Treadmill walking and running: kinematic, electromyographic and muscle synergies differences

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Abstract: The purpose of the current study is three-fold: it's proposed the analysis and the comment of the results obtained comparing both treadmill walking (1.4 m/s) and running (2.8 m/s) tasks, carried out by a group of 30 young adults, kinematically, electromyographically and focusing on the muscle synergy recruitment. In particular, the parameters taken into account among the investigation are: stance, swing and cadence (kinematic), time-normalized to gait-cycle length sEMG envelopes (electromyography) and motor modules and primitives (muscle synergies). Focusing on time-dependent parameters, performing a t-test it has been found a statistical significance ($p<0.001$) between gait variables in the two tasks. Also EMG-based differences have been evaluated through a t-test, which made it possible to appreciate statistical significances ($p<0.001$), for all of the 13 ipsilateral muscles analyzed; considering the entire gait, the envelopes that have shown more statistical differences are those related to VL and VM (respectively 81.1% and 82.5% of gait cycle). Muscle synergy recruitments have been compared performing a cosine-similarity measure (CS) to motor modules and a t-test to motor activation coefficients: muscle synergy weights are 93.7% to 98.3% similar between different tasks, and motor primitives showed statistical significance ($p<0.001$) mostly in the propulsion synergy.

Introduction

The central nervous system (CNS) plays the main role in motor control since it activates each muscle modulating the magnitude, the timing and the pattern of activation. Even though the way CNS controls and coordinates the contraction of different muscles during a specific motor tasks, generating by all muscle units in all muscles the appropriate amount of torque in each joint, still represents an open issue in the motor control field, a long-standing idea is that motor control may be simplified by a modular organization. Under this hypothesis, the control problem is reduced to modulating an appropriate selection of an adequate number of motor modules, also called muscle synergies, resulting in a simplified control of movement [3].

This way, the CNS combines a small number of muscle synergies (smaller than the number of muscles) to generate motor commands, showing regular patterns in the motor output and bypassing the musculoskeletal system redundancy.

In fact, it consists of a greater number of muscles than of joints; hence, a wide range of combinations of muscle patterns can produce the same movement. The aim of this study is to evaluate and compare the muscle patterns showed both in the walking and running task, in order to uncover possible differences or analogies in the recruitment of muscle synergies by the CNS during different locomotion conditions.

Moreover, the proposed investigation wants to analyze these differences from a kinematic and electromyographic point of view too.

In fact, as well as for synergic motor control, it is interesting to understand whether there are similarities in the level and timing of activation of a single muscle (electromyography) or to quantify the relationship between the phase of stance and swing (kinematics) performing the task at different speeds. Lastly, in addition to the determination of differences from a single point of view (e.g. the kinematic one), the purpose of this study is to determine whether these analyses are closely related or not, so if a difference showed in a given study is necessarily reflected to another one.

Materials

Population: Thirty healthy and regularly active young adults (15 females and 15 males, height 173 ± 10 cm, body mass 68 ± 12 kg, age 28 ± 5 years) composed the study population. None of them was using orthotic insoles, had any history of neuromuscular or musculoskeletal disorders, or any head or spine injury at the time of the measurements or in the previous 6 months. All the volunteers completed a self-selected warm-up running on a treadmill, typically lasting 3-5 minutes. The experiment protocol consisted of walking at 1.4 m/s and running at 2.8 m/s on a single-belt treadmill equipped with a pressure plate recording the plantar pressure distribution at 120 Hz.

Data organization: The available data are organized in two structures. The first one contains the gait cycle breakdown: it is defined as a structure with 60 fields (one for each trial and for each subject). Each field is structured as a N × 2, where N is the number of gait cycles acquired for each trial. The first column contains the touchdown incremental times in seconds, the second column contains the duration of each stance phase in seconds. The second structure contains the raw sEMG data and it is made up of 60 fields (one for each trial and for each subject); electromyographic signals, recorded with an acquisition frequency of 2 kHz, refer to the following 13 ipsilateral (right side) muscles: Fasciae Latae (FL), Rectus Femoris (RF), Vastus Medialis (VM), Vastus Lateralis (VL), SemiTendinosus (ST), Biceps Femoris (BF), Tibialis Anterior (TA), Peroneus Longus (PL), Gastrocnemius Medialis (GM) Gastrocnemius Lateralis (GL), and Soleus (SOL).

Methods

Firstly, the investigation started with a data pre-processing step: data structures (sEMG_RAW and TIME_CYCLES) have been re-organized selecting only the desired gait cycles. In particular, due to the initial and final 10 gait cycles related to the adaption phenomenon on treadmill, and because of the choice to study muscle synergies for a smaller number of gait cycles, 30 (15^{th} to 45^{th}) gait cycles have been considered.
To speed up the following steps, the samples related to the beginning and end of the stance-phase for a certain step, for each subject and task, were derived from the incremental values contained in structure TIME_CYCLES.

Kinematic analysis: The kinematic analysis is based on the extraction and comparison of stance, swing and cadence values between walking and running tasks. As mentioned above, stance time intervals were present in available data structures; swing durations have been derived subtracting, from the i_{th} touchdown time instant (when the gait cycle starts), the instant the $i\text{-}1_{th}$ stance phase ends. Lastly, cadence parameter have been extracted as the steps done in a minute: $cadence = \frac{2 \cdot n_{cycles} \cdot 60\ s}{total\_time\_acquisition}$, where the denominator is equal to the time interval which comprises the selected gait cycles.
In order to quantify the differences of such parameters in different tasks, for each patient, average values over 30 gait cycles are calculated. A t-test is performed between 30 (number of subjects) average gait parameters of walking task and 30 of running task.

EMG signals post-processing: It's been performed a sequence of processing on raw EMG signals to make it possible the comparison between the two tasks electromyographically and from a muscle synergy recruitment point of view.

Signals were high-pass filtered using an 8^{th} order IIR Butterworth zero-phase filter with cut-off frequency 35 Hz, full- wave rectified, and lastly low-pass filtered using a 5^{th} order IIR Butterworth zero-phase filter with cut-off frequencies 12 Hz to obtain the envelops of the signals [1].

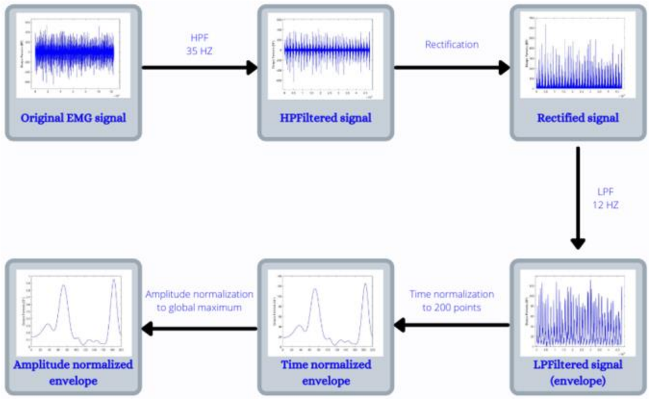

Figure 1 | EMG signal post processing pipeline

To eliminate the small oscillations present, which cause the presence of some negative values, they were imposed to zero, obtaining positive envelopes. The following step of the EMG post-processing pipeline consists in time and amplitude normalization. In particular, every muscle envelope has been time-normalized to 200 points, assigning 100 points to the stance and 100 points to the swing phase. Time normalization has been carried out by segmenting the entire signal in gait cycles (exploiting the knowledge of the moments in which each cycle begins) and interpolating every segment from its initial number of samples to the desired one.
The reason for this choice is two-fold: firstly, dividing the gait cycle into two macro-phases helps the reader to understand the temporal contribution of the different synergies, diversifying between stance and swing; secondly, the normalization of stance and swing durations to the same number of points for all participants (and for all the recorded gait cycles of each participant) makes the interpretation of the results independent from the absolute duration of the gait events. Furthermore, a lower number of points contained in every gait cycle allows lower computational time and cost.
Then, amplitudes of time-normalized envelopes were normalized with respect to its global maximum for each muscle [2].

EMG analysis: Obtaining for each patient and muscle 30 gait cycles envelopes from the previous processing pipeline, electromyographic analysis has been conducted performing a t-test, for each patient and muscle, between the 30 (number of gait cycles) envelopes of the walking and of the running tasks. For each of 30 (number of patients) tests performed to each muscle, statistical significance percentage (ratio between number of non-null hypothesis and total hypothesis) has been calculated. Finally, for each muscle, their mean and standard deviation have been calculated.

Muscle Synergy Extraction: Muscle synergies of each and every subject have been extracted from the recorded EMG activity using the classical Gaussian Non-Negative Matrix Factorization (NNMF) algorithm, implemented by the functions in the library *myMuscleSyneriesLibrary*. It was chosen 30 gait cycles for each subgroup (imposed to 1), a minimum number of synergies equal to one and a maximum number of synergies equal to 8, since the number of muscle synergies is always less than the number of muscles (in this case, 13). This algorithm doesn't require an a-priori knowledge of the optimal number of synergies involved in a certain task by the CNS but, iteratively, it reconstructs the EMG signal matrix (containing all muscles envelopes) by the product between motor modules and motor primitives' vectors. After the NNMF implementation, it was required a quality assessment method in order to select the optimal number of synergies for each subject (VAF method – threshold imposed to 90%).

Results and discussion

Kinematic results: Gait parameters (stance and swing time intervals and cadence) are reported in Figure 2-3-4. By performing a t-test, it was found that, in running, swing phase duration and cadence are significantly lower and stance phase duration is significantly higher in respect to values assumed in walking task ($p < 0.001$).

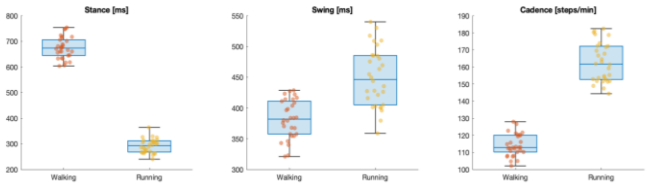

Figure 2-3-4| Boxplots of the gait parameters (stance, swing, cadence) related to all the 30 patients, for both treadmill walking and running tasks.

Electromyographic results: As mentioned in 'Methods' section, a t-test was performed to envelopes related to the selected 30 gait cycles to compare the two tasks for each patient. Once obtained the results, they have been averaged across the population to allow an overview of the differences between EMG activations during different speeds of the same task. The results are showed in Table 1, as the percentage of statistical significance with respect to the entire vector considered (percentage of non-null hypothesis in a set of samples, 200 for gait cycle and 100 for stance and swing phases). All muscle activations are statistically significant, for at least the 60% and up to a maximum of 82.5% of the total gait cycle.

In particular, VL and VM showed the highest number of non-null hypothesis (81.1% and 82.5% respectively). After highlighting a statistically significant difference between the two tasks, it was seen that the muscle

Statistical significance percentage			
Muscle	**Gait cycle**	**Stance phase**	**Swing phase**
ME	66.9 ± 1.9 %	70.3 ± 2.3 %	63.7 ± 2.3 %
MA	80.1 ± 1.9 %	78.3 ± 2.1 %	81.8 ± 2.2 %
FL	62.8 ± 2.4 %	67.8 ± 2.8 %	57.8 ± 2.8 %
RF	75.9 ± 2.2 %	74.9 ± 2.5 %	77.0 ± 2.5 %
VM	82.5 ± 1.8 %	77.6 ± 2.1 %	87.5 ± 2.3 %
VL	81.1 ± 2.8 %	76.5 ± 2.3 %	85.8 ± 2.3 %
ST	64.1 ± 1.9 %	68.9 ± 2.8 %	59.3 ± 2.8 %
BF	64.5 ± 1.8 %	65.9 ± 2.7 %	63.2 ± 2.7 %
TA	69.0 ± 2.2 %	72.5 ± 2.4 %	65.5 ± 2.4 %
PL	64.3 ± 2.1 %	70.9 ± 1.9 %	57.7 ± 2.0 %
GM	67.9 ± 2.4 %	71.9 ± 2.7 %	64.0 ± 2.7 %
GL	75.3 ± 2.6 %	79.4 ± 1.9 %	71.2 ± 2.0 %
SOL	78.2 ± 2.5 %	79.8 ± 2.3 %	76.6 ± 2.3 %

Table 1| Statistical significance percentage (μ ± SE) for each muscle envelope over the population; data are reported for the total gait cycle and considering only the stance phase and considering only the swing phase

activations related to stance phase are more rapid and intense in running than in walking (where there are more gradual activations), probably due to a shorter stance phase, as seen in the kinematic results. Figure 5 is showed as example.

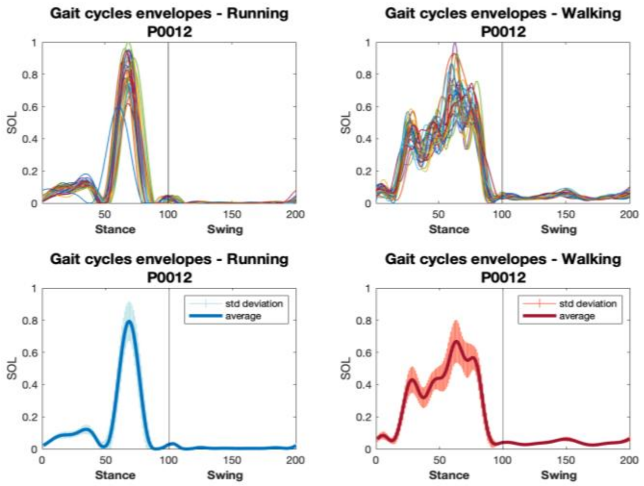

Figure 5| Example of gait cycles envelopes and averaged envelopes of P0012 (SOL) in walking and running tasks

Muscle Synergies results: After the choice of VAF as quality assessment method for the optimal number of muscle synergies, it was temporarily selected the minimum n (number of synergies) that overcame the imposed threshold (90%). Analysing its trend across the population, the minimum number of synergies necessary to reconstruct the EMG data was 4.2 ± 0.7 for running and 4.3 ± 0.6 for walking (μ ± SD). In order to evaluate differences and analogies in recruitment between the two tasks, it was decided to consider the same number of synergies for every subject: the mode of the distribution (that was also equal to its mean) was selected ($n = 4$). To understand the quality of this choice, the average VAF and the standard deviations were calculated, obtaining 91.0% ± 2.4% for running and 90.4% ± 1.9% for walking. In addition, local VAF has been implemented to evaluate whether average values were correctly distributed among all muscles or not: it's noticeable, for the running task, a number of muscles of 1 up to 4 presented a lower value than the imposed local threshold (75%) for a few subjects (mainly for the Soleus, probably due to its deep localization). Then, the following step consisted in the muscle synergies sorting, which was essential to the consequent averaging operation across the population (to obtain average motor modules and motor primitives). Sorting was carried out implementing the cosine similarity measure, $CS = \frac{W_i \cdot W_j}{||W_i|| \cdot ||W_j||}$, which quantifies the similarity between the two vectors W_i and W_j. Hence, it was firstly selected a reference subject (P0007), featured by a correct sorting of its synergies (in respect to average results found in literature); then each of its weights was compared through CS with all weight vectors of another subject. This way, assuming a greater similarity between two motor modules that must occupy the same position in the final order, the correct sorting was achieved. For those subjects that had a higher variability in the motor modules, and consequently that led to an incorrect sorting through CS, the matching was carried out manually.

Four fundamental synergies were clustered in all gait conditions. In both walking (Figure 6) and running (Figure 7), the first synergy functionally referred to the body weight acceptance, with a major involvement of knee extensors and hip extensors and abductors. The second synergy described the propulsion phase, to which the plantarflexors mainly contributed. The third synergy identified the early swing, showing the involvement of foot dorsiflexors. The fourth and last synergy reflected the late swing and the landing preparation, highlighting the relevant influence of knee flexors (in both walking and running), and foot dorsiflexors [2]. Finally, statistical comparisons could be carried out. It was implemented a CS measure to evaluate the similarity between average weights (calculated among the population) of different tasks.

CS_{S1}	CS_{S2}	CS_{S3}	CS_{S4}
0.968	0.937	0.944	0.983

Table 2| CS measure of muscle synergy weights between different tasks

The results, which are presented in Table 2, show a high similarity between the synergies weights despite the different task, emphasizing that the muscles involved are not dependent on the speed with which the motor task is carried out.

To assess statistical differences of activation coefficients, it was performed a t-test to motor primitives of all subjects (for a given synergy) at different speeds: the results are showed in Table 2, as the percentage of statistical significance with respect to the entire vector considered (percentage of non-null hypothesis in a set of samples, 200 for gait cycle and 100 for stance and swing phases).

Here, in opposition to the motor modules analysis, the study was conducted for each point of the activation coefficients vector, to understand statistical significances with a higher resolution. The synergies extracted, between the two tasks, showed, on average, statistical significance of about 25% (except synergy 3

Synergy	S_1	S_2	S_3	S_4
Gait	26.5%	32.5%	7%	21%
Stance	37%	65%	10%	40%
Swing	16%	0%	4%	2%

Table 3| Percentage of statistical significance for each synergy with respect to the total length of gait cycle, to stance phase and to swing phase

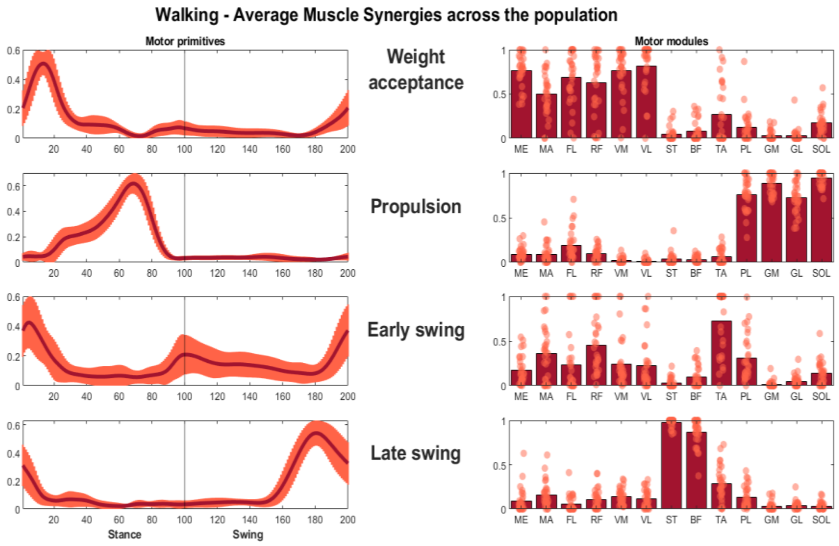

Figure 6| Plot of averaged motor primitives and motor modules extracted during the walking task

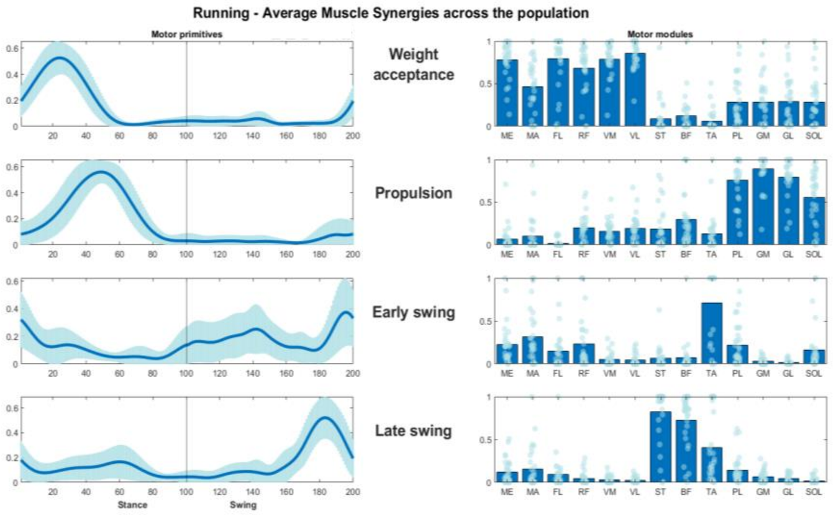

Figure 7| Plot of averaged motor primitives and motor modules extracted during the running task

which is statistically different in only the 7% of samples); in particular, each of them presents a higher value during the stance phase than the swing one (Table 3).

Finally, in order to have an overall view of the recruitment in the time of the four muscle synergies, the heat-maps of both tasks have been reported in Figure 8-9.

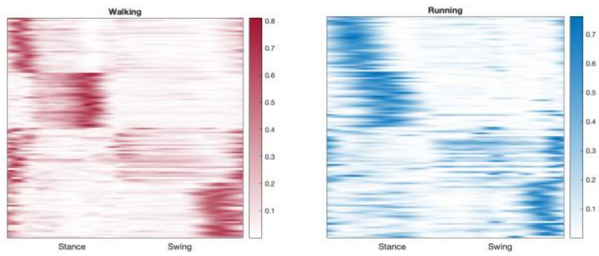

Figure 8-9| Heat-maps of muscle synergies

The samples belonging to the gait cycle are represented on the columns, the synergies of single subjects are represented on the rows.
As shown in Figure 8-9, the timing of muscle synergy activation is quite similar between the walking and running task; small differences can be observed in the first two synergies starting from the top, as the longer activation in the weight acceptance and the almost constant activation in propulsion (running), in contrast to a very early peak of activation in the first synergy and a growing one in propulsion (walking). These results reflect the statistical significances of motor primitives observed mainly in the stance phase.

Conclusions

The proposed study aims to analyze the kinematic, electromyographic and muscle synergy recruitment features in both walking and running tasks over a population of thirty subjects, and to quantify and evaluate their differences.
After the re-organization of the available data structures and a EMG signal processing, it was conducted the extraction of the most important parameters, such as stance and swing duration and cadence (for a kinematic point of view), envelopes (for the EMG-based analysis) and motor modules and primitives (for the muscle synergy extraction). Several comparisons have been performed. The investigation highlighted clear differences between the two tasks for kinematic and electromyographic parameters, while it led to more similar results for synergistic recruitment.
In fact, the weights of the four synergies have a high coefficient of similarity and the activation coefficients show lower statistically significant differences (which are mainly in the stance phase, as shown in the heat-maps).
Thus, the results obtained allow to conclude that the recruitment of muscle synergies is not highly variable with the task performed, such as kinematic and electromyographic parameters.

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