Influence of foot-landing positions at initial contact on knee flexion angles for single-leg drop landings

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Abstract

Purpose: Small knee flexion angles are associated with increased non-contact anterior cruciate ligament (ACL) injury risks. The purpose of this study was to provide insights into how ankle plantar flexion angles influenced knee flexion angles at initial contact during single-leg drop landings. **Method:** Thirteen male recreational basketball players performed single-leg drop landings from a 30-cm high platform using three randomized foot-landing positions (natural, fore-foot, and flat-foot). A motion capture system and a force plate were used to measure lower extremity kinematics and vertical ground reaction force (VGRF). A one-way repeated measures Analysis of Variance and the Friedman test were conducted (α = .05). **Results:** Foot-landing position had a significant effect on knee flexion angles at initial contact (p < .001). As compared to flat-foot landings (18° (*SD* 5°), significantly smaller knee flexion angles were found for natural (mean 12° (*SD* 5°), p = .004) and fore-foot landings (mean 12° (*SD* 3°), p = .001). There was however significantly smaller knee flexion range of motion in flat-foot landings (mean 42° (*SD* 3°), as compared to natural landings (mean 48° (*SD* 4°) (p < .001). Flat-foot landings also resulted in a 1.4 times higher maximum VGRF than fore-foot landings (p < .001) and 1.3 times higher than natural landings (p = .005). **Conclusions:** Natural and fore-foot landings are better for force absorption but are associated with smaller knee flexion angles at initial contact. These findings have important implications for non-contact ACL injuries.

Keywords: ACL, Vertical ground reaction force, Range of motion, Kinematics

Introduction

The anterior cruciate ligament (ACL) is the most commonly injured ligament in the knee for the general population (Bollen, 2000), playing a vital role in providing knee stability, especially in the anterior-posterior direction (Hartigan, Lewek, & Snyder-Macker, 2011). An ACL injury frequently takes place during non-contact situations (Dragoo, Braun, Durham, Chen, & Harris, 2012; Gianotti, Marshall, Hume, & Bunt, 2009; Waldén, Hägglund, Magnusson, & Ekstrand, 2011) which are caused by the player's own manoeuvres, rather than caused by contact with another player or object (Marsall, Padua, & McGrath, 2007). The manoeuvres with the highest number of ACL injuries differ between sports. In Australian football, injuries were mostly found during side-cutting and landing (Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007). Injuries during landing were most commonly found in basketball and badminton (Krosshaug et al., 2007; Kimura et al. 2010). Non-contact ACL injuries in soccer frequently occur during direction change or cutting with deceleration, jump landing, or pivoting with small knee flexion angles (Alentorn-Geli et al., 2009).

Small knee flexion angles are often associated with higher non-contact ACL injury risks (Lin, Liu, Garrett, & Yu, 2008). This association of small knee flexion angles with higher ACL injury risks was well supported by video analyses (Cochrane, Lloyd, Buttfield, Seward, & McGivern, 2007; Koga et al., 2010; Krosshaug et al., 2007; Olsen, Myklebust, Engebretsen, & Bahr, 2004), cadaver studies (Li et al., 1999; Renström, Arms, Stanwyck, Johnson, & Pope, 1986) and biomechanical explanations (Yu & Garrett, 2007). At small knee flexion angles of less than 15°, the addition of hamstring activation did not significantly reduce ACL loading, as compared to when quadriceps were activated alone (Li et al., 1999). Particularly for single-leg landings, knee flexion angles at initial contact are of a more significant concern than at maximum vertical ground reaction force (VGRF). The focus should, therefore, be on initial contact because peak ACL strain was found to occur at initial contact (Lamontagne, Benoit, Ramsey, Caraffa, & Cerulli, 2008). Athletes are often requested to land using 'soft' landings to achieve higher knee flexion angles (DeVita & Skelly, 1992; Garrett & Yu, 2007; Laughlin et al., 2011). However, 'soft' landing instructions are subjective, and a more definite way of influencing knee flexion angles should be identified. In a study by DiStefano, Padua, Brown, and Guskiewicz (2008), use of an ankle brace was found to significantly increase (brace = 12° ± 9°, no brace = 9° ± 9°, $p < .05$) knee flexion angles at initial contact. This result suggests that foot landing positions could influence knee flexion angles. In another study, Cortes et al. (2007) investigated the effects of foot-landing positions in the sagittal plane (fore-foot, rear-foot, and natural) on lower extremity kinematics. At initial contact, knee flexion angles during fore-foot landing were significantly smaller than the other landing positions (Cortes et al., 2007). Their study results further suggest that knee flexion angles could vary with different foot landing positions in the sagittal plane. However, Cortes et al. (2007) only focused on double-leg landings and not single-leg landings, whereby more non-contact ACL injuries take place (Donnelly et al., 2012). Results in double-leg landings might not accurately reflect that of single-leg landings as they result in different knee kinematics and muscle activation levels (Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007). Investigating the influence of ankle plantar flexion angles on knee flexion angles during single-leg landings would, therefore, provide further insights into ACL injury risk reduction.

This study aimed to examine the influence of fore-foot, natural, and flat-foot landing positions on non-contact ACL injury risks during single-leg drop landings. Knee flexion angles, knee flexion range of motion, and maximum VGRF were studied under the different landing conditions. Drawing conclusions from double-leg landing results (Cortes et al., 2007), it was hypothesized that fore-foot landing positions would result in significantly smaller knee flexion angles at initial contact; higher knee flexion range of motion; and were better for force attenuation, as compared to flat-foot landings.

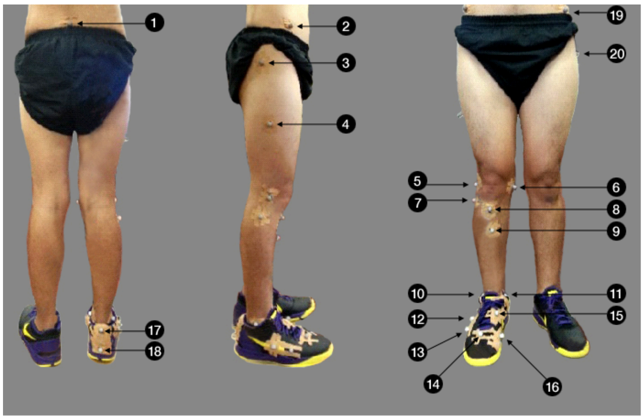
Methods

Participants

A power analysis was performed using G*Power (v3.1.9.2, Heinrich-Heine University, Düsseldorf, Germany). With no prior studies conducted for single-leg drop landings, a large effect size (f = .40) was used based on pilot study results for a one-way repeated measures Analysis of Variance (Cohen, 1988). A sample size of 12 was required (Power 1-β = .80, α = .05). During single leg landings, ACL injuries most commonly occur for male basketball players (Krosshaug et al., 2007). Thus, only male basketball players were recruited. Fifteen participants, through advertisements across the different universities in XXX *(blinded for review)*, agreed to take part in the study. Data of two participants were discarded as one dropped out of the study, and the other was incomplete. The remaining 13 male recreational basketball players (age: mean 23.3 (*SD* 1.8) years, height: mean 1.74 (*SD* .07) m, body mass: mean 67.4 (*SD* 10.7) kg) had no previous ACL injuries or lower extremity injuries in the previous six months before the study. They also played basketball minimally once a week for at least two years before the study. The study was approved by the XXX *(blinded for review)* Institutional Review Board. All participants signed written consent forms before taking part in the study.

Procedures

Participants wore the same type of basketball shoes (Nike Zoom Attero, Nike, Inc., Beaverton, OR, USA) to eliminate possible confounding effects due to differences in footwear. Standard shoe sizes of US 9, 10, 11, and 13 were adopted, and participants were pre-screened to confirm their shoe sizes. Shoes were secured tightly to minimize relative motion between the shoes and feet. When necessary, off-the-shelf insoles were provided to ensure an optimal fit. Retroreflective markers were attached to the lower extremity, mainly at the dominant limb using tape (Figure 1). The dominant leg was stated by participants as the leg that kicked a ball furthest (Hewett et al., 2005).



Marker	Landmark
1	Sacrum
2	Right anterior superior iliac spine
3	Right greater trochanter of femur
4	Mid-thigh
5	Lateral knee epicondyle
6	Medial knee epicondyle
7	Head of fibula
8	Tibial tuberosity
9	Anterior aspect of shin
10	Lateral malleolus
11	Medial malleolus
12	Base of fifth metatarsal
13	Head of fifth metatarsal
14	Toe (between second and third metatarsal heads)
15	Base of first metatarsal
16	Head of first metatarsal
17	Proximal calcaneus (CAL2)
18	Distal calcaneus (CAL1)
19	Left anterior superior iliac spine
20	Left greater trochanter of femur

Fig. 1. Positions of marker placement.

Three-dimensional motion data were recorded at 200 Hz using eight digital cameras and the Cortex software (version 1.1.4.368, Motion Analysis Corp, Santa Rosa, CA, USA). VGRF were recorded with a force plate (Kistler type 9287CA, Winterthur, Switzerland) at 1000 Hz, in synchronization with the digital cameras (Figure 2).

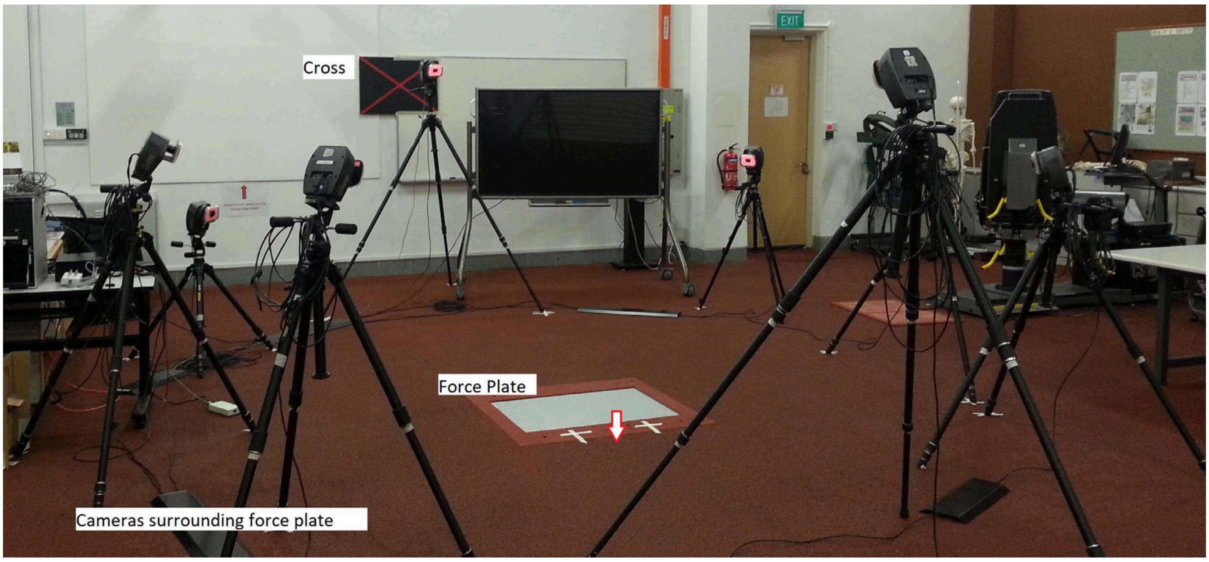


Fig. 2. Laboratory set-up of the cameras, force plate, and the board with the cross on the wall. The 30-cm platform was placed at the position, marked by the red arrow.

Each participant first did warm-up exercises for 5 min, targeting the lower extremity muscles. The exercises involved stretching of the hamstrings, quadriceps and calves, tip-toe exercises, ankle rotation, squats, low lunges, jumps, jogging on the spot and jogging with high knees. A standing trial was then captured with the participant standing shoulder-width apart, with the arms crossed over the chest, on the force plate. After that, participants practised the single-leg drop-landing manoeuvres. Participants were allowed as many practice trials as required until they could consistently perform the drop-landing manoeuvre comfortably. Participants first stood with their dominant limb on a 30-cm height platform, while facing a board marked with a red cross, placed at the front of the lab (Figure 2). Participants had to maintain the vertical height across the different landing trials by not bending down or jumping up prior to dropping from the platform (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003). The vertical height was monitored using the z-axis position of the sacrum marker, a proxy for the centre of mass, using the Cortex software. Participants stood with their dominant foot at the edge of the platform, with the non-dominant limb stretched out (Figure 3a). Participants then drop-landed onto a force plate using the same dominant leg and held still for 2 s to obtain stable readings in their final landing position (Figure 3b). They were requested to keep their hands crossed over their chests throughout the drop-landing manoeuvre to reduce any variability from swinging arms (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003).

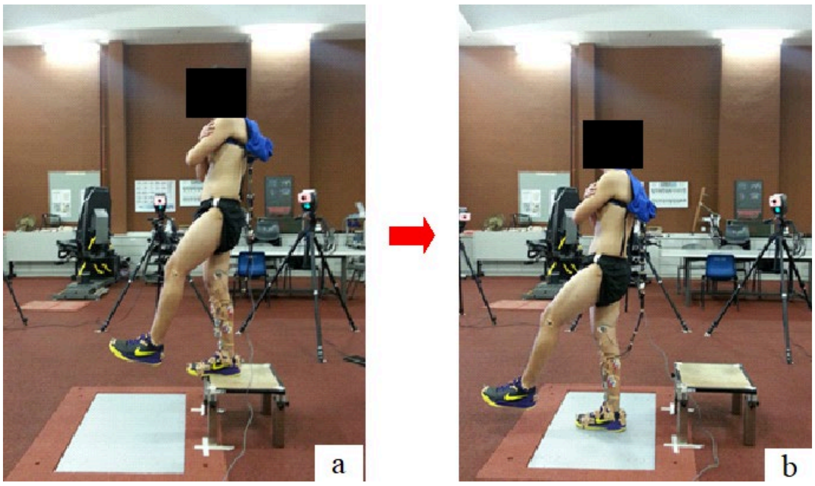


Fig. 3. Participants a) stood with their dominant foot at the edge of the platform, with the non-dominant limb stretched out and b) then drop-landed onto a force plate using the same dominant leg.

The protocol was conducted for natural, fore-foot and flat-foot landings at initial contact, with plantar flexion angles denoting as positive. There was no control over how participants should land during natural landings, which was defined as how participants usually landed from a jump. At each condition, three-dimensional kinematic data in the sagittal plane and maximum VGRF were collected for three trials. During data collection, two-dimensional kinematics in the Cortex software was used to monitor if participants landed in the required stipulated foot positions. After data were processed offline, ankle flexion angles were further verified with the processed data obtained using the Visual3D program (v5.00.33, C-Motion, Germantown, PA, USA) for fore-foot (>25° ankle flexion angles) and flat-foot (<5° ankle flexion angles). The ankle flexion angles were pre-determined during the pilot study to be representative of each fore-foot and flat-foot landing position. The sequence of conditions was randomized for each participant to minimize the influence due to fatigue, learning effects, or boredom. Drop-landing trials that met requirements for the stipulated positions were considered successful. The average of successful trials of each condition was used for analysis.

Data Reduction

The motion and VGRF data were filtered through a fourth-order, zero-phase-shift Butterworth low-pass filter at cut-off frequencies of 9 Hz and 50 Hz, respectively. A single-leg model was built consisting of the pelvis, thigh, shank, and foot segments using the Visual3D program (Teng, Kong & Leong, 2017). Markers used in the model were the sacrum, right and left anterior superior iliac spine, lateral and medial knee epicondyles, tibial tuberosity, head of fibula, lateral and medial malleoli, CAL1, CAL2 and toe (Figure 1). A virtual landmark marker was also created, with an axial location obtained from at a point between the lateral and medial epicondyles. Its anterior-posterior and medial-lateral locations were obtained by projecting the tibial tuberosity marker onto the plane formed by the head of fibula, lateral and medial malleoli markers. Cardan rotation sequence XYZ was used for the joint angle computation. Knee flexion angles, ankle flexion angles, and VGRF were extracted for analysis. Knee flexion angle was defined as the shank segment's orientation, relative to that of the thigh segment. Ankle flexion angle, at initial contact, was defined as the orientation of the foot segment relative to the shank segment. The static standing trial was used as each participant's neutral (zero) ankle flexion angle position. Normalization of the ankle flexion angles was carried out in the Visual3D program by setting the same orientation for the shank and the foot segments in the standing trial but different tracking markers in the respective segments. Knee flexion and ankle plantar flexion angles were denoted as positive. Knee flexion range of motion was computed by subtracting the knee flexion angles at initial contact from the maximum knee flexion angles. Initial contact with the ground was defined as the first point that VGRF exceeded 10 N (Ford et al., 2006). Maximum VGRF was normalized to body weight (BW). The landing phase was defined as the time from initial contact to the point of maximum knee flexion angle.

Statistical Analyses

All statistical analyses were performed using the statistical package SPSS (IBM SPSS 22.0, IBM Corp., NY, USA). Data are expressed as mean (standard deviation). Normality of data was tested using the Shapiro-Wilk test ($p > .05$). Since normality could not be assumed for ankle flexion angle and maximum VGRF, non-parametric Friedman test was conducted ($\alpha = .05$) for these two variables. For the knee flexion angles, normality could be assumed, and a one-way repeated measures Analysis of Variance was conducted at $\alpha = .05$. Assumption of sphericity was tested using the Mauchly's test of sphericity, and this assumption was not violated. For both parametric and non-parametric tests, post-hoc tests with Bonferroni adjustment were made with a family-wise $\alpha = .05$. Effect size was calculated as Cohen's d for the parametric data and as Pearson's correlation coefficient, r, for the non-parametric data (Field, 2013) between paired variables. The values of .20, .50 and .80 were defined as small, medium and large effects, respectively for d; while the values of .10, .30 and .50 were defined as small, medium and large effects, respectively for r (Cohen, 1988).

Results

Ankle plantar flexion angles were observed during natural landings, suggesting that participants naturally landed with fore-foot landings (Table 1). Foot-landing position had a significant effect on knee flexion angles at initial contact and maximum VGRF (Table 1). As compared to flat-foot landings, 5° smaller knee flexion angles were found for natural landings (p = .004, d = 1.17 (large effect)) and 6° smaller knee flexion angles were found for fore-foot landings (p = .001, d = 1.41 (large effect)). The representative postures of fore-foot and flat-foot landings at initial contact with the ground are shown in Figure 4.

Table 1. Kinematic Results of Foot Landing Positions in the Sagittal Plane (mean (standard deviation)) (n = 13)

Variables	Natural	Fore-foot	Flat-foot	F-values*	p-value
At Initial Contact					
Ankle Plantarflexion Angles (°)	$29(8)^{c}$	$34(4)^{c}$	$-3(4)^{a,b}$		< .001
Knee Flexion Angles (°)	$12(5)^{c}$	$12(3)^{c}$	$18(5)^{a,b}$	18.328	< .001
At Maximum VGRF					
VGRF (BW)	$4.5(.4)^{c}$	$4.3(.5)^{c}$	$5.9(.4)^{a,b}$		< .001
Knee Flexion Angles (°)	$30(6)^{c}$	$33(6)^{c}$	$21(5)^{a,b}$	54.875	< .001
Range of Motion #					
Knee Flexion Angles (°)	$48(4)^{c}$	45(4)	$42(3)^{a}$	13.984	< .001

VGRF, vertical ground reaction force; BW, body weight;
[a] Significantly different from natural landing.
[b] Significantly different from fore-foot landing.
[c] Significantly different from flat-foot landing.
Range of motion was calculated by subtracting the knee flexion angles at initial contact from maximum knee flexion angles.
* F-values were only available for parameters that were analysed using one-way repeated measures Analysis of Variance. F-value for 'ankle plantarflexion angles' and 'VGRF(BW)' were not available as they were analysed using Friedman test.

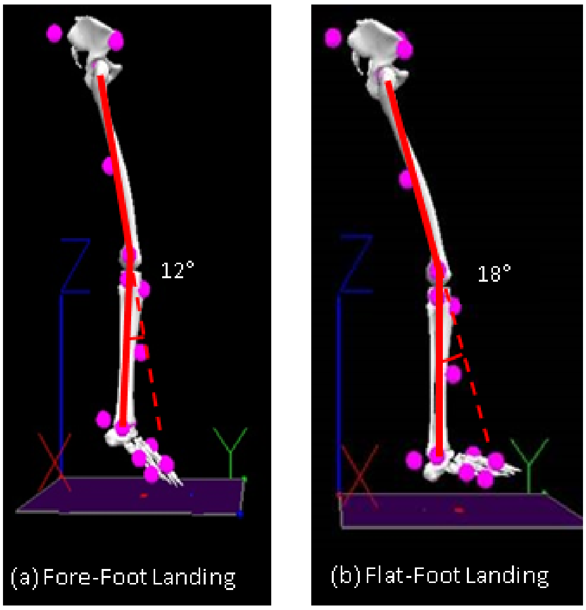


Fig. 4. Representative knee flexion angles at initial contact with the ground for (a) fore-foot landings and (b) flat-foot landings (Image created using Visual3D).

At maximum VGRF, flat-foot landings resulted in significantly smaller knee flexion angles (9°) as compared to natural landings ($p < .001$, $d = 2.65$ (large effect)). Similarly, flat-foot landings resulted in significantly smaller knee flexion angles (12°) as compared to fore-foot landings ($p < .001$, $d = 2.32$ (large effect)) at maximum VGRF. Knee flexion angles were not significantly different between natural and fore-foot landings at initial contact ($p = .715$, $d = .34$ (small effect)) or at maximum VGRF ($p = .078$, $d = .70$ (medium effect)).

Figure 5 shows the mean ensemble curves of the VGRF, knee, and ankle flexion angles over the entire landing phase. Maximum VGRF occurred earlier in the landing phase for flat-foot landing (around 10%), as compared to the other plantar-flexed foot landing positions (around 30 – 40%).

Foot-landing position had a significant effect on knee flexion range of motion ($p < .001$; Table 1). There was significantly higher knee flexion range of motion (6°) in natural landings, as compared to flat-foot landings ($p < .001$, $d = 1.58$ (large effect)). Knee flexion range of motion between fore-foot and flat-foot landings (3°) was however not found to be significantly different (p = .064, d = .73 (medium effect)). Finally, flat-foot landings (*Mdn* = 6.0 BW) resulted in significantly higher maximum VGRF than fore-foot landings (*Mdn* = 4.5 BW), Z = 3.180, p < .001, r = .62 (large effect) and natural landings (*Mdn* = 4.6 BW), Z = 3.180, p = .005, r = .62 (large effect)).

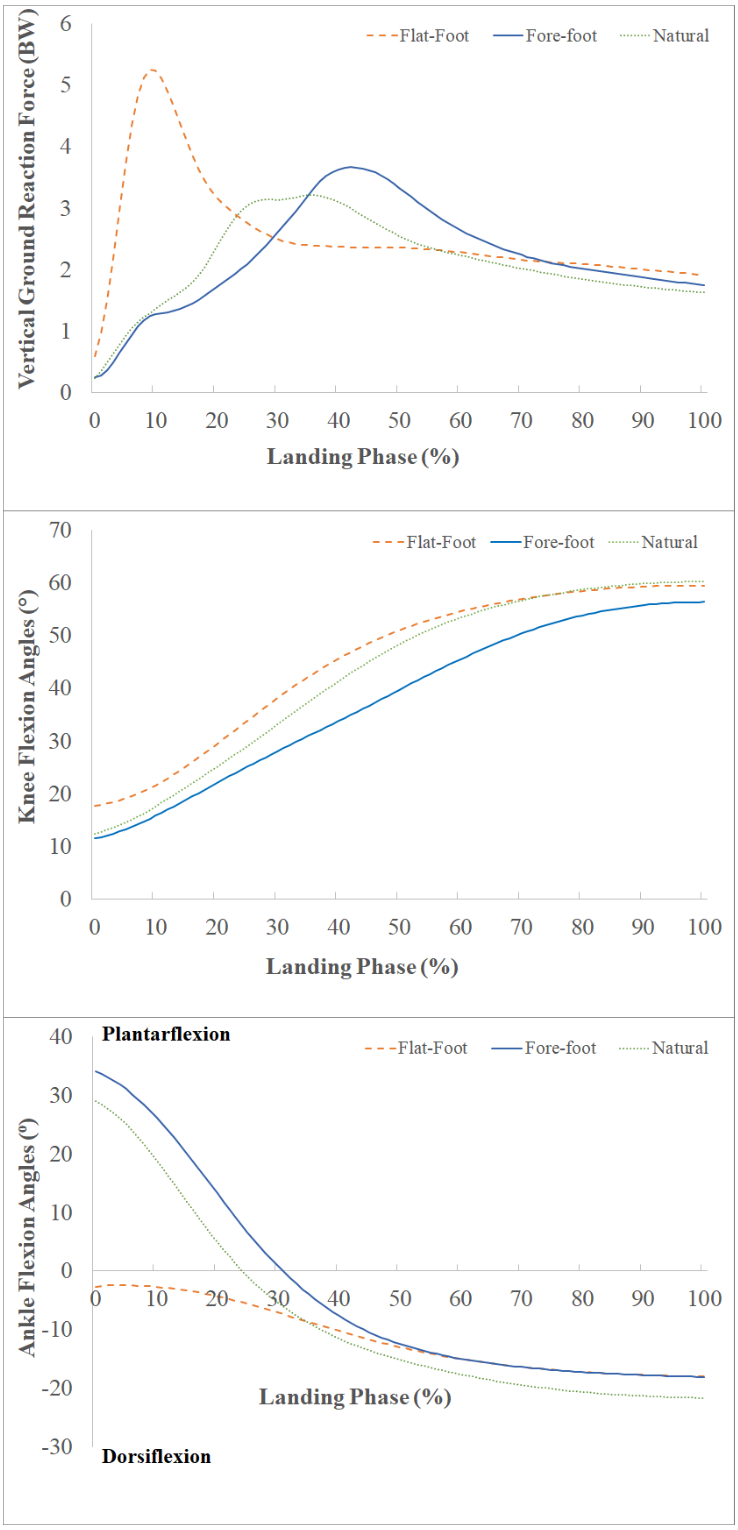


Fig. 5. Mean ensemble curves of vertical ground reaction force, normalized to body weight (BW), knee flexion angles, and angle flexion angles (n = 13). The whole landing phase was defined as the time from initial contact with the ground (0%) to the point of maximum knee flexion angle (100%).

Discussion

The aim of this study was to investigate the influence of foot landing positions (fore-foot, natural and flat-foot) on knee flexion angles and maximum VGRF. Results from this study would have implications for non-contact ACL injury risks. As hypothesized, fore-foot landing positions resulted in significantly smaller knee flexion angles at initial contact, lower maximum VGRF, greater knee range of motion and resulted in better force absorption, as compared to flat-foot landings.

Influence of Foot Landing Positions on Knee Flexion Angles at Initial Contact

Fore-foot landings have been encouraged to allow for greater force attenuation (Alentorn-Geli et al., 2009). In our study, fore-foot landing positions are nevertheless found to be associated with significantly smaller knee flexion angles at initial contact (Table 1). Small knee flexion angles are, in turn, associated with increased non-contact ACL injury risks (Lin, Liu, Garrett, & Yu, 2008). Cortes et al. (2007) also found significantly smaller knee flexion angles for fore-foot landings at initial contact, as compared to rear-foot and natural foot landing positions for double-leg landings. Mean knee flexion angles for double-leg fore-foot landing (18°, 95% CI [13 to 22]) (Cortes et al., 2007) was however higher than that found in our single-leg landing study (12°, 95% CI [10 to 14]). Similar to our findings, small knee flexion angles of less than 15° were also found during single-leg landings in the literature (Kim & Jeon, 2016; Orishimo, Liederbach, Kremenic, Hagins, & Pappas, 2014; Weinhandl, Joshi, & O'Connor, 2010). During small knee flexion angle positions of less than 15°, hamstring activation becomes less effective in countering quadriceps' anterior pull on the tibia, resulting in higher ACL loading (Li et al., 1999). Smaller knee flexion angles also further increase ACL strain by increasing the patella tendon-tibia shaft angle, as well as, ACL elevation and deviation angles (Yu & Garrett, 2007). Such small knee flexion angles could be contributory reasons to higher ACL injury risks for single-leg, rather than double-leg drop landings (Donnelly et al., 2012).

Conversely, fore-foot landing positions resulted in significantly higher knee flexion angles at maximum VGRF than flat-foot landings (Table 1). However, peak ACL strain occurred at initial contact for single-leg landings, rather than at maximum VGRF for cutting manoeuvres (Lamontagne, Benoit, Ramsey, Caraffa, & Cerulli, 2008). Small knee flexion angles during fore-foot landing might contribute to this increased ACL strain at initial contact for single-leg landings. It is therefore important to focus on initial contact, rather than at maximum VGRF as perceived (Alentorn-Geli et al., 2009), when considering the implications of foot-landing positions on ACL injury risks.

Influence of Foot Landing Positions on Force Absorption

Flat-foot landings resulted in significantly smaller knee flexion angles at maximum VGRF and less knee flexion range of motion than natural landings (Table 1); and significantly higher maximum VGRF than the other landing positions (Table 1). These results are expected as the impulse is the same under similar conditions but impact time is shorter for flat-foot landings, resulting in greater impact forces. Similarly, reduced dorsiflexion and knee flexion range of motion, together with higher maximum VGRF, were observed by DeVita and Skelly (1992) for stiffer double-leg landings. Figure 4 also shows that maximum VGRF occurred earlier in the landing phase for flat-foot landing than for the other foot landing positions. Similar to our study's findings for natural landings, maximum VGRF occurred around 30% of the landing phase during single-leg landings in the study by Weinhandl, Joshi, and O'Connor (2010). With maximum VGRF occurring earlier during flat-foot landings, this could have implications on whether the body can respond in time for force absorption. Flat-foot landings should be avoided to allow for greater force absorption, Natural landings resulted in a significantly higher knee flexion range of motion, as compared to flat-foot landings (Table 1). The mean range of motion (48°) recorded in our study for natural landings is similar to that recorded by Weinhandl, Joshi, and O'Connor (2010) (50°, 95% CI [45 to 55]) [32]. A greater range of motion would help in force absorption, and this is supported by significantly lower maximum VGRF, as compared to flat-foot landings. Similar to findings by Alentorn-Geli et al. (2009), a plantar-flexed landing is still found to be a better landing position in terms of force attenuation and should be adopted.

Need for Optimal Range of Fore-foot Landing Positions

In our study, foot landing positions in the sagittal plane were found to have a significant effect on knee flexion angles. Thus, athletes could change foot landing techniques to vary knee flexion angles. It would otherwise be more subjective to ask athletes to adopt ‘soft’ landings to increase knee flexion angles (DeVita & Skelly, 1992; Garrett & Yu, 2007; Laughlin et al., 2011), especially during competitions and under stressful conditions. While fore-foot landing positions are recommended for force attenuation, there is a need to determine an optimal range of ankle plantar flexion angles at initial contact because ankle angles can alter knee flexion angles. When ankle plantar flexion angles are too high, knee flexion angles may decrease, inferring from the natural and fore-foot landing results in our study. There were significantly smaller knee flexion angles at initial contact for natural and fore-foot landings, as compared to flat-foot landings. Cortes et al. (2007) found smaller knee flexion angles with increased mean ankle plantar flexion angles for fore-foot landings, as compared to natural landings. The role of the gastrocnemius could provide insights to this decrease in knee flexion angles with increased ankle plantar flexion angles. During small knee flexion angles of less than 15°, hamstring activation becomes less effective in reducing ACL loading (Li et al., 1999). Morgan, Donnelly, and Reinbolt (2014) found that average peak gastrocnemius forces were significantly higher than the hamstring forces during the weight acceptance phase of single-leg jump landings. They suggested that the gastrocnemius instead supported the hamstring muscles in countering the forces from the quadriceps. The gastrocnemius, another knee flexor, and a plantar flexor becomes less of a knee flexor as the ankle is plantarflexed (Hartigan, Lewek, & Snyder-Macker, 2011). With greater ankle plantar flexion angles, the gastrocnemius could act less as a secondary knee flexor, possibly resulting in smaller knee flexion angles.

Based on our results (Table 1), this optimal range of fore-foot landing positions could fall slightly below the natural ankle plantarflexion angle of 29°, to allow for a minimum knee flexion angle of 15°. Minimum knee flexion angle of 15° would then allow hamstring activation to be effective in countering quadriceps' anterior pull on the tibia, thus lowering ACL loading (Li et al, 1999).

Limitations

The main limitation of this study was that drop-landing tasks might not be as representative of natural landing conditions, as compared to landings following a jump (Collings et al., 2019). Future studies could be carried out using functional tasks such as during a stop-jump. Secondly, this study only recruited male participants. There is a higher incidence of ACL injury in females than in males (Waldén, Hägglund, Magnusson, & Ekstrand, 2011). Given that sex effects on knee flexion angles at initial contact were not significant for landings (Cortes et al., 2007; Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007), comparison between sexes was not the focus of this study. Thirdly, muscle activation patterns at initial contact were not recorded in this study. Morgan, Donnelly, and Reinbolt (2014) suggested the supportive role of the gastrocnemius in countering the quadriceps, and this could be recorded in the future. Lastly, skin-mounted markers may have more skin artefacts than marker clusters and may introduce noise to the data. Since this study focused on angles in the sagittal plane and not in the frontal and transverse planes, which are associated with lower signal-to-noise ratios (Hamill, Selbie, & Kepple, 2014), the errors resulting from skin artefacts would be smaller. Furthermore, the mean knee flexion angles at initial contact (13°) for natural landings in our study were similar to those for single-leg landing (12° to 14°) reported in the literature (Kim & Jeon, 2016; Orishimo, Liederbach, Kremenic, Hagins, & Pappas, 2014; Weinhandl, Joshi, & O'Connor, 2010). This further supports the results found in the present study.

Conclusion

A more plantar-flexed landing position was better in force absorption, as compared to flat-foot landings. Nevertheless, participants were found to land in small knee flexion angles of less than 15° at initial contact during single-leg drop landings. Small knee flexion angles were associated with increased non-contact ACL injury risks. Thus, while fore-foot landings are preferable to reduce impact forces, the resulting small knee flexion angles at initial contact is a concern as it may increase ACL injury risks during single-leg drop landings.

What does this study add?

To our best knowledge, this is the first study to manipulate various foot-landing positions during single-leg landings. This study offers a new perspective on how knee flexion angles could be influenced by foot-landing positions, rather than by subjective instructions to have ‘soft’ landings. These findings also have important implications for non-contact ACL injury risks and could provide new insights for biomechanists, coaches, physicians, podiatrists, and researchers in the area of ACL injury risk prevention.

Funding: Institute for Sports Research

IRB Application: Approved by Nanyang Technological University Institutional Review Board (IRB-2013-09-003)

Acknowledgments: The authors would like to thank Gu Kaiyi and Bai Jinhao for their contributions in data collection.

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