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# Biomechanical Factors Affecting Energy Cost During Running Utilizing Different Slopes

Running head: Biomechanical Factors Affecting Energy Cost during different slope running

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## ABSTRACT

This study aimed to examine the characteristics of electromyography (EMG) and kinematics of the supporting leg affecting energy cost while running at incline, level, and decline slopes. Twelve male Japanese middle- and long-distance runners volunteered for this study. The subjects were asked to run at  $13.5 \text{ km}\cdot\text{h}^{-1}$  on a treadmill under three slope conditions. Sagittal plane kinematics and the EMG of the lower limb muscles, respiratory gases were recorded. Energy cost differed significantly between slopes, being the lowest in decline slope and the greatest in incline slope. Integrated EMG (iEMG) of leg extensor muscles was greater in the incline slope than in the decline slope, and iEMG of the gastrocnemius and soleus muscles correlated positively with energy cost. The knee and ankle joint kinematics were associated with energy cost during running. In incline slope, the knee and ankle joints were more extended (plantarflexed) to lift the body. These movements may disturb the coordination between the ankle and knee joints. The gastrocnemius muscle would do greater mechanical work to plantarflex the ankle joint rather than transfer mechanical energy as well as greater mechanical work of mono-articular muscles. These muscular activities would increase energy cost.

Key words: running economy, kinematics, EMG, inclination,

## INTRODUCTION

Running economy is one of the most important determinant factors for distance running performance (Conley and Krahenbuhl, 1980; di Prampero, Atchou, Bruckner, & Moia, 1986; Midgley, McNaughton, & Jones, 2007). Several biomechanical factors significantly affect running economy (Kyröläinen, Belli, & Komi, 2001; Moore, 2016; Saunders, Pyne, Telford, & Hawley, 2004; Williams and Cavanagh, 1987). Williams and Cavanagh (1987) indicated that 54% of the inter-individual variance in running economy can be explained by biomechanical factors, although factors affecting running economy have reportedly been inconsistent. For example, a study reported that rearfoot strike is more economical (Ogueta-Alday, Rodriguez-Marroyo, & Garcia-Lopez, 2014), whereas other studies reported that no difference exists between running economy and footstrike patterns (Ardigo, Lafortuna, Minetti, Mognoni, & Saibene, 1995; Gruber, Umberger, Braun, & Hamill, 2013).

On the other hand, the finding that less vertical displacement of the center of mass (CoM) is associated with good running economy is consistent with many studies (Egbonu, Cavanagh, & Miller, 1990; Moore, Jones, & Dixon, 2014; Tseh, Caputo, & Morgan, 2008; Williams and Cavanagh, 1987). Teunissen, Grabowski, & Kram (2007) investigated determinant factors of energy cost, which is

an index of running economy, during running while altering body weight. They reported that generating force to support body weight is a primary determinant of energy cost. Heise and Martin (2001) also reported that the vertical impulse of ground reaction force correlates positively with running economy. In addition, Arellano and Kram (2011) suggested that dividing the energy cost induced by body weight support (potential energy) and forward propulsion (kinetic energy) is difficult. Therefore, they concluded that the work required for body weight support and forward propulsion could be approximately 80% of the total energy cost of running. However, the approach of Arellano and Kram (2011) and Teunissen, et al. (2007) might lead to unnatural alterations of running motion.

Seki, Kyröläinen, Numazu, Ohyama-Byun, & Enomoto (2019) suggested that changing exercise condition could alter movement and energy cost naturally and could clarify the biomechanical factors affecting energy cost during endurance exercise. They reported biomechanical factors affecting energy cost during repeated vertical jumping using this approach. Eriksson, Halvorsen, & Gullstrand (2011) attempted to manipulate vertical displacement and step frequency using visual and auditory feedback systems. They reported that the mechanical power of the CoM of the body can be reduced by a decrease in vertical displacement, but they did not measure metabolic variables. DeVita, Janshen,

Rider, Solnik, & Hortobágyi (2008) reported that vertical displacement was 19% greater in decline than in incline during ground contact. Thus, changing the slope of running surfaces is one of the possible manners to alter vertical displacement that could be associated with energy cost. Minetti, Moia, Roi, Susta, & Ferretti (2002) reported that the energy cost increases by 45% with incline running and decreased by -20% with decline running. Vertical displacements could be changed using various slopes of the running surface that might alter running movement with its energy cost naturally and could provide further insight regarding the biomechanical factors affecting energy cost.

Kinetics is also an important information to consider in the relationship between the movement of the lower limbs and energy cost. However, obtaining kinetics during treadmill running is difficult, because most treadmills cannot correctly measure ground reaction force. Gottschall and Kram (2005) measured the ground reaction force during incline and decline running, but they did not report running economy. Despite that Roberts and Belliveau (2005) measured joint kinetics during incline running, they did not investigate the relationship between joint kinetics and running economy. Instead of joint kinetics, electromyography (EMG) is another technique that provides beneficial information regarding how muscular activity influences energy cost. EMG is also known as a predictor of energy

expenditure (Arnaud, Zattara-Hartmann, Tomei, & Jammes, 1997; Bigland-Ritchie and Woods, 1976; Blake and Wakeling, 2013). Kyröläinen, et al. (2001) emphasized the importance of EMG to interpret the increase in energy cost, as it might be possible to evaluate the co-activation of agonists and antagonists and activities of bi-articular muscles. Earlier studies reported that energy cost between slopes differs significantly (e.g., Minetti et al., 2002), but its biomechanical mechanisms have not been explained yet. Vernillo et al. (2016) reviewed biomechanical and physiological studies of slope running and stated that important gaps in our biomechanical and physiological understandings of incline running still exist.

Therefore, examining the changes in EMG and kinematics of the lower limbs simultaneously with energy cost by manipulating the slope of the running surface would provide a new insight into the biomechanical factors of energy cost. Manipulating the slope of the running surface, which means changing vertical movement of the body as mentioned above, could clarify the work for body weight support from a kinematic aspect. The findings of the present study would be useful for improving running economy even in level running. Thus, the purpose of the present study was to quantify the characteristics of EMG and kinematics of the supporting leg affecting energy cost while

running at three different slopes. We hypothesized that changing slope affects changes in energy cost to lift and maintain the body position.

## MATERIALS AND METHODS

### Subjects

The subjects were 12 male Japanese middle- and long-distance runners (age:  $21.9 \pm 0.8$  years, height:  $1.71 \pm 0.05$  m; body mass:  $60.1 \pm 4.2$  kg). They were track athletes of the university and provided written informed consent prior to participation in the present study. They had over 60  $\text{mL}\cdot\text{kg}^{-1}\cdot\text{min}^{-1}$  of  $\text{VO}_2\text{max}$ . We asked the subjects to avoid intense exercise the night before the experiment. This study was approved by the ethics committee of the Faculty of Health and Sport Sciences, University of Tsukuba, Japan in accordance with the Declaration of Helsinki.

### Procedure and measurements

The subjects were asked to run on a treadmill (ORK-HS40-PRO; Ohtake Root Kogyo Co., Ltd, Japan) at  $13.5 \text{ km}\cdot\text{h}^{-1}$  under three slope conditions: +6% (incline), 0% (level), and -6% (decline).



Trial order was randomized on subject by subject basis. We employ only one running speed because mechanical work and many biomechanical variables may potentially differ between different running speeds. The running speed of  $13.5 \text{ km}\cdot\text{h}^{-1}$  was close to their lactate threshold. They had at least 5 minutes of rest between trials to decrease blood lactate level below  $2 \text{ mM}\cdot\text{L}^{-1}$ . The trial continued for 3 minutes to reach the steady-state level. They were wearing the short tights and the same footwear (Wave Cruise 9, Mizuno, Japan) to avoid effects of shoes.

Running kinematics were recorded using a motion capture system (Vicon MX, Vicon Motion Systems, UK) at 250 Hz. Thirty-nine reflective markers were attached to body landmarks (hand, wrist, elbow, shoulder, toe, fifth metatarsal bone, heel, lateral malleolus, shank, lateral condyle, thigh, greater trochanter, anterior superior iliac spine, posterior superior iliac spine, head, ear, suprasternal notch, seventh cervical vertebrae, xiphoid process, tenth thoracic vertebrae, and lower end of rib) and reflective tape was attached to the belt of the treadmill to convert body landmarks' coordinates to it on the absolute coordinate system like running on the track. The markers were placed on the skin excluding a few markers: the markers of the greater trochanter, anterior iliac spine, and posterior superior iliac spine were placed on the short tights, and the markers of the toe, 5<sup>th</sup> metatarsal bone, and heel were

attached on the shoes. Supporting leg kinematics applies to the right side for all subjects.

EMG was recorded with active surface electrodes (SX-230; Biometrics, UK) at a sampling frequency of 1 kHz from the rectus femoris (RF), vastus lateralis (VL), gluteus maximus (GM), biceps femoris (BF), tibialis anterior (TA), gastrocnemius medial (GA), and soleus (SO) muscles of the right leg. The inter-electrode distance was 20 mm. The electrodes were placed longitudinally over the muscle bellies between the center of the innervation zone and the distal tendon of each muscle in accordance with the SENIAM guidelines (Hermens et al., 1999). The EMG data were synchronized with kinematics data using the Vicon Nexus Software (Vicon Motion Systems, UK). Before running trials, the subjects performed maximal voluntary contraction (MVC) for normalizing EMG signals.

Respiratory gases were continuously analyzed using the breath-by-breath method and computerized standard open circuit technique (Iwayama et al., 2015) (AE-301s; Minato Medical Science, Japan). Blood lactate levels were collected from the fingertip and analyzed with Lactate Pro 2 (Arkay, Japan).

Analysis

Respiratory gas was analyzed during the last minute of the '3 minutes exercise period'.

Energy expenditure was calculated as an energy equivalent of 20202 J·L<sup>-1</sup> of oxygen, which refers to the respiratory exchange ratio (R) of 0.82 and a change of ±0.01, which corresponds to respective ±50 J changes in energy expenditure (Kyröläinen, et al., 2001). If R was 0.85, energy equivalent was 20352 J·L<sup>-1</sup>. When blood lactate levels were more than 2.0 mM·L<sup>-1</sup> (3 mL O<sub>2</sub>·kg<sup>-1</sup>·mM<sup>-1</sup>), energy expenditure was then calculated on the basis of an equivalent of 60 J·kg<sup>-1</sup>·mM<sup>-1</sup> (3 mL O<sub>2</sub>·kg<sup>-1</sup>·mM<sup>-1</sup>) (Kyröläinen, et al., 2001). This value was added to the overall energy expenditure. The energy cost was calculated as the energy expenditure divided by the running speed.

The two-dimensional coordinates in the sagittal plane were smoothed using a Butterworth low-pass digital filter at 10 Hz. Ground contact phase was detected based on the distance between the belt surface of the treadmill and following three markers: toe, fifth metatarsal bone, and heel. A rigid-body model consisting of 15 body segments (head, upper part of torso, lower part of torso, hand forearm, upper arm, foot, shank, and thigh) was constructed using two-dimensional coordinates of anatomical landmarks. We focused on the sagittal plane because vertical movement can be analyzed in the sagittal plane. The mass and center of mass location of each segment were estimated by the coefficients of Ae,

Tang, & Yokoi (1992). Then body's center of mass location (CoM) was obtained as a resultant center of mass of all body segments. The vertical displacement was defined as difference between the lowest height of the CoM during the support phase and highest height during following flight phase. The external work ( $W_{total}$ ) during ground contact was calculated using Equations 1–3 (Keir, Zory, Boudreau-Lariviere, & Serresse, 2012).

$$\Delta E_p = m \cdot g \cdot \Delta h_{CoM} \quad (\text{Eq. 1})$$

$$\Delta E_k = \frac{1}{2} \cdot m \cdot \Delta v_{CoM}^2 \quad (\text{Eq. 2})$$

$$W_{total} = |\Delta E_p| + |\Delta E_k| \quad (\text{Eq. 3})$$

where  $E_p$  is potential energy,  $m$  is body mass,  $g$  is acceleration of gravity,  $E_k$  is kinetic energy, and  $v_{CoM}$  is velocity of the CoM of the body. In addition, positive ( $W_{pos}$ ) and negative ( $W_{neg}$ ) mechanical works were calculated as sum of positive and negative  $\Delta E_p$  and  $\Delta E_k$ , respectively (Eq. 4 and 5).

$$W_{pos} = \Delta E_p^+ + \Delta E_k^+ \quad (\text{Eq. 4})$$

$$W_{neg} = \Delta E_p^- + \Delta E_k^- \quad (\text{Eq. 5})$$

EMG signals were high-pass filtered using a Butterworth digital filter at 10 Hz to eliminate the low-frequency motion artifact. Thereafter, integrated EMG (iEMG) during the ground contact was

calculated by rectifying and low-pass filtering EMG signals using a Butterworth digital filter at 15 Hz.

iEMG were normalized by the respective EMG values recorded during MVC and expressed as %MVC.

Kinematics and EMG were averaged during the contact phase of 10 running cycles of each trial. These steps were selected from the last 30 s of each 3-min period. For comparing energy cost in different slopes, vertical displacement, external work, and iEMG were normalized by a unit of running distance covering one meter.

## Statistics

Results are presented as the mean  $\pm$  standard deviation (SD). Normality of the variables was evaluated using the Kolmogorov-Smirnov test prior to any analysis. A one-way analysis of variance for repeated measurements was used to test the main effects of the slope. Homogeneity of variances was evaluated using Mauchly's test of sphericity. Lack of sphericity was treated by adjusting the degree of freedom before performing an F-test. If the main effects were significant, multiple analysis was conducted using the Tukey-Kramer method to test the difference between conditions. Pearson's correlation coefficient was used to determine the relationship between energy cost and biomechanical

variables. The statistical significance level was set at 5%.

## RESULTS

Table 1 demonstrates that the energy cost differed significantly between the slopes, being the greatest in incline slope and the lowest in decline slope. Positive and negative mechanical works, but not total mechanical work, also differed significantly between the slopes. The greatest positive and negative mechanical works were observed in incline and decline slopes, respectively.

\*\*\*\* Table 1 near here\*\*\*\*

Figure 1 shows iEMGs of the RF, VL, GM, BF, TA, GM, and SO muscles during the support phase of treadmill running in each slope. iEMGs of the RF, VL, GM, BF, GA, and SO muscles differed significantly between the slopes, being the greatest in incline slope and the lowest in decline slope ( $p < 0.01-0.001$ ). However, iEMG of the TA muscle did not differ between slopes.

\*\*\*\* Figure 1 near here \*\*\*\*

Figure 2 shows angles of the ankle, knee, and hip joints during the contact phase in each running slope. The ankle joint dorsiflexed during the first half of the contact and plantarflexed during the latter half. Plantarflexion angular displacement was significantly greater in incline slope and smaller in decline slope than in level (Table 2). However, maximal plantarflexion angle during the contact phase did not differ between slopes (Table 2). The knee joint flexed during the first half and, thereafter, extended during the latter half. The knee extension angular displacement did not differ between the slopes, but maximal knee extension angle at the toe-off was significantly greater in incline slope and smaller in decline slope than in level (Table 2). In addition, vertical displacement of CoM differed significantly between the slopes (Table 2) and was the greatest in incline slope and the lowest in decline slope.

\*\*\*\* Figure 2 and Table 2 near here \*\*\*\*

Figure 3 demonstrates significant correlations between energy cost and iEMGs of the GA ( $r = 0.43$ ,  $p < 0.01$ ) and SO muscles ( $r = 0.46$ ,  $p < 0.01$ ), plantarflexion angular velocity ( $r = 0.79$ ,  $p < 0.001$ ), and extension angular displacement of the knee joint ( $r = 0.58$ ,  $p < 0.001$ ). iEMG of the other studied muscles did not correlate significantly with energy cost.

\*\*\*\* Figure 3 near here \*\*\*\*

## DISCUSSION

The major findings of the present study can be summarized as follows: First, statistically significant differences were found in energy cost between the slopes. Second, no significant difference was observed in total mechanical work between the slopes. Third, vertical displacement was greater in incline slope than in decline slope. Fourth, the knee extension and plantarflexion increased significantly with the slopes. Fifth, iEMGs of the leg the extensor muscles were greater in incline slope than in decline slope. Six, energy cost correlated positively with iEMG of the GA muscle. These findings are well in line with our hypotheses. We hypothesized that changing slope affects changes in energy cost



in order to lift and maintain the speed. As expected, energy cost and vertical displacement were changed according to the slopes of the running surface.

Total mechanical work did not differ between the slopes, whereas energy cost was significantly greater in the incline slope than in the level and decline slopes. In decline slope, positive and negative mechanical works were the lowest and greatest, respectively. Previous studies (Aura and Komi, 1988; Kyröläinen and Komi, 2011) reported that negative mechanical work is economical as compared with positive mechanical work. This is one of the explanations for the difference in energy cost. Nevertheless, total mechanical work did not differ between slopes, but their mechanisms are difficult to explain. The present study refers to external work as mechanical work because measuring the ground reaction force during treadmill running is difficult. Winter (1979) demonstrated that external work is underestimated as it ignores movements of the limb segments about the CoM of the body. Furthermore, Holt, Roberts, & Askew (2014) have suggested that the reduction in mechanical work may not necessarily decrease energy cost in isolated muscles. If it could be adopted in this study, the difference in energy cost between the slopes is possible to explain, although the total mechanical work of the lower limb joints is the same. Sawicki, Lewis, & Ferris (2009) indicated that the lower limb joints

have different mechanical efficiencies, because of architectural differences in the muscle-tendon structure described as the proximo-distal gradient (Biewener and Daley, 2007). The muscle and/or tendon might store and recoil a part of mechanical energy as elastic energy, and therefore, efficiency could be possibly improved (Kyröläinen and Komi, 2011). The ankle joint is generally better suited for storing and returning elastic energy (Sawicki, et al., 2009), and has greater efficiency as compared with the knee and hip joints.

Vertical displacement of the CoM for 1 km differed significantly between slopes, being the lowest in decline slope and the greatest in incline slope. It suggests that the lower limb joints did greater mechanical work to lift the body in incline than in decline slope. A significant difference was found in angular displacement of extension of the knee joint and plantarflexion differed significantly between slopes, being lowest in decline slope and the greatest in incline slope. iEMG of the VL and SO muscles, which are mono-articular extensor, was greater in incline slope than in decline slope. Although the mechanical work done by the lower limb joints could not be calculated in the present study, these facts imply that the knee and ankle joints have done greater mechanical work in incline slope than in decline slope. van Ingen Schenau et al. (1995) reported that mono-articular muscles are considered as

generators. The greater muscular activity of the VL and SO muscles in incline slope implies that these muscles seemed to exert greater extension and plantarflexion torques. In addition, plantarflexion velocity correlated positively with energy cost. Plantarflexion velocity implies a shortening velocity of the plantarflexor muscles. Earlier studies have reported that greater shortening velocity of the muscle is correlated with lower efficiency during sledge jumping (Aura and Komi, 1988; Kyröläinen and Komi, 2011). These results were well in line with some previous studies (Aura and Komi, 1988; Kyröläinen and Komi, 2011). Furthermore, extension displacement of the knee joint was correlated positively with energy cost. Mechanical work done by the knee extensors is suggested to have an association with greater energy cost. Greater mechanical work of the knee joint was one of the reasons to increase energy cost (Seki, et al., 2019). Extension and plantarflexion of the lower limb joints were important movements to lift and accelerate the body. Maximal extension angle of the knee (at toe-off) was greater in incline than in decline slope. Although the maximal angle of the ankle and hip joints during the ground contact was also shown at toe-off, these angles did not differ between the slopes. Thus, extension of the knee joint might be a determinant factor of vertical displacement of the whole body and energy cost.

Interestingly, iEMG of the GA muscle had a relationship with energy cost. In addition, iEMG of the GA muscle was slightly greater in incline slope. Bi-articular muscles, such as GA, are assumed to transfer mechanical energy from the knee to ankle joint or vice versa (Swanson and Caldwell, 2000). The functional role of the GA muscle may also promote effective use of mechanical energy, and its length changes as a function of the ankle and knee angles. In incline slope, the ankle joint dorsiflexed more than other conditions at the middle of the contact phase, and the knee joint extended more in comparison with other conditions toward the end of contact phase. This suggests that shortening of the GA muscle was greatest in incline slope. If the GA muscle acted isometrically during knee extension, the GA muscle would couple knee extension to plantarflexion (van Ingen Schenau, 1989). In this situation, the knee extensor would extend the knee joint and plantarflex the ankle joint. This phenomenon is called the transfer of mechanical energy from the knee to the ankle (van Ingen Schenau, 1989) and likely to lead reduction in energy cost. However, if the GA muscle acted concentrically during knee extension, the ankle joint would be plantarflexed by shortening of the GA muscle. Furthermore, Holt, et al. (2014) reported that the energy cost of shortening is greater than that of isometric and stretching conditions. Although the ankle maximal plantarflexion angle did not differ

between the slopes, the knee maximal extension angle was greater in incline slope than decline slope.

This implies that coordination between the knee and ankle joints to transfer mechanical energy was disturbed in the incline slope. Landin, Thompson, & Reid (2015) have reported that the GA muscle acts mainly as a plantarflexor when the knee joint is in more extended position. Therefore, in incline slope, the GA muscle plantarflexed by itself rather than transferred mechanical energy, thus, increase in energy cost was observed.

In conclusion, the knee and ankle joint kinematics were associated with energy cost during running. In incline slope, the knee and ankle joints were more extended (plantarflexed) to lift the body. These movements may disturb the coordination between the ankle and knee joints. The GA muscle would perform greater mechanical work to plantarflex the ankle joint rather than transfer mechanical energy as well as greater mechanical work of mono-articular muscles. These muscular activities would increase energy cost. The findings could be applied in practice to improve running techniques, for which smaller vertical displacement of CoM may be associated with less extension of the lower limb joints and therefore, affecting coordination between the joints. For achieving proper movement patterns and running techniques, different drills should be performed. As a consequence, better running techniques

would lead to reduced energy cost, i.e. improved running economy, during running.

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#### DISCLOSURE OF INTEREST

The authors declare no conflicts of interest associated with this manuscript.

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## Figure captions

Figure 1. Mean ( $\pm$ SD) relative iEMG of the rectus femoris (RF), vastus lateralis (VL), gluteus maximus (GM), biceps femoris (BF), tibialis anterior (TA), gastrocnemius (GA), and soleus (SO) muscles during the contact phase in each slope. \*:  $p < 0.05$ , \*\*:  $p < 0.01$ , \*\*\*:  $p < 0.001$ .

Figure 2. Averaged patterns of the ankle, knee, and hip joint angles during the contact phase in each slope during running. Time was normalized by ground contact.

Figure 3. Relationships between energy cost and iEMG of the gastrocnemius and soleus muscles, maximal plantar flexion velocity of the ankle joint, and knee extension angular displacement.

Table 1 Mean ( $\pm$ SD) energy variables and results of ANOVA.

Variable	Slope			F	Multiple comparison
	-6%	0%	+6%		
$\dot{V}O_2$ [ml $\cdot$ kg $^{-1}\cdot$ min $^{-1}$ ]	34.1 $\pm$ 4.1	43.8 $\pm$ 3.9	53.9 $\pm$ 4.8	215.2***	-6% < 0% < +6%
Blood lactate level [mM $\cdot$ L $^{-1}$ ]	1.17 $\pm$ 0.43	1.80 $\pm$ 1.08	4.41 $\pm$ 2.23	32.96***	-6% < 0% < +6%
Energy cost [J $\cdot$ kg $^{-1}\cdot$ m $^{-1}$ ]	3.09 $\pm$ 0.36	4.15 $\pm$ 0.57	6.42 $\pm$ 0.81	338.62***	-6% < 0% < +6%
Positive mechanical work [J $\cdot$ kg $^{-1}\cdot$ m $^{-1}$ ]	0.95 $\pm$ 0.09	1.17 $\pm$ 0.09	1.50 $\pm$ 0.12	268.01***	-6% < 0% < +6%
Negative mechanical work [J $\cdot$ kg $^{-1}\cdot$ m $^{-1}$ ]	1.51 $\pm$ 0.12	1.22 $\pm$ 0.09	0.92 $\pm$ 0.10	316.70***	+6 < 0% < -6%
Total mechanical work [J $\cdot$ kg $^{-1}\cdot$ m $^{-1}$ ]	2.46 $\pm$ 0.16	2.39 $\pm$ 0.11	2.41 $\pm$ 0.18	1.45	

\*\*\*:  $p < 0.001$

Table 2. Mean ( $\pm$ SD) kinematic results measured in three different inclines and results of ANOVA (F).

Variable	Slope			F	Multiple comparison
	-6%	0%	+6%		
Step length [m]	1.29 $\pm$ 0.06	1.26 $\pm$ 0.05	1.24 $\pm$ 0.05	14.76***	+6% < 0% < -6%
Step frequency [Hz]	2.89 $\pm$ 0.14	2.95 $\pm$ 0.14	3.01 $\pm$ 0.14	20.35***	-6% < 0% < +6%
Contact time [s]	0.25 $\pm$ 0.01	0.26 $\pm$ 0.01	0.26 $\pm$ 0.01	23.52***	-6% < 0% < +6%
Vertical displacement to cover 1 m [m]	0.04 $\pm$ 0.01	0.07 $\pm$ 0.01	0.10 $\pm$ 0.01	2155.86***	-6% < 0% < +6%
Maximal plantarflexion angle during the contact phase [deg]	98 $\pm$ 7	98 $\pm$ 5	98 $\pm$ 4	0.12	
Maximal knee extension angle during the contact phase [deg]	148 $\pm$ 6	151 $\pm$ 4	154 $\pm$ 4	28.99***	-6% < 0% < +6%
Maximal hip extension angle during the contact phase [deg]	168 $\pm$ 13	167 $\pm$ 13	169 $\pm$ 15	0.56	
Plantarflexion displacement during the contact phase [deg]	22 $\pm$ 4	25 $\pm$ 4	27 $\pm$ 3	50.54***	-6% < 0% < +6%
Knee extension displacement during the contact phase [deg]	19 $\pm$ 3	21 $\pm$ 3	25 $\pm$ 3	81.38***	-6% < 0% < +6%
Hip extension displacement during the contact phase [deg]	21 $\pm$ 4	23 $\pm$ 4	27 $\pm$ 9	6.48*	-6% < +6%
Maximal plantarflexion velocity during the contact phase [deg·s <sup>-1</sup> ]	601 $\pm$ 27	682 $\pm$ 30	746 $\pm$ 41	98.27***	-6% < 0% < +6%
Maximal knee extension velocity during the contact phase [deg·s <sup>-1</sup> ]	260 $\pm$ 30	281 $\pm$ 29	325 $\pm$ 36	54.77***	-6% < 0% < +6%
Maximal hip extension velocity during the contact phase [deg·s <sup>-1</sup> ]	212 $\pm$ 31	219 $\pm$ 31	279 $\pm$ 149	2.46	

\*:  $p < 0.05$ , \*\*\*:  $p < 0.001$

Figure 1

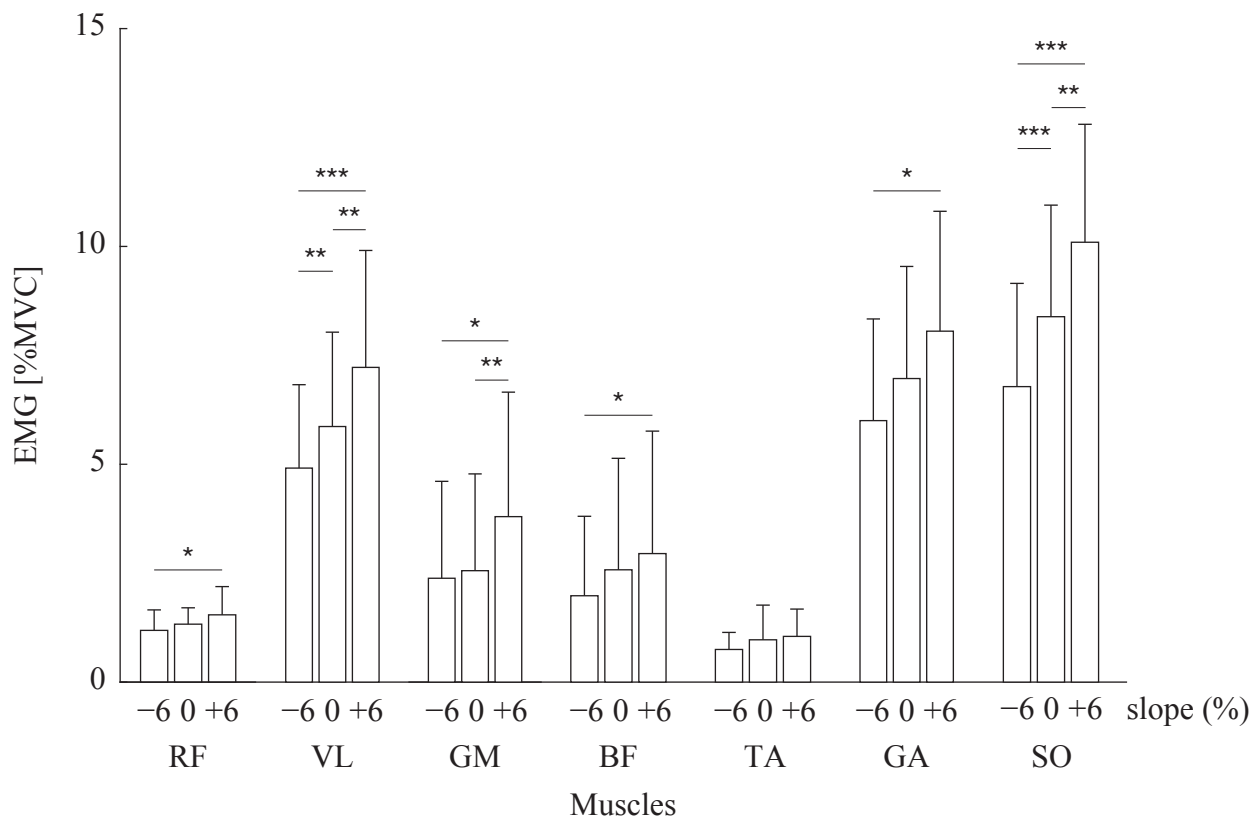


Figure 2

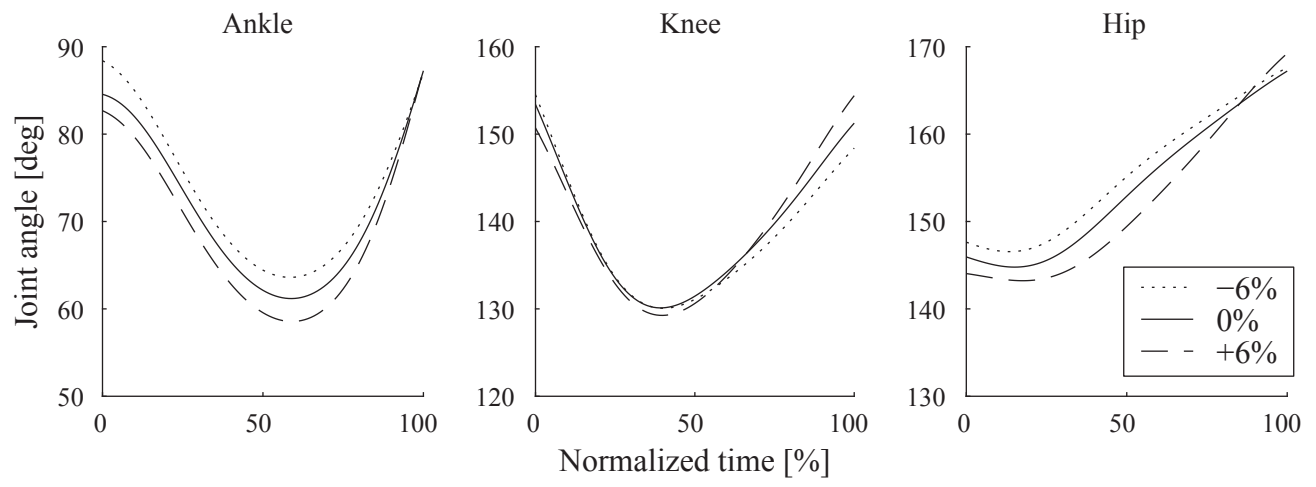


Figure 3

