Comparison of muscle synergies before and after 10 minutes of running

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Abstract. [Purpose] The purpose of this study was to investigate the modular control of locomotor tasks and compared the modules before and after a running intervention. [Subjects and Methods] Electromyographic measurements were performed on eight young, healthy males engaged in a 60s run on a treadmill at 2.8 m/s before and immediately after the 600s of running intervention. Electromyographic data for 15 trunk and lower-limb muscles on the right side were recorded. Muscle synergies were extracted from the electromyography signals using non-negative matrix factorization. [Results] Four modules explained the electromyographic activity of all muscles and had the functions of load acceptance (module 1), push-off (module 2), preparation of landing (module 3), and trunk-stabilization activity during the stance phase (module 4). Modules 1, 2, and 3 matched the basic modules reported in previous studies; whereas, module 4 was different before and after the intervention. [Conclusion] Before the intervention, module 4 engaged the trunk muscles and it was activated in the stance phase during running. However, after the intervention, module 4 engaged the muscles around the pelvis and it was activated after landing. This result suggests that the posture control changes from the trunk muscles to the lower-limb muscles after 10 minutes running.

Key words: Modular control, Non-negative Matrix Factorization, Muscle synergy

INTRODUCTION

Human movements are highly complex in terms of both neural activation and biomechanical output because the brain needs to control many musculoskeletal functions. Bernstein proposed the concept of “muscle synergy,” which posits that human motion is controlled through the cooperation of several muscles, rather than through the individual muscles1). The cooperation of several muscles control can reduce information processing in the Central Nervous System (CNS) because the CNS commands only a few modules responsible for muscle synergy2, 3). The synergy can be identified from electromyographic (EMG) patterns recorded from numerous muscles via an algorithm with two components2, 4–7), “muscle weighting,” which represents the relative weighting of each muscle within each module, and “activation coefficient,” which represents the relative activation of the muscle weights (Fig. 1). Based on the previous investigation of muscle synergies while running, it was concluded that there are five identified modules. These modules have the function, respectively; load acceptance, push-off to end the stance phase, trunk muscles during the transition from the stance to the swing phase, forward swing of the lower limbs during the swing phase, and landing preparation to stabilize the hip and trunk8). These five modules are believed to be the basic modules activated while running9).

The incidence of running related injury is 29–56% and the majority of injury occurs due to overuse10–13). One of the risk factors of running related injury is running distance per week14, 15). As running distance increases, muscle fatigue accumulates. In fact, it has been reported that muscle fatigue of the lower limbs occurs because of extended running; this
subsequently changes muscle activity\(^{16}\). The results of the aforementioned study indicated that thigh and calf muscle activity in the fatigued condition was lower than before running and the authors suggested that the alteration of muscle activity might facilitate injury\(^{16}\). As the modules are related to muscle activity, the modules activated while running may change with prolonging running time. However, there has been no study investigating module changes before and after an exercise intervention. We hypothesized that the modules, especially “muscle weighting”, during running changes before and after a running intervention. Thus, the purpose of this study was to investigate whether muscle synergy is changed because of prolonged running.

**SUBJECTS AND METHODS**

Eight young, healthy males (age: 22.4 ± 2.9 years, height: 174.1 ± 6.5 cm, weight: 65.3 ± 6.3 kg) participated in this study. They all performed physical activities two or three times a week at the recreational level. Exclusion criteria included a history of lower limb disorder, neurological disorder, or lower limb surgery. This study was approved by the Ethics Committee of our university (Ethics ID: 2013-033). All subjects provided informed consent to participate in the study.

Subjects performed stretching and running on a treadmill (R-16S, Alpen Co., Ltd. Japan) as a warm-up for 600s. The running pace was set by the subjects. After warm-up, the EMG measuring equipment was set up. Muscle activity during a 60s run at 2.8 m/s on the treadmill was measured before and immediately after a running intervention (defined as pre and post session, respectively). In the running intervention session, subjects ran for 600s and the intensity of the intervention was set at 70% of maximal oxygen consumption using the Karvonen formula. The target heart rate was calculated with the following formula: target heart rate = 0.7 × (220 – subject age – resting heart rate) + resting heart rate. The heart rate was measured using a cardiometer watch (FT2, Polar Co., Japan).

The activities of the trunk and lower extremity muscles were measured using a wireless EMG telemeter system (BioLog DL-5000, S&ME Co., Japan) at 1,000 Hz. Before the surface electrodes were attached, the skin was rubbed with a skin abrasive and alcohol to reduce skin impedance to a level below 2,000 Ω. Pairs of disposable Ag/AgCl surface electrodes (BlueSensor N-00-S, METS Co., Japan) at 1,000 Hz. Before the surface electrodes were attached, the skin was rubbed with a skin abrasive and alcohol to reduce skin impedance to a level below 2,000 Ω. Pairs of disposable Ag/AgCl surface electrodes (BlueSensor N-00-S, METS Co., Japan) were placed on the rectus abdominis (RA), external oblique (EO), internal oblique/ transversus abdominis (IO/TrA), the erector spinae (ES), rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), gluteus maximus (GMA), gluteus medius (GME), biceps femoris long head (BFL), biceps femoris short head (BFS), tensor fascia lata (TFL), adductor (ADD), tibialis anterior (TA), and gastrocnemius (GAS) muscles on the right side because the right foot was dominant in all the subjects. Before measurement in the pre session, subjects performed maximum voluntary contraction (MVC) tests using manual resistance to normalize the EMG data. Additionally, the electrodes were remained in place during pre and post sessions and intervention.

To analyze the EMG data, analysis software (BIMUTAS-Video, Kissei Comtec Co., Ltd., Japan) was used. The raw data were band-pass filtered between 20 to 500 Hz and full-wave rectified. The EMG data was normalized relative to the MVC data. Afterward, we extracted three running cycles with less noise, starting from one right heel contact to the next right heel contact, using coordinate data for the markers. This data was extracted with less disturbance of the waveform during the pre and post sessions. Each running cycle was interpolated to 200 time points. Then, as previously described, non-negative matrix factorization (NMF) was performed to extract muscle synergies based on the study by Lee and Seung\(^{17}\). Matrix factorization minimizes the residual Frobenius norm between the initial matrix and its decomposition, given as: $E=W \times C + e$, where, $E$ is a $p$-by-$n$ initial matrix ($p=$number of muscles and $n=$number of time points), which represents the normalized EMG matrix, $W$ is a $p$-by-$s$ matrix ($s=$number of synergies), which represents the “muscle weighting,” $C$ is an $s$-by-$n$ matrix, which represents the “activation coefficient,” and $e$ is a $p$-by-$n$ matrix, which is the residual error matrix.
The main findings of this study were that four modules were active while running and that module 4 was different between the pre and post sessions. Module 4 pre engaged the trunk muscles and it was similar to the basic module. However, the SP for module 4 between pre and post sessions was low; hence, module 4 activity was different between the pre and post sessions. Module 4 pre engaged the trunk muscles and it was activated in the stance phase. Researchers reported that trunk muscles are activated to enhance the rigidity of the trunk, stabilize the lumbar spine and pelvic girdle, and transmit ground reaction forces efficiently during push-off, which ends the stance phase. From the activation coefficient of module 4, the timing of the trunk muscle activity was similar to that reported by Saunders et al. Therefore, module 4 pre may be activated to stabilize posture through the trunk muscles during the stance phase. Conversely, module 4 post engaged the muscles around the pelvis and the activation coefficient indicated that it was activated after landing. Thus, the role of module 4 post may be related to postural control for landing impact using the muscles around the pelvis. From our findings, postural control function changed from the trunk muscles to the muscles around the pelvis because of 10 minutes running. In a study targeting long distance running as marathon, it has been indicated that running distance is a risk factor of running related injuries. Therefore, the alteration of module due to running may be relevant to running related injuries of lower limb.

The function of module 3 was to prepare for landing. These modules were roughly similar to the basic modules reported by Cappellini et al.; therefore, these three modules are speculated to be essential for running because there were no changes in the number of modules occurred. Module 1 was activated to absorb the weight load during the stance phase, and module 2 was activated for push-off. The function of module 3 was to prepare for landing. These modules were roughly similar to the basic modules reported by Cappellini et al.; therefore, these three modules are speculated to be essential for running because there were no changes following the intervention. Module 4 pre was similar to the basic module. However, the SP for module 4 between pre and post sessions was low; hence, module 4 activity was different between the pre and post sessions. Module 4 pre engaged the trunk muscles and it was activated in the stance phase. Researchers reported that trunk muscles are activated to enhance the rigidity of the trunk, stabilize the lumbar spine and pelvic girdle, and transmit ground reaction forces efficiently during push-off, which ends the stance phase. From the activation coefficient of module 4 pre, the timing of the trunk muscle activity was similar to that reported by Saunders et al.; therefore, module 4 pre may be activated to stabilize posture through the trunk muscles during the stance phase. Conversely, module 4 post engaged the muscles around the pelvis and the activation coefficient indicated that it was activated after landing. Thus, the role of module 4 post may be related to postural control for landing impact using the muscles around the pelvis. From our findings, postural control function changed from the trunk muscles to the muscles around the pelvis because of 10 minutes running. In a study targeting long distance running as marathon, it has been indicated that running distance is a risk factor of running related injuries. Therefore, the alteration of module due to running may be relevant to running related injuries of lower limb.

Turpin et al. reported that there was no change for the modules due to muscle fatigue during rowing exercise although...
our results showed a couple of different modules. Rowing exercise is performed in a stable condition in the sitting position. Our data indicated that the module with an altered function was the posture control module. Running may be a less stable exercise than rowing because of the inherent double floating phase during which the body is unsupported. Additionally, our intervention was not a fatigue protocol. Thus, our data was different from those obtained by Turpin et al\textsuperscript{29}). As a limitation of this study, there was a possibility that the intensity of intervention was low. Before this study, we intervened the high intensity exercise than this study as preliminary experiments. However, we were unable to attach and maintain the surface electrodes because the subjects perspired while running. Therefore, we performed the study with 70% running intensity and intervened for 10 minutes. Actually, the exercise intensity during sports competitions is higher and the exercise period is longer than the intervention in this study. Our results indicated that module 4 pre, which had the role of posture control through the trunk muscles, changed to module 4 post, which engaged muscles around the pelvis with 10 minutes running. Therefore, the increased contribution from module 4 post may induce running related injury because most running related injuries are caused by overuse\textsuperscript{12, 13}). Future study should investigate the relationship between the modules and prospective injury survey.

In conclusion, we investigated muscle synergy during running using NMF. The module that engaged the trunk muscles to control initial posture while running changed to engage the muscles around the pelvis for posture control after 10 minutes running. The change of posture control function from trunk muscles to muscles around the pelvis may be a risk for running related injuries.

### Table 1. The mean of VAF of each number of modules

| Number of modules | 1     | 2     | 3     | 4     | 5     | 6     | 7     |
|-------------------|-------|-------|-------|-------|-------|-------|-------|
| VAF (%) pre       | 69.8 ± 5.3 | 81.5 ± 2.4 | 86.5 ± 2.1 | 90.5 ± 1.6 | 93.0 ± 1.4 | 95.1 ± 1.1 | 96.2 ± 0.9 |
| VAF (%) post      | 71.0 ± 7.8 | 81.6 ± 5.7 | 86.8 ± 4.1 | 90.5 ± 2.6 | 93.0 ± 2.2 | 95.2 ± 1.6 | 96.6 ± 1.4 |

The number of modules is decided when the VAF exceeds 90% for the first time.

VAF: variance accounted for

![Fig. 2. Comparison of extracted muscle weighting and activation coefficient between pre and post session running](image)

Muscle weighting represents the relative weighting of each muscle within each module and activation coefficient represents the relative activation of the muscle weights.

The vertical axis of muscle weighting is the proportion of muscle activity (from 0 to 1) and that of activation coefficient is the activity value of muscle weighting.

SP: scalar-product, if SP>0.75, these modules are judged to be identical. In this figure, only module 4 is different between pre and post.

P-flex: planter flexor, K-ext: knee extensor, K-flex: knee flexor, H-ext: hip extensor
REFERENCES

1) Bernstein N: The co-ordination and regulation of movement. 1967.
2) d’Avella A, Saltiel P, Bizzi E: Combinations of muscle synergies in the construction of a natural motor behavior. Nat Neurosci, 2003, 6: 300–308. [Medline] [CrossRef]
3) Bizzi E, Cheung VC: The neural origin of muscle synergies. Front Comput Neurosci, 2013, 7: 51. [Medline] [CrossRef]
4) d’Avella A, Fernandez L, Portone A, et al.: Modulation of phasic and tonic muscle synergies with reaching direction and speed. J Neurophysiol, 2008, 100: 1433–1454. [Medline] [CrossRef]
5) Dominici N, Ivanenko YP, Cappellini G, et al.: Locomotor primitives in newborn babies and their development. Science, 2011, 334: 997–999. [Medline] [CrossRef]
6) Lacquaniti F, Ivanenko YP, Zago M: Patterned control of human locomotion. J Physiol, 2012, 590: 2189–2199. [Medline] [CrossRef]
7) Berger DJ, Gentner R, Edmunds T, et al.: Differences in adaptation rates after virtual surgeries provide direct evidence for modularity. J Neurosci, 2013, 33: 12384–12394. [Medline] [CrossRef]
8) d’Avella A, Saltiel P, Bizzi E: Combinations of muscle synergies in the construction of a natural motor behavior. Nat Neurosci, 2003, 6: 300–308. [Medline] [CrossRef]
9) 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]
10) Clement DB, Taunton JE, Smart GW, et al.: A survey of overuse running injuries. Phys Sportsmed, 1981, 9: 47–58. [Medline] [CrossRef]
11) Taunton JE, Ryan MB, Clement DB, et al.: A retrospective case-control analysis of 2002 running injuries. Br J Sports Med, 2002, 36: 95–101. [Medline] [CrossRef]
12) van Mechelen W: Running injuries. A review of the epidemiological literature. Sports Med, 1992, 14: 320–335. [Medline] [CrossRef]
13) Ballas MT, Tytko J, Cookson D: Common overuse running injuries: diagnosis and management. Am Fam Physician, 1997, 55: 2473–2484. [Medline]
14) Cowan DN, Jones BH, Frykman PN, et al.: Lower limb morphology and risk of overuse injury among male infantry trainees. Med Sci Sports Exerc, 1996, 28: 945–952. [Medline] [CrossRef]
15) Macera CA, Pate RR, Powell KE, et al.: Predicting lower-extremity injuries among habitual runners. Arch Intern Med, 1989, 149: 2565–2568. [Medline] [CrossRef]
16) Weist R, Eils E, Rosenbaum D: The influence of muscle fatigue on electromyogram and plantar pressure patterns as an explanation for the incidence of metatarsal stress fractures. Am J Sports Med, 2004, 32: 1893–1898. [Medline] [CrossRef]
17) Lee DD, Seung HS: Algorithms for non-negative matrix factorization. In Advances in Neural Information Processing Systems, 2001, 556–562.
18) Torres-Oviedo G, Macpherson JM, Ting LH: Muscle synergy organization is robust across a variety of postural perturbations. J Neurophysiol, 2006, 96: 1536–1546. [Medline] [CrossRef]
19) 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 1985, 1986, 10: 1727–1736. [Medline] [CrossRef]
20) Hug F: Can muscle coordination be precisely studied by surface electromyography? J Electromyogr Kinesiol, 2011, 21: 1–12. [Medline] [CrossRef]
21) Cheung VC, Turolla A, Agostini M, et al.: Muscle synergy patterns as physiological markers of motor cortical damage. Proc Natl Acad Sci USA, 2012, 109: 14652–14656. [Medline] [CrossRef]
22) Vaz JR, Ostad HB, Cabri J, et al.: Muscle coordination during breaststroke swimming: comparison between elite swimmers and beginners. J Sports Sci, 2016, 34: 1941–1948. [Medline] [CrossRef]
23) van den Brand R, Heutschi J, Barraud Q, et al.: Restoring voluntary control of locomotion after paralyzing spinal cord injury. Science, 2012, 336: 1182–1185. [Medline] [CrossRef]
24) Reisman DS, Block HJ, Bastian AJ: Interlimb coordination during locomotion: what can be adapted and stored? J Neurophysiol, 2005, 94: 2403–2415. [Medline] [CrossRef]
25) Ting LH, Chiel HJ, Trumbower RD, et al.: Neuromechanical principles underlying movement modularity and their implications for rehabilitation. Neuron, 2015, 86: 38–54. [Medline] [CrossRef]
26) Saunders SW, Rath D, Hodges PW: Postural and respiratory activation of the trunk muscles changes with mode and speed of locomotion. Gait Posture, 2004, 20: 280–290. [Medline] [CrossRef]
27) Saunders JB, Inman VT, Eberhart HD: The major determinants in normal and pathological gait. J Bone Joint Surg Am, 1953, 35-A: 543–558. [Medline] [CrossRef]
28) Tang PF, Woolacott MH, Chong RK: Control of reactive balance adjustments in perturbed human walking: roles of proximal and distal postural muscle activity. Exp Brain Res, 1998, 119: 141–152. [Medline] [CrossRef]
29) Turpin NA, Guével A, Durand S, et al.: Fatigue-related adaptations in muscle coordination during a cyclic exercise in humans. J Exp Biol, 2011, 214: 3038–3101. [Medline] [CrossRef]