Research Article

The Effect of Two Types of Foot Orthoses on the Knee Valgus Angle Within Single-Leg Landing: Implications for ACL Damage Hazard Decrease

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Abstract

Background: Non-contact anterior cruciate ligament (ACL) injuries commonly occur when athletes land in high risk positions such as knee valgus. The impact of foot orthoses during exercises that recreate a non-contact ACL harm system (i.e., landing) in any case will be obscured.

Objectives: The purpose of the current study research was to evaluate the effect of two foot orthoses (semi-hard foot orthoses and hard foot orthoses) on knee valgus angle during single-leg drop landing.

Methods: Twenty male leisure volleyball gamers performed landing in one-leg step from 30 cm height in 3 conditions (without foot orthoses, mid-hard foot orthoses, and hard foot orthoses). A motion capture system was used to measure lower extremity kinematics. Two risk factors of ACL injury, maximum knee valgus angle (KVA), and maximum knee flexion was measured. ANOVA was used for statistical analysis ($P < 0.05$).

Results: With mid-hard foot, orthoses provide the maximum level of knee flexion and the minimum level of knee valgus during single-leg drop landing.

Conclusions: It may be concluded showed that foot orthoses affect knee kinematics. More knee flexion and less knee valgus brought about by mid-hard foot orthoses can reduce injuries of the anterior cruciate ligament (ACL).

Keywords: Non-contact, ACL, Foot Orthoses, Landing, Three Dimensional Kinematics

1. Background

Landing is one of the most commonly used sporting movements, which can produce an impact force between 2 to 12 times the body weight, the most common cause mechanism of lower limb injury (1, 2). This mechanical impact should be moderated via the musculoskeletal system. Increasing impact forces and repetition of these forces facilitate damage to the soft tissue surrounding the joint during landing. Single-leg landing (SLL) is common in sports such as volleyball, soccer, basketball, and badminton (1). The alignment of the lower limb has the main responsibility for absorbing pressure during contact and modulating the load (3). Lack of knee joint muscles' ability to absorb force during landing may lead to changes in the kinematics of this joint, including increased knee valgus angle (KVA) and knee flexion (4, 5). About 70% of knee injuries, especially anterior cruciate ligament (ACL), occur in non-contact in-juries (6, 7). Increased KVA and decreased knee flexion, tibia spin, internal rotation, and hip adduction within cutting and landing maneuvers can usually be the mechanism of damage to ACL injuries (6-10). which can increase the strain on the ACL (11, 12). Studies showed that SLL leads to an increase in damage to the ACL in a non-contact position compared to double-leg landing (DLL) (13, 14). They concluded that increased knee valgus and decreased knee flexion followed. A study by Yeow et al. showed that single-leg landing, compared to double-leg landing, significantly reduced knee flexion and increased the risk of anterior cruciate ligament damage (14).

Foot orthoses are usually prescribed to improve lower limb and functional impairment (15, 16). Researchers have considered the effect of orthoses on lower limb biomechanics during walking and running (17-19). When using foot orthoses during walking and running, a decrease in the movement of the lower limbs was observed in both

the frontal and transverse planes (19). The effect of foot orthoses in sports activities such as landing, jumping, and cutting is still unknown, and most studies have investigated the effect of foot orthoses during walking and running (16, 19). Few studies have focused on the effect of foot orthoses on sports activities, such as landing, and have found different results (20, 21). Foot orthoses have increased activity in the gluteus medius during slow movements (20). On the contrary, foot orthoses have also shown an increase in movement in the frontal plane in the ankle and an increase in lateral pressure on the foot during the landing after the layup in basketball (21). The use of foot orthoses causes reduced lower limb injury in sports activities (15). It has been suggested that foot orthoses can change the movement of the lower extremities, which is responsible for reducing damage.

2. Objectives

The purpose of this research was to consider the effect of two types of foot orthoses (semi-hard foot orthoses and hard foot orthoses) on knee valgus angle during single-leg drop-landing.

3. Methods

3.1. Participants

Findings from a pilot study (n = 8) demonstrated the variability expected for each variable of interest. To achieve 80% or above power with two degrees of freedom at a significant level of 0.05, we calculated a needed sample size of 20 subjects (22). Twenty male recreational volleyball players with the mean ± standard deviation of age: of 22.3 ± 1.45 years; height of 1.75 ± 15.51 meters; body mass: 69.05 ± 0.51 kg participated in this study. All participants had no history of ACL injury as well as lower limb injury in the past six months. All players had played volleyball at least once a week during the past three years. All participants signed a written agreement form, before beginning this study.

3.2. Equipment

Participants used similar shoes (ASICS Gel-Rocket 7, Men's Volleyball Shoes, model B405N, Japan) to prevent the effect of footwear. In the current study, all participants were right-leg dominant. Nine reflective markers (15 mm in diameter) were placed on the anatomical points of the dominant leg. The anatomical points included the anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), medial and lateral condyles of the femur, medial and lateral condyles of the tibia, medial and lateral malleoli, and the second metatarsal head (on the shoe).

Markers were placed on the anatomical points before measurements. Three-dimensional (3D) motion data was captured at a frequency of 200 Hz using six digital cameras and Cortex software (Motion Analysis Corp., Santa Rosa, CA, USA) version 2.5.0.1160. Two types of foot orthoses (semi-hard and hard) with four layers were used in the study. The orthoses were similar except for the first and upper layers. The second to fourth layers in both foot orthoses were made of (polypropylene, bone POLYFOAM, and hard POLYFOAM), respectively. The first layer in semi-hard foot orthoses was made from semi-hard foam, while the first layer of hard foot orthoses was made from hard foam. The thicknesses of the layers of foot orthoses were: semi-hard foam and hard foam 5 mm, polypropylene was 1.5 - 2 mm, and hard polyfoam was 7 mm. Also, in general, the thickness of foot orthoses under the heel was 6 to 10 mm, and in the center of the inner longitudinal arc, it reached 25 to 30 mm. It should be noted that the molecular composition of the material used to make the foams was the same. But the factor that made hard foam, for example, harder in terms of hardness was that the amount of material poured into the mold was higher during the manufacturing of hard foam (Figure 1).

3.3. Procedures

Subjects were required to do a 5-minute warm-up and given time to practice drop-landing maneuvers before recording data. Participants were requested to stand on a 0.3 m high step on two legs, with both hands on iliac crests. The drop-landing operation was performed with the dominant leg at 20 cm from the front edge of the platform. Subjects were trained to easily roll off the platform and perform the most natural possible landing. Subjects were trained to maintain their balance for at least 2 seconds after landing. Drop-landing was performed in three positions in a randomized manner: including without foot orthoses, with semi-hard foot orthoses, and with hard foot orthoses. A successful landing was defined as a landing without jumping up or forward or landing on both legs. Six successful trials were recorded for each orthosis.

3.4. Measures

The maximum knee angle was measured on the sagittal and coronal planes. Kinematics data were automatically filtered by a fourth-order zero-lag Butterworth 12-HZ low-pass filter (23). A kinematics model was proposed based on the lower limb to describe knee joint angles during landing. The hip joint center was estimated based on the methods described by Vaughan et al. (24). In the standing trial, joint centers for the knee and ankle were determined (25). The femur's center was defined as the mid-
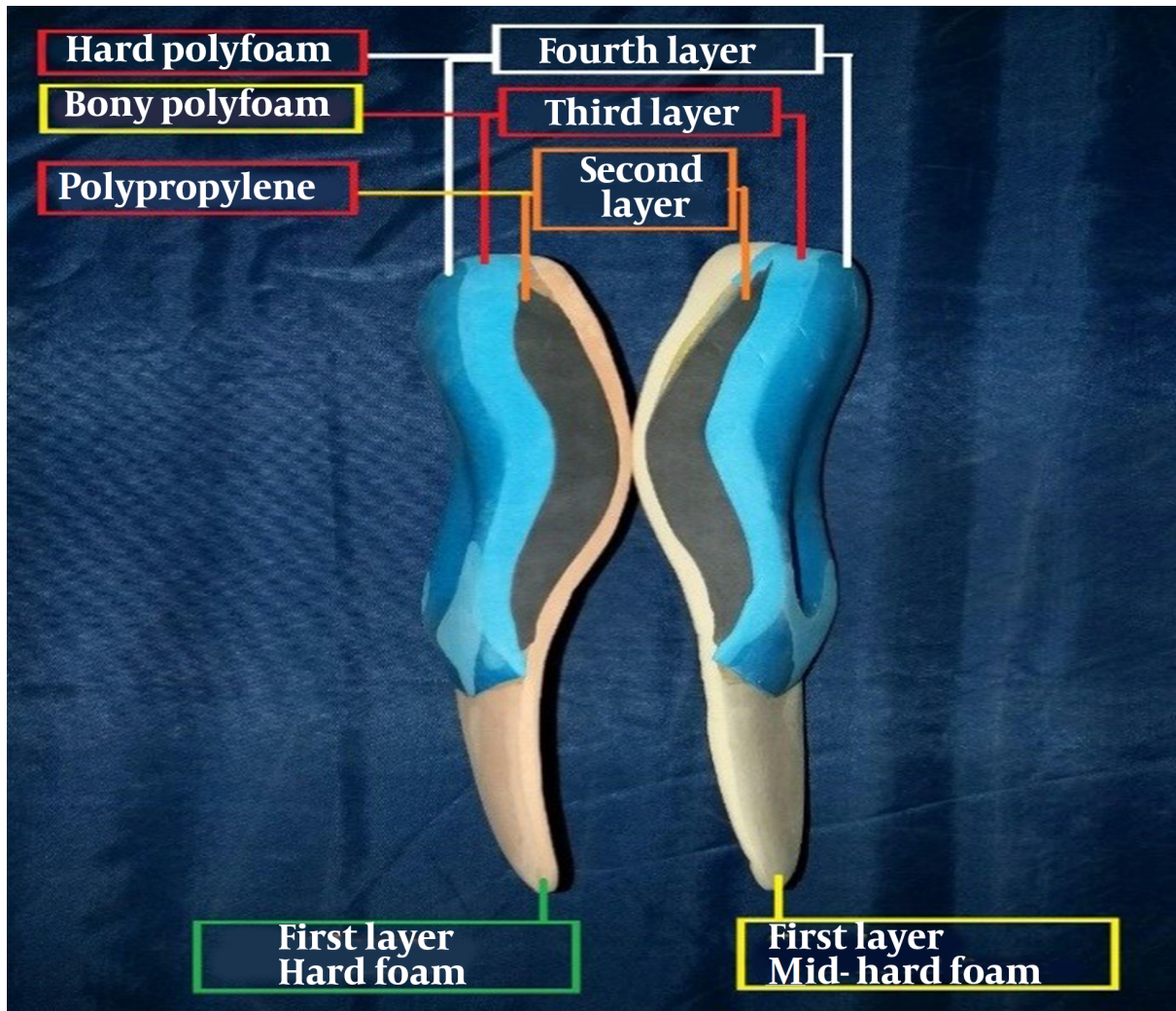
Figure 1. The right image of the semi-hard foot orthoses and the left-hand side of the hard foot orthoses with a four-layer profile used in this study.

point between the medial and lateral condyles of the femur. The center of the tibia was defined as the midpoint between the medial and lateral condyles of the tibia. The center of the talocrural joint was defined as the midpoint between medial and lateral malleoli. Knee flexion and valgus angles were calculated based on the GROOD and SUNTAY method (26). Maximum knee flexion and maximum knee valgus angles within landing were calculated for each trial during the landing phase. The fool landing phase was defined from the primary ground contact of the dominant foot until maintaining balance without motion and with no ground contact in the opposite foot for at least 2 seconds.

3.5. Statistical Analysis

The variables of interest were maximum knee valgus angle and maximum knee flexion angle during the whole phase of landing. The mean of the six measurements in each trial condition (without foot orthoses, semi-hard foot orthoses, and hard foot orthoses) was calculated. The mean values were used for the analysis. The Shapiro-Wilk test ($P > 0.05$) was used to check the normality of continuous variables. Continuous variables were presented using mean and standard deviation (SD). One-way repeated measures analysis of variance (ANOVA) was used to compare the three trials. Pairwise comparisons were made using Bonferroni correction. The η^2 was calculated as effect size. The η^2 value greater than 0.25 is considered as large effect size, while moderate and small effect size are defined as η^2

vale between, 0.09 and 0.01 and lesser than 0.01, respectively (27). Data was analyzed using the statistical package for social sciences (SPSS) software (IBM Inc, USA) version 22.

4. Results

4.1. Knee Valgus Angle

There was a significant difference in knee joint valgus angle between the three trials (P < 0.001). Post-hoc tests showed that knee valgus angle was significantly smaller in the mid-hard foot orthoses compared to hard foot orthoses (P < 0.001) or without orthosis trials (P < 0.001). Furthermore, the post hoc test revealed a significant difference in knee valgus angle between without foot orthoses and hard foot orthoses conditions (P < 0.001) (Table 1).

4.2. Knee Flexion Angle

There was a significant difference in knee flexion angle between the three trials (P < 0.001). Post-hoc tests revealed greater knee flexion angle in the mid-hard foot orthoses trial, compared to hard foot orthoses (P = 0.000) or without orthoses trials (P = 0.000). Furthermore, the post-hoc test revealed a significant difference in knee flexion angle between without foot orthoses and hard foot orthoses trials (P = 0.002) (Table 1).

5. Discussion

To the best of our knowledge, this study was the first research that assessed the effect of two different foot orthoses (mid-hard foot orthoses and hard foot orthoses) on knee valgus angle during single-leg drop-landing. As we hypothesized, the mid-hard foot orthoses resulted in a significant reduction in maximum knee valgus angle and a significant increase in maximum knee flexion angle during single-leg landing. These findings indicate that foot orthoses alter the kinematics of knee during single-leg landing.

5.1. Effects of Foot Orthoses on Knee Valgus Angle

The results of this study demonstrated that mid-hard foot orthoses reduced knee valgus angle during single-leg drop landing compared to hard foot orthoses or without foot orthoses trials. No study has yet compared lower limb kinematics during landing with different foot orthoses. Several studies have reported kinematic changes in lower extremity joints during walking and running due to foot orthoses (16, 19, 28-30).

Subjects wearing foot orthoses were shown to exhibit kinematic changes in the frontal (knee), and transverse (ankle and hip) planes. Foot orthoses were shown to reduce knee joint valgus (knee abduction) in a group of female basketball players (31). Previous studies showed that little changes in knee valgus angle could extremely alter knee valgus load. McLean et al. showed that a two-degree change in knee valgus angle resulted in 40 N.M enhancement in knee valgus moment (32). Furthermore, Chaudhari and Andriacchi reported that knee abduction (frontal plane position) as little as 2° has the potential to cause ACL injury (33). Our findings suggest that landing with mid-hard foot orthoses could decrease the valgus load on the knee through decreasing the knee valgus angle.

5.2. Effects of Foot Orthoses on Knee Flexion Angle

Our research also showed the maximum knee flexion angle was achieved when landing with a semi-hard foot orthosis. Previous studies showed that the maximum damage to ACL occurs at an angle close to full extension (8). Increase in knee flexion during sports activities reduces the force on the ACL. The anterior shear force is the main predicting factor of the load on ACL (12). When the knee flexion increases, the angle between the patella and tibial tendon increases, and the quadriceps contraction force produces less anterior shear force in the proximal tibia (34). Therefore, the increase in knee flexion is accompanied by a decrease in force on the ACL. On the other hand, the increase in knee flexion is accompanied by a reduction in the posterior ground reaction force. This finding is due to the direct relation between anterior shear force and posterior ground reaction force (35). Therefore, it can be concluded that the increase in knee flexion will reduce the force on the ACL. Southard et al. used a biomechanical model and compared biomechanical factors in both normal landing and landing with a bending knee and showed that tension on ACL is reduced by 10% in landing mode (36). Concerning findings of previous studies, the findings suggest that landing with a semi-hard foot orthosis can reduce tension on the ACL.

5.3. Conclusions

Enhanced knee valgus angle increased the risk of ACL damage during landing among athletes. Our study showed that mid-hard foot orthoses decreased KVA. The results of this study were based on a small number of subjects. There is need for more studies on the effectiveness of foot orthoses on kinematics in coronal and sagittal planes. A foot orthosis that has the potential to optimally change lower limb motion or neuromuscular recruitment models will be promising in decreasing the incidence of noncontact ACL damage in a male sportsman.

Table 1. Mean (SD) of Knee Valgus and Flexion Angles

Variables [a]	Without Foot Orthoses	Mid-hard Foot Orthoses	Hard Foot Orthoses	P-Value	η^2
Knee valgus (°)	11.34 (0.91) [a, b]	7.18 (1.27) [b, d]	9.31 (1.61) [a, d]	< 0.001	0.64
Knee flexion (°)	38.98 (2.05) [a, b]	48.78 (2.53) [b, d]	46.36 (1.94) [a, d]	< 0.001	0.79

[a] η^2 was measured as effect size. The cut-off values for large, moderate, and small effect size are 0.25, 0.09, and 0.01, respectively.
[b] Significant difference with mid-hard foot orthoses trial.
[c] Significant difference with hard foot orthoses trial.
[d] Significant difference with without foot orthoses trial.

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Footnotes

Authors' Contribution: Seyed Ali Abbassi Khaboushan designed the conception and design of the study. All authors provided critical feedback and helped shape the research, analysis, and manuscript. All authors discussed the results and contributed to the final manuscript. All authors confirmed final approval of the version to be submitted.

Conflict of Interests: All authors declare that they have not received any direct or indirect payment in any stage of this study and preparation and publication of this paper.

Ethical Approval: The process of this research has been approved by the bio-research ethics committee of Ferdowsi University of Mashhad with the ethics code IR.UM.REC.1399.059 (link: ethics.research.ac.ir/EthicsProposalView.php?id=151913).

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Informed Consent: We informed all participants verbally about the study, and those that agreed to their data being used signed a written informed consent form. Informed consent was obtained from the participants.

References

1. Dufek JS, Bates BT. Biomechanical factors associated with injury during landing in jump sports. *Sports Med.* 1991;**12**(5):326–37. [PubMed ID: 1763250]. https://doi.org/10.2165/00007256-199112050-00005.

2. McNair PJ, Prapavessis H, Callender K. Decreasing landing forces: effect of instruction. *Br J Sports Med.* 2000;**34**(4):293–6. [PubMed ID: 10953904]. [PubMed Central ID: PMC1724204]. https://doi.org/10.1136/bjsm.34.4.293.

3. Hargrave MD, Carcia CR, Gansneder BM, Shultz SJ. Subtalar Pronation Does Not Influence Impact Forces or Rate of Loading During a Single-Leg Landing. *J Athl Train.* 2003;**38**(1):18–23. [PubMed ID: 12937467]. [PubMed Central ID: PMC155506].

4. Bates NA, Ford KR, Myer GD, Hewett TE. Impact differences in ground reaction force and center of mass between the first and second landing phases of a drop vertical jump and their implications for injury risk assessment. *J Biomech.* 2013;**46**(7):1237–41. [PubMed ID: 23538000]. [PubMed Central ID: PMC3644482]. https://doi.org/10.1016/j.jbiomech.2013.02.024.

5. Myer GD, Ford KR, Hewett TE. New method to identify athletes at high risk of ACL injury using clinic-based measurements and free-ware computer analysis. *Br J Sports Med.* 2011;**45**(4):238–44. [PubMed ID: 21081640]. https://doi.org/10.1136/bjsm.2010.072843.

6. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med.* 2004;**32**(4):1002–12. [PubMed ID: 15150050]. https://doi.org/10.1177/0363546503261724.

7. Krosshaug T, Nakamae A, Boden BP, Engebretsen L, Smith G, Slauterbeck JR, et al. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med.* 2007;**35**(3):359–67. [PubMed ID: 17092928]. https://doi.org/10.1177/0363546506293899.

8. Boden BP, Dean GS, Feagin JJ, Garrett WJ. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* 2000;**23**(6):573–8. [PubMed ID: 10875418]. https://doi.org/10.3928/0147-7447-20000601-15.

9. Hewett TE, Torg JS, Boden BP. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *Br J Sports Med.* 2009;**43**(6):417–22. [PubMed ID: 19372088]. [PubMed Central ID: PMC4062295]. https://doi.org/10.1136/bjsm.2009.059162.

10. Koga H, Nakamae A, Shima Y, Iwasa J, Myklebust G, Engebretsen L, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med.* 2010;**38**(11):2218–25. [PubMed ID: 20595545]. https://doi.org/10.1177/0363546510373570.

11. Berns GS, Hull ML, Patterson HA. Strain in the anteromedial bundle of the anterior cruciate ligament under combination loading. *J Orthop Res.* 1992;**10**(2):167–76. [PubMed ID: 1740734]. https://doi.org/10.1002/jor.1100100203.

12. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* 1995;**13**(6):930–5. [PubMed ID: 8544031]. https://doi.org/10.1002/jor.1100130618.

13. Pappas E, Hagins M, Sheikhzadeh A, Nordin M, Rose D. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. *Clin J Sport Med.* 2007;**17**(4):263–8. [PubMed ID: 17620779]. https://doi.org/10.1097/JSM.0b013e31811f415b.

14. Yeow CH, Lee PV, Goh JC. Sagittal knee joint kinematics and energetics in response to different landing heights and techniques. *Knee.* 2010;**17**(2):127–31. [PubMed ID: 19720537]. https://doi.org/10.1016/j.knee.2009.07.015.

15. Finestone A, Novack V, Farfel A, Berg A, Amir H, Milgrom C. A prospective study of the effect of foot orthoses composition and fabrication on comfort and the incidence of overuse injuries. *Foot Ankle Int.* 2004;**25**(7):462–6. [PubMed ID: 15319103]. https://doi.org/10.1177/107110070402500704.

16. MacLean C, Davis IM, Hamill J. Influence of a custom foot orthotic intervention on lower extremity dynamics in healthy runners. *Clin Biomech (Bristol, Avon)*. 2006;**21**(6):623–30. [PubMed ID: 16603287]. https://doi.org/10.1016/j.clinbiomech.2006.01.005.

17. Johnston LB, Gross MT. Effects of foot orthoses on quality of life for individuals with patellofemoral pain syndrome. *J Orthop Sports Phys Ther*. 2004;**34**(8):440–8. [PubMed ID: 15373007]. https://doi.org/10.2519/jospt.2004.34.8.440.

18. Ferber R, Davis IM, Williams D3. Gender differences in lower extremity mechanics during running. *Clin Biomech (Bristol, Avon)*. 2003;**18**(4):350–7. [PubMed ID: 12689785]. https://doi.org/10.1016/s0268-0033(03)00025-1.

19. Mündermann A, Nigg BM, Humble RN, Stefanyshyn DJ. Foot orthotics affect lower extremity kinematics and kinetics during running. *Clin Biomech (Bristol, Avon)*. 2003;**18**(3):254–62. [PubMed ID: 12620789]. https://doi.org/10.1016/s0268-0033(02)00186-9.

20. Hertel J, Sloss BR, Earl JE. Effect of foot orthotics on quadriceps and gluteus medius electromyographic activity during selected exercises. *Arch Phys Med Rehabil*. 2005;**86**(1):26–30. [PubMed ID: 15640985]. https://doi.org/10.1016/j.apmr.2004.03.029.

21. Yu B, Preston JJ, Queen RM, Byram IR, Hardaker WM, Gross MT, et al. Effects of wearing foot orthosis with medial arch support on the fifth metatarsal loading and ankle inversion angle in selected basketball tasks. *J Orthop Sports Phys Ther*. 2007;**37**(4):186–91. [PubMed ID: 17469671]. https://doi.org/10.2519/jospt.2007.2327.

22. Montgomery DC. *Design and Analysis of Experiments*. 2nd ed. New Jersey, USA: John Wiley & Sons Inc; 1984.

23. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon)*. 2010;**25**(2):142–6. [PubMed ID: 19913961]. [PubMed Central ID: PMC2815098]. https://doi.org/10.1016/j.clinbiomech.2009.10.005.

24. Vaughan CL, Davis BL, O'connor JC. *Dynamics of human gait*. **2**. Illinois, USA: Human Kinetics Publishers Champaign; 1992.

25. Wu G, Siegler S, Allard P, Kirtley C, Leardini A, Rosenbaum D, et al. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion–part I: ankle, hip, and spine. International Society of Biomechanics. *J Biomech*. 2002;**35**(4):543–8. [PubMed ID: 11934426]. https://doi.org/10.1016/s0021-9290(01)00222-6.

26. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J Biomech Eng*. 1983;**105**(2):136–44. [PubMed ID: 6865355]. https://doi.org/10.1115/1.3138397.

27. Hanna D, Dempster M. Effect sizes and power. *Psychology statistics for dummies*. New Jersey, USA: John Wiley & Sons; 2012.

28. Brannigan AE, Kerin MJ, McEntee GP. Gilmore's groin repair in athletes. *J Orthop Sports Phys Ther*. 2000;**30**(6):329–32. [PubMed ID: 10871144]. https://doi.org/10.2519/jospt.2000.30.6.329.

29. McPoil TG, Cornwall MW. The effect of foot orthoses on transverse tibial rotation during walking. *J Am Podiatr Med Assoc*. 2000;**90**(1):2–11. [PubMed ID: 10659526]. https://doi.org/10.7547/87507315-90-1-2.

30. Nester CJ, van der Linden ML, Bowker P. Effect of foot orthoses on the kinematics and kinetics of normal walking gait. *Gait Posture*. 2003;**17**(2):180–7. [PubMed ID: 12633779]. https://doi.org/10.1016/s0966-6362(02)00065-6.

31. Joseph M, Tiberio D, Baird JL, Trojian TH, Anderson JM, Kraemer WJ, et al. Knee valgus during drop jumps in National Collegiate Athletic Association Division I female athletes: the effect of a medial post. *Am J Sports Med*. 2008;**36**(2):285–9. [PubMed ID: 17977999]. https://doi.org/10.1177/0363546507308362.

32. McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech (Bristol, Avon)*. 2005;**20**(8):863–70. [PubMed ID: 16005555]. https://doi.org/10.1016/j.clinbiomech.2005.05.007.

33. Chaudhari AM, Andriacchi TP. The mechanical consequences of dynamic frontal plane limb alignment for non-contact ACL injury. *J Biomech*. 2006;**39**(2):330–8. [PubMed ID: 16321635]. https://doi.org/10.1016/j.jbiomech.2004.11.013.

34. Nunley RM, Wright D, Renner JB, Yu B, Garrett WE. Gender Comparison of Patellar Tendon Tibial Shaft Angle with Weight Bearing. *Res Sports Med*. 2006;**11**(3):173–85. https://doi.org/10.1080/15438620390231193.

35. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. *Knee Surg Sports Traumatol Arthrosc*. 2003;**11**(5):307–11. [PubMed ID: 14523613]. https://doi.org/10.1007/s00167-003-0403-6.

36. Southard J, Kernozek TW, Ragan R, Willson J. Comparison of estimated anterior cruciate ligament tension during a typical and flexed knee and hip drop landing using sagittal plane knee modeling. *Int J Sports Med*. 2012;**33**(5):381–5. [PubMed ID: 22318556]. https://doi.org/10.1055/s-0031-1299750.