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Global lower limb muscle coactivation during walking at different speeds: Relationship between spatio-temporal, kinematic, kinetic, and energetic parameters

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ABSTRACT

Muscle coactivation is the mechanism that regulates the simultaneous activity of antagonist muscles around the same joint. During walking, muscle joint coactivation varies within the gait cycle according to the functional role of the lower limb joints. In the present study, we used a time-varying multi-muscle coactivation function (TMCf) with the aim of investigating the coactivation of 12 lower limb muscles and its relationship with the gait cycle, gait speed (low, self-selected, and fast), ground reaction force, gait variability, and mechanical energy consumption, and recovery in a sample of 20 healthy subjects. Results show that the TMCf is speed dependent and highly repeatable within and between subjects, similar to the vertical force profile, and negatively correlated with energy recovery and positively correlated with both energy consumption and balance-related gait parameters. These findings suggest that the global lower limb coactivation behavior could be a useful measure of the motor control strategy, limb stiffness, postural stability, energy efficiency optimization, and several aspects in pathological conditions.

1. Introduction

Muscle coactivation is the mechanism that regulates the time and amplitude of simultaneous activity of antagonist muscles around the same joint (Le et al., 2017; Rosa et al., 2014; Olney, 1985). During ground surface linear walking, muscle joint coactivation varies within the gait cycle, according to the functional role of the lower limb joints along gait phases, reaching higher values during weight acceptance and transition from stance to swing subphases (Falconer and Winter, 1985), and lower values during mid-stance (Olney, 1985). In addition to gait phases, several other factors influence the rate of the muscle coactivation during locomotion, including age (Franz and Kram, 2013; Hortobagyi et al., 2009), speed (Peterson and Martin, 2010), and motor context, i.e., stable vs. unstable conditions (Martino et al., 2015). Increased coactivation has been reported in patients affected by several gait disorders characterized by reduced balance, reduced muscle strength, or increased joint laxity (Martino et al., 2014; 2015; Mari et al., 2014; Rinaldi et al., 2017; Serrao et al., 2016; Kitatani et al., 2016; Boudarham et al., 2016). Essentially, the Central Nervous System (CNS) employs muscle coactivation as a motor control mechanism to modulate joint stiffness and postural stability, optimize energy efficiency, enhance movement accuracy, and allow adaptation to

Abbreviations: CNS, Central Nervous System; sEMG, surface ElectroMyoGraphy; TMCf, time-varying multi-muscle coactivation function; GRFs, ground reaction forces; SS, comfortable self-selected speed; L, comfortable low speed; F, comfortable fast speed; $FWHM_{TMCf}$, full width at half maximum; CoA_{TMCf}, center of activity; CMC, coefficient of multiple correlation; CMC_{IS}, coefficient of multiple correlation_intra-subject; CMC_{TMCf_IS}, coefficient of multiple correlation_TMCf_intra-subject; CMC_{TMCf_BS}, coefficient of multiple correlation_TMCf_between-subjects; CMC_{TMCf_speed}, coefficient of multiple correlation_TMCf_speed; VF, vertical force; $FWHM_{VF}$, full width at half maximum vertical force; CoA_{VF}, center of activity vertical force; Normalized cross-correlation function, R_{xy}; Temporal shift, τ^*; R-step, energy recovery; TEC, total energy consumption (TEC); RoM, range of motion; SD, standard deviation

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environmental demands (Darainy and Ostry, 2008; Simmons and Richardson, 1988). However, increased coactivation implies reduced power and increased metabolic cost and compressive forces across the joints, which may in turn, lead to cartilage loss (Collins et al., 2011; Lewek et al., 2004).

Single joint coactivation has been quantified using mathematical tools derived from agonist–antagonist approaches, requiring an a priori sorting of the muscles depending on the moment generated at a given joint. According to this approach, a series of different methods have been proposed based on the ratio of overlapping or cross-sectional areas below the signals of surface electromyography (sEMG), as recorded from two antagonist muscles or muscle groups of the same joint (Falconer and Winter, 1985; Damiano et al., 2000; Rudolph et al., 2000; Kellis et al., 2003; Don et al., 2007; Brookham et al., 2011). Peterson and colleagues (Peterson and Martin, 2010) have investigated the coactivation of many muscles around both the shank and thigh. However, they used a mathematical method based on the sum of coactivations calculated for the single joints according to an agonist-antagonist approach. Nevertheless, analyzing muscle coactivation at a single joint, or as the sum of those calculated at single joints, does not fully explain the global strategy exerted by the CNS in controlling and modulating the neuromuscular output. The motor system coordinates muscles, combines and hierarchically controls muscle synergies, regulates local spinal interneuronal reflexes, and synchronizes the neural systems, throughout the CNS, into an integrated and adaptive motor behavior (Torres-Oviedo et al., 2006; Tresch, 2007).

Thus, a global characterization of lower limb muscle coactivation during walking may be helpful to understand the general strategy adopted by the CNS to control the whole lower limb depending on the motor context i.e., gait phases, balance, speed, and metabolic cost. We hypothesized that the simultaneous activation of the lower limb muscles was modulated by gait speed and torque production and correlated with energy cost and gait stability.

In the present study, we used a time-varying multi-muscle coactivation function (TMCf) (Ranavolo et al., 2015) with the aim of investigating the relationship between global lower limb muscle coactivation and gait cycle, speed, ground reaction force (GRF), gait variability, and mechanical energy consumption in a sample of healthy subjects.

2. Methods

2.1. Subjects

Twenty healthy subjects were recruited (8 women and 12 men; mean age, 40 ± 13.81 years; Body Mass Index (BMI), 24.86 ± 3.35 kg/m^2). None of the subjects had pathologies known to influence the normal gait pattern. All participants provided written informed consents, and the study design complied with the Declaration of Helsinki and was approved by the local ethics committee.

2.2. Experimental procedure

We recorded sEMG signals at 1000 Hz using a bipolar 16-channel wireless system (FreeEMG 300 System, BTS). After skin preparation, Ag/AgCl surface electrodes (Kendall ARBO) were placed over the muscle belly in the direction of the muscle fibers according to the European Recommendations for Surface Electromyography (Hermens et al., 2000). A pre-processing filtering and denoising procedure was performed (Hamming filter between 10 and 400 Hz and common mode reaction ratio equal to 100 dB). Pairs of electrodes (inter-electrode distance, 2 cm) were placed unilaterally on the dominant side of each participant on the gluteus medius, rectus femoris, vastus lateralis, vastus medialis, tensor fascia lata, semitendinosus, biceps femoris, tibialis anterior, gastrocnemius medialis, gastrocnemius lateralis, soleus, and peroneus longus.

GRFs were measured at the sampling rate of 680 Hz by eight dynamometric platforms (P6000, BTS).

Kinematic data were recorded by using an eight-infrared-camera optoelectronic motion analysis system at a sampling frequency of 340 Hz (SMART-DX 6000 System, BTS). Twenty-two reflective spherical markers were attached to the anatomical landmarks (Davis et al., 1991). Acquisition of sEMG, kinetic, and kinematic data was synchronized.

Subjects were asked to walk barefoot at comfortable self-selected (SS), low (L), and fast (F) gait speeds along a walkway approximately 10 m in length. Because we were interested in natural locomotion, only general, qualitative, verbal instructions (no analog or digital metronomes were used) were provided. Before the recording session, the subjects practiced for a few minutes to familiarize themselves with the procedure. Ten trials at three gait speeds each (total 30 trials) were recorded per subject.

2.3. Data analysis

The data were processed using a 3D reconstruction software (SMART Tracker and SMART Analyzer, BTS) and MATLAB (8.3.0.532, MathWorks). Electromyographic, kinetic, and kinematic data were time-normalized to the duration of the gait cycle (time between two successive foot contacts of the same leg) and interpolated to 201 samples using a polynomial procedure. Heel-strike and toe-off events were determined as in the study by Serrao et al. (2016).

2.3.1. Global coactivation of lower limb muscles

The raw sEMG signals were band-pass filtered using a zero-lag third-order Butterworth filter (20–400 Hz), full wave rectified, and low-pass filtered with a zero-lag fourth-order Butterworth filter (10 Hz). For each individual, the sEMG signal from each muscle was normalized to its peak value across all trials (Burden, 2010). From the processed sEMG signals, we calculated the simultaneous activation of the 12 lower limb muscles by considering the TMCf (Ranavolo et al., 2015; Serrao et al., 2016, Le et al., 2017). This sigmoid-weighted, time-dependent function for the inclusion of multiple muscles during walking was designed to receive full-wave-rectified, low-pass-filtered, and 0–100 amplitude normalized sEMG signals as inputs. Sample values of this function ranged from 0 to 100 and are calculated by the following equation:

$$TMCf\,(d(i),\,i) = \left(1 - \frac{1}{1 + e^{-12(d(i)-0.5)}}\right) \cdot \frac{\left(\sum_{m=1}^{M} EMG_m(i)/M\right)^2}{max_{m=1\cdots M}\left[EMG_m(i)\right]}$$

where M is the number of muscles considered, $EMG_m(i)$ is the sEMG sample value of the m_{th} muscle at the instant i, d(i) is the mean of the differences between each pair among the twelve $EMG_m(i)$ samples at the instant i:

$$d(i) = \left(\frac{\sum_{m=1}^{M-1}\sum_{n=m+1}^{M} |EMG_m(i) - EMG_n(i)|}{(M!/(2!(M-2)!))}\right)$$

$M!/2!(M-2)!$ is the total number of possible differences between each pair of $EMG_m(i)$. The 201 samples $TMCf\,(d(i),\,i)$has the following properties: inverse relationship with the mean of the differences d(i), values close to the mean activation of the m(i) muscle sample values considered when d(i) is close to 0, and values close to 0 when d(i) is close to 1. In particular, the smaller the differences in muscle sample activation, the closer the d(i) values are to 0 and the closer the sigmoid-coefficient values are to 1, leaving the $TMCf\,(d(i),\,i)$value close to the value of its mean. Inversely, the greater the differences in muscle activations, the more d(i) increases and the more the sigmoid coefficient decreases, thereby reducing the TMCf (d(i),i) values. For each subject and gait speed, data over individual strides were calculated and then averaged across cycles.