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Gender Differences in Lower Extremity Kinematics and Kinetics of the Vertical Ground Reaction Force Peak in Drop-landing by Flatfooted Subjects

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Abstract. [Purpose] The purpose of this study was to investigate the kinematics and kinetics of the vertical ground reaction force peak during a drop-landing by flatfooted males and females. [Subjects] Twenty subjects (ten male and ten female subjects) participated. [Methods] Subjects performed a drop-landing task from 40 cm, and knee, hip, and ankle kinematics and kinetics were recorded using 12 cameras, the Vicon motion system (Vicon, Oxford, England) and 2 force platforms. [Results] A significant difference in the hip, knee, and ankle joint angles was observed in the sagittal plane between the male and female groups. Significant differences in F(x) and F(z) were observed between males and females, but no significant difference was found for F(y). [Conclusion] A higher risk of not only an anterior cruciate ligament injury but also ankle ligament injury may arise from hyper-inversion flatfooted females, compared to flatfooted males, because of lower hip joint control.
Key words: Flatfoot, Gender, Kinematics, Kinetics, Landing

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INTRODUCTION

Landing with a load affects the foot, in contact with the ground, as well as the trunk and spine through the lower limbs[1)], and loads on the lower limbs may be 10 times or more the maximum pressure of body weight depending on the height of the drop[2)]. Drop-landing affects ground reaction force and muscle activation and can cause injury in various lower limb joints, ligaments and muscles as well as fatigue fracture, and chondral destruction[3, 4)]. If the drop-landing occurs in a state in which the lower limb joints are extended, it may cause anterior cruciate ligament (ACL) rupture[5, 6)].

According to a recent study of gender differences, the flexion angle is less in females; thus, more load is borne on the ACL, leading to a higher incidence of ACL injury in females[7)]. The body absorbs impacts by coordinating the joints to reduce musculoskeletal injury[8)]. When impact is not absorbed, the vertical ground reaction force (VGRF) becomes higher inflicting a greater load on the body leading to possible lower limb injury.

Usually, the muscles are controlled optimally and posture is maintained with a hip, knee, and ankle strategy to absorb impact and maintain balance. Because the feet are a distal part of the lower limbs and have a small support base, even a small mechanical change can affect postural control[9)]. Flatfoot is the most common general foot deformation. It causes muscle and myofascial tension, and increases internal rotation of the hip joint and lordosis of the lumbar spine in a closed chain posture. This postural change in the lumbosacral complex increases the risk of low back pain[10)], meaning additional efforts are required to control posture and maintain foot balance, because of the hypermobility of the midfoot. These indicate that flatfooted subjects have reduced dynamic balance through malalignment of the body[11)].

Studies of reduced balance control ability arising from flatfeet have been conducted, but a biomechanical study has not been performed of the VGRF peak of flatfooted subjects in drop-landings, which take place frequently. Therefore, the purpose of this study was to investigate the kinematics and kinetics of the VGRF peak in drop-landings performed by flatfooted males and females.

SUBJECTS AND METHODS

The subjects were ten males (age, 23.1±2.0 years; height, 173.2±4.1 cm; Weight, 69.6±7.2 kg; navicular drop, 13.2±2.0 mm) and ten females (age, 22.7±2.2 years; height, 161.3±5.3 cm; Weight, 57.9±5.2 kg; navicular drop, 12.4±1.2 mm) whose feet were classified as flat by radiological and physical examinations. The subjects had not undergone any operation on the lower limbs, feet, or ankles. If the difference in the navicular tuberosity height of the feet was 10 mm or more following the navicular drop test, the feet were classified as flat. The navicular drop test was devised by Brody and measures the height of the navicular tuberosity on the right foot from the floor between resting and neutral standing position[12]. For the test, subjects sat with both feet on the floor and hip, knee, and ankle joint flexed at 90 degrees without weight bearing, and the rater measured the height of the navicular tuberosity from the floor. Then the subject was asked to stand with full weight bearing, and the height of navicular tuberosity on the foot from the floor was measured. The navicular drop was calculated as the distance between the navicular height with and without weight bearing[11]. The navicular drop test has excellent intra-rater (0.90–0.99) and inter-rater (0.85–0.96) reliabilities[13, 14]. The subjects were given sufficient explanation about the purpose and method of the study before participating in the experiment and willingly consented to participation.

A 100 × 50 cm wood foot rest was prepared for the drop-landing, which was performed from a box higher than the ground to the floor. The height of the box was 40 cm. For the drop-landing, the subjects were asked to stand with their feet shoulder width apart, looking straight ahead, and to land with both feet touching the two force plates simultaneously. The subjects were given sufficient practice in performed the task, and, after taking a 5minute rest, they performed the drop-landing three times. Data were analyzed using the mean values of the three measurements. The maximum joint angles at the hip (between the pelvis and the thigh), knee (between the thigh and the shank), and ankle (between the lower leg and the foot) joint in the sagittal plane (flexion/extension) were extracted from the recorded data. By convention, zero angle at each joint in the sagittal plane corresponds to the standing posture with the trunk, thigh and lower leg in a straight line. The GRF and joint angle were measured during the drop-landing.

We used 12 Vicon MX-F40 infrared cameras, and the Vicon motion system (Vicon, Oxford, England), comprising a data station, a control PC, and 25 mm luminescent markers, and two force plates (AMTI, Advanced Mechanical Technology, Watertown, USA) to provide the data for the kinematic and kinetic analysis of the lower limbs. The luminescent markers were attached to the pelvis and segments of the lower limbs. Fifteen were attached to the lower limbs according to the plug-in gait marker set for a kinematic segment axis model. The Vicon motion system computes not only the positional data of the individual markers but also individual segment values. Two-dimensional images are captured by the individual cameras at 120 Hz, and a three-dimensional image is reconstructed using Woltering filtering. Mechanical analysis of the individual joints is made possible by using Euler's method to derive the joint angles. The two force plates measured the GRF during the drop-landings, and the GRF was normalized to each subject's body weight. The GRF data is composed of the vertical (z-axis), anterior/posterior (y-axis), and medial/lateral (x-axis) force components. The maximum GRF and torque were recorded on the computer at 1000 Hz. The analog signal was input through the Ultranet system, which synchronizes data from the cameras and force plates, and the data were combined using the Vicon Nexus software. All motion data were analyzed using Polygon software (Vicon).

Table 1. Peak joint angles in the sagittal plane of the lower extremity at peak vertical ground reaction force (GRF) (°)

	Male	Female
Hip (flexion)*	46.3 ± 11.5	36.3 ± 4.7
Knee (flexion)*	74.9 ± 6.2	67.1 ± 8.5
Ankle (dorsiflexion)*	32.0 ± 4.6	39.0 ± 4.3

*p<0.05

Table 2. Ground reaction force at peak vertical ground reaction force (GRF) during drop-landing

	Male	Female
F(x) *	24.7 ± 12.2	15.3 ± 6.0
F(y)	24.5 ± 13.6	16.6 ± 14.2
F(z) *	367.9 ± 40.0	404.1 ± 33.8

*p<0.05; F(x), medial-lateral direction of the GRF; F(y), anterior-posterior direction of the GRF; F(z), vertical direction of GRF

The collected data are presented as the mean (SD), and were statistically processed using SPSS 15.0 software. The independent *t*-test was performed to compare the changes in kinematics and the GRF values depending on gender after confirming normality with the Shapiro–Wilks test. A p value <0.05 was considered significant.

RESULTS

The kinematic results during the drop-landing are shown in Table 1. A significant difference in the peak hip, knee, and ankle joint angles was observed in the sagittal plane between the male and female groups ($p < 0.05$) (Table 1). The peak hip and knee joint angles of the male group were greater than those of the female group in the sagittal plane, but the female group showed a greater peak ankle joint angle in the sagittal plane than the male group.

The kinetic results during drop-landing are shown in Table 2. A significant difference in F(x) and F(z) was observed between the male and female groups ($p < 0.05$), but no significant difference was found for F(y) (Table 2). F(x) for the male group was greater than that for the female group, but F(z) was greater for females than for males.

DISCUSSION

During a landing, the lower limb joint functions to reduce and control momentum by flexion. If the lower limb flexion is not properly implemented, the GRF increases and a greater load is borne by the lower limb joints[15]. In this study, the GRF and the joint angles of the lower limbs at the GRF peak were determined when subjects performed a drop-landing from a height of 40 cm. A study in which the range of motion was measured when normal adults performed a drop-landing reported that the flexion angle of the knee joint is smaller for females than for males; thus, knee joint injury should occur more frequently in females[16]. According to the results of our study, the male subjects showed a greater knee joint angle when absorbing the impact than the female subjects.

A study of the role of the hip joint in the sagittal plane during drop-landing reported that hip joint flexion brings the center of mass closer to the base of support (BOS)[17]. In our study, the hip and knee joint flexion angles were larger for the male subjects than for the female subjects, but the ankle joint flexion angle was larger for the females than for the male subjects during the drop-landing. The hip and knee joints of the females were more extended during the impact, therefore, the ACL had to bear a greater loads. In terms of neuromuscular function, muscle activation of the hamstring is inadequate in females[7]. A more erect landing strategy is mechanically disadvantage to the hamstring and the quadriceps pull the tibia with greater ACL tension[16, 18], so females are at greater risk of ACL injury during drop-landing. Also, weakness of the posterior tibialis in flatfooted subjects results in loss of mechanical force, decreasing stability of the medial longitudinal arch, and weakening the ankle joint[19]. Because flatfooted subjects have weakness of the posterior tibialis, postural balance is effected by ankle plantarflexion, resulting in an unstable drop-landing posture. Our results are consistent with the results of Horak et al.[20] who reported that elderly subjects maintain balance using hip joint movement, and, if it fails, an overload is borne by the ankle joint. Flatfooted males with a weak ankle stabilizer may maintain balance using the proximal muscles at the GRF peak[21]. However, action of the proximal muscles in females may not occur rapidly enough the deliver quick movement of the center of gravity (COG), resulting in excessive ankle joint movement.

The GRF values also supportour interpretation of the kinematic results. The VGRF (z-axis) was higher for the flatfooted females than for the flatfooted males, indicating that the load was conveyed more vertically to the body of female subjects as they used less hip and knee joint flexion than the male subjects, so the load was not be distributed and increased in specific regions of the foot[22]. The medial/lateral GRF (x-axis) show that more load was conveyed to the medial part of the foot in males than in females. This means that the flatfooted males made the BOS wider for balance control when performing drop-landings than the flatfooted females, who had more difficulty with balance control. A landing maneuver of medially directed GRF usually promotes hip adduction and knee and ankle abduction to absorb the impact[23]. These findings of differences between males and females may be explained by different strengths of ligaments and muscles possibly, arising from hormonal differences, lower initial fitness, and anthropometric factors[24]. It has also been reported that there are gender differences in the neuromuscular responses of the quadriceps and hamstring, the quadriceps activation in females being earlier and rising more rapidly than the hamstring at relatively small knee flexion angles[25].

The control of balance explains the mechanism by which a flatfooted subject controls him/herself during a drop-landing. Because control of the ankle joint is lower than that of normal adults, the GRF of a drop-landing is decreased, primarily due to control by the quadriceps, hamstring and trunk muscles at peak GRF to increase body stability. We assume this lower extremity configuration at peak GRF during a drop-landing is the strategy adopted by flatfooted subjects to prevent ACL injury due to excessive load or ankle injury due to ankle hyper-inversion, since they have muscle weakness around the ankle. Therefore, flatfooted females are at higher risk not only of ACL injury but also ankle ligament injury through hyper-inversion than flatfooted males because of their lower control of the hip joint. Therapists should consider evaluating and treating females with flatfoot because of their increased risk of physical injury.

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