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SPECTRAL PROPERTIES OF MULTIPLE MYOELECTRIC SIGNALS: NEW INSIGHTS INTO THE NEURAL ORIGIN OF MUSCLE SYNERGIES

JULIEN FRÈRE *

University of Lorraine, Laboratory "Development, Adaption and Disability" (EA 3450), Faculty of Sports Sciences, 30 rue du Jardin Botanique, CS 30156, F-54603 Villers-lès-Nancy, France

Abstract—It is still unclear if muscle synergies reflect neural strategies or mirror the underlying mechanical constraints. Therefore, this study aimed to verify the consistency of muscle groupings between the synergies based on the linear envelope (LE) of muscle activities and those incorporating the time–frequency (TF) features of the electromyographic (EMG) signals. Twelve healthy participants performed six 20-m walking trials at a comfort and fast self-selected speed, while the activity of eleven lower limb muscles was recorded by means of surface EMG. Wavelet-transformed EMG was used to obtain the TF pattern and muscle synergies were extracted by non-negative matrix factorization. When five muscle synergies were extracted, both methods defined similar muscle groupings whatever the walking speed. When accounting the reconstruction level of the initial dataset, a new TF synergy emerged. This new synergy dissociated the activity of the *rectus femoris* from those of the *vastii* muscles (synergy #1) and from the one of the *tensor fascia latae* (synergy #5). Overall, extracting TF muscle synergies supports the neural origin of muscle synergies and provides an opportunity to distinguish between prescriptive and descriptive muscle synergies.

Key words: locomotion, motor module, neural control, non-negative matrix factorization, wavelet analysis.

INTRODUCTION

Low-dimensional motor modules formed by muscles activated simultaneously, named muscle synergies, have been proposed to simplify the construction of motor behaviors (Ivanenko et al., 2003; d'Avella and Bizzi, 2005; Ting and McKay, 2007; Torres-Oviedo and Ting, 2007; Ting and Chvatal, 2010). To face the great amount of degrees of freedom of the human body and for a given motor task, the synchronous muscle synergies allow a decrease in the number of variables controlled by the central nervous system (CNS). Thus, rather than individual muscles, it seems that the primary neural element to produce movement is muscle synergy, which is itself controlled by a higher neural command that functionally modulates the pattern of activation of multiple muscles (Rana et al., 2015). In human locomotion, it has been found that a set of a limited number of muscle synergies (four to five) explain the multi-muscle activation and are found to represent functional subtasks of the gait cycle (Ivanenko et al., 2004; Neptune et al., 2009; Chvatal and Ting, 2012). Across a variety of constraints, it has been suggested that the time-varying modulation of similar muscle groupings (*i.e.*, motor modules) may represent the integration of sensory inflows (Cheung et al., 2005; Hug et al., 2011; Safavynia and Ting, 2012, 2013; van den Hoorn et al., 2015). Also, the number of muscle synergies extracted for a given task has been suggested to express the complexity of the neuromuscular control of the motor behavior (Clark et al., 2010). Therefore, muscle synergies may be an integrative, useful tool to analyze the neural structures (spinal cord, brainstem, motor cortex) underlying motor behaviors and to quantify changes related to motor deficit or to the efficiency of any given therapy or rehabilitation program (Safavynia et al., 2011; Ting et al., 2012, 2015; Routson et al., 2013; Roemmich et al., 2014; Wenger et al., 2016).

However, the neural origin of muscle synergies is still a matter of debate within the current literature. It is unclear if muscle synergies effectively reflect the CNS strategies (Bizzi and Cheung, 2013) or simply mirror the underlying mechanical constraints (*i.e.*, descriptive synergies) (Kutch and Valero-Cuevas, 2012; de Rugy et al., 2013). For instance, muscle synergies may be movement-related since non-neural constraints, such as a low-dimensional space of muscle–tendon length change, may explain the dimensionality reduction of multi-muscle activations (Kutch and Valero-Cuevas, 2012). According to Valero-Cuevas (2016) "*The question then is, how can one infer prescriptive synergies (i.e., the existence of synergies of neural origin) from experimental data that naturally exhibit descriptive synergies? This is the heart of the debate in this area at the moment.*" Using the spectral properties of the surface electromyographic (EMG) signals has been found to be another approach to determine the neural structures underlying the muscle activation. More specifically, frequency bands from

*Fax: +33 372 746 712.
E-mail address: julien.frere@univ-lorraine.fr
Abbreviations: BF, *biceps femoris*; CNS, central nervous system; CV, coefficient of variation; EMG, electromyographic; G_{max}, *gluteus maximus*; LE, linear envelope; LG, *gastrocnemius lateraleris*; MG, *gastrocnemius medialis*; NMF, *non-negative matrix factorization*; RF, *rectus femoris*; SOL, *soleus*; ST, *semitendinosus*; TA, *tibialis anterior*; TF, time–frequency; TFL, *tensor fascia latae*; VAF, *variance accounted for*; VL, *vastus lateralis*; VM, *vastus medialis*; wt, *wavelet*.

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EMG-EMG coherence might reflect subcortical [~10 Hz; Grosse and Brown (2003), Boonstra et al. (2009)] or cortical [20–60 Hz; Grosse et al. (2002)] pathways. For instance, during a postural task Danna-Dos-Santos et al. (2014, 2015) found significant peaks of intermuscular coherence within the low-frequency bands (0–5 and 5–20 Hz) among muscles grouped in functional synergies. These results corroborated the neural origin hypothesis of muscle synergies to lower the dimensionality of the neuromuscular control. In combination with the extraction of muscle synergies during a pedaling task, De Marchis et al. (2015) determined that solely the knee extensors muscle synergy had a significant peak of EMG-EMG coherence within the 30–60-Hz frequency band, likely reflecting a cortically mediated muscle synergy to produce power during the descending phase of the pedaling cycle. This result also suggested that the other muscle synergies would be descriptive of the mechanical constraints of the pedaling task.

Therefore, coupling intermuscular coherence analysis with the extraction of muscle synergies might be a promising approach to discriminate prescriptive muscle synergies from descriptive ones. However, such a method does not allow investigating any change in frequency as function of time of activation. Indeed, the extraction of muscle synergies is a time-domain analysis while the EMG-EMG coherence provides correlates solely in the frequency-domain. Moreover, it has been showed that a similar EMG envelope could be explained by different underlying time–frequency patterns (Wakeling, 2004; Hodson-Tole and Wakeling, 2007; Frère et al., 2012a). Consequently, a method able to extract synergies composed of muscles sharing similar time–frequency features would provide new evidences relative to their potential neural origin.

The aim of this study was to propose a new method of muscle synergies extraction that incorporates the spectral properties (*i.e.*, time–frequency domain) of multiple muscle activities and to verify the consistency of muscle groupings with muscle synergies based on the global muscle activities (*i.e.*, time domain) during human gait. In considering that the muscle synergies are of neural origin, it was hypothesized that the muscle vectors (*i.e.*, motor modules) were similar across the two methods of extraction, whatever the walking velocity. In case of discrepancy between the methods of muscle synergy extraction, one might consider the time–frequency muscle synergies as a new tool to distinguish prescriptive from descriptive muscle synergies.

EXPERIMENTAL PROCEDURES

Participants

Twelve volunteers (10 men and 2 women, age: 31.9 ± 9.3 years, height: 178 ± 8 cm, body mass: 77 ± 10.8 kg) participated in this study. They were informed of the purpose of the study and methods used before providing written consent. The experimental procedure was carried out in accordance with the principles of the Declaration of Helsinki.

Protocol

Participants were asked to walk overground within a corridor at a self-selected speed. Two walking self-selected speeds were assessed: a comfort condition and a fast condition. For each condition, each participant walked at least 20 m three times, in order to assess 10 walking cycles per trial (the first and last walking cycles were not retained). At least, 30 walking cycles were recorded and analyzed for each condition. All participants began with a comfort trial but the order of the five following trials was randomized. A walking cycle was defined as the time between two consecutive heel strikes of the same foot.

Materials and data collection

The 20-m walking distance was materialized by means of three pairs of ground cone markers, each 10-m apart. The 20-m walking time was manually recorded with a digital chronometer between two instants: when the foot left the ground at the walking initiation and when the participant crossed the last pair of ground markers.

The activity of eleven muscles of the right side of the body was simultaneously recorded: *tibialis anterior* (TA), *soleus* (SOL), *gastrocnemius lateraleris* (LG), *gastrocnemius medialis* (MG), *vastus lateralis* (VL), *rectus femoris* (RF), *vastus medialis* (VM), *biceps femoris* (long head, BF), *semitendinosus* (ST), *tensor fascia latae* (TFL), and *gluteus maximus* (G_{max}). The EMG activity was recorded using wireless electrodes (Delsys Trigno™ Wireless System, Boston, MA, USA) with an inter-electrode distance of 10 mm. The electrodes were placed longitudinally with respect to the underlying muscle fiber arrangement (de Luca, 1997) and were located according to recommendations of Surface EMG for Non-Invasive Assessment of Muscles [SENIAM, Hermens et al. (2000)]. Before applying electrodes, the skin was shaved and cleaned with alcohol to minimize impedance. Raw EMG signals were preamplified (gain 300, bandwidth 20–450 Hz) at a sample rate of 2000 Hz. Two triaxial accelerometers (Delsys Trigno™ Wireless System, Boston, MA) were placed at the level of the third metatarsal and of the heel (sampling rate 148.18 Hz). All the raw signals (EMG and accelerations) were synchronized and stored in digital format using EMGworks® Acquisition software (Delsys, Boston, MA). Data were then processed offline with custom-built Matlab® scripts (The Mathworks, Natick, Massachussetts, USA).

Data analysis

From the 20-m walking time, the mean walking velocity (in $m.s^{-1}$) was calculated. The longitudinal acceleration (*x*-axis) of the heel accelerometer and the frontal acceleration (*z*-axis) of the metatarsal accelerometer were used to determine each heel contact with the ground. Both signals of acceleration were rectified and smoothed with a zero lag low-pass filter (5 Hz, Butterworth filter, 2nd order). Both signals had common peaks which corresponded to the heel strikes and thus