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Similarity of muscle synergies extracted from the lower limb including the deep muscles between level and uphill treadmill walking

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ABSTRACT

This study aimed to examine muscle synergies involving the deeper muscles of the lower limb during level and uphill treadmill walking. Seven men and five women walked on a treadmill at three speeds (60, 80, and 100 m/min) and two grades (level and 10% grade). Surface electromyographic (EMG) signals were recorded from 10 muscles of the lower limb, including vastus intermedius, adductor magnus, and adductor longus. Muscle synergies were extracted applying non-negative matrix factorization, and the relative co-activation across muscles and the temporal information of synergy recruitment were identified by the muscle synergy vector and synergy activation coefficient, respectively. Correlation coefficients between a pair of synergy vectors during level and uphill walking were analyzed as a similarity index, with the similarity criterion at $r = 0.76$. Changes in synergy activation coefficients between the walking conditions were evaluated by cross-correlation analysis. The mean number of synergies ranged from 3.8 to 4.0 across all conditions, and they were not significantly different between level and uphill walking conditions. Similarity between walking conditions was high ($r > 0.76$) for three muscle synergies, but not for one synergy that mainly consisted of the quadriceps femoris. The inter-condition similarity of the synergy activation coefficients was high for the four synergies, and a significant lag time for *synergy 2*, which consisted mainly of the activity of medial gastrocnemius, was found at 60 and 80 m/min. The muscle synergies extracted from the lower limb involving the deeper muscles appear to be consistent during level and uphill treadmill walking.

1. Introduction

One of the key topics of interest in neurophysiology, motor control, and biomechanics studies is how a large number of skeletal muscles is activated by the central nervous system to produce human movement. It is accepted that the central nervous system simplifies a large number of degrees of freedom using a few low-dimensional modules, due to the redundancy of the human musculoskeletal system [1,2]. The concept of muscle synergies involves muscle groups that are activated in a fixed ratio by a single input signal. The existence of the muscle synergies is still under discussion, and muscle synergy may reflect task constraints rather than the neural control strategy during an isometric finger force control task [3,4]. However, many researchers have discussed human gait in terms of muscle synergies, and it has been suggested that muscle synergies concisely predict the neural control strategies for activating muscles during gait [2,5–7]. In fact, a small number of muscle synergies could account for the majority of activation patterns of 16 muscles in the trunk and lower limb during walking over a wide speed range [5]. Hence, the muscle synergy hypothesis provides a better understanding of basic motor patterns during gait that are centrally organized [5,6].

Electromyographic (EMG) profiles recorded from active muscles have been used to identify muscle synergies applying a decomposition algorithm [1,8]. Although it has been demonstrated that any muscle synergies account for the surface EMG patterns from numerous activities of the leg and trunk muscles during walking [7–9], muscle synergies estimated from the decomposition algorithm are sensitive to the number and function of muscles [10]. As a technical limitation of synergy identification, the EMG recordings have been limited to the superficial muscles. Regarding the deeper muscles of the thigh, we have developed surface EMG recordings for vastus intermedius (VI), adductor magnus (AM), and adductor longus (AL) [11,12], since the previous reports have shown that the deep thigh muscles represent a

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specific activation pattern [12–14], and the neuromuscular activation pattern including the deep muscles of the lower limb may demonstrate that specific muscle synergies exist during gait.

The muscle synergies have been addressed across a variety of speeds, with multi-directional and inclined locomotion [15,16]. Gonzalez-Vargas et al. [15] showed that the synergies extracted from 15 neuromuscular activation patterns were consistent across 25 walking conditions including 5 speeds and 5 ground elevations. However, there is little information about muscle synergies identified from the lower limb involving deeper muscles during walking. From the functional viewpoint, the deep muscles in the thigh play a key role during locomotion; VI and AM contribute to limb stability during the stance phase, and AL acts as a hip flexor during the swing phase of walking [17]. Since dramatic changes of joint torque and neuromuscular activation in the knee and hip were observed between level and uphill walking conditions [18], the present results may show that task-specific neuromuscular activation of the deep muscles of the thigh in response to changes in the environment is fundamental to the control of gait.

In the present study, neuromuscular activation of the lower limb including the deeper muscles of the thigh was recorded during treadmill walking, and muscle synergies were extracted using a decomposition algorithm. More specifically, treadmill walking was carried out on level and inclined ground over a moderate speed range to evaluate the task-dependency of neuromuscular activation and muscle synergies during walking. The aim was to test the hypothesis that there is a specific muscle synergy consisting principally of activation of VI and/or hip adductors during walking and its task-specific modulation in response to ground elevation.

2. Methods

2.1. Subjects

Seven men and five women were recruited for the present study. The respective physical characteristics of the men and women were as follows: age, 24.5 ± 4.3 and 25.2 ± 6.8 years; height, 174.0 ± 6.0 and 155.9 ± 5.5 cm; and body mass, 67.5 ± 12.6 and 48.5 ± 4.6 kg. None of the subjects reported any previous history of lower limb surgery or pathology. The procedure, purpose, risks, and benefits associated with the present study were explained to the subjects, and their written, informed consent was obtained. The Institutional Review Board of the Research Center of Health, Physical Fitness & Sports at Nagoya University approved the experimental protocols, which were conducted in accordance with the guidelines in the Declaration of Helsinki.

2.2. Experimental protocol

Subjects walked on a motorized treadmill (TREAD-MILL, Nishikawa Iron Works, Kyoto, Japan) set to 60 m/min and level for 3 min prior to testing as a familiarization protocol. The subjects completed a walking protocol that consisted of six different combinations of speeds and grades, i.e. 60, 80, and 100 m/min at level and a grade of 10%. The walking trial was performed twice for each condition, with each trial lasting 1 min. Speed-grade combinations were conducted in randomized order for each subject. Rest intervals between the trials were at least 1 min.

2.3. Surface electromyography recording

Surface EMG signals were recorded during walking from 10 muscles of the right lower limb: VI, vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), long head of biceps femoris (BF), semimembranosus (SM), AM, AL, medial head of gastrocnemius (MG), and tibialis anterior (TA) muscles. EMG sensors consisting of two silver bar electrodes (0.1 × 1 cm each), with 1-cm inter-electrode distance, were used for EMG acquisition from each muscle. The DE-2.1 sensor preamplifier and main amplifier with a bandpass filter at 20–450 Hz (Bagnoli, Delsys, Boston, MA, USA), were set at gains of 10- and 100-fold, respectively, providing 1000-fold amplification. Signals were sampled at 2000 Hz using an AD converter (Power Lab, ADInstruments, Melbourne, Australia). Timings of heel contact were identified by a foot-switch (DL-250, S & ME, Tokyo, Japan) taped to the heel of the right shoe. The gait cycle was determined using the footswitch attached to the heel for the EMG analysis. One right heel contact to next right heel contact was defined as one gait cycle. Electrical signals from the footswitch were synchronized with surface EMG signals on a personal computer using software (Chart 7, ADInstruments).

After shaving, abrading, and cleaning the skin with alcohol, electrodes were positioned at specific locations for each muscle [19]. The superficial regions of VI, AL, and AM were determined using an ultrasound apparatus (Logiq e, GE Healthcare, Duluth, MN, USA) [11,12]. The VI electrode was positioned on the skin where the VI superficial region overlapped at 0° and 65° knee flexed positions [13,14,20]. Electrodes for AM and AL were placed on the skin between the SM and the gracilis and sartorius muscles, respectively [12]. A reference electrode was placed over the iliac crest.

2.4. Post-processing and extraction of muscle synergies

Data analyses were performed with custom software (Matlab, Mathworks, Natick, MA, USA). During the constant phase of each trial, 10 consecutive gait cycles were sampled for analysis, which are sufficient to produce a representative EMG pattern of the gait [21]. The EMG signals were rectified and smoothed with a low-pass filter at 9 Hz using a fourth-order Butterworth filter [21]. The EMG envelopes of each cycle were interpolated to 100 time points and normalized to the maximum value of EMG amplitudes over all conditions within the same subject for each muscle. Thus, the EMG scales ranged from 0 to 1.

Muscle synergies were extracted from each EMG data matrix using non-negative matrix factorization (NMF) [22]. E is a $p \times n$ matrix (where p is the number of muscles, and n is the number of time points). The NMF minimizes the residual between the initial matrix and its decomposition, given as follows:

$$E = WH + e,$$

where W is a $p \times k$ matrix of the synergy vectors, containing the spatial information of muscle coactivation, H is a $k \times n$ matrix of the synergy activation coefficients, involving the temporal information of synergy recruitment, k is the number of extracted muscle synergies, and e is the residual error matrix. To avoid local minima, the algorithm was repeated 20 times for each subject [23]. The muscle synergy vectors were normalized by their maximums under the synergy to which they belong.

Iterative analysis was performed by varying the number of synergies between one and 10, and then the least number of synergies k that accounted for > 90% of the variance accounted for (VAF) in each subject [24] was selected:

$$VAF = 1 - \frac{\sum_{i=1}^{p} \sum_{j=1}^{n} (e_{i,j})^2}{\sum_{i=1}^{p} \sum_{j=1}^{n} (E_{i,j})^2}$$

VAF was defined as the uncentered Pearson correlation coefficient.

2.5. Statistical analysis

Two-way (speed × condition) repeated measures analysis of variance (ANOVA) was used to compare the gait cadence and the number of muscle synergies. Differences in the VAFs across level and uphill walking conditions were compared using two-way (synergy number × condition) repeated measures ANOVA at each muscle synergy.