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Similarity of muscle synergies in human walking and cycling: preliminary results*

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Abstract— **Recent investigations on how the Motor System coordinates different tasks in humans have indicated that a low-dimensional structure of muscle synergies is sufficient to explain specific spatiotemporal components underlying such behaviors. In this work, we tested the hypothesis that pedaling and walking share common modular features by using the muscle synergies paradigm. Seven healthy subjects walked on a treadmill at their maximum speed and also cycled in an ergometer, set at the same walking cadence. EMG activity was recorded from 10 muscles of the most dominant leg. A Non-Negative Matrix Factorization algorithm was applied to extract synergies. Four synergies were sufficient to explain 90% of the EMG variability during walking and cycling. There were statistically significant correlations (higher than 71%) across similar synergies for each task (walking and pedaling). These preliminary results support the hypothesis of modular control across different human motor tasks and may indicate that some synergies are shared amongst different rhythmic movements.**

I. Introduction

Over the last two decades, experiments in animals and humans have demonstrated that the analysis of muscle synergies can reveal a modular structure that mediates muscle activity during multi-limb movements. Such modular organization suggests the presence of a low-dimensional library of muscle groups which, when adequately combined, can lead to high-dimensional coordinated movements. Similarly, there is evidence that the same synergies are shared across different biomechanical conditions, such as variations in speed [1], posture [2] and load [3] variations.

In this work, we aim to test the hypothesis that muscle synergies are also shared across different motor functions, such as walking and pedaling. Cheung et al. [4] suggested that the majority of the synergies used for generating locomotor behaviors are centrally organized, but their activation may be modulated by sensory feedback so that the final motor outputs are adapted to the external environment [8]. In this study this hypothesis was tested by identifying similarities in muscle synergies weights (i.e. the time-invariant contribution of each muscle within a muscle group), that might reflect the activation of central control mechanisms.

Previous work already suggests that different forms of rhythmic movements may share common neuromuscular pattern. According to the "common core hypothesis" proposed by Zehr [5], pedaling, stepping and walking may have common central control mechanisms. Also Pacheco et al. [6] show that rehabilitation treatments based on combined hand and pedaling movements may have positive outcomes on walking. Should similarities in neuromuscular behavior between walking and pedaling be demonstrated, pedaling might gain potential relevance as a diagnostic and rehabilitation scenario for people with impaired locomotion.

During walking, only 4-5 muscle synergies are required to account for whole muscle activity of several lower leg muscles [7] [3]. During pedaling, muscle activation can be explained by the combination of three muscle synergies among trained cyclists [8] and four muscle synergies among non-professional subjects [9]. When walking and pedaling are analyzed separately, muscle synergies are maintained across different mechanical constraints (torque, velocity, posture), showing only a few timing adjustments [10] [9], and the inter-individual variability of EMG patterns observed does not represent differences in the adopted locomotor strategy [8].

A first attempt to compare modular control of walking and pedaling can be achieved by comparing different studies (e.g. [7] and [10]). It is noteworthy that the same muscle groups are activated, even for the different tasks, which supports the hypothesis of shared synergies across different rhythmic movements. To our knowledge, the work presented is the first study to evaluate the possible existence of shared modular control between walking and cycling by using the muscle synergies analysis paradigm.

II. Materials and Methods

A. Participants

Seven healthy subjects (4 males and 3 females; age: 27 ± 2.9 years; weight: 71.6 ± 13.6 Kg; height: 175.4 ± 8.85 cm) with no neurological injuries or gait disorders volunteered to participate in this research study. Before giving their written consent to participate, they were informed about the procedures and possible discomfort associated with the experiments. A local committee provided ethic approval for the experimental design of this study.

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B. Experimental protocol

Subjects exercised on a treadmill (DOMYOS TC-450 Motorised Treadmill, Decathlon, Villeneuve d'Ascq, France) and on an ergometer (MOTOmed viva2, RECK, Betzenweiler, Germany). For each subject, the experiment was divided into two sessions, performed in the same day. Before starting the first session, each subject determined his/her maximum walking speed (MWS) on the treadmill.

The first session aimed at extracting muscle synergies while walking at MWS. Surface EMG activity was continuously recorded during this session. This session began with a warm-up (walk at a self-selected speed on the treadmill) period of five minutes. After this acclimation period, subjects performed one walking trial at his/her MWS of 30-s duration on a treadmill. After performing the trial, it was calculated the cadence at which each subject walked, in order to set the same cadence for the second session.

The second session was performed 15 minutes afterwards and aimed at extracting muscle synergies while pedaling. Surface EMG activity was also continuously recorded during this session. After a brief period (1-2 minutes) of pedaling warm-up in the ergometer, subjects performed a cycling trial, at the very same frequency obtained at walking, with 30-s duration. Gear was set to the same value for all the subjects, because according to De Marchis *et al* [9], muscle synergies are consistent when pedaling under different biomechanical demands. Subjects were asked to maintain a constant pedaling rate, set by a metronome, which was used to synchronize subject's cadence.

In order to eliminate possible effects of the initial acceleration or final deceleration on EMG records, only the ten central gait/cycling cycles of each trial were analyzed, for each condition (walking or cycling) and subject.

C. Data collection

Surface EMG activity was amplified at 2K and continuously sampled at 2048 Hz during both sessions, by using an EMG acquisition system (EMG-USB, OT Bioelettronica, Torino, Italy). For that purpose, bipolar electrodes (Ag-AgCl, Ambu® Neuroline 720, Ambu, Ballerup, Denmark) were attached to the skin with a 2-cm interelectrode distance on the following 10 muscles (see Figure 1) in the most dominant leg: Gluteus Maximus (GMax), Gluteus Medius (GMed), Tensor Fasciae Latae (TFL), Rectus Femoris (RF), Vastus Lateralis (VL), Biceps Femoris (BF), Semitendinosus (ST), Gastrocnemius Medialis (GM), Soleus (SOL) and Tibialis Anterior (TA).

Electrodes were placed longitudinally with respect to the underlying muscle fiber arrangement and were located according to the SENIAM [11] recommendations. Before attaching the electrodes, the skin was shaved and cleaned with alcohol to minimize impedance. A minimum period was taken to allow the alcohol to vaporize in order to dry the skin before placing the electrodes. The wires connected to the electrodes were well secured with adhesive tape to avoid movement-induced artifacts.

A setup constituted by an IMU (Technaid S.L. MCS system) placed on the crank of the pedal was used to detect the bottom dead center (BDC; lowest position of the corresponding pedal of the dominant leg). In the treadmill condition, a footswitch was placed beneath the heel of the dominant leg and the status of the contact of the heel with the ground was extracted applying a threshold to its analog signal. IMU's and footswitch's data were used for identification and segmentation in cycling and stride cycles, respectively. Therefore, each stride cycle started at each heel strike moment and ended at the next heel strike moment of the same foot; a pedaling cycle was defined as a complete revolution of the corresponding pedal of subject's dominant leg, starting from the lowest pedal position (BDC).

IMU, EMG and footswitch data were synchronized by applying a trigger signal. Data were analyzed offline through Matlab R2011a (The Mathworks, Natick, MA) and IBM SPSS Statistics 20 software (IBM).

D. Muscle synergies extraction

Raw EMG data (for both conditions of walking and cycling) were band-passed filtered (3^{rd} order Butterworth digital, pass-band 20-400 Hz) to attenuate DC offset, motion artifacts and high frequency noise. Filtered EMG was then demeaned. After that, EMG signals were smoothed using a 50-point root mean squared (RMS) algorithm [12]. For each condition (walking and cycling) and for each muscle, EMG data were normalized to the average of its peaks across cycles and resampled at each 1% of the stride/cycling cycle [8]. For each subject and condition (walking or cycling), normalized EMGs were combined into an m x t matrix (EMG_0), where m indicates the number of muscles (ten in this case) and t is the time base (t = no. of strides (10) x 100) [7]. By including consecutive walking/cycling cycles as previously done by Clark et al. [7] and Hug et al. [8], cycle-to-cycle variability is taken into account.

The non-negative matrix factorization (NNMF) algorithm used by Lee and Seung [13] was applied to each EMG_0 matrix for extraction of muscle weightings vectors (W) and activation signals (H) from each subject and for each condition. The number of synergies n was specified a priori (dimensionality two, three, four, five and six). The NNMF algorithm found the properties of the synergies by populating two matrices: an m x n synergy matrix W, which specifies the time-invariant weights of all the muscles within each synergy, and an n x t activation matrix (H), which specifies the activation timing of each muscle synergy [7]. These two matrices were multiplied to produce an m x t matrix (EMG_r). EMG_r was compared to EMG_0 by calculating the sum of the squared errors $(EMG_0 - EMG_r)^2$ and the result was used for iterative optimization until it converged on the muscle weighting vectors and the activation signals that minimized the error. The algorithm was repeated 10 times for each subject, in order to avoid local minima. The lowest cost solution was kept (i.e., minimized squared error between EMG_0 and EMG_r). Finally, muscle weightings (W) vectors were normalized by their maximum under the synergy to which they belong [8] and the corresponding activations signals (H) were scaled by the same quantity [9].

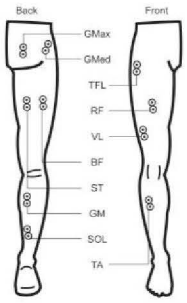

Figure 1. Location of the electrodes on each studied muscle

E. Muscle synergy comparison

The variation accounted for (VAF) was calculated to determine the minimum number of synergies needed to adequately reconstruct EMG_0 of each subject, for all conditions. VAF was calculated as the ratio of the sum of the squared error values to the sum of the squared EMG_0 values [$VAF = 1 - (EMG_0\text{-}EMG_r)^2/EMG_0$] [7]. VAF was calculated for each muscle and for each condition within the gait/pedaling cycle. A minimal VAF value of 90% was required to consider the reconstruction quality very good.

Furthermore, we performed Pearson's correlations to assure the criterion of similarities for muscle weightings vectors extracted from walking and cycling conditions of the group average. The same correlation was previously used for each subject to order synergies.

III. RESULTS

As represented in Table I, all the subjects were capable of maintaining a very similar cadence between walking and pedaling trials. Also their maximum walking speeds were very similar. Therefore, it is reasonable to test similarities between walking and pedaling motor control, with similar cadences.

TABLE I. WALKING CADENCE AND SPEED AND CYCLING CADENCE AMONG SUBJECTS.

	Walking (stride cycles) / (Km/h)	Cycling
Subject 1	73 / 7.0	77
Subject 2	73 / 7.2	74
Subject 3	66 / 6.5	65
Subject 4	65 / 7.3	65
Subject 5	71 / 8.0	74
Subject 6	68 / 7.0	69
Subject 7	64 / 6.5	68
Average (Mean ± SD)	68.57 +- 3.5 / 7.33 +-0.47	70.29 +- 4.4

Four synergies were identified as sufficient to reconstruct the EMG signals from all the analyzed muscles during walking and cycling, with a VAF higher than 90%. The extracted synergies are very similar to those reported by Clark [7] for walking and De Marchis [9] for pedaling.

These preliminary results show a statistically significant correlation across corresponding muscle weightings vectors (mean r = 79.8% ± 6%) from each task (walking and pedaling), as depicted in Figure 2. A minimum correlation value of 71% was obtained for muscle weightings vector for synergy 2 and a maximum value of 89% was obtained for muscle weightings vector for synergy 1.

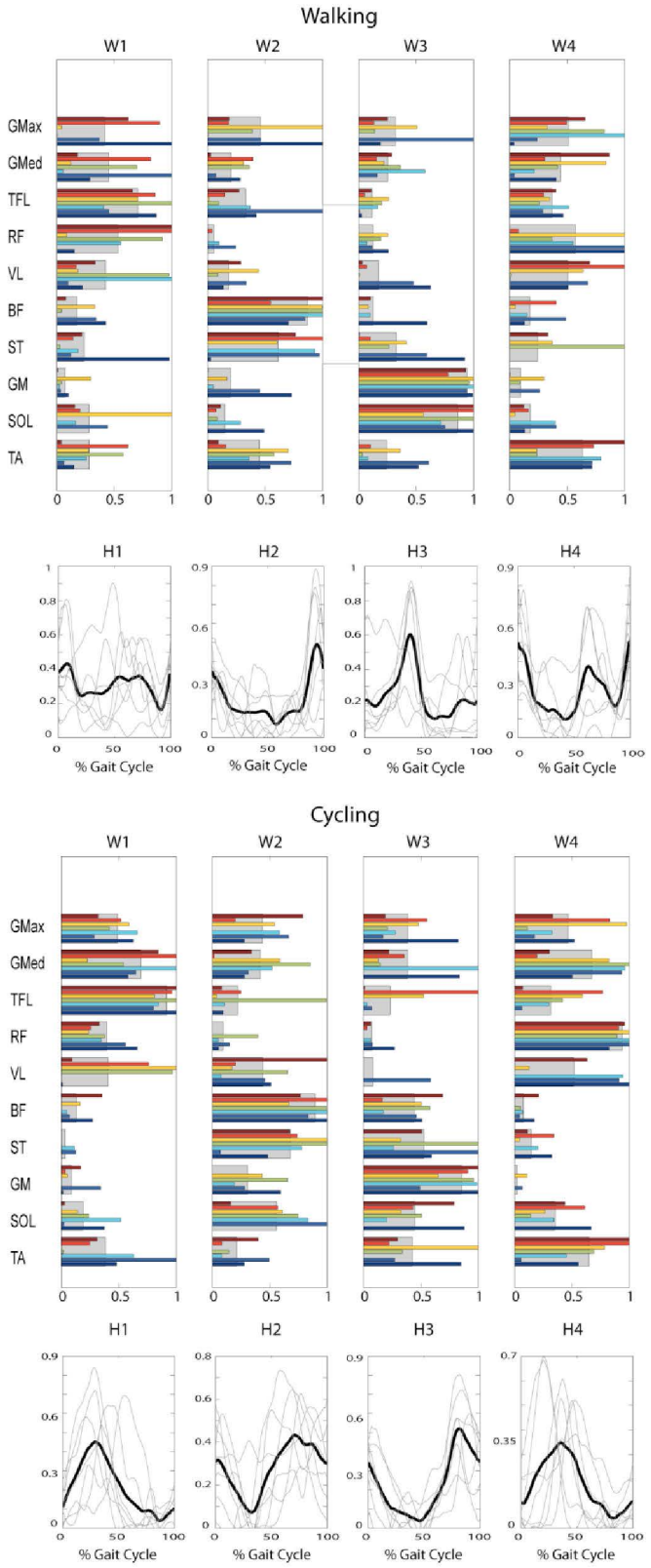

Figure 2. Upper plot: Muscle weightings vectors (W) and activation signals (H) extracted during the walking condition. Low plot: W and H extracted during cycling condition. For the muscle weighting vectors (W1-W4), each different colour represents each subject, and the goup average is represented in grey. In relation to the activation signals (H1-H4), grey lines represent the results of each studied subject, and black lines represent the group average.

It is notorious the burst-like structure in activation signals H (see Figure 2), as reported by Gizzi et al. [14] for walking in a gait robotic trainer at different walking speeds. Each synergy seems to be activated at different phases of walking and cycling, except synergies 1 and 4, which seem to be activated at similar moments.

IV. Discussion and Conclusions

The preliminary results presented here support the hypothesis that walking and cycling share a similar spatially fixed modular organization of muscle activity.

Synergy 1 actuates mainly in the activation of GMax (hip adductor), GMed (hip abductor), TFL (hip adductor and flexor, also assists knee extension), RF (hip flexor and knee extensor) and VL (knee extensor). During walking, this synergy is active mostly during early stance phase, providing body support during weight bearing [7] [15]. During cycling, we observed the existence of a similar synergy, activating the same muscles, allowing the production of force to facilitate the upstroke phase of cycling.

Synergy 2 is mostly responsible for the activation of the hamstrings (ST and BF, hip extensors and knee flexors), and therefore involved in the acceleration of the leg at late swing and propulsion of the body during early stance in walking [7] [15]. During cycling, this synergy activates the very same muscles, during the downstroke phase of pedaling.

Synergy 3 is mainly related with the activation of GM (knee flexor, ankle plantarflexor) and SOL (ankle plantarflexor), remaining active during the late stance phase of walking. This synergy has a main role in body support, forward propulsion and swing initiation [7] [15]. The synergistic activation of GM and SOL is also observed in pedaling. Here, this synergy allows the typical ankle plantarflexion during the final part of the downstroke phase of cycling.

Synergy 4 is mainly related with the activation of TA (ankle dorsiflexor), as well as the activation of GMax, GMed, TFL, RF and VL. In walking, this synergy is mainly active during the early stance phase (responsible for dorsiflexion) and throughout the swing phase (contributing to foot clearance) [7] [15]. In cycling, this synergy contributes, together with synergy 1, to the leg movements during the upstroke phase. Besides, the activation of TA allows the ankle dorsiflexion, typical of this phase of pedaling.

As the muscle weighting vectors (W) are very similar across the two analyzed motor activities, activation signals (H) may seem to represent adaptations of the motor system to the specific task (modulated by sensory feedback [4]), resulting in the observed variability of the EMG envelope during walking and cycling.

The presented results will need to be statistically confirmed by a more extensive analysis from a higher number of subjects and across different biomechanical conditions, namely by changing walking and cycling speed. If similarities in neuromuscular behavior between walking and pedaling are demonstrated with further analysis, pedaling can be envisioned as a potential tool for the quantitative assessment of common neural mechanisms of walking.

Following a rehabilitative perspective, future efforts will be devoted to create a database of synergies from non-injured subjects – for different mechanical constraints and rhythmic movements – to be used as reference target data for rehabilitation paradigms, based on FES or some other kind of afferent input (e.g. visual biofeedback).

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References

[1] G. Cappellini, Y. P. Ivanenko, R. E. Poppele, and F. Lacquaniti, "Motor patterns in human walking and running," J Neurophysiol, 95(6): pp. 3426–3437, June 2006.

[2] G. Torres-Oviedo and L. Ting, "Muscle Synergies Characterizing Human Postural Responses," Journal of Neurophysiology, 98: pp. 2144-56, 2007.

[3] Y. P. Ivanenko, R. E. Poppele, and F. Lacquaniti, "Five basic muscle activation patterns account for muscle activity during human locomotion," J Physiol, 556(Pt 1): pp. 267–282, April 2004.

[4] V. C. Cheung, L. Piron, M. Agostini, S. Silvoni, A. Turolla, and E. Bizzi, "Stability of muscle synergies for voluntary actions after cortical stroke in humans," Proc Natl Acad Sci U S A, 106(46); pp. 19563–19568, 2009.

[5] E. P. Zehr, J. E. Balter, D. P. Ferris, S. R. Hundza, P. M. Loadman, and R. H. Stoloff. "Neural regulation of rhythmic arm and leg movement is conserved across human locomotor tasks," J Physiol; 582; pp. 209–227, 2007.

[6] L. E. A. Pacheco, R. J. Oglivie, S. Chong, and V. K. Mushahwar, (2011, November). "Arm and leg cycling with functional electrical stimulation improves walking after incomplete spinal cord injury," Poster session presented at Neuroscience 2011, Washington, DC.

[7] D. J. Clark, L. H. Ting, F. E. Zajac, R. R. Neptune, and S. A. Kautz. "Merging of healthy motor modules predicts reduced locomotor performance and muscle coordination complexity post-stroke," J Neurophysiol 103 (2), pp. 844–857, 2010.

[8] F. Hug, N. A. Turpin, A. Guével, and S. Dorel, "Is interindividual variability of EMG patterns in trained cyclists related to different muscle synergies?," J Appl Physiol; 108; pp. 1727–1736, 2010.

[9] C. De Marchis, A. M. Castronovo, D. Bibbo, M. Schmid, and S. Conforto, "Muscle Synergies are Consistent when Pedaling Under Different Biomechanical Demands," in 34th Annual International Conference of the IEEE EMBS, pp. 3308–3311, 2012.

[10] F. Hug, N. A. Turpin, A. Couturier, and S. Dorel1, "Consistency of muscle synergies during pedaling across different mechanical constraints," J Neurophysiol; 106: pp. 91–103, 2011.

[11] H. J. Hermens, B. Freriks, and R. Merletti. "European recommendations for surface electromyography: results of the SENIAM project," Biomedical and health research program. Roessingh Research and Development, 1999.

[12] J. M. Hidler and A. E. Wall. "Alterations in muscle activation patterns during robotic assisted walking," Clin Biomech (Bristol, Avon), 20(2), pp. 184–193, February 2005.

[13] D. D. Lee, and H. S. Seung, "Learning the parts of objects by non-negative matrix factorization," Nature 401 (6755), pp. 788–791, 1999.

[14] L. Gizzi, J. F. Nielsen, F. Felici, J. C. Moreno, J. L. Pons, and D. Farina. "Motor modules in robot-aided walking. Journal of NeuroEngineering and Rehabilitation," 9:76, 2012.

[15] R. R. Neptune, D. J. Clark, and S. A. Kautz. "Modular control of human walking: a simulation study," Journal of Biomechanics, 42(9), pp. 1282–1287, 2009.