Original Article

Association between Muscle Synergy and Stability during Prolonged Walking

Keisuke Suzuki, MS, RPT[1, 2)*], Yusuke Nishida, PhD, RPT[1)], Kazuhiko Mitsutomi, RPT[2)]

[1)] *Department of Rehabilitation Sciences, Seirei Christopher University: 3453 Mikatabara-cho, Kita-ku, Hamamatsu, Shizuoka 433-8558, Japan*
[2)] *Department of Rehabilitation, Iwata City Hospital, Japan*

Abstract. [Purpose] The purpose of this study was to examine whether changes in muscle synergy could affect gait stability or muscle activity by comparing muscle activity before and after prolonged walking. [Subjects and Methods] Twelve healthy male subjects walked on a treadmill for 10 min as a warm-up. Data were recorded from the participants during the first and last 1 min during 90 min of walking at 4.5 km/h. Electromyographic (EMG) activity was recorded for 7 leg muscles, and patterns of coordination were determined by principal component analysis (PCA). The patterns of activity within the anatomic muscle groups were additionally determined by repeating PCA. iEMG was calculated using the mean EMG for each cycle step during the 1 min walking periods. The largest Lyapunov exponent was calculated to quantify each subject's inherent local dynamic stability. [Results] The patterns for each of the 7 muscles showed no change between the start and end periods. However, the end period showed a higher co-activation of the triceps surae, lower iEMG of the medial gastrocnemius, and a smaller largest Lyapunov exponent of the mediolateral and anteroposterior directions than those observed during the start period. [Conclusion] The increase in triceps surae co-activation may be associated with gait stability.
Key words: Muscle synergy, Muscle activity, Gait stability

INTRODUCTION

Compromised gait function can increase the morbidity and mortality of affected individuals[1)]. Several reports have indicated that the energy cost needed for gait is associated with gait stability[2, 3)]. Increased gait stability reduces the energy needed by the muscles involved in postural control during gait[2)]. Therefore, a stable gait is important for improving gait endurance.

Gait is a complex exercise pattern created by neuromuscular, biomechanical, and kinesiological activities[4)]. Furthermore gait is achieved by the concerted action of muscles located throughout the body[5)]. These muscles not only produce or dissipate the work required for each movement but also redistribute the work among different body segments[6)]. Thus, many muscles are necessary for a coordinated motion, even if their primary functions do not appear to support gait stability. Muscle synergy represents the enhanced activity of muscles working at the same time and can simplify coordination in complex movements by activating muscles appropriate for the intended movement. For example, 5 muscle synergies account for the majority

of variance in the surface EMG signals recorded from 32 muscles during walking[7, 8)] and running[7)]. Muscle synergy patterns, however, can change with prolonged exercise[9)]. Similarly, gait stability can be affected by prolonged exercise[10)]. Therefore, changes in gait stability that occur due to prolonged exercise may result from changes in muscle synergy patterns.

The function of individual muscles is largely determined by their architecture, moment arms, and fiber-type composition[11)]. As the pennation angle of a muscle increases, the cross-sectional area (area perpendicular to the fiber direction) increases, and it is typically accompanied by a decrease in the ratio of the muscle fiber length to the whole muscle length. These changes predispose the muscle to generate greater forces with reduced fascicle strain[11)]. When the same work is performed, synchronous contraction of anatomic muscles leads to a decrease in muscle activity[12–14)]. Joint motion is regulated by external muscle activity during gait[15)]. External muscle activity results in a strong contraction and dynamic movement of a joint[16)]. Therefore, we speculated that increased muscle synergy of anatomic muscle groups leads to increased gait stability. Nevertheless, little is known about the relationship between gait stability and muscle synergies. The purpose of this study was to determine whether changes in muscle synergies affect gait stability and muscle activity.

SUBJECTS AND METHODS

Twelve healthy male subjects volunteered for this study

*Corresponding author. Keisuke Suzuki (E-mail: keisukedondon@gmail.com)

(age range: 21–27 years, mean age: 23.0 ± 2.0 years, body mass: 64.3 ± 6.8 kg, height: 172.1 ± 6.0 cm). This study conformed to the Declaration of Helsinki, and informed consent was received from all the participants following the protocol of the Ethics Committee of the Seirei Christopher University.

Subjects walked on a level treadmill (AUTORUNNER AR-200, MINATO, Japan) while wearing a safety harness that allowed a natural arm swing. Subjects initially walked at 4.5 km/h until they were accustomed to the treadmill. Next, they walked at the same speed for 1 min as a control. Finally, the subjects continued to walk on the treadmill at 4.5 km/h for 90 min, and data of the first and last 1 min were recorded.

Electromyographic (EMG) activity was recorded of 7 muscles in the subject's right leg: the tibialis anterior (TA), vastus medialis (VM), rectus femoris (RF), soleus (SOL), medial gastrocnemius (MG), semitendinosus (ST), and gluteus medius (GM). EMG activity was recorded using active electrodes (Blue Sensor M,M-00-s, Mets, Japan), which were applied to lightly abraded skin over the respective muscle belly. Electrode placement was carefully chosen to minimize cross-talk from adjacent muscles. The signals were amplified, filtered (20–450 Hz, Tele Myo G2, EM-601, NORAXON Inc., USA), sampled at 1,500 Hz, and recorded to disc.

A triaxial accelerometer (WAA-001, Wireless Technologies Inc., Japan) was clipped tightly on a band tightened around the subjects' waist. It was located approximately at the center of the spine at the L3 level to measure acceleration of the back and waist in mediolateral (Ax), vertical (Ay), and anteroposterior (Az) directions[10]. The signals were sampled at 200 Hz.

The gait cycle was defined by consecutive foot contacts, as determined by signals from a foot contact switch. The EMG data were rectified and subsequently interpolated to 100 evenly spaced values for each instance of the foot switch being triggered. Next, EMG datum traces were normalized by their largest value for each step and by the mean value at each of the 100 time points within each step.

EMG data recorded during the walking trials were examined using a principal component analysis (PCA) to identify coordinated patterns among the 7 test muscles[17]. PCA is a technique in which a large set of multivariate data is reduced to a smaller set of common variables. In this case, each stride was represented by 700 values (from the 7 muscles and 100 time points per stride). PCA reduced these 700 values to 3 principal components (PCs) that were further analyzed. Each of these components contains information from all recorded muscles and time points; thus, the major components can be used to evaluate the features of the co-ordinated muscle patterns. This was achieved by arranging the data into a p × N matrix A, where p = 700 values per pattern (7 muscle × 100 time points for the EMG data), and N = 360 steps analyzed in both the start and end periods. The covariance matrix B was calculated from the data in A, and the PC weightings determined from the eigenvectors ξ covariance matrix B. The importance of each PC was given by the eigenvalues for each eigenvalue–eigenvector pair

with the greatest absolute eigenvalues corresponding to the primary PCs. The proportion of EMG patterns explained by each PC was given by ξ' B ξ, and the loading scores for each PC for the N time points were given by ξ' A. Each instantaneous pattern of activity can be reconstructed from the vector product of the PC weightings and the PC loading scores. The patterns of activity within the anatomic muscle groups (SOL-MG, VM-RF) were additionally determined by repeating PC analysis using p = 200, corresponding to whether each anatomic group had 2 muscles, respectively[18]. The co-activation of muscles within anatomic groups was indicated by the contribution ratio.

iEMG was calculated from the mean EMG data for cycle steps during the 1 min of walking. EMG signals were integrated over the 1 min of walking at the same speed during the start and end periods. Each iEMG signal was expressed as a percentage of the iEMG reading obtained during the 1 min control period.

Humans naturally experience many small perturbations during normal walking. These include both external (e.g., environmental and/or sensory) and internal perturbations created by neuromuscular noise. Local dynamic stability, as quantified by the largest Lyapunov exponent, refers to the sensitivity of a dynamic system to infinitesimally small perturbations[19]. The Lyapunov exponent is the mean exponential rate of divergence of nearby points, and it quantifies local dynamical stability[19, 20]. We estimated the largest finite-time Lyapunov exponent of the down-sampled (200 Hz) Ax, Ay, and Az using the algorithm proposed by Sano-Sawada[21]. The attractor dynamics were reconstructed using embedded-delayed samples[22]. The time lag was determined by the time at which the autocorrelation dropped to 1/e of the value at time zero. The embedding dimension was set to 5 using the result of false nearest neighbor analysis[23]. The largest Lyapunov exponent was estimated by calculating the slope of the linear fit to the log-scaled curve of the divergence distance between neighboring trajectories in the reconstructed phase space[24]. To exclude the effect of different gait cycle times among the data, the time axis for the divergence curve was rescaled by dividing by the mean gait cycle time. A decrease in the largest Lyapunov exponent corresponds to an increase in gait stability.

Contribution ratio, iEMG, and the largest Lyapunov exponent were compared between the start and end gait periods using the paired t-test. The significance of all statistical tests was accepted for values of p< 0.05.

RESULTS

The patterns of activations among muscles can be determined by PCA of their time-varying EMG activity. In the present study, more than 80% of the EMG activity patterns were explained by the first 3 PCs. PC1 partitioned the muscles into anatomic groups, with the triceps surae giving negative weightings, whereas TA, quadriceps, ST, and GM gave positive weightings in both the start and end periods (Table 1). PC2 partitioned the muscles into groups with TA and ST giving negative weightings, whereas the triceps surae, quadriceps, and GM gave positive weightings

Table 1. Principal component (PC) weightings calculated for the start and end periods to determine the synergies among the 7 muscles tested

	PC1		PC2		PC3	
	Start	End	Start	End	Start	End
Tibialis anterior	0.706	0.709	−0.044	−0.025	0.201	0.424
Vastus medialis	0.820	0.802	0.044	0.418	−0.065	−0.005
Rectus femoris	0.658	0.712	0.449	0.460	−0.482	−0.209
Soleus	−0.636	−0.560	0.669	0.682	0.148	0.331
Medial gastrocnemius	−0.696	−0.668	0.582	0.584	0.271	0.336
Semitendinosus	0.573	0.526	−0.195	−0.352	0.705	0.628
Gluteus medius	0.502	0.480	0.628	0.650	0.214	−0.142
Contribution ratio (%)	43.7	41.7	23.3	24.9	13.0	13.3

The percentage of the contribution ratio explained by each PC is shown.

Table 2. The contribution ratio, iEMG, and the largest Lyapunov exponent of the start and end periods

		Start	End
Contribution ratio (%)	Triceps surae	92.6 ± 1.9	94.1 ± 2.6*
	Quadriceps	86.6 ± 11.3	88.8 ± 7.2
iEMG (%)	Tibialis anterior	97.4 ± 3.7	98.5 ± 5.8
	Vastus medialis	105.1 ± 17.0	154.6 ± 128.3
	Rectus femoris	97.5 ± 11.9	90.3 ± 27.0
	Soleus	98.4 ± 5.3	104.8 ± 14.5
	Medial gastrocnemius	99.1 ± 3.1	92.0 ± 12.0*
	Semitendinosus	98.0 ± 11.1	96.1 ± 30.2
	Gluteus medius	106.3 ± 20.1	128.0 ± 71.3
Largest Lyapunov exponent	Mediolateral (Ax)	0.132 ± 0.016	0.124 ± 0.015*
	Vertical (Ay)	0.124 ± 0.008	0.128 ± 0.013
	Anteroposterior (Az)	0.114 ± 0.009	0.107 ± 0.006*

Values are expressed as the mean ± SD. *$p < 0.05$, significant difference between the start and end periods.

in both the start and end periods (Table 1). PC3 partitioned the muscles into anatomic groups with RF giving negative weightings, whereas TA, triceps surae, and ST, gave positive weightings in both the start and end periods (Table 1). Thus, the PC weighting showed coupling of activity among muscles within anatomic groups (Table 1). Each contribution ratio indicated there was no change in PCs between the start and end periods (Table 1).

The triceps surae co-activation during the end period was significantly higher than that observed in the start period ($p < 0.05$). The quadriceps co-activation, however, did not change between the start and end periods (Table 2). The MG iEMG in the end period was significantly reduced in the start period ($p < 0.05$), but no changes were observed for the other muscles (Table 2). The mediolateral and anteroposterior directions for the largest Lyapunov exponent in the end period were significantly reduced from the start period ($p < 0.05$) but showed no changes in the vertical direction (Table 2).

DISCUSSION

The aim of the present study was to examine whether a change in muscle synergies affects gait stability or muscle activity. We found that muscle synergies did not change for the 7 muscles tested; however, triceps surae co-activation was increased by prolonged gait. In addition, decreased MG activity and increased gait stability occurred with increased triceps surae co-activation during gait. This suggests that anatomic muscle group synergies correlate with an increase in gait stability.

A number of muscle activation patterns, which are termed synergies, occur during walking, and combine to enable the necessary muscle coordination required for locomotion[7]. The PCA determines the dominant components of the signal; therefore, any unbiased or random noise will be included in the lower order components. This study considered the 3 major PCs of the EMG activity patterns; the remaining components (accounting for 20% of the EMG signal) were excluded. PC1 comprised TA and quadriceps, which were active at initial contact. PC2 comprised triceps surae and GM, and these muscles were active during midstance and heel lift-off. PC3 comprised ST, which was active during the late swing phase. These findings indicate that the muscle synergies did not change for these 7 muscles. However, studies reporting the activity of these muscles individually have shown the relative levels of EMG activity varies among muscles in prolonged cyclic exercise[25, 26].

In this study, the individual muscles showed different responses to prolonged exercise and fatigue. This included a decrease in the iEMG of MG (Table 2) and advances in the timing of SOL and MG (Table 2). Prolonged cyclic exercise increases co-activation[27] and coincident firing[13] in anatomic muscle groups. Increased anatomic muscle group co-activation decreases with external muscle activity[14, 28]. These results raise the possibility that the decrease in MG activity which we observed was the result of increased triceps surae co-activation.

The largest Lyapunov exponent is an index of local gait stability[19, 20]. The largest Lyapunov exponent influences muscle synergies and muscle activity[3, 10, 19]. When the same work is performed, synchronous contraction of anatomic muscles leads to a decrease in muscle activity[14, 28]. Joint motion is regulated by external muscle activity during gait[15]. External muscle activity results in a strong contraction and dynamic movement of a joint[16, 29]. A decrease in external muscle activity leads to an increase in the stability of joint movement and gait. Thus, an increase in the co-activation of the triceps surae would lead to a decrease in the activity of the medial gastrocnemius, which in turn would lead to an increase in gait stability. We hypothesized that an increase in muscle synergies among anatomic muscle groups would lead to increased gait stability. Our data raise the possibility that an increase in muscle synergies among anatomic muscle groups leads to an increase in gait stability.

REFERENCES

1) Guralnik JM, Ferrucci L, Pieper CF, et al.: Lower extremity function and subsequent disability: consistency across studies, predictive models, and value of gait speed alone compared with the short physical performance battery. J Gerontol A Biol Sci Med Sci, 2000, 55: M221–M231. [Medline] [CrossRef]

2) Ackermann M, van den Bogert AJ: Optimality principles for model-based prediction of human gait. J Biomech, 2010, 43: 1055–1060. [Medline] [CrossRef]

3) Manor B, Li L: Characteristics of functional gait among people with and without peripheral neuropathy. Gait Posture, 2009, 30: 253–256. [Medline] [CrossRef]

4) Son S, Noh H: Gait changes caused by the habits and methods of carrying a handbag. J Phys Ther Sci, 2013, 25: 969–971. [Medline] [CrossRef]

5) Safavynia SA, Torres-Oviedo G, Ting LH: Muscle Synergies: Implications for Clinical Evaluation and Rehabilitation of Movement. Top Spinal Cord Inj Rehabil, 2011, 17: 16–24. [Medline] [CrossRef]

6) Cappellini G, Ivanenko YP, Poppele RE, et al.: Motor patterns in human walking and running. J Neurophysiol, 2006, 95: 3426–3437. [Medline] [CrossRef]

7) Ivanenko YP, Poppele RE, Lacquaniti F: Five basic muscle activation patterns account for muscle activity during human locomotion. J Physiol, 2004, 556: 267–282. [Medline] [CrossRef]

8) Ivanenko YP, Poppele RE, Lacquaniti F: Motor control programs and walking. Neuroscientist, 2006, 12: 339–348. [Medline] [CrossRef]

9) Gates DH, Dingwell JB: The effects of neuromuscular fatigue on task performance during repetitive goal-directed movements. Exp Brain Res, 2008, 187: 573–585. [Medline] [CrossRef]

10) Yoshino K, Motoshige T, Araki T, et al.: Effect of prolonged free-walking fatigue on gait and physiological rhythm. J Biomech, 2004, 37: 1271–1280. [Medline] [CrossRef]

11) Lieber R: Skeletal Muscle Structure, Function, and Plasticity. Philadelphia, 2002.

12) Hunter SK, Rochette L, Critchlow A, et al.: Time to task failure differs with load type when old adults perform a submaximal fatiguing contraction. Muscle Nerve, 2005, 31: 730–740. [Medline] [CrossRef]

13) Corcos DM, Jiang HY, Wilding J, et al.: Fatigue induced changes in phasic muscle activation patterns for fast elbow flexion movements. Exp Brain Res, 2002, 142: 1–12. [Medline] [CrossRef]

14) Padua DA, Arnold BL, Perrin DH, et al.: Fatigue, vertical leg stiffness, and stiffness control strategies in males and females. J Athl Train, 2006, 41: 294–304. [Medline]

15) Nozaki D: Torque interaction among adjacent joints due to the action of biarticular muscles. Med Sci Sports Exerc, 2009, 41: 205–209. [Medline] [CrossRef]

16) Wakeling JM, Uehli K, Rozitis AI: Muscle fibre recruitment can respond to the mechanics of the muscle contraction. J R Soc Interface, 2006, 3: 533–544. [Medline] [CrossRef]

17) Wakeling JM, Rozitis AI: Spectral properties of myoelectric signals from different motor units in the leg extensor muscles. J Exp Biol, 2004, 207: 2519–2528. [Medline] [CrossRef]

18) Wakeling JM, Horn T: Neuromechanics of muscle synergies during cycling. J Neurophysiol, 2009, 101: 843–854. [Medline] [CrossRef]

19) Kang HG, Dingwell JB: Dynamics and stability of muscle activations during walking in healthy young and older adults. J Biomech, 2009, 42: 2231–2237. [Medline] [CrossRef]

20) Lockhart TE, Liu J: Differentiating fall-prone and healthy adults using local dynamic stability. Ergonomics, 2008, 51: 1860–1872. [Medline] [CrossRef]

21) Sano M, Sawada Y: Measurement of the Lyapunov spectrum from a chaotic time series. Phys Rev Lett, 1985, 55: 1082–1085. [Medline] [CrossRef]

22) Takens F: Detecting Strange Attractors in Turbulence: Lecture Notes in Mathematics. Berlin: Rand DA, Young LS, 1981.

23) Kennel MB, Brown R, Abarbanel HD: Determining embedding dimension for phase-space reconstruction using a geometrical construction. Phys Rev A, 1992, 45: 3403–3411. [Medline] [CrossRef]

24) Kang HG, Dingwell JB: Effects of walking speed, strength and range of motion on gait stability in healthy older adults. J Biomech, 2008, 41: 2899–2905. [Medline] [CrossRef]

25) Turpin NA, Guével A, Durand S, et al.: Effect of power output on muscle coordination during rowing. Eur J Appl Physiol, 2011, 111: 3017–3029. [Medline] [CrossRef]

26) Hug F, Turpin NA, Guével A, et al.: Is interindividual variability of EMG patterns in trained cyclists related to different muscle synergies? J Appl Physiol, 2010, 108: 1727–1736. [Medline] [CrossRef]

27) Selen LP, Beek PJ, van Dieën JH: Fatigue-induced changes of impedance and performance in target tracking. Exp Brain Res, 2007, 181: 99–108. [Medline] [CrossRef]

28) Tamaki H, Kitada K, Akamine T, et al.: Alternate activity in the synergistic muscles during prolonged low-level contractions. J Appl Physiol, 1998, 84: 1943–1951. [Medline]

29) Shirasawa H, Kanehisa H, Kouzaki M, et al.: Differences among lower leg muscles in long-term activity during ambulatory condition without any moderate to high intensity exercise. J Electromyogr Kinesiol, 2009, 19: e50–e56. [Medline] [CrossRef]