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ORIGINAL

The Effect of Foot Landing Type on Lower-extremity Kinematics, Kinetics, and Energy Absorption during Single-leg Landing

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Objective: The aim of this study was to examine the effect of foot landing type (forefoot vs. rearfoot landing) on kinematics, kinetics, and energy absorption of hip, knee, and ankle joints.

Method: Twenty-five healthy men performed single-leg landings with two different foot landing types: forefoot and rearfoot landing. A motion-capture system equipped with eight infrared cameras and a synchronized force plate embedded in the floor was used. Three-dimensional kinematic and kinetic parameters were compared using paired two-tailed Student's *t*-tests at a significance level of .05.

Results: On initial contact, a greater knee flexion angle was shown during rearfoot landing ($p < .001$), but the lower knee flexion angle was found at peak vertical ground reaction force (GRF) ($p < .001$). On initial contact, ankles showed plantarflexion, inversion, and external rotation during forefoot landing, while dorsiflexion, eversion, and internal rotation were shown during rearfoot landing ($p < .001$, all). At peak vertical GRF, the knee extension moment and ankle plantarflexion moment were lower in rearfoot landing than in forefoot landing ($p = .003$ and $p < .001$, respectively). From initial contact to peak vertical GRF, the negative work of the hip, knee, and ankle joint was significantly reduced during rearfoot landing ($p < .001$, all). The contribution to the total work of the ankle joint was the greatest during forefoot landing, whereas the contribution to the total work of the hip joint was the greatest during rearfoot landing.

Conclusion: These results suggest that the energy absorption strategy was changed during rearfoot landing compared with forefoot landing according to lower-extremity joint kinematics and kinetics.

Keywords: Anterior cruciate ligament, Forefoot, Rearfoot, Injury, Biomechanics

INTRODUCTION

Sport-related lower-extremity injuries, such as anterior cruciate ligament (ACL) tears or ankle sprains, often occur during landing (Beynnon, Vacek, Murphy, Alosa, & Paller, 2005). Landing is a common and fundamental action in various sports, including basketball and volleyball. The impact generated on landing is transferred to the knee and hip joints, and if this impact is not effectively controlled and absorbed, it can cause acute or chronic musculoskeletal injury to the lower extremity or spine (Dufek & Bates, 1990). Thus, movement strategies are required during landing that can efficiently absorb energy and minimize the load on joints, and there have been a number of studies on the kinematics, kinetics, and mechanisms of shock absorption in the joints of the lower extremity during landing for the sake of injury prevention (Cho, Kim, & Koe, 2012; Cortes et al., 2007; Decker, Torry, Wyland, Sterett, & Steadman, 2003; Norcross et al., 2013; Yeow, Lee, & Goh, 2010).

Shock absorption during landing is mostly mediated by the joints of the lower extremity (Coventry, O'Connor, Hart, Earl, & Ebersole, 2006). Joint work is calculated by integrating the product of the joint moment and angular velocity; here, the negative work represents the energy absorbed by eccentric muscle contraction (Winter, 1990). Thus, the negative work at the joint is an important measure of energy absorption. In previous studies, increased impact was associated with an increase in joint work to absorb this impact (Zhang, Bates, & Dufek, 2000). It has been reported that impact-related injury can be prevented by effective impact absorption and that this can only be achieved by increasing flexion of the joints or by adjusting the distribution of the total work upon landing across individual joints (Coventry et al., 2006; Yeow et al., 2010).

Until now, landing has been researched in terms of various factors, such as landing height, fatigue, with or without shoes, sex, and landing type (Conventry et al., 2006; Decker et al., 2003; Devita & Skelly, 1992; Lee, Lee, & Choi, 2001; Shultz, Schmitz, Tritsch, & Montgomery, 2012; Zhang et al., 2000). Here, landing type is divided into forefoot landing, in which the toes touch the ground first, and rearfoot landing, in which the heel touches the ground first. A previous study that analyzed video of ACL injuries during athletic competition found that athletes landed rearfoot at the time of injuries, and reported that rearfoot landing

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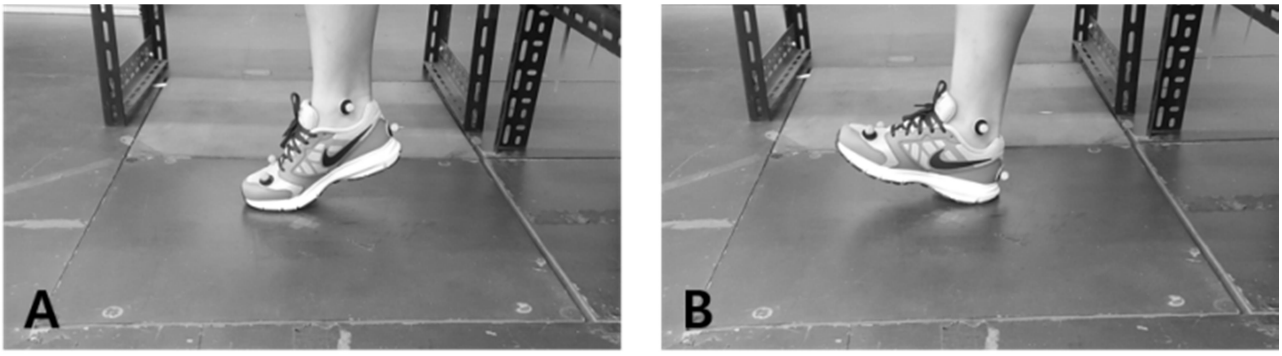

Figure 1. Lateral view of foot position at initial contact of the forefoot landing (A) and rearfoot landing (B).

could increase the risk of injury because the triceps surae, which is an ankle plantarflexor, does not absorb sufficient energy (Boden, Torg, Knowles, & Heweett, 2009). Other research comparing the biomechanical characteristics of the lower extremity during forefoot and rearfoot landing found that the joint kinematics and the contribution of each joint to the total work varied according to the landing type in double-leg landing and that the ground reaction force (GRF) was larger in rearfoot landing than forefoot landing (Cortes et al., 2007; Kovacs et al., 1997). However, over 70% of ACL injuries occur during single-leg landing or change-of-direction movements (Boden et al., 2009; Nagano, Ida, Akai & Fukubayashi, 2009). Even though single-leg landing presents a higher risk of injury than double-leg landing (Nagano et al., 2009), there have not yet been any studies on the effects of landing type on biomechanical characteristics during single-leg landing. Hence, the present study compared lower-extremity kinematics, kinetics, and joint work during single-leg landing across different landing types, with the aim of examining differences in the biomechanical characteristics of the lower-extremity joints between forefoot and rearfoot landing. To this end, we proposed and tested the following hypotheses: that the lower extremity joint flexion angles and work would differ between the two landing types on initial contact and at peak vertical GRF and that there would be a difference in the relative contribution of each joint to the total work.

METHODS

1. Participants

The subjects were 25 healthy Korean male adults (age: 22.6 ± 2.6 years; height: 175.9 ± 2.6 cm; weight: 74.1 ± 9.0 kg; body mass index: 24.2 ± 2.8 kg/m^2). We excluded subjects who had signs or a history of musculoskeletal disease requiring surgery as well as subjects who had experienced an injury within the last 6 months that restricted physical activity for at least 2 weeks. This experiment was assessed beforehand by the Sogang University Institutional Review Board, and all subjects gave their consent to participate.

2. Measurements

Subjects were instructed to stand on a 30-cm-high platform placed in front of the GRF measuring device with at least half of each foot feet over the edge and then to step off with their dominant foot (defined as the preferred foot for kicking a ball), without any horizontal or vertical leap, and fall directly onto the middle of the GRF measuring device, maintaining their position and balance after falling for 1~2 sec. Subjects were instructed, in turn, to perform forefoot landing, in which the toes touch the ground before the heel, and rearfoot landing, in which the heel touches the ground before the toes (Figure 1). To minimize error in the lower-extremity joint movements due to movement of the arms, both arms were fixed to the upper body during the experiment. To eliminate the effects of footwear, all subjects wore the same running shoes (Nike Downshifter 6; NIKE Inc., Beaverton, OR, USA). Each type of landing was repeated until the subject successfully completed the movement twice. A successful completion was defined as a trial where the appropriate part of the foot touched the ground first at initial contact, and the subject maintained proper balance and posture after landing, such that their foot did not move on the ground.

To measure lower-extremity joint movements by landing type, a three-dimensional motion analysis system consisting of 10 infrared cameras (Eagle; Motion Analysis Inc., Santa Rosa, CA, USA) was used to measure movements during single-leg landing with a sampling rate of 400 Hz. Reflective markers with a diameter of 12.5 mm were attached to the following anatomical landmarks: the bilateral anterior superior iliac spines, the sacrum, the greater trochanter, the midpoint of the femur, the medial and lateral epicondyles of the femur, the medial and lateral plateau of the tibia, the midpoint of the tibia, the medial and lateral malleoli, the calcaneus, and the first and fifth metatarsal heads (Jeong & Shin, 2016). After inserting a GRF measuring device (9260AA6; Kistler, Winterthur, Switzerland) linked to the motion analysis system into the floor, measurements were taken at a sampling rate of 1,200 Hz.

3. Data processing

According to previous studies, the greatest strain on the ACL occurs at the moment of peak vertical GRF, and the load on the ACL can be

predicted based on its correlation with the peak anterior force on the tibia (Cerulli, Benoit, Lamontagne, Caraffa & Liti, 2003; Yu, Lin, & Garrett, 2006). Therefore, the data analysis interval was defined as the time from when the foot first contacted the ground with a vertical GRF of at least 20 N until the time of peak vertical GRF. To remove noise, the collected kinematic and kinetic data were passed through a fourth-order Butterworth low-pass filter with a cutoff frequency of 10 Hz. To calculate joint movement, the anatomical coordinates of the body segments, excluding the feet, were defined according to the method proposed by Dyrby and Andriacchi (2004). The coordinates for the feet were defined by the following new method proposed by Hong and Shin (2015). The superoinferior axis was defined as the line passing through the calcaneus, perpendicular to the plane containing the calcaneus and the first and fifth metatarsals. The mediolateral axis was generated by taking the cross-product of the superoinferior axis and the vector from the calcaneus to the midpoint between the medial and lateral malleoli. The anteroposterior axis was generated by taking the cross-product of the superoinferior axis and the mediolateral axis. The coordinates of GRF were oriented such that the superior direction was +Z, the anterior direction was +Y, and +X was the cross-product of

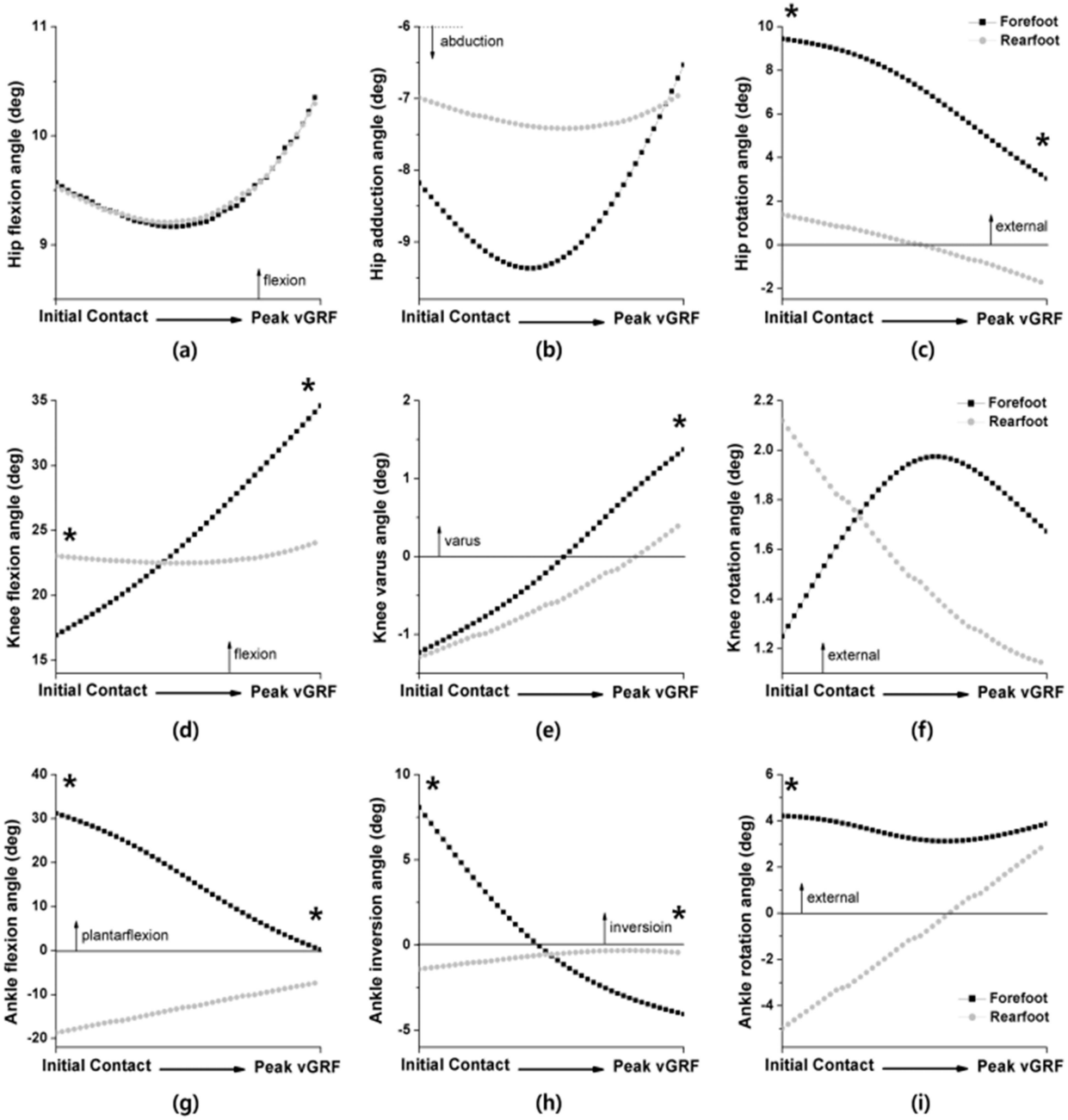

Figure 2. The ensemble mean kinematics of hip, knee, and ankle joints from initial contact to peak vertical ground reaction force (vGRF) for forefoot and rearfoot landing during single-leg landing. Positive angles are indicated. The black and gray lines represent the data during forefoot landing and rearfoot landing, respectively.

Table 1. Comparison of three-dimensional kinetics of hip, knee, and ankle joints at peak vertical GRF and ROM in the sagittal plane from initial contact to peak vertical GRF between forefoot and rearfoot landing during single-leg landing

		Forefoot	Rearfoot	*p*-value
Peak vertical GRF (N/BW)		3.77 ± 0.82	4.56 ± 0.72	**<.001**
Hip	Extension (+)/flexion (−) moment	15.01 ± 5.40	17.27 ± 12.72	.280
	Adduction (+)/abduction (−) moment	-15.44 ± 6.44	-13.46 ± 10.19	.148
	External (+)/internal (−) rotation moment	-1.39 ± 1.61	-1.15 ± 1.80	.435
Knee	Flexion (+)/extension (−) moment	-11.48 ± 3.46	-8.29 ± 5.99	**.003**
	Varus (+)/valgus (−) moment	-3.44 ± 3.38	-2.88 ± 4.72	.422
	External (+)/internal (−) rotation moment	1.31 ± 0.79	0.50 ± 0.60	**<.001**
Ankle	Plantar (+)/dorsi (−) flexion moment	12.93 ± 2.99	2.47 ± 2.70	**<.001**
	Inversion (+)/eversion (−) moment	0.27 ± 2.46	-0.30 ± 2.09	**.042**
	External (+)/internal (−) rotation moment	0.61 ± 0.59	0.50 ± 0.78	.597
ROM from initial contact to peak vertical GRF in sagittal plane				
Hip ROM (deg)		1.1 ± 0.9	0.9 ± 0.7	.271
Knee ROM (deg)		13.0 ± 3.8	0.8 ± 0.9	**<.001**
Ankle ROM (deg)		28.3 ± 10.0	8.7 ± 3.0	**<.001**

The values are presented as mean ±standard deviation (mean ± SD). Moment values were normalized for each subject relative to the product of their height and body weight: %[NM/(BW × height)].
Abbreviations: GRF, ground reaction force; BW, body weight; ROM, range of motion.

the +Y and +Z vectors. The local coordinate system was such that, for the right leg, the inferior direction was +Z, the anterior direction was +Y, and the medial direction was +X. Joint moment was calculated using the Newton-Euler dynamic equations (Winter, Patla, Frank, & Walt, 1990). The work at each joint was obtained by integrating the power, which was calculated as the product of the moment and the angular velocity (Yeow et al., 2010). Positive work and negative work indicate, respectively, the energy production from concentric muscle contraction and the energy absorption by eccentric muscle contraction (Winter, 1990). Thus, the work was calculated by integrating only the interval in which negative power was observed. Joint moment and work were normalized for each subject relative to the product of their height and weight; GRF data were normalized relative to the subject's weight (Favre, Hayoz, Erhart-Hledik, & Andriacchi, 2012).

4. Statistical analysis

To identify differences in the characteristics of forefoot landing and rearfoot landing, MATLAB (Version R2014b; The Mathworks, Natick, MA, USA) was used to compare the means for the two sets of data using a paired, two-tailed *t*-test (significance level = .05). Lower-extremity joint kinematics and kinetics were compared between the time of initial contact and the time of peak vertical GRF.

RESULTS

1. Kinematics/kinetics at initial contact and peak vertical GRF

In forefoot landing compared with rearfoot landing, the hip joint showed a significantly larger external rotation angle at both the initial contact and peak vertical GRF, but there was no significant difference in flexion or adduction angles (both $p < .001$, Figure 2a-2c). At initial contact, forefoot landing showed approximately 20% less knee flexion than rearfoot landing (forefoot landing: 18.2° ± 3.0°, rearfoot landing: 22.7° ± 3.3°, $p < .001$, Figure 2d), but this was reversed at peak vertical GRF, where forefoot landing showed 33% more knee flexion than rearfoot landing (forefoot landing: 31.2° ± 4.2°, rearfoot landing: 23.5° ± 3.5°, $p < .001$, Figure 2d). In the ankle joint, forefoot landing showed plantarflexion (29.0° ± 7.7°), inversion (5.8° ± 4.6°), and external rotation (4.1° ± 5.5°) at first contact, whereas rearfoot landing showed dorsiflexion (16.3° ± 6.5°), eversion (1.0° ± 3.6°), and internal rotation (3.4° ± 6.3°) (all $p < .001$, Figure 2g-2i). The peak vertical GRF was approximately 17% larger for forefoot landing than for rearfoot landing ($p < .001$, Table 1). At peak vertical GRF, forefoot landing showed a larger extension moment in the knee and ankle joints than rearfoot landing ($p = .003$ and $p < .001$, respectively, Table 1). In addition, from initial contact to peak vertical GRF, forefoot landing showed a larger range of motion (ROM) of the knee and ankle joints than rearfoot landing (both $p < .001$, Table 1), but there was no significant difference in the

ROM of the hip joint (p = .271, Table 1).

2. Negative work during initial impact phase

The joint work from initial contact to peak vertical GRF was significantly larger for forefoot landing than for rearfoot landing for the hip, knee, and ankle joints (p < .001, Table 2). In particular, considering the proportion of total work, the ankle joint performed the most work in forefoot landing (hip: 16.9%; knee: 22.5%; ankle: 60.6%), whereas the ankle performed the least work and the hip performed the most work in rearfoot landing (hip: 54.5%; knee: 36.4%; ankle: 9.1%, Figure 3).

Table 2. Comparisons of negative work of hip, knee, and ankle joints from initial contact to peak vertical GRF

	Forefoot	Rearfoot	p-value
Hip negative work	-1.20 ± 0.41	-0.63 ± 0.31	**<.001**
Knee negative work	-1.59 ± 0.58	-0.43 ± 0.29	**<.001**
Ankle negative work	-4.25 ± 1.86	-0.13 ± 0.13	**<.001**

The values are mean ± standard deviation. Moment values were normalized for each subject relative to the product of their height and body weight: %[J/(BW × height)].
Abbreviation: GRF, ground reaction force.

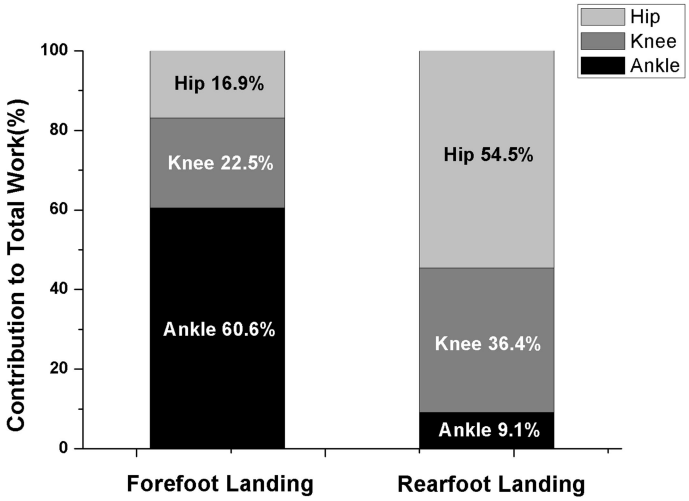

Figure 3. Contribution to total negative work of hip, knee, and ankle joints from initial contact to peak vertical ground reaction force.

DISCUSSION

This study aimed to investigate differences in the biomechanical characteristics of the lower-extremity joints between forefoot and rearfoot landing; hence, we compared the lower-extremity kinematics, kinetics, and joint work in single-leg landing for these two landing types. We found that rearfoot landing was associated with stiff movement, with more restricted ROM of the knee and ankle joints. Moreover, the total work from initial contact to peak vertical GRF was higher for rearfoot landing than for forefoot landing, and the relative contribution of each joint to the total work also differed between landing types.

We found that forefoot landing showed significantly greater joint flexion than rearfoot landing from initial contact to peak vertical GRF, which is the interval during which the impact is absorbed during single-leg landing. In forefoot landing, knee flexion from initial contact to peak vertical GRF was 13° and ankle dorsiflexion was 28.3°, whereas for rearfoot landing, knee flexion was 0.8° and ankle plantarflexion was 8.7°, which was consistent with our hypothesis that the two landing types would show differences in flexion of the lower-extremity joints. To effectively absorb the impact during landing, flexion needs to increase in each joint, and this has been reported to prevent injury from the impact being transferred to the body (Devita & Skelly, 1992; Yeow et al., 2010; Zhang et al., 2000). In other words, since the results of our study show greater knee and ankle flexion for forefoot landing, the knee extensors and ankle plantarflexors are able to continue absorbing energy, reducing the load acting on the joints; conversely, for rearfoot landing, more restricted flexion of the knee and ankle joints impairs the ability to properly absorb the impact of the GRF, which is larger than in forefoot landing, resulting in a higher risk of knee or ankle injury. Furthermore, previous studies have reported that a hard landing, in which the peak knee flexion is small, shows greater activation of the vastus lateralis than a soft landing, in which the peak knee flexion is larger (Lee et al., 2001; Pollard, Sigward, & Powers, 2010). Thus, in rearfoot landing where there is greater knee extension than in forefoot landing, the large quadriceps femoris muscles are expected to be more active, pulling the tibia anteriorly and causing significant forces and strain on the ACL, which in turn could increase the risk of ACL injury (Li et al., 1999; Pandy & Shelburne, 1997).

Our study revealed that rearfoot landing, compared with forefoot landing, was associated with significantly smaller extension moment and work for both the ankle and the knee joint, which was consistent with our hypothesis that there would be a difference between the two landing types in lower-extremity joint work. During landing, the knee joint shows flexion and the ankle joint shows dorsiflexion, which results in a rapid downward shift in the body's center of weight; in order to prevent the center of weight descending too rapidly, the extensor muscles at each joint act to reduce the velocity by eccentric contraction (Devita & Skelly, 1992). Decreased extension moment and joint work during rearfoot landing indicates a decrease in the capacity of the knee and ankle extensors (respectively, quadriceps femoris and triceps surae) to absorb energy. A large vertical GRF accompanied by a reduced capacity for eccentric contraction can increase the risk of overuse injury in the lower extremity (Bus, 2003; Hargrave, Carcia, Gansneder & Shultz, 2003), and therefore, persistent rearfoot landing is expected to increase the risk of future lower-extremity joint injury.

To absorb the impact generated during landing, all the joints in the lower extremity need to act simultaneously. However, the proportion of the impact absorbed by each joint differs according to landing type. Previous studies have reported that, during a hard landing, where knee ROM is restricted, the ankle joint makes a large contribution to energy

absorption and the hip extensors make almost no contribution to energy absorption and that this was associated with hip flexion (DeVita & Skelly, 1992; Zhang et al., 2000). In the present study, in rearfoot landing, where the angle of the knee is restricted, the joint that contributed most to energy absorption was the hip joint (54.5%), and the joint that contributed less was the ankle joint (9.1%). Likewise, a study by Kovacs et al. (1998) also reported that the ankle joint made the smallest contribution in rearfoot landing. This is thought to be because the ankle joint is in dorsiflexion during rearfoot landing, which makes it unable to achieve sufficient plantarflexion for energy absorption, and therefore, energy absorption occurs at the hip joint instead of the ankle joint.

Our study found that, unlike rearfoot landing, forefoot landing showed plantarflexion and inversion at first contact. Wright, Neptune, van den Bogert, & Nigg (2000) reported that increased plantarflexion at first contact increases the risk of ankle sprain, and ankle inversion has also been reported to result in acute lateral ankle sprains (Donovan & Feger, 2017). In rearfoot landing, the ankle is stiff, and therefore, even though it contacts the ground in a position of dorsiflexion and eversion, the ankle is unable to efficiently absorb energy, resulting in a greater impact on the ankle than in forefoot landing. A large impact on the ankle and foot joints can cause ankle or calcaneal fracture, which is reported to account for one in three foot injuries (Herscovici & Scaduto, 2012). In other words, the large ankle ROM in forefoot landing is effective at absorbing the impact upon landing, but the position of the ankle joint at first contact can increase the risk of lateral ankle sprain (Wright et al., 2000). Thus, it seems that strengthening or controlling the lower leg muscles involved in ankle movements could enable better control of the ankle at first contact during forefoot landing, which would help reduce ankle injuries. Further research on the relationship between forefoot landing and ankle injuries will be required. Meanwhile, the large impact generated by the stiff ankle joint in rearfoot landing can cause fractures (Herscovici & Scaduto, 2012), and therefore, in the future, it will be essential to study methods to prevent the ankle injuries that can develop in rearfoot landing.

Because the present study was conducted on healthy male adults in their 20s, the results cannot be easily generalized. Since there are likely to be sex differences in the effect of landing type on biomechanical characteristics of the joints, additional research will be required in female subjects.

CONCLUSION

Our study aimed to elucidate the biomechanical characteristics of forefoot and rearfoot landing by comparing lower-extremity kinematics, kinetics, and joint work according to landing type. In forefoot landing, increased knee flexion and ankle dorsiflexion are effective at absorbing energy, and this can reduce the load on the joints. In rearfoot landing, reduced extension moment and joint work decreased the capacity of the knee and ankle extensors to absorb energy, which is thought to increase the risk of later overuse injury in the lower extremity. The contribution of each joint to total work, in order to absorb the impact upon landing, also showed differences according to landing type. Because the ankle joint is in a state of dorsiflexion in rearfoot landing, the inability to achieve sufficient plantarflexion impairs energy absorption, which in turn results in greater energy absorption at the hip joint, rather than the ankle joint.

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