode). Although the neural basis of CKC at movement frequency versus its first harmonic is still controversial, there are some evidences that the two frequencies (i.e. F0 and F1) could reflect partly different aspects of cortical proprioceptive processing [9]. However, further experiments need to be performed to test this hypothesis and clarify the neuronal mechanisms.

It has been shown previously that CKC is stronger for Laplacian EEG spatial derivation compared to average

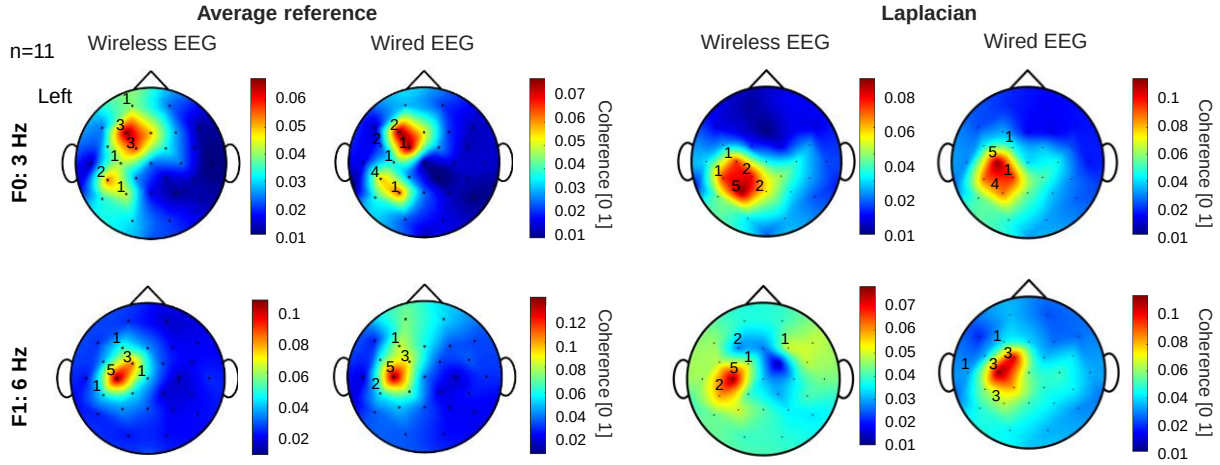

Figure 3. Scalp topographies of CKC distribution at stimulation frequency (F0) and its first harmonic (F1) averaged across participants (n=11) for EEG recording systems and for spatial filters. The superimposed numbers indicate the occurrences among participants of CKC peak in each electrode location. Please note that the colorbar range varies across recording systems, frequencies and spatial derivations.

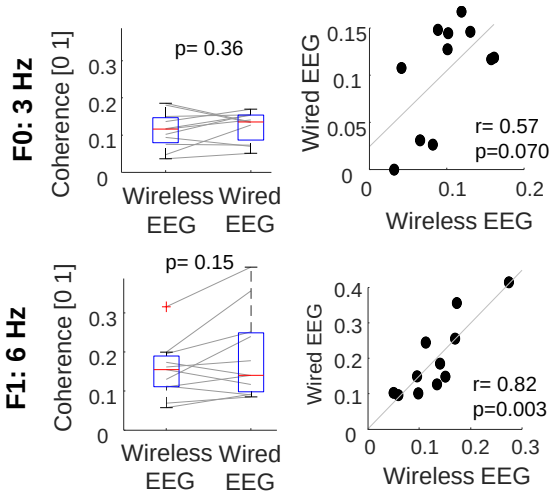

Figure 4. On the left panel: boxplot of grouped CKC strength for individual CKC peaks at movement frequency (first row) and its first harmonic (second row) for EEG recording systems. Gray lines connect individual CKC values. On the right panel: scatterplots of individual CKC values of wireless and wired EEG recordings. Corresponding linear regression lines and Spearman correlation coefficients are superimposed.

reference one at F0 [8]. Also in our data, the Laplacian derivation provided stronger CKC than the average referenced signals. However, this result was not always found at F1 (see Table I and Fig. 3). This may be explained by the operating principle of the Laplacian filter itself, being the result from a linear combination of five electrodes. Therefore, if the coherent source in the brain is widespread across the five electrodes area (as in the case of F0), it is reasonable to expect that the Laplacian-based CKC estimation will be overall stronger in the subtended area. On the contrary, if the coherent source in the brain is more focal (as in the case of F1), then the Laplacian-based CKC results in a less focal source and weaker strengths. Not surprisingly, this actual effect of the Laplacian derivation is strictly dependent on the density of the electrodes cap. The more dense the electrodes, the higher the probability to find multiple electrodes in the same coherent brain area, as in the case of the abovementioned study [7]. As a result, detailed experiments could be carried out to extensively explore which could be the optimal spatial EEG derivation in case of using a less dense 30-electrodes cap.

Fig. 4 illustrates group and individual CKC values that were comparable between the EEG devices with no statistically significant differences ($p = 0.37$ at F0 and $p = 0.15$ at F1). The scatterplots show a positive correlation between the CKC values obtained with the EEG systems, indicating a good agreement between the two devices.

IV. CONCLUSION

In this study we used an innovative wireless-EEG acquisition system to examine cortical proprioceptive processing using CKC method. The comparison between our wireless device and a standard, wired recording system demonstrated the suitability of the wireless EEG amplifier to quantify CKC for proprioceptive stimulation of the hand. This initial validation in laboratory environment was aimed at validating the performance of our system with respect to state-of-the-art devices, laying the foundations for future investigations of the human sensorimotor functions in naturalistic conditions and tasks.

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