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The effects of downhill slope on kinematics and kinetics of the lower extremity joints during running

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ABSTRACT

Background: The purpose of this study was to investigate how lower extremity kinematics and kinetics change when running downhill.
Methods: Fifteen male recreational runners ran on an instrumented treadmill with three different slope conditions [level (0°), moderate (−6°), and steep (−9°)] at a controlled speed of 3.2 m/s. Ten consecutive steps were selected for analysis for each of the slope conditions and the order of slope conditions was randomized. Synchonized motion analysis and force plate were used to determine joint kinematics and kinetics.
Results: Compared to level running, participants demonstrated significantly larger knee flexion but smaller ankle plantar-flexion and hip flexion during downhill running ($Ps < 0.05$). Significantly smaller peak propulsive ground reaction forces and posterior impulses were found during downhill running ($Ps < 0.05$). Furthermore, participants experienced significantly larger extension moment and negative joint power at the knee ($Ps < 0.05$) but smaller plantar-flexion moment and negative joint power at the ankle during downhill running ($Ps < 0.05$). Negative net joint work increased for all joints with increased declinations and the knee joint showed the greatest increase in negative net joint work amongst the three joints ($Ps < 0.05$).
Significance: These findings indicate that runners modify their running mechanics resulting in greater kinetic demand on the knee during downhill running. Differences in lower extremity injury mechanisms with different running slopes may be linked to the changes in loading at the knee but further investigation using clinical trials is needed to support the potential relationship.

1. Introduction

Distance runners on a trail match their running mechanics to accommodate various surface and slope conditions. Runners often experience uphill and downhill slopes during a run and thus biomechanical adaption is essential to control the body efficiently and safely [1]. Particularly, during downhill (DH) running, greater eccentric activation of knee and ankle extensor muscles is required compared with level running to decelerate and maintain balance on a declined surface [2]. It has been suggested that the increased eccentric muscle activities of the lower extremities during DH running would induce greater post-exercise muscle soreness and damage [2], which may cause future running related injuries.

DH running requires less metabolic energy [3,4] on a DH slope of up to 20% [5]. However, it has been suggested that the ability of the lower extremities to attenuate and dissipate the impacts from landing would decrease due to high stress at the joints resulting from greater eccentric muscle activities during DH running [6,7]. As a consequence of higher demand on the lower extremity during DH running, runners have to change their running strategies through the changes in stride length [8] and step frequency [4]. Furthermore, runners often switch their heel strike patterns from heel to midfoot/forefoot to reduce impact forces [9] with an increased range of knee flexion [7] on a DH slope.

Recently, studies have investigated an adaptation in kinetic variables of runners to understand potential injury mechanisms of DH running [1,10,11]. It has been suggested that accumulated high impact forces and joint loading is directly related to the early onset of fatigue [7], development of stress fractures [12] and patellofemoral pain in runners [13]. Increased peak impact and braking forces [10] but decreased propulsive forces [11] and energy generation [14] have been

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found during moderate and steep sloped DH running compared to a level condition. Furthermore, increased power absorption at the knee [15] and decreased power generation at the knee and hip were found during DH running compared with uphill and level running [1]. These findings may indicate that through more eccentric activities, the knee joint would play a major role in attenuating high stress and redistributing the loads within the lower extremity joints during the weight acceptance phase of DH running [3].

However, most previous studies primarily investigated kinematic changes and ground reaction force (GRF) components of runners during DH running compared with level running, while only few studies have considered kinetic variables, especially resultant joint moments, which provide an indication of the amount of loading experienced by the joint [1,15]. In addition, there was no consensus on whether the loading of the lower extremity joints changes in response to DH conditions, therefore the clinical relevance for injury prevention and rehabilitation were limited. Increased eccentric muscle activation of the lower extremity extensor muscles to decelerate the body during the stance phase of DH running is speculated to induce greater muscle damage and delayed onset of muscle soreness [2]. The quantification of joint loads during the eccentric phase in DH running would help in understanding the changes in kinetic demands on the lower extremity joints and their relationship to the potential risk of running related injuries [1].

Thus, the aim of this study was to investigate how kinematics and kinetics of the lower extremity change during moderate and steep DH running compared with level running. We hypothesized that runners would increase extension moments of the lower extremity joints and increase negative joint power and work in order to accommodate higher demands on the joints of the lower extremity during DH running compared with level running.

2. Methods

2.1. Participants

Fifteen male recreational runners (mean age: 25.6 ± 4.27 yrs; mean mass: 75.38 ± 5.02 kg; mean height: 177.0 ± 5.0 cm) with no history of lower extremity symptoms or injuries were recruited from local running clubs. All participants ran more than 10 km per week for the previous 6 months. The experimental procedure was approved by the university ethics committee (ID: 20161006). Written consent was obtained from each participant prior to the start of the study.

2.2. Procedure

Fifteen reflective markers were attached to the following anatomical positions: left and right posterior superior iliac spines, left and right anterior superior iliac spines, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, calcaneus, first and fifth metatarsal heads, styloid process, second metatarsal base, and calcaneus of the right leg. Two additional clusters of four markers were placed on the right thigh and leg. The participants were asked to stand in the middle of the treadmill with feet pointing anteriorly and shoulder width apart for a static trial. Each participant was given a five-minute adaptation period of running on an instrumented treadmill (Bertec Corp, Columbus, USA, sampling frequency = 1000 Hz) prior to each of the three slope conditions [Level (0°), Moderate (−6°), and Steep (−9°), [11,14] (Fig. 1). The proper calibration to reduce measurement errors and resulting accuracy of the treadmill were applied [16]. Participants ran at 3.2 m/s for 30-seconds and the last ten successful steps were selected for analysis [17]. Each participant wore his own running shoes for the test and the order of slope conditions was randomized (generated at www.random.org). The marker trajectories were collected using a 7-camera motion capture system (ProReflex MCU 240, Qualysis, Sweden) at a sampling frequency of 100 Hz. A spline interpolation was performed for minor missing data using three frames of

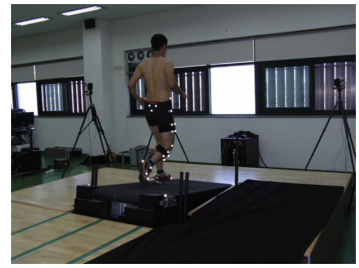

Fig. 1. Experimental setup (downhill running).

data before and after the missing data point. The kinematic and kinetic data were synchronized and filtered using low-pass Butterworth filters with the cutoff frequencies of 9 Hz and 100 Hz, respectively. The sampling and cutoff frequencies of kinematic and kinetic data were determined based on previous studies [18,19]. Both the kinematic and kinetic data were imported into software (Visual 3D, C-Motion, USA) to calculate three-dimensional joint angles, moments, powers and work for further statistical analyses. The contact phase of each running step was identified as the period from initial contact of right foot to take off, as determined by the instrumented treadmill. Joint forces and moments were calculated with inverse dynamics. Muscle power is defined as the product of net muscle moment and angular velocity about joint and net joint work was calculated by integrating the area under the joint power-time curve. The laboratory coordinate system was defined as follows: X-axis was medial–lateral, y-axis was anterior–posterior and z-axis was vertical. Joint angles were computed using Euler angles with an XYZ rotation sequence (X: flexion-extension, Y: abduction-adduction, Z: internal-external rotation).

2.3. Statistical analysis

A repeated measures ANOVA was performed to examine if there was any significant difference between slope conditions (level, moderate, and steep). Bonferroni Post-hoc comparisons were applied when appropriate. Statistical alpha levels were set to 0.05. The statistical analyses were computed using SPSS software (version 15, SPSS Inc., USA).

3. Results

Tables 1 and 2 show the changes in gait parameters and GRF components for each of the slope conditions. For temporal spatial parameters, no significant differences were found in most of the gait parameters except contact time among the three slope conditions (Table 1). There were differences between moderate and steep slope conditions in posterior impulse and maximum posterior force, indicating decreased propulsive GRF force components in the steeper conditions (Table 2).

Table 1
Comparisons of gait parameters between slope conditions.

Variables	Level (0°) M(SD)	Moderate (-6°) M(SD)	Steep (-9°) M(SD)	F(df)[p]	p-value		
					0° vs. −6°	0° vs. −9°	−6° vs. −9°
Contact time (s)	0.27 (0.03)	0.26 (0.03)	0.25 (0.03)	4.92(1, 14) [.04]	1.00	.13	**.02**
Swing time (s)	0.45 (0.02)	0.46 (0.03)	0.47 (0.04)	5.35(1, 14) [.04]	.20	.11	.40
Stride length (m)	2.28 (0.07)	2.32 (0.08)	2.32 (0.10)	2.84(1, 14) [.11]	.06	.34	1.00
Stride frequency (Hz)	1.41 (0.04)	1.38 (0.05)	1.38 (0.06)	2.68(1, 14) [.12]	.06	.37	1.00

Note: Bolded numbers indicate significant differences.

Table 2
Comparisons of kinetic parameters between slope conditions.

Variables	Level (0°) M(SD)	Moderate (−6°) M(SD	Steep (−9°) M(SD)	F(df)[p]	p-value 0° vs. −6°	0° vs. −9°	−6° vs. −9°
Anterior impulse (Ns)	−12.25 (2.38)	−13.37 (7.45)	−14.29 (9.63)	.75(1, 14) [.40]	1.00	1.00	.98
Posterior impulse (Ns)	13.16 (2.06)	11.62 (4.33)	9.99 (4.65)	7.00(1, 14) [.02]	.50	.06	< **.01**
Vertical impulse (Ns)	262.43 (26.45)	261.65 (26.57)	262.59 (26.61)	.09(1, 14) [.77]	1.00	1.00	1.00
Max. anterior force (N)	−153.98 (22.58)	−164.11 (55.10)	−169.51 (70.75)	.84(1, 14) [.38]	1.00	1.00	1.00
Max. posterior force (N)	150.51 (17.34)	132.83 (42.31)	117.40 (51.51)	5.79(1, 14) [.03]	.37	.10	**.01**
Max. vertical force (N)	1732.19 (125.05)	1717.38 (102.60)	1750.87 (125.29)	.21(1, 14) [.65]	1.00	1.00	.12

Note: Bolded numbers indicate significant differences. Max.: maximum.

Table 3
Comparisons of joint kinematic parameters between slope conditions.

Joint	Variables	Level (0°) M(SD)	Moderate (−6°) M(SD)	Steep (−9°) M(SD)	F(df)[p]	*p*-value 0° vs. −6°	0° vs. −9°	−6° vs. −9°
Ankle	Max. dorsi-flexion(°)	22.6 (3.4)	18.3 (3.9)	15.7 (3.7)	119.41(1, 14) [.00]	< **.01**	< **.01**	< **.01**
	Max.Plantar-flexion(°)	12.8 (5.4)	6.6 (5.4)	5.1 (5.6)	33.85(1, 14) [.00]	< **.01**	< **.01**	.48
	ROM (°)	35.4 (5.1)	24.8 (5.3)	20.9 (4.9)	203.92(1, 14) [.00]	< **.01**	< **.01**	< **.01**
	Dorsi-flexion at HC (°)	12.0 (4.4)	10.4 (5.0)	9.9 (5.6)	3.69(1, 14) [.08]	.31	.23	.98
Knee	Max. flexion (°)	39.5 (5.)	42.0 (5.1)	44.9 (6.6)	12.19(1, 14) [.00]	< **.01**	**.01**	.19
	Min. flexion (°)	14.2 (5.5)	12.4 (5.8)	12.3 (6.3)	6.13(1, 14) [.03]	.06	.08	1.00
	ROM (°)	25.3 (4.4)	29.7 (4.4)	32.6 (4.2)	24.31(1, 14) [.00]	< **.01**	< **.01**	**.04**
	Flexion at HC (°)	14.3 (5.5)	12.2 (5.8)	12.2 (6.3)	7.03(1, 14) [.02]	**.03**	.06	1.00
Hip	Max. flexion (°)	34.1 (8.9)	32.2 (7.7)	31.6 (8.2)	11.25(1, 14) [.01]	**.01**	**.02**	.97
	Min. flexion (°)	5.1 (8.2)	7.8 (8.3)	10.5 (9.7)	51.71(1, 14) [.00]	< **.01**	< **.01**	< **.01**
	ROM (°)	28.9 (4.4)	24.3 (4.9)	21.1 (5.9)	64.68(1, 13) [.00]	< **.01**	< **.01**	< **.01**
	Flexion at HC (°)	34.0 (8.9)	31.6 (8.3)	30.5 (8.3)	26.97(1, 13) [.00]	< **.01**	< **.01**	.22

Note: Bolded numbers indicate the differences. Max.: maximum, Min.: minimum, HC: heel contact.

Range of motion (ROM) was significantly decreased by about 30% on the moderate slope and 41% on the steep slope at the ankle ($P < 0.01$) and about 16% on the moderate slope and 27% on the steep slope at the hip ($P < 0.01$), as the slope declination increased. However, increased ROM of the knee, about 17% on the moderate slope and 29% on the steep slope ($P < 0.01$), and greater maximum knee flexion, about 6% on the moderate slope and 14% on a steep slope ($P < 0.01$), were found compared with running on the level condition (Table 3).

For joint moments, maximum knee extension moment was increased by about 7% on the moderate slope and 23% on the steep slope ($P < 0.05$), but maxium ankle plantar-flexion moment was decreased by about 16% on the moderate slope and 28% on the steep slope ($P < 0.01$) compared with running on the level condition (Fig. 2). For joint energetics, decreased maximum positive power at the ankle, knee and hip joints were found as the slope declination increased. On the other hand, participants exhibited smaller maximum negative power at the ankle, about 43% on the moderate slope and 60% on the steep slope ($P < 0.01$), but larger maximum negative power at the knee, about 52% on the moderate slope and 104% on the steep slope ($P < 0.01$), with an increase in slope declination (Fig. 3b). In addition, net joint work, which was calculated by integrating the area under the joint power-time curve (Fig. 3a), decreased for all joints with increased declinations ($P < 0.01$). Finally, greater increment of negative net joint work at the knee was observed compared with ankle and hip joints ($P < 0.01$, Fig. 3c).

4. Discussion

4.1. Gait parameters and GRF components

This study examined the adaptations in kinematics and kinetics of running across three different slope conditions (level, moderate and steep slopes). The current results indicate that there were no differences in gait parameters such as swing time, stride length and stride frequency across slope conditions, which is in line with some previous studies on DH running at a similar speed [1,10]. However, a slight decrease in contact time during DH running may result from the use of

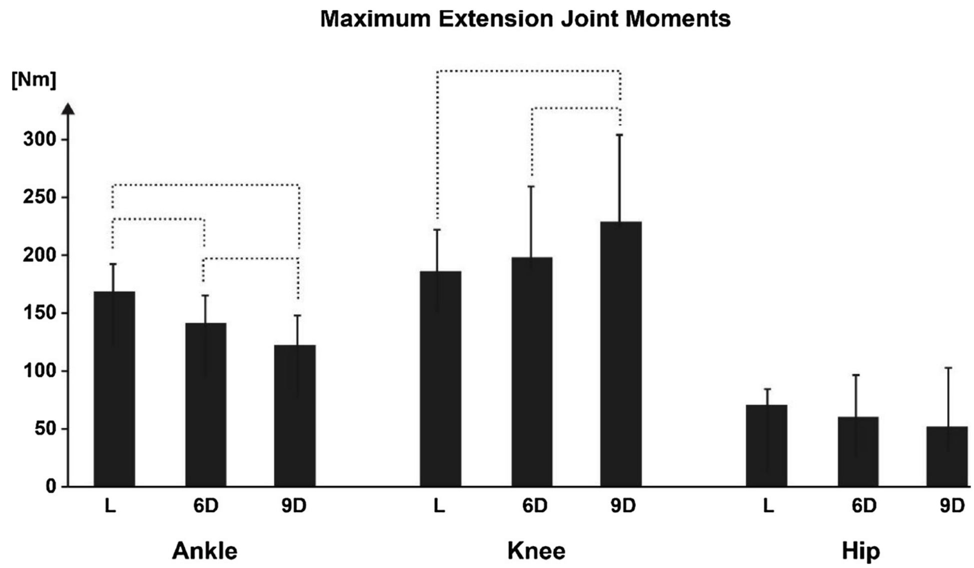

Fig. 2. Comparisons of maxium extension joint moments between slope conditions. The dotted line indicates significant difference between slope conditions.

potential energy from one step to the next step, while uphill running may require increased contact time to generate more muscle work in order to lift the center of mass to a higher position for the next step of the running cycle [20].

While a decreased propulsive component of GRF was observed during DH running, no significant changes in impact and braking components of GRF were found between level and DH running conditions. These findings contradict the results of previous studies on downhill running [1,10], which showed a significantly increased impact peak and braking force of GRF during DH running. It is expected that the type of foot strike pattern (e.g. heel, midfoot or forefoot contact at initial contact during running) of the runner may influence ground reaction forces during DH running [9,11,21]. For example, most experienced runners who are rearfoot strikers on a level surface might switch to a non-rearfoot strike (i.e. forefoot or midfoot strikes) in running DH [4]. When compared to a rearfoot strike landing, a non-rearfoot strike landing would reduce leg stiffness and impact loading [22]. We did not find consistent patterns of impact peaks in runners, which may be explained by the differences in runners' foot strike patterns depending on their skill levels or running styles [9,11]. Thus, there was a minimum effect of declination of running slope on gait parameters while significant reduction of propulsive GRF during DH running was observed.

4.2. Kinematic and kinetic changes

Runners decreased their ankle and hip motion but increased knee range of motion with an increased DH slope. This finding is partially supported by Dick and Cavanagh [23], who showed that runners displayed an increased range of motion at the knee in DH running compared with level running. Additionally, more extended knee and hip joint angles at intial ground contact were observed (i.e. stiffer leg or less cushioned knee [7]), which may place the body in a more vulnerable position during impact when DH running compared with running on a level surface [6]. However, the total range of knee flexion was signficanltly increased by about 29%, from 25.3° on a level surface to 32.6° in DH running (9°), which contributes to increased joint work and joint power because of greater knee flexion velocity. This increase in the total range of knee flexion may play a major role in response to the higher kinetic demands on the lower legs required during DH running.

During DH running, runners experienced greater peak knee extension moments while smaller peak ankle plantarflexion moments across an increase in DH inclination. To the best of our knowledge, this study is the first research to show significantly greater joint moments at the knee during DH running compared with level running, while other studies did not show differences due to slightly smaller downward slope conditions in the studies [1] or they have not investigated joint moments [3,4,6,7,15,21]. Previous studies have suggested that increasing resultant joint moments is a risk factor for overuse joint injuries in running and other sport activities [13,24]. Our results show that running downhill at 9° increased knee extension moments by about 19%. This is a major finding of the study showing a significant increase in knee extension moments during DH running compared with level running and this trend increases as DH slope increases. Therefore, our hypothesis was partially supported by the results, showing an increased extension moment with increased negative joint power and work during DH running in the knee but not for the ankle and hip joints. Repetitive high levels of knee extension moments may stress the extensor muscles of the knee, causing over-load and future joint injuries in the lower extremity during DH running.

4.3. Biomechanical and clinical significance of DH running

Increased eccentric activation of the knee during DH running is known to increase muscle damage and soreness, followed by fibre swelling and pain after excercise [2,23]. Negative external work is required to decelerate the body during DH running. A study have found increased negative joint power at the knee and hip during DH running, but among the lower extremity joints the contribution of the hip in absorbing the loads is relatively small (i.e. 10%) compared to the knee [1]. Furthermore, the contribution of the ankle joint to generate force also decreased with increased declinations (Fig. 3), which is indicated by the decreased positive ankle joint work. This implies that as downward running slopes increase, the ankle may lose its function as a force generator and absorber, transferring this function to the knee. This phenomenon is persuasive since the anatomy of the ankle joint and the surrounding muscles were not well designated to provide negative and positive work on sloped surfaces [25]. Therefore, during DH running, runners modify their running strategy with an increased range of knee flexion and greater negative knee joint power and work.

The findings of this study may provide additional insight for the deveopment of rehabilitation or training programmes for trail runners. It is speculated that increased knee joint loads and high eccentric activity during DH running would induce fatigue and over-load over a long period of excercise [7]. Mizrahi et al. investigated the relationship between impact and the adaptation in running kinematics due to

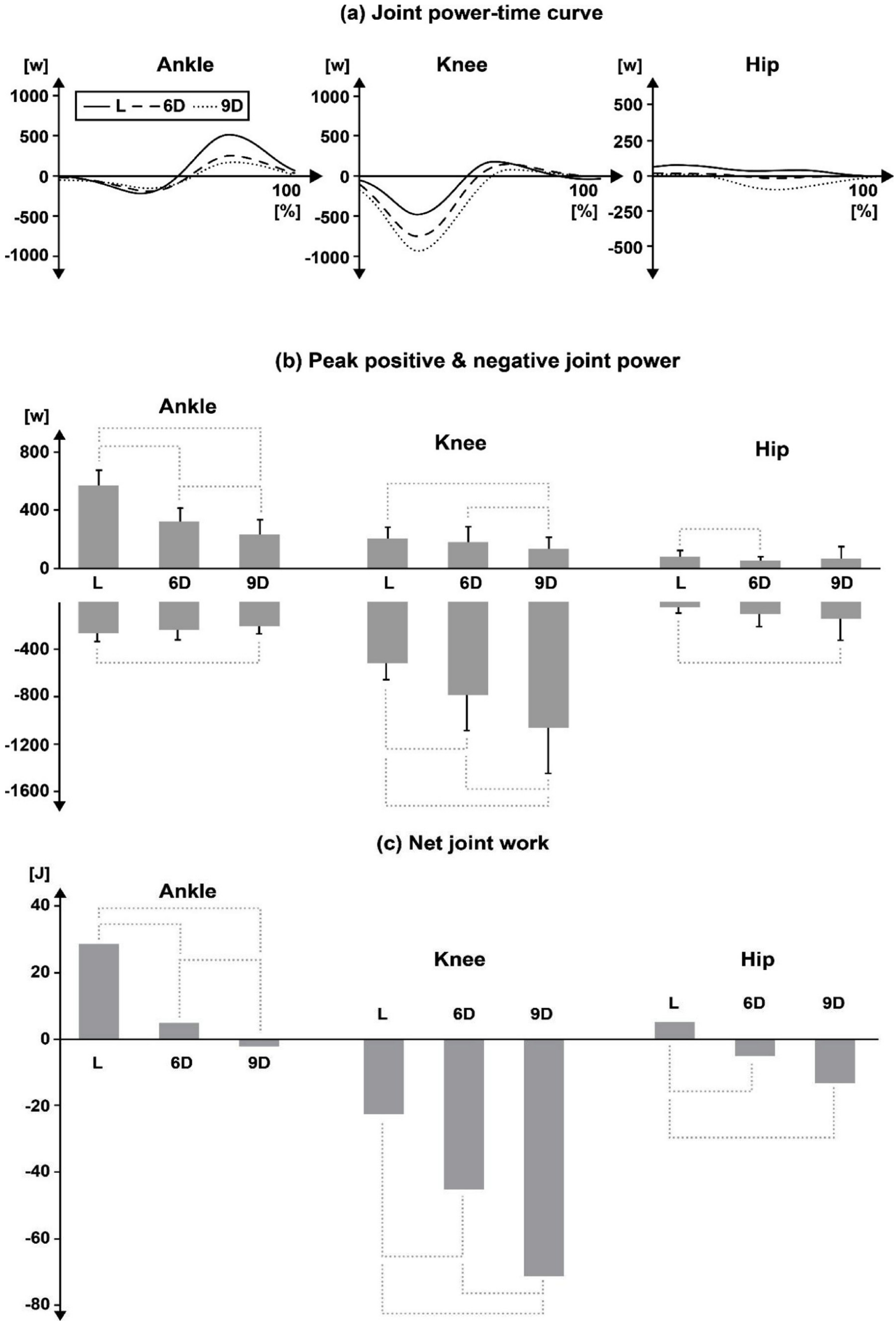

Fig. 3. Comparisons of (a) joint power-time curve, (b) peak positive and negative joint power and (c) net joint work between different slope conditions.The dotted line indicates significant difference between slope conditions.

fatigue using a 30 min DH running protocol at −4°. Their findings suggest that increased knee flexion and ankle dorsiflexion are requried for shock attenuation, especially after 15 min when fatigue gradually begins. However, the mechanism of how much joint loads and eccentric activity are directly related to a potential risk of injury have not been well established. It is known that a certain level of eccentric muscle activity of the lower extremity joints is necessary to maintain the stability of the body against gravity during walking and running. Furthermore, previous studies have suggested that a moderate level of eccentric resistance exercises for a patient population with anterior cruciate ligament reconstruction [26] and older adult population for fall prevention [27] has been beneficial. Therefore, a key concept when applying an eccentric exercise program for the purpose of training and rehabilitation is how to optimize both the intensity and duration of the exercise program.

4.4. Limitations

There are a few limitations in the methods of this study that need to be addressed. First, DH running was conducted on an instrumented treadmill under controlled speed and slope, which might elicit slightly different running mechanics from over-ground running. If less controlled, runners may adapt their running patterns to different sloped surfaces to reduce the risk of injury and improve efficiency of movement. However, our main interest was to determine how runners change kinetic running strategies under more controlled DH conditions (i.e. speed and slope) and it is difficult to collect force data on an overground sloped surface [11]. In addition, previous studies also suggest that most variables in running biomechanics on a treadmill are comparable to over ground running [28,29]. Second, the current study did not include lower extremity electromyography (EMG) measurements. The information from EMG measurements, especially the extensors such as quadriceps, may increase the understanding of the relationship between force generation and muscle recruitment patterns during eccentric contractions while DH running [30]. Lastly, we were not able to find consistent impact peaks in vertical GRF during DH running. This may suggest that differences in running experience and styles, such as foot strike patterns, of the participants played a role in different impact peak responses during DH running [9,11]. Studying different running speeds (preferred vs non-preferred) and running mileage, interplay with foot strike pattern and foot morphology, could provide a better understanding of DH running adaptations in future studies.

5. Conclusion

Compared to level running, runners alter running strategy by regulating predominantly the function of the knee to absorb energy and loads when running DH on a steep slope. The increment in knee extension moment (19%) may not be enough to reach an injury threshold during a short period of running. However, the accumulated joint load at the knee with increased muscle fatigue from a long period of exercise may compromise the functions of the joint, which increases the risk of injury. Therefore, proper adjustment of the intensity and duration of the eccentric exercises, taking into consideration of the experience and age of the runners, would be beneficial for rehabilitation after injury and for improvement of running performance on a DH slope. This warrants further research regarding training methods and interventions to reduce the risk of potential lower extremity injuries during DH running.

Conflict of interest disclosure

None.

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