Corticomuscular Coherence Variation throughout the Gait Cycle during Overground Walking and Ramp Ascent: A Preliminary Investigation*

Anna T Winslow, Justin Brantley, Fangshi Zhu, Jose L Contreras Vidal, *Senior Member*, *IEEE*, and He Huang, *Senior Member, IEEE*

Abstract— **Recent designs of neural-machine interfaces (NMIs) incorporating electroencephalography (EEG) or electromyography (EMG) have been used in lower limb assistive devices. While the results of previous studies have shown promise, a NMI which takes advantage of early movement-related EEG activity preceding movement onset, as well as the improved signal-to-noise ratio of EMG, could prove to be more accurate and responsive than current NMI designs based solely on EEG or EMG. Previous studies have demonstrated that the activity of the sensorimotor cortex is coupled to the firing rate of motor units in lower limb muscles during voluntary contraction. However, the exploration of corticomuscular coherence during locomotive tasks has been limited. In this study, coupling between the motor cortex and right *tibialis anterior* muscle activity was preliminarily investigated during self-paced over-ground walking and ramp ascent. EEG at the motor cortex and surface EMG from the *tibialis anterior* were collected from one able-bodied subject. Coherence between the two signals was computed and studied across gait cycles. The EEG activity led the EMG activity in the low gamma band in swing phase of level ground walking and in stance phase of ramp ascent. These results may inform the future design of EEG-EMG multimodal NMIs for lower limb devices that assist locomotion of people with physical disabilities.**

I. Introduction

Emerging powered lower limb exoskeletons and prostheses have shown great promise in restoring mobility in patients with lower limb disabilities [1]–[3]. Since these devices must function in coordination with the user's intent, a neural-machine interface (NMI) that can decode the neural signal and identify user intent is essential [4].

Electroencephalography (EEG) and electromyography (EMG) have been individually used in recent designs of neural-machine interfaces for lower limb assistive devices [4]. We have shown the feasibility of reconstructing the linear envelope of the EMG patterns of lower-limb muscles during over-ground walking with and without an exoskeleton suggesting the slow cortical potentials in the delta band (0.1 Hz to 3 Hz) contain information about motor signals during gait production [5]. Our group has also decoded EMG signals to identify the user's locomotive task [6]. Greater than 90% classification accuracy was reported for identifying seven locomotive tasks. EMG-based NMI has enabled seamless terrain transitions for amputees wearing advanced power prostheses [7].

EEG and EMG present different but complementary advantages as the neural control source. EEG movement-related potentials often precede movement onset by greater than one second [8] while EMG activity is usually observed approximately 100 ms before movement onset [9]. EMG signals, however, possess a much better signal-to-noise ratio than EEG signals. If the two sources can be combined appropriately, a more accurate and responsive NMI might be designed. Understanding the coupling of cortical activity and motor unit activity in locomotion may lead to effective design of an EEG-EMG multimodal NMI for lower limb assistive devices. While studies have reported evidence supporting cortical involvement in voluntary muscle contraction [10], cortical activation during locomotive tasks has only recently been studied due to advancements in powered wearable robots for gait assistance [1], [2].

A more relevant study directly investigated the functional coupling of cortical activity and lower limb muscle activity during treadmill walking using coherence analysis [11]. Significant coupling between the motor cortex and shank muscles was found in the low gamma band in swing phase during treadmill walking. However, only the swing phase was investigated. Additionally, the treadmill test condition was not truly characteristic of self-paced locomotion. For example, treadmill walking is known to influence subjects' cadence, as well as their joint kinematics and kinetics [12]. Furthermore, level ground walking is only a single terrain over which a NMI would need to successfully operate. Thus, towards the design of a NMI for lower limb devices that assist gait and balance in daily living, the aim of this study is to preliminarily investigate the modulation in coupling of low gamma band motor cortex activity and *tibialis anterior* (TA) muscle activity across the gait cycle of a single subject during self-paced over-ground walking and ramp ascent.

II. Methods

The protocol was approved by the IRB of The University of North Carolina at Chapel Hill and the University of Houston. One 180 cm tall male subject, aged 31, was enrolled in the study after giving informed consent.

A. Experimental Methods

The activation of the right *tibialis anterior* (RTA) was sampled at 1,000 Hz using active Ag/AgCl electromyogram

*This work was partly supported by NSF awards 1406750, 1361549, and IIS-1302339.

A.W. and H.H. are with the NCSU/UNC Department of Biomedical Engineering, NC State University, Raleigh, NC, 27695; University of North Carolina at Chapel Hill, Chapel Hill, NC 27599.

JLCV, JB and FZ are with the Department of Electrical and Computer Engineering at the University of Houston, Houston, TX 77204.

(EMG) electrodes (Motion Lab Systems, Inc., USA). The location of the RTA muscle belly was approximated relative to various anatomical landmarks [13]. Electrode locations were confirmed by palpating the identified muscle area as the subject performed ankle dorsiflexion and plantarflexion movements. Hair was removed from the skin above the electrode sites and the areas were cleaned with alcohol wipes prior to electrode placement.

EEG data were transmitted wirelessly to a data acquisition computer at 1,000 Hz (BrainVision LLC, USA). A sixty-four channel active Ag/AgCl electrode array was arranged on the subject's scalp in the 10/20 system using an EEG cap (actiCAP, Brain Products GmbH, DEU). The TP9, TP10, PO9, and PO10 electrodes were repositioned adjacent to the right and left eyes to capture eye blinks, as well as lateral and vertical eye movements. All electrodes were gelled to reduce their impedance, and all electrode impedances were below 15 kΩ during pre- and post-test impedance checks. Electrode wires were routed between electrodes and secured to the cap with athletic pre-wrap to minimize motion artifact due to cable pulling.

The subject was asked to walk a 7.9 m (26 ft) long section of level ground at their preferred walking speed, beginning from a standing rest. Additionally, the subject was asked to walk a 3.5 m (11.3 ft), 5.5 degree ramp at their preferred speed. One trial consisted of four repetitions of level walking and two repetitions of ramp ascent from rest. Twelve trials total were conducted. Toe off (TO) and heel contact (HC) gait event timings were extracted from bilateral foot pressure insole data sampled at 100 Hz (Novel GmbH, DEU). The subject's walking speed was extracted from an inertial measurement unit (IMU) based motion capture system (MVN, Xsens Technologies B.V., NLD). A synchronization pulse was sent to all data acquisition systems before and after each trial to enforce consistent timing. All trials were video recorded. The experimental setup is shown in Figure 1.

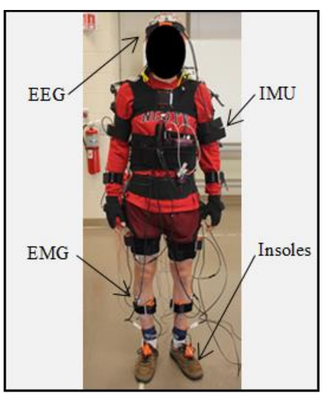

Figure 1. Experimental setup showing EEG, EMG, IMU, and pressure insole placement.

B. Pre-processing, Artifact Removal, and Segmentation

The raw EMG data were zero-phase bandpass filtered using a fourth order Butterworth filter with a 1-450 Hz passband. Raw EEG data were zero-phase high pass filtered with a 0.1 Hz cutoff. EEGLAB was used for EEG data pre-processing [14]. A commercial software package was used to perform the EMG pre-processing and coherence analysis (MATLAB 7.1, The MathWorks Inc., Natick, MA). EEG movement artifacts were identified by performing Independent Component Analysis (ICA) on the data using the EEGLAB *runica* algorithm. Components which displayed spatiotemporal and frequency characteristics typical of muscle artifact, eye blinks, or eye movement were discarded. The cleaned data were then re-referenced to a common average reference.

Steady level ground walking data and steady ramp ascent data were extracted from the total number of steps. The first gait cycle transitioning to and from rest were excluded from the analysis for every level walkway passage. Similarly, transitions between the ramp and level ground were excluded during the ramp ascent data extraction. Individual gait cycles were extracted from the steady level ground and ramp ascent data relative to right heel contact (RHC). A total of 144 gait cycles were retained for the level walking coherence analysis and 19 gait cycles were retained for the ramp ascent analysis.

C. Coherence Analysis

Periodogram-based coherence analysis has been established as a means of identifying functional coupling between physiological signals [10],[15]. In this work, coupling between the motor cortex EEG activity and the RTA EMG activity were investigated during over-ground walking and ramp ascent using coherence analysis. The coherency is the normalized cross-power spectral density for input signal x and output signal y at frequency λ, which can be written as

$$C_{xy}(\lambda) = \frac{f_{xy}(\lambda)}{\sqrt{f_{xx}(\lambda)f_{yy}(\lambda)}}. \qquad (1)$$

In this analysis, the cleaned EEG data were input signal x while the output signal y was the filtered RTA EMG activity. Using the periodogram method, estimates of the PSD and CPSD are obtained by averaging the Fourier transforms of short data windows which have the same timing relative to some event. In this analysis, spectrograms were constructed for individual gait cycles relative to each stride's duration (in ms), absolute event timings (in ms) were converted to normalized gait timings (unitless) for each gait cycle, and the spectrograms were then averaged across all cycles. Since the CPSD is a complex-valued function, the coherency function, (1), is also complex. The strength of the association between x and y at frequency λ is quantified by the squared modulus of the coherency, called the coherence, which is written as

$$\left|C_{xy}(\lambda)\right|^2 = \left|\frac{f_{xy}(\lambda)}{\sqrt{f_{xx}(\lambda)f_{yy}(\lambda)}}\right|^2. \qquad (2)$$

The argument of the coherency from (1) represents the phase between the input and output signals at frequency λ.

The statistical significance of the coherence values were calculated using established methods based on Fisher's Z-transformation [16]. The Z-transform is a variance-stabilizing transformation which permits variance estimation [15]. Halliday *et al* report that the variance of the Z-transformed modulus of coherency can be estimated as

$$var\left(arctanh\left(\left|C_{xy}(\lambda)\right|\right)\right) = \frac{1}{2N} \qquad (3)$$

where N is the number of averaged time windows from which the coherence was calculated [15]. The variance of the phase of coherency can be calculated from [15] as

$$var\left(arg\left(C_{xy}(\lambda)\right)\right) = \frac{1}{2N}\left(\frac{1}{|C_{xy}(\lambda)|^2} - 1\right) \quad (4)$$

In order to capture the time evolution of coherence across the gait cycle, coherency estimates were calculated using a 250 ms sliding Hamming window across each gait cycle, and a 50 ms time increment between windows was used. Significance was determined by conducting a Z-test on the parameters of interest for a null hypothesis of μ=0. The Z-scores were normalized relative to the square root of the variances from Eqs. (3) and (4). The resulting p-values were compared to the Bonferonni-corrected significance threshold, $\alpha = 0.05/E$, where E is the number of electrodes across which the comparison was made.

III. Results and Discussion

A single gait cycle of the filtered EMG activity during overground walking is plotted in Fig. 2a relative to RHC. The coherence of the central motor cortex electrode, Cz, with the RTA EMG activity is shown in Fig. 2b and the phase of the Cz-RTA coherency is shown in Fig. 2c. A positive phase value indicates that the coherent EEG activity led the corresponding EMG activity. Regions of interest are boxed in Fig. 2b and 2c, and insignificant results are dulled. Regions with significant coherency magnitude and phase which persisted a minimum of two consecutive time points are outlined in white.

During level ground walking, the RTA was most active after right heel contact (0% to 15% of the gait cycle) and in early swing phase (65% to 85% of the gait cycle), where dorsiflexion is employed to achieve toe clearance in swing. Significant coherence magnitude and phase were observed in early swing. These timings agree well with the activation profile of the RTA muscle in Fig. 2a.

The maximum swing phase coherence, 0.026, occurred at 20 Hz and persisted from 70% to 75% of the gait cycle. The frequency range of EEG and EMG coupling align with previous studies which found significant motor pool firing rate synchrony in the 15-35 Hz range during both voluntary muscle contraction and voluntary co-contraction [10], [17]. Coupling between cortical and muscle activity was also observed in the beta and low gamma (15-35 Hz) bands during voluntary movement studies involving transcranial magnetic stimulation (TMS) and magnetoencephalography (MEG) [18], [19].

As evidenced by the positive phase in Fig. 2c, the EEG activity led the EMG activity during early swing phase at 20 Hz. Previous studies have demonstrated that low frequency TMS over the motor cortex during treadmill walking decreases EMG activity in distal leg muscles, which suggests the involvement of the corticospinal tract in walking [20], [21]. Therefore, the corticospinal tract may be activated in early swing during level ground walking which contributes to the observed RTA EMG activity observed in early swing phase[11], [20], [21].

A single gait cycle of the filtered EMG activity during ramp ascent is plotted in Fig. 3a relative to RHC. The coherence of the central motor cortex electrode, Cz, with the RTA EMG activity is shown in Fig. 3b and the phase of the

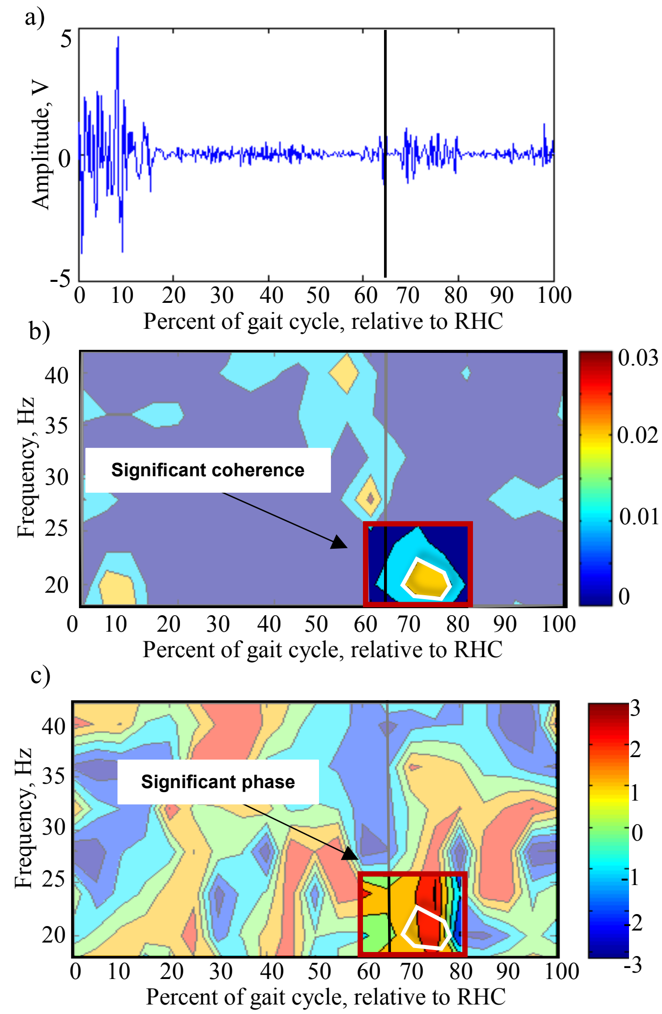

Figure 2: a) RTA EMG activity during a single gait cycle, b) Cz-RTA coherence, and c) the phase, in radians, of Cz-RTA coherency during overground walking. A vertical black line marks right toe off (TO) timing in each plot. Significant coherence, with EEG activity leading EMG activity, was observed in early swing at 20 Hz.

coherency is shown in Fig. 3c. The RTA activity during ramp ascent, Fig. 3a, was similar to that during level walking, Fig. 2a, with the exception that increased RTA activity was present late in the stance phase for ramp ascent (from 40% to 60% of the gait cycle). Interestingly, in contrast to overground walking, significant magnitude and phase of coherency were only observed during the stance phase of ramp walking.

The maximum coherence during stance phase, 0.38, occurred at 20 Hz at 45% of the gait cycle. The coherence had a positive phase (Fig. 3c), indicating that the EEG activity led the EMG activity in late stance. A similar result was observed in early stance at higher frequencies in the low gamma band (28 Hz).

The results of this coherence analysis provide evidence in support of cortical involvement in self-paced multi-terrain walking. Complementary to previous work which reported fairly consistent EEG-RTA EMG across the swing phase of walking [11], this observation indicates the time varying property of EEG and EMG coherence throughout the full gait cycle, as well as the terrain-dependence of EEG-RTA EMG coherence. Hence, a NMI decoding algorithm design incorporating both signals must take these properties into account and may need to be time dependent.

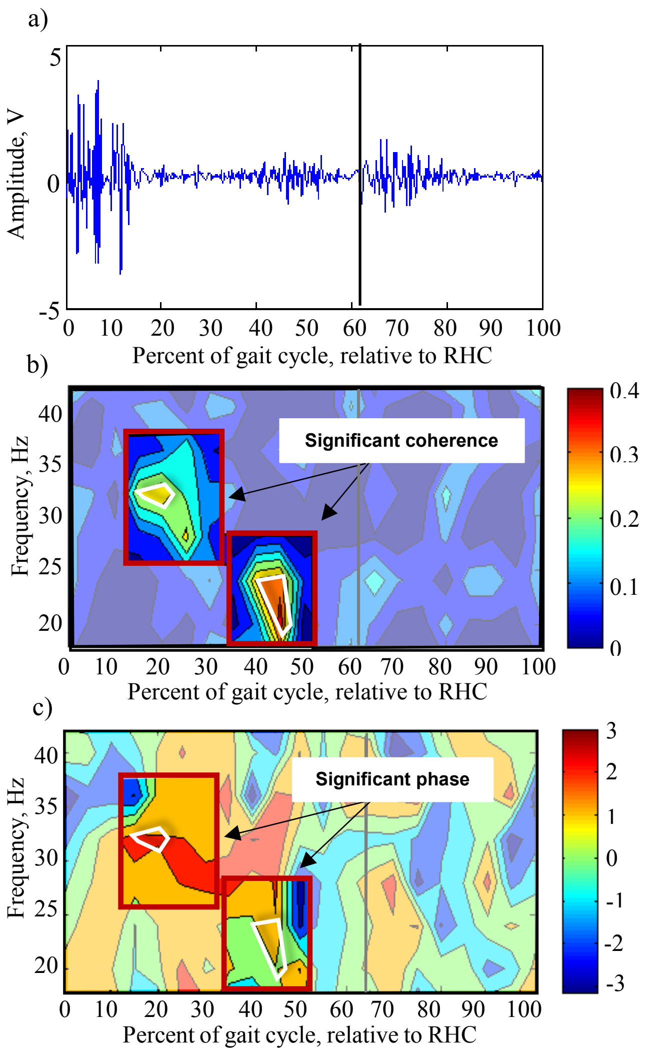

Figure 3: a) RTA EMG activity during a single gait cycle, b) Cz-RTA coherence, and c) the phase, in radians, of Cz-RTA coherency during ramp ascent. A positive phase indicates that EEG lead EMG. Results are plotted relative to right heel contact (RHC). A vertical black line marks right toe off (TO) timing in each plot. Significant coherence, with EEG activity leading EMG activity, was observed in early stance.

IV. Conclusion

Gait phase dependent coupling between the motor cortex and the right *tibialis anterior* was observed during self-paced level ground walking and ramp ascent in the low gamma band. The phase of the coherence indicated that the EEG activity led the EMG activity during the swing phase of level walking and the stance phase of ramp ascent. Based on these results, it is believed that EEG activity over the motor cortex may serve as a useful supplement to EMG-based locomotor assistive devices during level walking. Future work is required to assess the robustness of EEG-EMG coherence during self-paced locomotion across other terrains typical of daily living, such as stairs. Additionally, the study of muscles which actuate joints other than the ankle could identify the potential of EEG-EMG decoding in multi-joint control.

References

[1] M. Bortole *et al*, "The H2 robotic exoskeleton for gait rehabilitation after stroke : early findings from a clinical study," *J. Neuroeng. Rehabil.*, pp. 1–14, June 2015.

[2] M. Wehner *et al* , "A lightweight soft exosuit for gait assistance," in *IEEE ICRA*, pp. 3347–3354, May 2013.

[3] S. Au, M. Berniker, and H. Herr, "Powered ankle-foot prosthesis to assist level-ground and stair-descent gaits," *Neural Networks*, vol. 21, pp. 654–666, Apr. 2008.

[4] J. L. Contreras-Vidal*, A. Kilicarslan, H. Huang, and R. G. Grossman, "Human-Centered design of wearable neuroprostheses and exoskeletonstle," *AI Magazine*, pp. 12–22, 2015.

[5] Y. He *et al*, "An integrated neuro - robotic interface for stroke rehabilitation using the NASA X1 powered lower limb exoskeleton," in 36th Annual IEEE EMBC proceedings, pp. 3985–3988, 2014.

[6] H. Huang, T. A. Kuiken, and R. D. Lipschutz, "A strategy for identifying locomotion modes using surface electromyography," *IEEE Trans. Biomed. Eng.*, vol. 56, no. 1, pp. 65–73, Jan. 2009.

[7] F. Zhang, M. Liu, S. Harper, M. Lee, and H. Huang, "Engineering Platform and Experimental Protocol for Design and Evaluation of a Neurally-controlled Powered Transfemoral Prosthesis," *J. Vis. Exp.*, vol. 89, no. e51059, July 2014.

[8] O. Bai *et al*, "Prediction of human voluntary movement before it occurs," *Clin. Neurophysiol.*, vol. 122, no. 2, pp. 364–372, 2011.

[9] E. C. Wentink, V. G. H. Schut, E. C. Prinsen, J. S. Rietman, and P. H. Veltink, "Detection of the onset of gait initiation using kinematic sensors and EMG in transfemoral amputees," *Gait Posture*, vol. 39, pp. 391–396, 2014.

[10] D. M. Halliday, B. A. Conway, S. F. Farmer, and J. R. Rosenberg, "Using electroencephalography to study functional coupling between cortical activity and electromyograms during voluntary contractions in humans," *Neurosci. Lett.*, vol. 241, pp. 5–8, 1998.

[11] T. H. Petersen, M. Willerslev-Olsen, B. a Conway, and J. B. Nielsen, "The motor cortex drives the muscles during walking in human subjects.," *J. Physiol.*, vol. 590, no. Pt 10, pp. 2443–2452, May 2012.

[12] F. Alton, L. Baldey, S. Caplan, and M. C. Morrissey, "A kinematic comparison of overground and treadmill walking," *Clin. Biomech.*, vol. 13, no. 6, pp. 434–440, Sept. 1998.

[13] A. O. Perotto, *Anatomical Guide for the Electromyographer: The Limbs and Trunk*, 4th ed. Charles C. Thomas Publisher Ltd., 2005.

[14] A. Delorme and S. Makeig, "EEGLAB: an open source toolbox for analysis of single-trial EEG dynamics including independent component analysis," *J. Neurosci. Methods*, vol. 134, pp. 9–21, 2004.

[15] D. M. Halliday *et al*, "A framework for the analysis of mixed time series/point process data - theory and application to the study of physiological tremor, single motor unit discharges and electromyograms," *Prog. Biophys. Mol. Biol.*, vol. 64, no. 2, pp. 237–278, 1995.

[16] G. Nolte *et al*, "Identifying true brain interaction from EEG data using the imaginary part of coherency," *Clin. Neurophysiol.*, vol. 115, no. 10, pp. 2292–2307, Apr. 2004.

[17] S. Hansen, N. L. Hansen, L. O. D. Christensen, N. T. Petersen, and J. B. Nielsen, "Coupling of antagonistic ankle muscles during co-contraction in humans," *Exp. Brain Res.*, vol. 146, pp. 282–292, Oct. 2002.

[18] N. L. Hansen and J. B. Nielsen, "The effect of transcranial magnetic stimulation and peripheral nerve stimulation on corticomuscular coherence in humans.," *J. Physiol.*, vol. 561, pp. 295–306, Sept. 2004.

[19] B. A. Conway *et al*, "Synchronization between motor cortex and spinal motoneuronal pool during the performance of a maintained motor task in man.," *J. Physiol.*, vol. 489, no. 3, pp. 917–924, 1995.