About the cortical origin of the low-delta and high-gamma rhythms observed in EEG signals during treadmill walking

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HIGHLIGHTS

• Spectral and time–frequency EEG analysis was performed in ambulatory context.
• Motion artifacts may affect EEG signal integrity up to 15 Hz.
• EEG and accelerometer signals exhibit similar time–frequency properties.
• Cortical origin of low-delta and high-gamma bands during locomotion is put in doubt.

ARTICLE INFO

Article history:
Received 7 August 2013
Received in revised form 9 December 2013
Accepted 21 December 2013

Keywords:
Accelerometer
Brain control of locomotion
Electroencephalography
Motion artifacts

ABSTRACT

This paper presents a spectral and time–frequency analysis of EEG signals recorded on seven healthy subjects walking on a treadmill at three different speeds. An accelerometer was placed on the head of the subjects in order to record the shocks undergone by the EEG electrodes during walking. Our results indicate that up to 15 harmonics of the fundamental stepping frequency may pollute EEG signals, depending on the walking speed and also on the electrode location. This finding may call into question some conclusions drawn in previous EEG studies where low-delta band (especially around 1 Hz, the fundamental stepping frequency) had been announced as being the seat of angular and linear kinematics control of the lower limbs during walk. Additionally, our analysis reveals that EEG and accelerometer signals exhibit similar time–frequency properties, especially in frequency bands extending up to 150 Hz, suggesting that previous conclusions claiming the activation of high-gamma rhythms during walking may have been drawn on the basis of insufficiently cleaned EEG signals. Our results are put in perspective with recent EEG studies related to locomotion and extensively discussed in particular by focusing on the low-delta and high-gamma bands.

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1. Introduction

Recently, numerous experimental results have indicated a strong involvement of the brain during locomotion. Significant changes in motor and cognitive demands (i.e. spatial attention) have been observed in the context of bipedal walking in unknown or cluttered dynamic environments [8,12,21,25]. Functional neuroimaging studies have shown that the primary motor cortex is recruited during rhythmic foot or leg movements [9,11,16,17,19,26]. Additionally, the technique of functional near-infrared spectroscopy (fNIRS) has allowed to detect involvement of frontal, premotor and supplementary motor areas during walking [15,28].

All those results were obtained using imagery techniques which are characterized by a good spatial but poor temporal resolution. In contrast, electroencephalography (EEG) is a measurement technique offering a sufficiently good temporal resolution to study the dynamics of brain. However, EEG study of cortical activity elicited during walk is highly challenging: EEG signals are by essence noisy and may be affected by different artifacts generated either by extracerebral physiological activity or by the gait itself [7].

Two strategies have thus been developed in the literature in order to overcome these experimental difficulties. The static approach consists in focusing on simplified foot or leg movements which imply common cerebral processes with gait. In these experimental protocols, subjects are mainly static and produce only limited lower limb movements. On the other hand, the dynamic approach consists in recording EEG signals from subjects walking on a treadmill. In this case, a powerful analysis technique to discriminate the different artifact contributions from the real cortical signals is of course required. Regrettably, the results of those different analyses are most of the time partially, if not totally,
incompatible regarding both the location of the brain areas activated and the frequency bands of interest [5,6].

In this paper, the EEG signals recorded during treadmill walking are analyzed and compared with data acquired by an accelerometer placed on the head of each subject. Similarities between both types of signals are presented and extensively discussed in order to bring new clues in the general understanding of EEG signals recorded during human locomotion, in particular for the very low and very high frequency bands.

2. Materials and methods

2.1. Data collection

Seven healthy volunteers (5 males and 2 females) without any known physical or neurological disorders participated in this experiment (age-range: 25–33 years) whose protocol was extensively described elsewhere [7]. Basically, one of the objectives of this data collection was to assess the feasibility of developing a brain–computer interface under ambulatory conditions. Therefore, each subject walked bare feet on a treadmill at 1.5, 3 and 4.5 km/h wearing an EEG cap (32 passive electrodes) connected to the Advanced Neuro Technology analyzer (ANT, Enschede, The Netherlands) digitizing the signals at 512 Hz. Additionally, a piezoelectric accelerometer (Dytran 3100B) was fixed to a rigid plate mounted on a three-point linkage, firmly strapped on the head of the subject with an elastic band and plugged into the ANT system (see Fig. 3, supplementary material). This montage ensured the correct transmission of shocks to the accelerometer. Simultaneously, the kinematics of the lower limb movements was recorded using a system of six infrared cameras (Bonita, Vicon, Los Angeles, USA). Each EEG recording (i.e. 3 per subject) lasted about 12 min. All procedures were approved by the Université Libre de Bruxelles Internal Review Board and complied with the standards defined in the Declaration of Helsinki.

2.2. Pre-processing and spectral analysis

In a first step, the times of important gait events were determined with the kinematics data. Two principal events are defined in human locomotion: the heel strike, which is the time of the first contact of the foot with the ground, and the toe off, which is the last instant of contact of the foot with the ground. Consequently, 4 typical events follow one another during a gait cycle: the right heel strike (RHS), the left toe off (LTO), the left heel strike (LHS) and finally the right toe off (RTO) before the next RHS. During walking on the treadmill, the heel strike time is defined when the position of the malleolus marker is the most forward (in the treadmill axis direction), while the toe off time is defined when the fifth metatarsal marker is in the most backward position [33].

EEG signals were processed using the EEGLAB toolbox [10]. A standard spectral analysis (FFT) was made in order to compare the frequency contents of EEG electrodes and the accelerometer. The goal of such analysis was to check the possible presence of common harmonics associated to the stepping frequency of each subject. We focused on Cz, Oz and T8, for the diversity of their spatial localizations (top, back and left side of the head respectively) and thus for the diversity of the brain areas which are monitored.

2.3. Ensemble averaged time–frequency analysis

A time–frequency analysis was conducted in order to compare the signals coming from the EEG electrodes and the accelerometer, on a gait cycle basis. With this aim, EEG data were first detrended and then epoched by defining a time-window of 2 s around each left heel strike. Each epoch was visually inspected, rejected in case of obvious presence of eye or muscle artifacts.

As the lower limb movements during locomotion are only quasi-periodic, the stride length varies from one step to the other. The generation of precise ensemble averaged event-related spectral perturbation (ERSP) plots is thus not straightforward. We first computed spectrograms for each EEG channel during each epoch for each subject, as described in [14]. All the single-trial spectrograms were then linear time-warped so that the times of heel strikes and toe off events occurred at the same adjusted latencies. After this operation, spectrograms were ensemble averaged for all subjects. The average log spectrum for all movement cycles was subtracted from the log spectrogram for each movement cycle. The resulting changes from baseline are the ERSP plots presented in next section, as a function of the percentage of the normalized gait cycle. Significant ERSPs (p < 0.05) were computed using a bootstrapping method [10].

3. Results

Common harmonics were found in the spectra of EEG electrodes and the accelerometer. These harmonics correspond to the fundamental stepping frequency of the subjects, which ranges from about 0.6 Hz at 1.5 km/h to 1 Hz roughly at 4.5 km/h. Box-plots shown in Fig. 1 clearly indicate that the number of harmonics present in the spectra is monotonously increasing with the walking speed. Here, harmonics with Signal to Noise Ratio >2 are considered, the signal being the peak amplitude at frequency f (multiple of the fundamental) and the noise being the background amplitude evaluated in the [f–0.5, f+0.5] Hz interval. At 4.5 km/h, up to 15 harmonics are observed in the EEG electrode spectra and almost twice as much in the accelerometer spectra. More precisely, harmonics are produced up to 15 Hz and 30 Hz in EEG and accelerometer signals respectively. In these conditions, delta, theta, alpha and low beta bands are impacted. Also, the distributions of harmonic numbers are obviously differing from one electrode to the other, meaning that the phenomenon giving rise to the harmonics most likely depends on the electrode spatial localization.

Regarding the time–frequency analysis, it appears that an event-related desynchronization (ERS) is produced in the accelerometer signal at each heel strike, during the double support phase of gait, while an event-related synchronization (ERS) appears during each swing phase (cf. Fig. 2), regardless of the walking speed (see additional material). Both ERD and ERS occur in a large frequency interval ranging from low-delta band up to high-gamma band (150 Hz) without any discontinuity (although inaccurate for an accelerometer, we use the terms ERS and ERD to describe more easily our observations). The same characteristic alternation (i.e. ERD–ERS) is visible in EEG electrodes Cz, Oz and T8 at 3 and 4.5 km/h, but with discontinuities along the frequency axis which vary both in number and localization according to the EEG electrode. Interestingly, the results obtained for the lowest walking speed (1.5 km/h) look different. Indeed, a specific succession of ERS–ERD–ERS occurs below 10 Hz, starting at each gait event (i.e. both heel strikes and toe offs) in all EEG electrodes, while the standard ERD–ERS alternation found at higher walking speeds is seen in the accelerometer channel.

4. Discussion

In addition to “traditional” EEG artifacts (ocular, muscular, power line, . . .), EEG recordings realized in ambulatory conditions are degraded by specific sources of noise [6,7]. Triboelectric noise is generated by movement, friction and flexion of the cable components, resulting in a static or piezoelectric movement transducer.
The number of harmonics present in the EEG electrode and accelerometer spectra is monotonously increasing with walking speed. Also, the distributions are differing according to the spatial localization on the head. Examples of spectra for one subject walking at 4.5 km/h are inserted.

The ERSP analyses for both EEG electrodes (Cz on column 1, T8 on column 2, Oz on column 3) and the accelerometer (column 4) during walking indicate successive synchronizations (in red) and desynchronizations (in blue) along the gait cycle. Rows 1, 2, 3 correspond to 1.5, 3 and 4.5 km/h respectively. The color scale is in dB. (For interpretation of the references to color in text, the reader is referred to the web version of this article.)

Please cite this article in press as: T. Castermans, et al., About the cortical origin of the low-delta and high-gamma rhythms observed in EEG signals during treadmill walking, Neurosci. Lett. (2014). http://dx.doi.org/10.1016/j.neulet.2013.12.059
Electrode movements are produced by movements of the head, but also by the shocks undergone by the whole body at each step, which—albeit significantly attenuated—are transmitted to the head [24]. These movements modify the magnetic and capacitive coupling of the user and the electrode leads, leading to an alteration of the parasitic current flowing into the leads [20]. A resulting parasitic voltage drop is then produced in the electrode/gel/skin interface which interferes with the EEG signal [31]. Finally, electrode movements can also cause impedance variation which directly affects the electrode voltage offset [29].

Unfortunately, our results indicate that all these motion artifacts are not limited to a small spectral band, so they cannot be simply removed by frequency filtering. This first conclusion is in total agreement with Kerick et al. [18]. Therefore, motion artifacts should be eliminated using a specific treatment, which should be more sophisticated than a simple low-, high- or band-pass filter. In addition, the fact that distributions of harmonic numbers differ from one electrode to the other means that the way motion artifacts affect EEG spectra depends on the electrode spatial localization. This should also be taken into account when correcting motion artifacts.

With these considerations in mind, we are inclined to suspect that several published EEG analyses of walk may be strongly polluted by motion artifacts. For instance Presacco et al. [23] did not use any specific motion artifact cleaning in their EEG decoding. In order to reconstruct the kinematics of lower limbs during treadmill walking, they applied a simple band-pass filter between 0.1 and 2 Hz to the raw EEG signals. Note that this frequency range perfectly coincides with the stepping frequency of human locomotion, as mentioned in this paper and in [24]. Then Presacco et al. [23] used a linear Wiener filter to reconstruct lower limb kinematics during treadmill walking. From a mathematical point of view, though, it may not be surprising that lower limb elevation angles can be accurately reconstructed from a signal containing their first harmonics by determining a set of parameters. In order to rule out the presence of motion artifacts, the authors computed the phase-locking values (PLV) among sensors. The rationale was that potential motion artifacts due to EEG wires or the subject’s motion would affect all sensors equally. In their logic, the authors claim that mechanical artifacts did not play a role in their decoding because the PLV values they found ranged from about 0.55 in the walking condition (around 1 and 2 Hz) to 0.4 in the baseline (rest) condition. According to them, these low PLV values (found in both conditions) suggest lack of mechanical coupling due to concerted wire movement. However, our results indicate that motion artifacts do not affect the electrodes located on the top of the head and those on the head circumference in the same way. Thus, the low PLV values argument should not be considered as a direct proof of the absence of motion artifacts. Consequently, it still remains to be confirmed (or infirmed) that EEG delta band contains information about angular and linear kinematics of the lower limb during walk.

This last point is supported by a recent paper from Antelis et al. [2]. These authors have focused on several publications in which are reported successful reconstructions of different limb kinematics from EEG using the low frequency activity of the EEG and linear regression models [1,3,4]. Antelis et al. [2] showed that the mathematical properties of the linear regression model and of the correlation metric used in these studies could explain the good reported results. Moreover, they demonstrated that correlation results obtained with real EEG signals, shuffled or random EEG data were not statistically different. This means that the linear models developed in [1,3,4] are able to provide the same results irrespectively of the presence or absence of limb velocity information in EEG signals.

Other recent studies dedicated to EEG analysis during a locomotion task may be added to this discussion. Severens et al. [27] investigated the possibility of measuring ERD and ERSPs during walking on treadmill. After cleaning EMG artifacts using canonical correlation analysis (CCA), they found an ERD in the mu band above the central motor cortex (electrode Cz) and in the beta band above the lateral motor cortex (electrodes C3 and C4). In addition, they found that ERSPs in mu and beta bands were coupled to the gait cycle with significant differences between left swing, right swing and double support phase of the gait cycle. They did not report any signal of cortical origin at low frequency. Indeed, as the low frequency modulations they found in the ERSPs were also visible in the occipital channels, the authors explained these were very unlikely related to brain activity and probably due to remaining artifacts.

Wagner et al. [32] also showed that mu and beta rhythms are suppressed during active walking in the Lokomat, a robotic gait orthosis. They also provided evidence of modulations of the lower gamma band (25–40 Hz), localized in central midline areas and related to the phases of the gait cycle. For different reasons, the authors speculate that these activations and deactivations might be related to sensorimotor processing of the lower limbs in the complex motor pattern of human locomotion. Although their ERSPs plots exhibit ERD and ERS around and below 5 Hz, they neither comment them nor claim that these originate from cortical activity.

The first analysis of EEG during walk on treadmill was actually published by Gwin et al. [14]. In this study, successive synchronization (ERS) and desynchronization (ERD) were found for the first time, in phase with the gait cycle, present in different regions of the brain and in numerous frequency bands comprised between 3 and more than 150 Hz. By using a method based on independent component analysis (ICA) combined with an inverse modeling approach, the authors claimed they could discriminate electrophysiological sources, muscle sources and other artifacts from the raw EEG signals. However, in a previous study, the very same authors [13], using the very same dataset, clearly stated that:

"Unlike more spatially stationary artifacts in EEG signals arising from eye movements, scalp muscles, fMRI gradients, etc., which may be resolved by ICA decomposition into a subspace of one or more independent components, we found that gait-related movement artifact remained in many if not most of the independent components. This prevented us from removing only a small subset of components capturing the movement artifacts." For this reason, they considered the removal of motion artifacts from EEG during walking and running on treadmill using an artifact template subtraction method. Such method allowed to enhance the detection of P300 potentials in ambulatory conditions. Nevertheless, the study of cerebral processes involved in human locomotion is not possible using a subtraction method, as it would undoubtedly remove interesting signal from the EEG recordings. For this reason, the authors used only the ICA approach to clean the EEG signals in [14]. In this study, no mention is made of specific motion artifacts, nor of any particular treatment to reject them. Thus, it may be doubted that the time–frequency analysis plots shown in that paper do not contain any motion artifact contribution. Fig. 2 clearly shows that periodic power spectral changes over large frequency bands can be observed in the accelerometer signal, in a similar way to the results obtained after ICA by Gwin et al. [14].

Finally, it should be noted that spectral analysis does not allow to determine which cortical region is directly involved in the transmission of motor commands to the muscles. In contrast, coherence analysis reveals anatomical coupling between cortical activity and the motor output to the muscles by detecting common rhythmicities in EMG and EEG signals. According to the significant coherence values (24–40 Hz around Cz) found by Petersen et al. [22], the multiple ERP–ERS detected by Gwin et al. [14] in the 3–24 and 40–76 Hz bands are obviously not indicative of a direct corticospinal drive, at least, not to the tibialis anterior. Thus, one may think that these signals, if not affected by residual artifacts, would rather reflect the
control of sensory afferents (i.e., one of the hypotheses formulated by Gwin et al. [14] themselves). It is interesting to note that the studies by Wagner et al. [32] and Severens et al. [27] do not report multiple ERD–ERS in $\alpha$, $\beta$ and $\gamma$ bands and are in line with the coherence study made by Petersen et al. [22].

5. Conclusions

Despite the inherent difficulties arising when analyzing EEG signals under ambulatory conditions, several groups have recently published papers about EEG decoding of movements or the fundamental analysis of mechanisms taking place in the brain during locomotion. By simultaneously recording the data coming both from a conventional EEG cap and from an accelerometer placed on the head of subjects walking at different speeds on a treadmill, we demonstrated that motion artifacts in phase with the fundamental stepping frequency could exhibit harmonics impacting EEG signals up to $15 \text{ Hz}$. This pollution of signals is dependent on the electrode location, which renders the motion artifact clearing step more complex than applying a general band-pass filter. For this reason, we suspect that several studies published in the literature present results that are based on insufficiently cleaned EEG data.

As the EEG decoding of gait proposed by Presacco et al. [23] was realized on the basis of raw data, it should be considered in no way as a proof that EEG delta band contains information about kinematics of the lower limbs. Such statement still needs to be confirmed or infirmed. In addition, we have shown that our time–frequency analysis results for the real EEG electrodes and for the accelerometer data are similar and exhibit, in particular, large and rhythmic activities spreading over wide frequency bands (up to $150 \text{ Hz}$). This suggests that these large extensions in high gamma band, previously reported by Gwin et al. [14], might be due in fact to motion artifacts as well. Consequently, researchers conducting future EEG studies during locomotion should take a great care in the cleaning procedure, as data in this context are strongly affected by an entanglement of downgoing, upgoing and artefactual contributions[6].

Acknowledgments

M. Duvinage is a FNRS (Fonds National de la Recherche Scientifique) Research Fellow. This paper presents research results of the Belgian Network DYSCO (Dynamical Systems, Control, and Optimization), funded by the Interuniversity Attraction Poles Programme, initiated by the Belgian State, Science Policy Office. The scientific responsibility rests with its author(s).

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.neulet.2013.12.059.

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