Effects of unstable shoes on energy cost during treadmill walking at various speeds

Keiji Koyama [1], Hisashi Naito [2]✉, Hayao Ozaki [2] and Toshio Yanagiya [1]
[1] Department of Sports Biomechanics, and [2] Department of Exercise Physiology, Graduate School of Health and Sports Science, Juntendo University, Inzai, Chiba, Japan

Abstract

In recent years, shoes having rounded soles in the anterior-posterior direction have been commercially introduced, which are commonly known as unstable shoes (US). However, physiological responses during walking in US, particularly at various speeds, have not been extensively studied to date. The purpose of this study was to investigate the effect of wearing unstable shoes while walking at low to high speeds on the rate of perceived exertion (RPE), muscle activation, oxygen consumption (VO_2), and optimum speed. Healthy male adults wore US or normal walking shoes (WS), and walked at various speeds on a treadmill with no inclination. In experiment 1, subjects walked at 3, 4, 5, 6, and 7 $km \cdot h^{-1}$ (duration, 3 min for all speeds) and were recorded on video from the right sagittal plane to calculate the step length and cadence. Simultaneously, electromyogram (EMG) was recorded from six different thigh and calf muscles, and the integrated EMG (iEMG) was calculated. In experiment 2, RPE, heart rate and VO_2 were measured with the walking speed being increased from 3.6 to 7.2 $km \cdot h^{-1}$ incrementally by 0.9 $km \cdot h^{-1}$ every 6 min. The optimum speed, defined by the least oxygen cost, was calculated from the fitted quadratic relationship between walking speed and oxygen cost. Wearing US resulted in significantly longer step length and lower cadence compared with WS condition at any given speed. For all speeds, iEMG in the medial gastrocnemius and soleus muscles, heart rate, and VO_2 were significantly higher in US than WS. However, RPE and optimum speed (US, 4.75 ± 0.32 $km \cdot h^{-1}$; WS, 4.79 ± 0.18 $km \cdot h^{-1}$) did not differ significantly between the two conditions. These results suggest that unstable shoes can increase muscle activity of lower legs and energy cost without influencing RPE and optimum speed during walking at various speeds.

Key words: Rocker sole shoes, oxygen consumption, EMG, RPE, optimum speed.

Introduction

Physical activity and exercise are recommended to maintain and improve physical fitness and health. Many countries have set a guideline and introduced programs to promote exercise and to increase daily energy consumption for prevention of metabolic syndromes such as diabetes, hypertension and cardiovascular disease. The keys to increasing daily energy consumption are intensity and time of physical activity or exercise.

In recent years, some shoe makers have introduced shoes designed to increase the level of muscle activity when worn. These shoes have soles that are rounded in the anterior-posterior direction and thus become unstable during standing, giving these shoes names such as rocker sole shoes or unstable shoes (US). The unstable position that is produced as a result of wearing US is thought to require greater use of the lower extremity muscles, thereby increasing daily energy consumption without additional exercise. For example, maintaining a standing position when wearing US agitates the center of pressure compared with wearing normal walking shoes (WS) and leads to increased leg or foot muscles activity (Landry et al., 2010; Nigg et al., 2006). However, it has not been shown that the increased muscle activity actually leads to increased energy consumption while maintaining a standing position. Other studies have reported on the effects of wearing US during walking but the effectiveness of wearing these shoes is controversial. Romekes et al. (2006) found that wearing US increases muscle activation during walking, compared with wearing WS, while another study showed decreased activity in some muscles during walking (Nigg et al., 2006). Van Engelen et al. (2010) reported that wearing US during walking at normal speed increases oxygen consumption (VO_2), whereas Gjovaag et al. (2011) observed no increase in VO_2, and Hansen et al. (2011) found that VO_2 decreased.

The experimental designs in these studies differed in several aspects, such as the subjects, type of shoes, definition of walking speed, and method of walking, making comparisons among studies difficult. Since many of these previous studies were carried out using a single individual walking speed, the discrepancy as to the effect of US on VO_2 may partly be ascribed to the inconsistency in walking speed among studies. In other words, US may impact on VO_2 response differently depending on walking speed. It was therefore of our interest to determine VO_2 during walking in US vs. normal shoes at various speeds to detect the interaction between walking speed and VO_2 (or energy expenditure) response in a controlled experiment. The present study therefore examined, at various walking speeds, VO_2 as well as the rate of perceived exertion (RPE) with normal or unstable shoes. Through the speed-VO_2 data, the optimum walking speed, that would elicit the lowest oxygen cost, was calculated and compared between the two shoe conditions. To identify possible mechanisms for the anticipated changes in these values between the shoe conditions, step length and cadence, and EMG of lower legs were recorded for all given walking speeds.

Methods

Study design

This study was composed of two experiments using different subject samples. In experiment 1, we investigated

the effect of wearing US on step length, cadence, and EMG during walking. In the second experiment, the effect of US on RPE, VO_2, and optimum speed during walking was studied.

Subjects

A total of 14 healthy volunteers participated in the two experiments; six males (age, 26.3 ± 5.3 years; height, 1.72 ± 0.05 m; mass, 68.0 ± 6.1 kg) in experiment 1 and eight males (age, 24.0 ± 2.5 years; height, 1.70 ± 0.05 m; mass, 63.1 ± 4.7 kg) in experiment 2. They provided written informed consent after the experimental procedures, study design, and possible risks and benefits were explained. This study was approved by the ethics committee of Juntendo University, and the two experiments were conducted according to the Declaration of Helsinki.

Shoe conditions

Unstable shoes (US; Shape-ups, SKECHERS, USA) and normal walking shoes (WS) commercially available in Japan were used. The mean shoe mass was 1.00 kg for US and 0.54 kg for WS. The sole thicknesses of the fore foot and rear foot were 2.5 and 4.3 cm for US and 1.5 and 3.3 cm for WS, respectively. The ratio of flat area to total area of the sole was 60% for US and 80% for WS. The curvature of the arc had a 35-cm radius for US and a 60-cm radius for WS. According to the methods of Hansen et al. (2011), the radii are represented as a ratio to leg length, resulting in curvatures of approximately 40% (R40) and 70% (R70) for US and WS, respectively. Thus, the degree of curvature was greater for US than for WS. The hardness of the mid-sole was measured at the fore foot and rear foot using a durometer (WR-207E, Nishi Tokyo Seimitsu, Japan), where hardness was defined as the force required to push the durometer needle 2.5 mm. The hardness at the fore foot was 2.5 ± 0.2 N for US and 3.7 ± 0.3 N for WS, and that at the rear foot was 1.4 ± 0.2 N for US and 4.3 ± 0.2 N for WS. Thus, the sole material of the US was softer than that of the WS.

Walking test protocol

In experiment 1, subjects wearing US or WS performed a walking warm-up for 20 min on a treadmill and then walked on a treadmill at speeds of 3, 4, 5, 6, and 7 km·h^{-1} at 0% incline. Each subject walked for 3 min at each speed. After resting for at least 1 h to minimize fatigue effects, the subjects switched to whichever shoe type they had not worn the first time, and repeated the protocol. All trials were conducted on the same day.

In experiment 2, the subjects wearing US or WS performed the warm-up as in experiment 1 and then walked at 0% incline on a treadmill as the speed was increased from 3.6 to 7.2 km/h in increments of 0.9 km·h^{-1} every 6 min. US and WS trials were conducted on different days.

In all experiments, subjects wore a short-sleeved shirt and short pants. The order in which the US and WS were worn was randomized among the subjects. All experiments were carried out in a room with an ambient temperature of 20°C and a relative humidity of 50%.

Measurements and analysis
Contact time, cadence, and step length during walking

In experiment 1, the walking motion of the subjects was captured from the right sagittal plane using a video camera (Exlim EX-F1, Casio, Japan), with a frame rate of 300 fps and a shutter speed of 1/500 s. Captured images were recorded on a memory card. Contact time and cadence were quantified from the recorded images using a software (QuickTime Player 7.6.9, Apple, Japan). Step length was calculated by dividing the walking speed by double the cadence. A walking cycle was defined as the time from initial foot contact to the following ipsilateral initial contact. Contact time was calculated as a percentage of the walking cycle time. These values are represented as the average of five cycles.

Electromyogram (EMG)

In experiment 1, bipolar surface electrodes (Ambu, Balerup, Denmark; diameter, 10 mm; center-to-center distance, 20 mm) were placed over the rectus femoris, vastus lateralis, biceps femoris, tibialis anterior, soleus, and medial gastrocnemius muscles of the right leg. The longitudinal axes of the electrodes were in line with the direction of the underlying muscle fibers. The reference electrode was attached to the anterior-superior iliac spine. Inter-electrode resistance was maintained at <3kΩ by skin preparation. Pulling artifacts were avoided by properly fixing the electrode cables to the skin with tape.

Surface EMG signals were obtained using a telemetry system (WEB-5000, Nihon Kohden, Japan). The signals were passed through a locally ground pre-amplifier and sent to a receiving unit via a lightweight transmitter belt worn around the waist. The analogue signals were sent telemetrically to a recording computer (12-bit analog-to-digital converter) with a sampling frequency of 1 kHz. Band pass filtering (20–200 Hz) was used to remove movement artifacts and signal noise. EMG data were quantified by integrated the full-wave rectified EMG (iEMG) for 1 min. The iEMG of each muscle with US was calculated relative to the iEMG during walking with WS.

Heart rate, VO$_2$, and oxygen cost

In experiment 2, the heart rate was continually recorded every 5 s with a Polar portable device (CS400, Polar Electro, Finland). Simultaneously, VO_2 was recorded every 30 s with a metabolic cart (AE-300S, Minato Medical Science, Japan). The analyzer was calibrated with reference gases of known concentrations before each experiment. The values were averaged during the final 3 min of sampling at each walking speed. Oxygen cost was defined as the VO_2 required to walk 1 km at each speed. The optimum speed, defined by least oxygen cost, was calculated from the quadratic relationship between speed and oxygen cost using a curve fitting program (Excel 2010, Microsoft, WA, USA)

RPE

RPE was recorded using a Borg scale (Borg, 1974) at the end of walking at each speed.

Table 1. Step length, cadence and percentage of contact time during walking. Values are means (±SD).

Shoes conditions	Speed (km·h^{-1})	US	WS	
Step length (m)	3	.57 (.04)	.52 (.03)	*
	4	.65 (.05)	.63 (.03)	
	5	.74 (.04)	.72 (.03)	*
	6	.83 (.04)	.80 (.03)	*
	7	.88 (.05)	.85 (.05)	*
Cadence (step·min^{-1})	3	43.8 (3.2)	48.5 (2.9)	*
	4	51.3 (3.9)	53.1 (2.8)	*
	5	56.0 (2.9)	57.9 (2.1)	
	6	60.6 (2.7)	62.9 (2.4)	*
	7	66.8 (3.7)	69.2 (4.4)	*
Percentage of contact time (%)	3	67.1 (1.5)	68.2 (1.2)	
	4	66.2 (1.4)	66.3 (1.2)	
	5	64.5 (1.2)	64.4 (1.2)	
	6	63.3 (.6)	62.6 (.7)	
	7	61.8 (1.2)	61.8 (.9)	

US: unstable shoes, WS: walking shoes. * Significantly different from US.

Statistical analysis

All statistical analyses were performed using SPSS version 17.0 software (SPSS Inc., Chicago, IL, USA). The effects of shoe type on step length, cadence, RPE, VO_2, and oxygen cost across speeds were analyzed using a two-way repeated-measures analysis of variance. When an interaction was identified, Bonferroni corrected pairwise post hoc comparisons were made between individual shoe type and speed. Moreover, the difference in the optimum speed between shoe types was tested using a paired-*t*-test. P values of equal to or less than 0.05 was considered to be statistically significant, and all values are presented as mean ± standard deviation (SD).

Results

Step length, cadence, and percentage of contact time

Table 1 shows the step length, cadence, and percentage of contact time during walking while wearing US and WS. Step length was significantly longer (4–11%) for US than for WS. Cadence was 3–10% lower for US than for WS. The main effects of shoe type and speed on step length and cadence were significant (shoes, $p < 0.01$; speed, $p < 0.01$). The percentage of contact time did not differ significantly between US and WS.

EMG

Figure 1 shows the absolute and relative iEMG values for each muscle. For all muscles, the iEMG showed a tendency to be higher when walking with US than with WS.
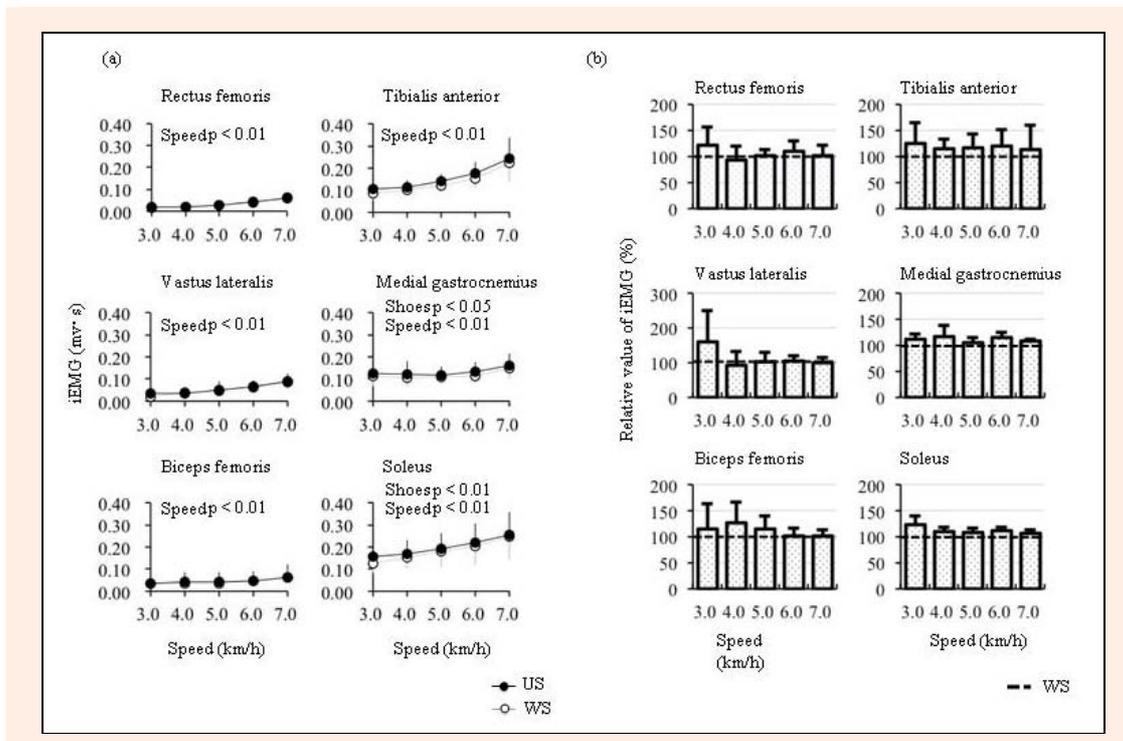

Figure 1. The absolute and relative integrated electromyogram (iEMG) values of the lower extremity muscles during walking at various speeds. **(a)** Absolute iEMG values. Black and white plots show US and WS, respectively. **(b)** Relative iEMG values. The iEMG of each muscle with US was calculated relative to the iEMG with WS. The dashed line at 100% indicates the level of WS. US: unstable shoes, WS: walking shoes. Values are means ± SD.

In particular, the medial gastrocnemius (6–16%) and soleus muscles (8–23%) had significantly higher iEMG values, and the main effects of shoe type and speed were significant for these two muscles (medial gastrocnemius: $p < 0.05$ and $p < 0.01$ for shoes and speed, respectively; soleus: $p < 0.01$ for each).
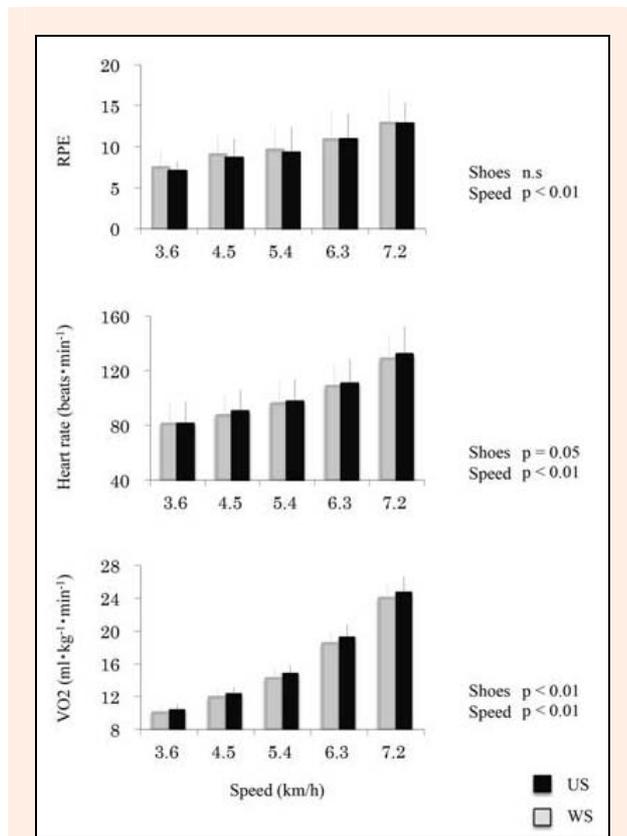

Figure 2. **Rating of perceived exertion (RPE), heart rate and oxygen consumption (VO$_2$) during walking at various speeds.** Black and gray bars show US and WS, respectively. US, unstable shoes; WS, walking shoes. Values are means ± SD.

RPE, heart rate, VO$_2$, and oxygen cost

Figure 2 shows the RPE, heart rate, and VO$_2$. The relationship between speed and oxygen cost is presented in Figure 3. Although RPE increased with increasing walking speed, it was not significantly different between US and WS. Heart rate was significantly higher (0.4–2.9%) for US than for WS. Moreover, VO$_2$ and oxygen cost were significantly higher (3.4–4.9%) for US than for WS. Shoe type and speed had significant effects on heart rate, VO$_2$, and oxygen cost (heart rate: $p = 0.05$ and $p < 0.01$ for shoes and speed, respectively; VO$_2$ and oxygen cost: $p < 0.01$ for both). Optimum speed did not differ significantly between US (4.75 ± 0.32 km·h^{-1}) and WS (4.79 ± 0.18 km·h^{-1}).

Discussion

To our knowledge, only two studies have examined the effects of wearing commercially available US on VO$_2$ while walking at one or two speeds (Gjovaag et al., 2011; van Engen et al., 2010). Therefore, earlier studies were able to identify only fragments of walking speed vs. VO$_2$, which presumably have a quadratic curve relation. For example, Van Engelen et al. (2010) reported that wearing US caused an increase of about 10% in metabolic energy cost compared with WS when walking at a fixed speed (4.5 km·h^{-1}). Gjovaag et al. (2011) measured VO$_2$ during walking at a self-selected speed (4.5 km·h^{-1}) and a fixed higher speed (5.8 km·h^{-1}) while wearing US and WS. The VO$_2$ at the self-selected speed for US and WS were 11.7 ± 1.3 and 11.4 ± 1.2 ml·min^{-1}·kg^{-1}, respectively, and the corresponding values for the higher speed were 15.3 ± 1.2 and 15.1 ± 0.8 ml·min^{-1}·kg^{-1}. Although the VO$_2$ for US was 1–2% higher than that for WS, the difference was not significant.
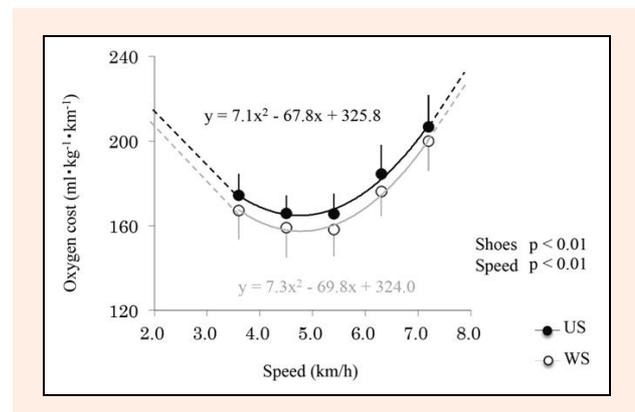

Figure 3. **Relationship between walking speed and oxygen cost during walking.** Black and white plots show US and WS, respectively. US: unstable shoes, WS: walking shoes. Values are means ± SD.

This study is the first to show that the quadratic relationship between walking speed and VO$_2$ shifted to a higher value in subjects wearing US, and that VO$_2$ increased by 3–5% in subjects wearing US while walking compared with subjects wearing WS. Optimum speed and RPE did not differ according to shoe type, but a decreased cadence and increased step length were significantly associated with wearing US while walking.

The general features of US, compared with WS, are that the sole is rounded, thicker, and softer, and the shoe mass is greater. These features might have changed the walking motion, thus leading to a change in muscle activity and/or VO$_2$.

The largest difference between US and WS was the shape of the sole. The flat area of the sole as a percentage of the total area on the bottom of the shoe was less for US (60%) than for WS (80%), and the radius of the curve was smaller for US (35 cm) than for WS (60 cm), indicating that the US had a pronounced curvature compared with the WS. It is possible that the increased instability attributable to the sole shape increased the activities of the thigh and calf muscles during standing, thereby contributing to increased energy consumption (Nigg et al., 2006). It has also been reported that wearing US, compared with WS, while walking at a self-selected speed decreased both cadence and step length, and increased muscle activity of the lower extremities (Romkes et al., 2006). Wearing US while walking at a fixed speed (5.0 km·h^{-1}) increased

medial gastrocnemius muscle activity, but the changes in muscle activity varied among different muscles (Nigg et al., 2006). In contrast, it has been suggested that the increased curvature of US may decrease oxygen consumption. Hansen et al. (2011) made shoes with soles of different curvatures and compared the VO_2 during walking at a self-selected speed (4.03 km/h). The VO_2 (ml·min^{-1}·kg^{-1}) was 11.9 for flat-soled shoes, 10.9 for shoes with R25 curvature (25% of leg length), 10.7 for shoes with R40 curvature (40% of leg length), and 11.0 for shoes with R55 curvature (55% of leg length). These results indicate that it was easier to walk in curved-sole shoes, as VO_2 decreased by 10% compared with flat-soled shoes (p < 0.05). However, no significant difference in VO_2 was observed among the different curvatures (R25-R55). In the present study, the ratios of the radius to leg length for US and WS corresponded to about 40% (R40) and 70% (R70), respectively. We cannot directly compare the results obtained in our experiment using commercially available shoes and the results obtained using purely experimental shoes. Nevertheless, the findings that the flat area of the US (60%) was less than that of the WS (80%) and that the curvature of the US (R40) was greater than that of the WS (R70) suggest that the curved shape of the shoes would not affect VO_2. Therefore, the 4% greater VO_2 for US cannot be explained by their curved sole.

The thickness of the sole differed between US (fore foot, 2.5 cm; rear foot, 4.3 cm) and WS (fore foot, 1.5 cm; rear foot, 3.3 cm), making the leg length 1 cm longer when wearing US compared with WS. This represents only 1% increase of a subject's leg length in the present study. Meanwhile, the step length increased 2 to 5 cm when wearing US compared with WS corresponding to 3–13% increases of the step length. A previous study by van Engelen et al., 2010 demonstrated that step length during walking on the ground at 4.5 km·h^{-1} was 1cm longer for US than WS, which appeared much less than the 3–13% increase observed in the present study, and therefore makes it difficult for only 1% increase of the leg length to explain the entire magnitude of increase in step length. One of the possible mechanisms may be that the backward force produced by the treadmill may have influenced increased step length in the present study.

In the present study, as the shoe mass differed between US (1.0 kg) and WS (0.54 kg), the total weight of shoes was 1kg higher in US than WS, equivalent to about 1% body weight. This increase might cause 1% increase in energy expenditure if they walked at the same speed. Therefore it is possible that this difference might have influenced muscle activity and VO_2. However, it is well known that a difference of 0.5 kg in shoe mass had little effect on VO_2 during walking at normal speed (Hettinger and Muller, 1953). In addition, one foot must be on the ground during walking. Therefore, the greater mass of US may not directly influence muscle activity or VO_2 very much.

Thus, the 4% greater VO_2 for US cannot be clearly explained by sole curvature, sole thickness, or shoe mass, leaving sole softness (or hardness) as a possible key factor underlying the increased VO_2. The US sole was softer

than the WS sole (US hardness: fore foot, 2.5 ± 0.2 N and rear foot, 1.4 ± 0.2 N; WS hardness: fore foot 3.7 ± 0.3 N and rear foot, 4.3 ± 0.2 N). During normal walking, the heel contacts the ground with dorsiflexion, which then becomes plantar flexion at toe off (Romkes et al., 2006). Van Engelen et al. (2010) reported that the force developed for propulsion is attenuated before it is conveyed to the ground when the sole hardness is lower, resulting in increased metabolic energy cost during walking at the later supporting phase. Stewart et al. (2007) demonstrated that lower sole hardness of an US results in increased plantar pressure on the fore foot during walking. In the present study, the iEMG of the lower leg was higher (soleus, 7–23% higher; medial gastrocnemius, 6–16% higher) when walking while wearing US compared with WS. These suggest that because the US sole material was softer than that of WS, the propulsion force of the push-off phase was reduced for US. Therefore, to maintain a given walking speed, the medial gastrocnemius and soleus muscles must develop greater activity when wearing US. Although the increased iEMG of these muscles could be caused by an increase of the contact time rather than the intensity of muscle contraction, there was no significant difference in the percentage of contact time between US and WS over a given time period (Table 1). This suggests that the increased iEMG with US could be largely accounted for by increased intensity of muscle activity.

Taken together, our results indicate that the increased step length and increased muscle activity per unit time in the lower leg, particularly the calf, during walking at all speeds while wearing US contributed to the increase in VO_2 and oxygen cost. Nevertheless, these observations are insufficient to explain the entire increase in VO_2. The present study measured muscle activity only in the lower body. Increased activity of muscles in the body trunk or changes in the motion of the upper extremities might have occurred because of changes in posture, or unknown factors might have contributed to the increase in VO_2. A previous study showed that when step length was extended by 20% while walking at a constant speed, the mechanical work of the ankle joint increased, the vertical movement of the center of mass increased by 24%, and metabolic power increased by 36% (Gordon et al., 2009). The present study demonstrated an increase of 3–13% in step length during walking at various speeds while wearing US compared with WS, which might have led to the increase in VO_2. Furthermore, because walking propulsion force is developed by muscle tension and utilization of the elastic energy of the tendinous tissues (Fukunaga et al., 2001; Ishikawa et al., 2005), the effect of tendinous tissue behavior on VO_2 during walking should be investigated in the future.

We observed little change in RPE despite the significant increase in VO_2 in subjects wearing US. Gjovaag et al. (2011) reported that while wearing US, RPE increased only at the fastest walking speed with a 10% inclination of the treadmill. Wang and Hansen (2010) reported the possibility that wearing curved-sole shoes instead of flat-sole shoes may bring a sense of easier walking because of the changes in ankle joint movement during walking. In the present study, although heart rate

and VO$_2$ increased, an increase in RPE might have been suppressed by the feeling of easier walking while wearing US.

Conclusion

Unstable shoes can increase muscle activities of lower leg and energy cost without changes in RPE and optimum speed during walking at various speeds.

References

Borg, G.A. (1974) Perceived exertion. *Exercise and Sport Science Reviews* **2,** 131-153.

Fukunaga, T., Kubo, K., Kawakami, Y., Fukashiro, S., Kanehisa, H. and Maganaris, C.N. (2001) In vivo behaviour of human muscle tendon during walking. *Proceedings. Biological Sciences* **7,** 229-33.

Garber, C.E., Blissmer, B., Deschenes, M.R., Franklin, B.A., Lamonte, M.J., Lee, I.M., Nieman, D.C. and Swain, D.P. (2011) American College of Sports Medicine position stand. Quantity and quality of exercise for developing and maintaining cardiorespiratory, musculoskeletal, and neuromotor fitness in apparently healthy adults: guidance for prescribing exercise. *Medicine and Science in Sports and Exercise* **43,** 1334-1359.

Gjøvaag, T.F., Dahlen, I., Sandvik, H. and Mirtaheri, P. (2011) Oxygen Uptake and Energy Expenditure during Treadmill Walking with Masai Barefoot Technology (MBT) Shoes. *Journal of Physical Therapy Science* **23,** 149-153.

Gordon, K.E., Ferris, D.P. and Kuo, A.D. (2009) Metabolic and mechanical energy costs of reducing vertical center of mass movement during gait. *Archives of Physical Medicine and Rehabilitation* **90,** 136-144.

Hansen, A.H. and Wang, C.C. (2011) Effect of rocker shoe radius on oxygen consumption rate in young able-bodied persons. *Journal of Biomechanics* **44,** 1021-1024.

Hettinger, T.H. and Muller, E.A. (1953) Der EinifluB des Schuhgewichtes auf den Energieumsatz beim Gehen und Lastentragen. *Arbeitsphysiologie* **15,** 33-40.

Ishikawa, M., Komi, P.V., Grey, M.J., Lepola, V. and Bruggemann, G.P. (2005) Muscle-tendon interaction and elastic energy usage in human walking. *Journal of Applied Physiology* **99,** 603-608.

Landry, S.C., Nigg, B.M. and Tecante, K.E. (2010) Standing in an unstable shoe increases postural sway and muscle activity of selected smaller extrinsic foot muscles. *Gait & Posture* **32,** 215-219.

Nigg, B., Hintzen, S. and Ferber, R. (2006) Effect of an unstable shoe construction on lower extremity gait characteristics. *Clinical Biomechanics* **21,** 82-88.

Romkes, J., Rudmann, C. and Brunner, R. (2006) Changes in gait and EMG when walking with the Masai Barefoot Technique. *Clinical Biomechanics* **21,** 75-81.

Stewart, L., Gibson, J.N. and Thomson, C.E. (2007) In-shoe pressure distribution in "unstable" (MBT) shoes and flat-bottomed training shoes: a comparative study. *Gait & Posture* **25,** 648-651.

van Engelen, S.J., Wajer, Q.E., van der Plaat, L.W., Doets, H.C., van Dijk, C.N. and Houdijk, H. (2010) Metabolic cost and mechanical work during walking after tibiotalar arthrodesis and the influence of footwear. *Clinical Biomechanics* **25,** 809-815.

Wang, C.C. and Hansen, A.H. (2010) Response of able-bodied persons to changes in shoe rocker radius during walking: changes in ankle kinematics to maintain a consistent roll-over shape. *Journal of Biomechanics* **43,** 2288-2293.

Key points

- During walking at various speeds, wearing unstable shoes results in longer step length and lower cadence compared with wearing WS.
- Wearing unstable shoes increases muscle activities of lower leg.
- Wearing unstable shoes shifts the quadratic relationship between walking speed and oxygen cost upward and increases energy cost about 4% without changes in RPE and optimum speed.

AUTHORS BIOGRAPHY

Keiji KOYAMA
Employment
Ph.D. Student, Graduate School of Health and Sports Science, Juntendo University, Japan. Lecturer, Faculty of Culture and Sport Policy, Toin University of Yokohama, Japan
Degree
MSc
Research interests
Biomechanics of sports movement, footwear design and ergonomics, muscle-tendon structure and function
E-mail: koyakei@yahoo.co.jp

Hisashi NAITO
Employment
Professor, Graduate School of Health and Sports Science, Juntendo University, Japan
Degree
PhD
Research interests
Muscle adaptation, stress proteins, genes, youth fitness
E-mail: naitoh@sakura.juntendo.ac.jp

Hayao OZAKI
Employment
Ph.D. Student, Graduate School of Health and Sports Science, Juntendo University, Japan
Degree
PhD
Research interests
Exercise physiology, sport and training science
E-mail: ozaki.hayao@gmail.com

Toshio YANAGIYA
Employment
Associate Professor, Graduate School of Health and Sports Science, Juntendo University, Japan
Degree
PhD
Research interests
Biomechanics of sports movements, skeletal muscle structure and function, training science
E-mail: yanagiya@sakura.juntendo.ac.jp

✉ **Hisashi Naito**
Department of Exercise Physiology, Graduate School of Health and Sports Science, Juntendo University, Inzai, Chiba, 270-1695, Japan