



Muscle synergies underlying sit-to-stand tasks in elderly people and their relationship with kinetic characteristics



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ABSTRACT

Background: Physiological evidence suggests that the nervous system controls motion by using a low-dimensional synergy organization for muscle activation. Because the muscle activation produces joint torques, kinetic changes accompanying aging can be related to changes in muscle synergies.

Objectives: We explored the effects of aging on muscle synergies underlying sit-to-stand tasks, and examined their relationships with kinetic characteristics.

Methods: Four younger and three older adults performed the sit-to-stand task at two speeds. Subsequently, we extracted the muscle synergies used to perform these tasks. Hierarchical cluster analysis was used to classify these synergies. We also calculated kinetic variables to compare the groups.

Results: Three independent muscle synergies generally appeared in each subject. The spatial structure of these synergies was similar across age groups. The change in motion speed affected only the temporal structure of these synergies. However, subject-specific muscle synergies and kinetic variables existed.

Conclusions: Our results suggest common muscle synergies underlying the sit-to-stand task in both young and elderly adults. People may actively change only the temporal structure of each muscle synergy. The precise subject-specific structuring of each muscle synergy may incorporate knowledge of the musculoskeletal kinetics.

1. Introduction

Standing from a seated position is crucial for human activities because standing upright on one's feet is a vital prerequisite for bipedal walking. The sit-to-stand ability is acquired before walking in the developmental process of humans (Avery et al., 2003). Because our entire body participates in the sit-to-stand task, kinetic coordination is required (Schenkman et al., 1990). Additionally, because we require a large amount of energy to perform the sit-to-stand task (Hortobágyi et al., 2003), multi-muscle coordination is vital.

Regarding this multi-muscle coordination, “muscle synergy” exists as one of the neural control hypotheses (Bernstein, 1996). Muscle synergy is a functional unit; a higher neural center is assumed to unite muscles into groups and then use one parameter per group to modify activation levels of all muscles within the group in parallel. Each muscle synergy can be regarded as a low-level feedforward controller producing joint torques that can be related to global biomechanical and/or kinetic variables (d'Avella et al., 2003; Torres-Oviedo and Ting, 2010).

Technological advances have allowed us to identify the linear combination of a small number of muscle synergies underlying natural behaviors. However, no studies have verified the muscle synergy underlying the sit-to-stand task at different speeds in elderly subjects. In the elderly, both nerve and muscle tissue degenerate. Due to this degeneration, it becomes difficult for the elderly to perform the sit-to-stand task quickly. Therefore, some kinetic variables of sit-to-stand tasks differ depending on age or speed (Hanke et al., 1995; Pai et al., 1994; Vander Linden et al., 1994). Such kinetic changes accompanying aging or speed can be related to changes in muscle synergies.

The purpose of this study was to provide basic knowledge on muscle coordination underlying the sit-to-stand task by examining the relationships between biomechanics and muscle synergies.

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Table 1
Characteristics of the subjects analyzed in the study.

Group	Age (years) [†]	Height (m) ^{**}	Mass (kg) ^{**}
	Mean ± SD	Mean ± SD	Mean ± SD
Young (n = 4)	22.5 ± 1.2	1.70 ± 0.01	69.75 ± 3.09
Elderly (n = 3)	72.0 ± 2.0	1.65 ± 0.04	54.77 ± 7.19

Comparison between groups: [†]p < 0.001, ^{**}p > 0.05.

2. Methods

2.1. Subjects

Seven male subjects participated in this experiment. Four younger subjects were recruited from physical therapy students at Saitama Prefectural University. Three older subjects were recruited from the local community. In this study, we decided the number of subjects based on previous studies that conducted similar experiments to measure body kinematics (Pai and Rogers, 1990), and muscle synergies during sit-to-stand tasks (An et al., 2013a, 2013b). Individuals with a history of myocardial infarction, stroke, fracture, or symptomatic arthritis of the lower extremity were excluded from this study. We explained the experiments in detail and obtained written consent from all subjects. Table 1 presents characteristics of the subject groups.

2.2. Procedure

The starting position was standardized, with the subjects seated in a chair without a back or armrests. Subjects started with their trunk in a vertical position and hands on their chests, keeping their hands in this position throughout the movement. The seat height corresponded to each subject's knee height, determined as the distance from the knee joint to the ground. Each subject performed the sit-to-stand task under the following two conditions: (1) rising at one's natural speed; (2) rising "as fast as possible." Each subject was given a few practice trials to familiarize themselves with the commands and the study's protocol. Five trials were then recorded for each of the conditions, with the order of the conditions being randomized.

2.3. Data collection

Surface electromyography (EMG) data from seven targeted muscles in the left leg were collected at 1000 Hz by a commercial EMG system (Noraxon USA, Scottsdale, AZ, USA). The seven muscles were the tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), vastus lateralis (VL), rectus femoris (RF), semitendinosus (ST), and gluteus maximus (GM). Each skin site was cleaned with alcohol prior to electrode placement. A camera motion capture system (Vicon Motion Systems, Oxford, UK) was used to determine the spatial location of body segments, and to calculate the body's center of mass (CoM) during the task. Thirty-five passive retroreflective markers were placed over bony landmarks according to the plug-in-gait model implemented in the camera motion capture system (bilaterally on the 2nd MTP head, heel, ankle, knee, thigh, anterior superior iliac crest, posterior superior iliac crest, shoulder, upper arm, elbow, radial and ulna wrist, 2nd finger, forehead, and posterior head; single markers were placed on jugular notch, inferior sternum, C7, T10, and right scapula). Data were sampled at a rate of 100 Hz. Two force platforms (Kistler Instrumente AG, Winterthur, Switzerland) were also employed. One platform was located beneath the stool and the other beneath the subject's feet. These were used to measure the time at which each subject lost contact with the stool. Data were sampled at a rate of 1000 Hz.

All data were synchronized using Vicon Workstation v4.5 software and saved to disk for offline analysis. Data were analyzed using R 3.0.2 (The R Foundation for Statistical Computing, Vienna, Austria) software.

2.4. Data processing

Muscle activation signals were band-pass filtered (20–500 Hz) with a zero-lag fourth-order Butterworth filter, demeaned, rectified, and then smoothed with a zero-lag fourth-order low-pass (10 Hz) Butterworth filter. Spatial location of body segments and the ground reaction forces were low-pass filtered at 20 Hz.

To compare the different trials of the subjects, we normalized the movement based on each movement time as 100%. The sit-to-stand task phases were divided as follows (An et al., 2013b; Schenkman et al., 1990): Phase 1 (the flexion momentum phase) began with the first shoulder movement in the horizontal direction; Phase 2 (the momentum transfer phase) began at contact loss with the stool; Phase 3 (the extension phase) began when the shank segment tilted forward to the maximum; and Phase 4 (the posture stabilization phase) began when the vertical shoulder position achieved its maximum height. The duration of Phase 4 was determined by extending the time series an additional 20% of the duration of Phases 1–3.

To apply the non-negative matrix factorization method, subject-specific EMG data matrices were generated for each speed condition. EMG data from all trials were concatenated rather than averaged to create data matrices that were 7 (number of muscles) × 5 (number of trials) × 100 (number of data points in one trial) in size for each participant. To allow for comparison between subjects, we normalized the EMG data for each subject to maximum muscle EMG activity for a given muscle across all trials, such that the data ranged from 0 to 1. Before extraction, each muscle was normalized to unit variance such that each muscle's variability was equally weighted in the extraction. This normalization was removed after extraction (Sawers et al., 2015).

2.5. Non-negative matrix factorization

A non-negative matrix factorization was applied to each data matrix to extract muscle synergies (An et al., 2013b; Dominici et al., 2011; Lee and Seung, 1999). This decomposed the EMG data matrices (M) into two components, spatial structure (W) and temporal structure (C). Spatial structure is defined as muscle synergy, and it denotes the relative activity ratio of multiple muscles. Temporal structure is defined as the weighting signal, which denotes the activation profiles of each muscle synergy. This is expressed as the following equation:

$$M \cong WC$$

Fig. 1 shows the actual M, W, and C in a representative subject.

To determine the number of muscle synergies needed to account for the recorded EMG data at each speed condition, we first extracted synergies 1–7. The preciseness of the fit of the data reconstruction for each muscle synergy was then quantified by the variance accounted for (VAF). The VAF describes how much of the variability in the original EMG data is accounted for by the EMG reconstructed from the muscle synergies and their weighting signals (Zar, 1999). To help ensure consistency in selecting the number of muscle synergies embedded within the EMG data sets, we calculated the 95% confidence interval (CI) for the VAF of the reconstructed EMG at each synergy number (1–7). This was accomplished by implementing a bootstrapping procedure where the EMG data sets were resampled 500 times with replacement, and the VAF of the reconstructed EMG was recalculated after each resampling. Ninety-five percent CIs were then constructed from the bootstrapped VAF values at each synergy number, and the number of synergies was selected as the minimum number of synergies at which the lower bound of the 95% CI exceeded 90% VAF (Cheung et al., 2009; Sawers et al., 2015).

2.6. Cluster analysis

In this study, hierarchical cluster analysis (Ward's method) was used to classify extracted muscle synergies based on the spatial structure

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