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ORIGINAL

CUSHIONING OF VERTICAL JUMPS ACCORDING TO THE PURPOSE OF POSTERIOR SPORT MOVEMENT

AMORTIGUACIÓN DE LOS SALTOS VERTICALES SEGÚN EL PROPÓSITO DEL MOVIMIENTO DEPORTIVO POSTERIOR

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ABSTRACT

The main objective of this study was to identify biomechanical differences among three landing styles: a) discrete landing (DL) from a drop jump; b) preparatory landing preceding jumping (LPJ), and c) landing preceding running (LPR). The sample was composed of 29 athletes who perform jumps routinely. Sagittal plane jump parameters were recorded using a camera synchronized with a force plate. Peak reaction forces were attenuated more efficiently in LPJ and LPR, as compared to DL. The kinematic model used for LPJ was similar to that for DL. Yet, neural and reflex activity during muscle flexion-extension during DL makes impact absorption more efficient. LPR is associated with an increased risk for injury as a result of the forward displacement of the tibia.

KEYWORDS: Biomechanics, drop jump, landing, forces

RESUMEN

El principal objetivo ha sido detectar las diferencias biomecánicas de tres modelos de aterrizaje: a) aterrizaje discreto (AD), con misión de amortiguar un salto vertical, b) aterrizaje preparatorio al salto vertical (APS) y c) aterrizaje que precedía a una carrera (APC). Han participado 29 deportistas donde el salto constituye una habilidad básica. Se ha utilizado una plataforma de fuerza sincronizada a una cámara de vídeo que registraba el plano sagital de los saltos. Los resultados indican que los APS y APC amortiguan más los picos de fuerza de reacción vertical que los AD. El modelo cinemático utilizado para los APS fue similar al AD, aunque la mayor actividad neural y refleja propia del ciclo estiramiento-acortamiento para realizar el posterior salto vertical haría más eficiente la absorción de impactos. La estrategia utilizada para los APC muestra un incremento del riesgo de lesión como consecuencia del desplazamiento hacia delante de la tibia.

PALABRAS CLAVE: Biomecánica, Salto vertical, aterrizaje, fuerzas.

INTRODUCTION

The strategies used to attenuate ground reaction forces during landing in sports such as basketball, volleyball or soccer have been the object of intensive research. Some authors have focused on the prevention or treatment of injuries (Cortes et al., 2007; Lobiatti, Coleman, Pizzichillo & Merni, 2010; Decker, Torry, Wyland, Sterett & Steadman, 2003; Rowley & Richards, 2015). Other authors have analyzed impact absorption mechanisms to improve post-landing task performance (Gutiérrez-Dávila, Campos y Navarro, 2009; Peng, 2011; Waller, Gersick, & Holman, 2013; Gutiérrez-Dávila, Giles, González, Gallardo & Rojas, 2015).

Caster (1993) classified landings in sports into two groups: a) *Discrete landing* defined as a phase subsequent to an airborne activity, which purpose is attenuating force impact; and b) *Preparatory landing* defined as a phase prior to an airborne activity. The purpose of this landing style is activating the muscle flexion-extension cycle or canceling out the impulse of the vertical speed of landing to create forward momentum for acceleration. We analyzed the effects of three types of landings from a 0.5m drop jump — a) Discrete landing (DL); b) preparatory landing followed by a drop jump (PL); and c) landing followed by a run (LR)— on ground reaction forces (GRF), joint goniometry, and contribution of body segments to the vertical displacement of the center of mass (CM).

Research has consistently shown that athletes may adopt unsafe individual GRF attenuation strategies in an attempt to improve subsequent task performance. These strategies may increase the risk for injury. Thus, there is evidence that alterations in body segment positions and knee angular velocities

during landing may affect the magnitude of momentum, which causes joint stress (Cowling y Steele, 2001; Decker et al., 2003; Lacquaniti, 1992; Zhang, Bates y Dufek, 2000). The magnitude of force momentum is determined by ground reaction forces (GRF) and the position adopted by body segments. Hence, increased GRFs and joint positions far from the optimal trajectory of CM acceleration would increase force momentum. The anterior cruciate ligament (ACL) prevents the forward displacement of the tibia. High GRF and excessive forward gliding of the tibia increase tibial acceleration during landing from a vertical jump and put strain on the ACL (McNair & Marshall, 1994; Ericksen, Gribble, Pfile & Pietrosimone, 2013).

Kulig, Fietzer, & Popovich (2011) analyzed GRF and knee mechanics during a specific leap in ballet, where aesthetic constraints determine the landing style. The authors concluded that aesthetic constraints influence impact absorption strategies and increase joint stiffness during weight acceptance. These two mechanisms reduce angular displacement during landing and increase force momentum and joint structural stress.

Studies on neuro-muscular events have revealed the presence of neural and reflex mechanisms that modulate musculo-skeletal stiffness before and after landing (Sampello, 2005). Muscles have been found to preactivate (coactivate) shortly before initial contact, which predisposes the musculo-skeletal system to absorb the impact more efficiently. The timing of coactivation and level of preactivation are regulated by predictive control mechanisms. The levels of coactivation and preactivation increase linearly based on the expected impact force magnitude (Sampello, McDonagh y Challis, 2001; Sampello, 2005). These mechanisms are also involved in the activation of the flexion-extension cycle that starts in preparatory landing prior to a jump (Gollhofer and Kyröläine, 1991; Komi, 1992). During impact absorption, extension reflexes and voluntary muscle activity remain activated to modulate joint angles to the expected impact force. Then, muscle stiffness is regulated for a safe GRF absorption (Sampello, 2005).

Impact force attenuation is not only modulated by the position of the knees and ankles (Decker et al., 2003), but it requires the coordination of the whole body (McNitt-Gray, 2000). There is evidence that proper arm coordination reduces peak force at initial contact, increases braking distance and improves stability during landing (Niu, Zhang and Zhao, 2013; Pancorbo, Olivares Rojas & Gutiérrez-Dávila 2016). A better understanding of body segment contribution to landing would provide a picture of the segmental coordination required for impact force absorption. The role of each body segment could be determined based on their percentual contribution to the vertical displacement of the CM.

No conclusive data have been provided on the influence of the post-landing task on the risk for injury. Two contradictory hypotheses have been proposed: a) the post-landing task influences the landing model. Thus the preparatory position of some body segments augments GRF momentum, which increases the risk for injury; b) some post-landing tasks require the intensification of neural

control activity to reduce peak GRF during landing. A better understanding of the influence of post-landing tasks would contribute to the prevention of injuries.

METHODS

Participants

The study involved 29 male students of a School of Sports Sciences (age = 21.1 ± 1.7 years; height = 1.78 ± 0.06 m; mass = 70.6 ± 8.1 kg). Inclusion criteria were being a basketball, handball or soccer team player, as vertical jumps are a basic skill in these sports. In accordance with the guidelines of the Ethics Committee of the University of Granada, informed consent was obtained from all participants.

Materials and methods

The sagittal plane of jumps performed on a force plate was recorded via a Casio EX - FH20 camera set at 210 Hz synchronized with a force plate Dinascan/IBV set at 500 Hz. Participants performed a protocolized 10-min warm-up involving continuous run, active stretching exercises and vertical jumps. Then, participants were asked to perform three different styles of landing from a height of 0.5 m, namely:

- a) *discrete landing (DL)*: from an upright position, participants jumped onto the force plate with their arms straight over the head; arms moved freely during the phases of flight and impact absorption;
- b) *preparatory landing before a drop jump (LPJ)*, where participants had to perform a vertical jump at maximum height immediately after a drop jump; and
- c) *preparatory landing before a run (LPR)*, where participants were asked to perform a rapid horizontal displacement after landing from a drop jump.

Some practice trials were performed for each of the three experimental conditions and for adjustment of the motion capture system. Based on the Schmidt & Lee (2011) protocol, test trials were performed when total impact absorption times were consistent among trials. Total impact absorption time was defined as the period between initial contact and the instant at which the vertical velocity of the CM reached positive values. Five test trials were performed for each experimental condition. There was a 2 min rest period between trials and 10 min between experimental conditions. We only analyzed the values obtained in the trial where total absorption time was the median of the five valid trials. Experimental conditions were randomly presented to participants.

Data Analysis

Potential systematic errors of the force platform were determined from the average value of horizontal and vertical components ($F_{R(X)}$ and $F_{R(Y)}$, respectively) during the 20 records preceding initial contact. After subtracting the potential systematic error and participant's body weight, the two peak vertical GRF values reached at initial contact were recorded (1-vPGRF and 2-vPGRF).

Horizontal and vertical acceleration of the CM were determined from $F_{R(X)}$ and $F_{R(Y)}$ and participant's body mass. Finally, we determined CM velocities ($v_{(X)CM}$ and $v_{(Y)CM}$, respectively) and positions ($S_{(X)CM}$ and $S_{(Y)CM}$, respectively). Thus, the vertical and horizontal velocity components were determined by integration of acceleration-time and velocity-time functions using the trapezoidal method with a time increase of 0.002 s. Integration constants were derived from manual digitization of the video, as follows:

To simplify the process of manual digitization of the video, landings were considered a symmetric movement. Thus, calculations were based on sagittal-plane coordinates for the 11 sensors representing a simplified 8-segment human model. Sensors were attached to the anterior and posterior parts of the foot; ankle, knee, hip, elbows and wrist joints; the third metacarpal, the suprasternal gap, the maxillary joint and the vertex. Sensor position was determined by the model and inertial parameters proposed by Zatsiorsky & Seluyanov (1985) and adapted by Leva (1996). In this simplified model, the mass of the segments representing the limbs was calculated as the sum of the mass of the two corresponding segments. Following digitization, sagittal coordinates of the sensors were filtered at 8 Hz (Winter 1990) and interpolated at 500 Hz using fifth-order smoothing splines. Conversion of digital coordinates into real data was performed by a reference system composed of a 2x2x0.5 m cube. The constants of integration for CM position at initial contact with the force plate (between two frames) were determined with respect to the geometrical center of the feet when lying flat on the force plate. Components of the instantaneous velocity of the CM were determined from the first derivative of the vertical position of the CM with respect to time.

Successive angular positions of hip, knee and ankle joints were calculated from the scalar product of the vectors that defined the position of their respective segments. The foot segment was defined as the vector from the lateral malleolus to the end part of the foot. Following the methodology proposed by Gutiérrez-Dávila, Garrido, Amaro, Ramos & Rojas (2012), these coordinates were used to determine the contribution of each segment to the vertical displacement of the CM.

Time analysis was performed based on Schot (1994) and Lees (1981) contributions. According to Schot, the phase of impact (PI) was defined as the first 100 ms from first contact with the force plate. According to Lees, the phase of impact absorption (PIA) was defined as the first 150 to 200 ms. Based on these definitions, PIA was defined as the first 200 ms elapsed from first contact with the force plate. The first 100 ms were considered the PI.

Statistical analysis

An analysis of variance of repeated measures was performed to confirm test reliability in the three experimental conditions (five valid trials per condition). For such purpose, total absorption time was used as the dependent variable. No statistically significant differences were observed. The intraclass correlation coefficient (CCI) for this variable was 0.81 ($p < 0.001$) for DL; 0.92 ($p < 0.001$) for LPJ; and 0.93 ($p < 0.001$) for LPR. Means and SDs were calculated for each variable in each experimental condition. Data were normally distributed and intraclass correlation coefficient was high. Repeated-measures analysis of variance (ANOVA) was performed. Differences between groups were detected by multivariate analysis. A $p < 0.05$ was considered statistically significant. All statistical analyses were performed using the Statgraphic Plus 5.1 package for Windows.

RESULTS

Table 1 includes descriptive and inferential statistics for general variables related to impact force attenuation in the three experimental conditions. Differences were observed in peak GRF (2-PvGRF) expressed in relation to body weight ($p < 0.01$). Hypothesis tests applied to this variable revealed that peak force was higher in discrete landings as compared to preparatory landings. The mean time required to reach 2-PvGRF ($t_{2-PvGRF}$) was higher in landings preceding a jump (LPJ), as compared to the other two conditions ($p < 0.01$). During practice trials, the CM was higher in discrete landing, as compared to preparatory landings ($Y_{CM(P)}$; $p < 0.5$). As expected, the CM was more advanced in landings that precede running ($X_{CM-(FIRST CONTACT)}$; $p < 0.001$). Positive values in this variable suggest a forward displacement of the CM with respect to the geometric center of the feet when lying flat on the force plate. The opposite occurs when values are negative.

Table 1.-Descriptive and inferential statistics for general variables related to landing impact absorption in the three experimental conditions. Force values are expressed in newtons by Kg of body mass.

Variables	Discrete (DL)	Preceding jumping (LPJ)	Preceding running (LPR)	F
2-PMF (N/N)	4,941 ± 1,344	4,157 ± 1,211 ¹	3,936 ± 0,886 ¹	13,17**
Time 2-PvGRF $t_{(2-PvGRF)}$ (s)	0,050 ± 0,015	0,054 ± 0,015	0,066 ± 0,020 ^{1,2}	9,79**
Positions $Y_{CM(INITIAL CONTACT)}$ (m)	1,154 ± 0,041	1,136 ± 0,048 ¹	1,132 ± 0,053 ¹	3,35*
Position $X_{CG(INITIAL CONTACT)}$ (m)	-0,068 ± 0,018	-0,042 ± 0,029	0,110 ± 0,078 ^{1,2}	111,27***
Angle $\theta_{Ankle-(INITIAL CONTACT)}$ (°)	134 ± 7	132 ± 11	130 ± 8	2,93
Angle $\theta_{KNEE-(INITIAL CONTACT)}$ (°)	159 ± 6	154 ± 7 ¹	157 ± 7	4,05*
Angle $\theta_{HIP-(INITIAL CONTACT)}$ (°)	157 ± 6	151 ± 8 ¹	159 ± 9 ²	11,83***

*** $p < 0.001$; ** $p < 0.01$; * $p < 0.05$

Hip, knee and ankle angular positions at the approximate instant of initial contact with the force plate are also shown in this table. Some differences were observed between means in the angular position of the knee ($\theta_{KNEE-(INITIAL CONTACT)}$; $p < 0.05$). Multivariate analysis revealed that the mean knee angle was smaller in LPJ as compared to discrete landing (DL). Analysis also showed clear differences in the angular positions of the hip joints ($\theta_{HIP-(INITIAL CONTACT)}$; $p < 0.001$). The mean hip angle was smaller in LPJ as compared to the other landing modalities.

Table 2 shows descriptive and inferential statistics of the components of displacement of the CM during the impact and absorption phase (IP and AP, respectively). PI and PIA occur during the first 100 ms and 200 ms following initial contact with the force plate, respectively. Slightly significant differences were observed in vertical displacement of the CM during the impact phase between LPJ and the other landing modalities ($Sy_{CM(100)}$ $p < 0.05$). At the end of the absorption phase, differences increased ($Sy_{CM(200)}$, $p < 0.001$), with vertical displacement of the CM being shorter in LPR as compared to the other landing modalities.

As expected, the mean horizontal displacement of the CM was greater at the end of the impact and absorption phase in LPR ($Sx_{CM(100)}$, $Sx_{CM(200)}$, respectively ($p < 0.001$). A greater horizontal displacement compensates the shorter vertical displacement that occurs in LPR. As a result, braking distance was shorter in DL.

Table 2.-Descriptive and inferential statistics for displacement of the CM in the sagittal plane, and angular displacements of ankle, knee and hip joints at two time points (100 and 200 ms after initial contact with the force plate). Velocities reached by the CM at these two time points.

Variables	Discrete (DL)	Preceding Jumping (LPJ)	Preceding Running (LPR)	F
Displacement $Sy_{CM(100)}$ (m)	0,248 ± 0,023	0,262 ± 0,024 ¹	0,250 ± 0,025 ²	7,71*
Displacement $Sy_{CG(200)}$ (m)	0,380 ± 0,055	0,400 ± 0,053	0,338 ± 0,058 ^{1,2}	19,81***
Displacement $Sx_{CM(100)}$ (m)	0,034 ± 0,006	0,035 ± 0,005	0,051 ± 0,006 ^{1,2}	34,69**
Displacement $Sx_{CM(200)}$ (m)	0,037 ± 0,015	0,047 ± 0,016	0,154 ± 0,063 ^{1,2}	64,35***
$\Delta\theta_{ANKLE(100)}$ (°)	49 ± 8	48 ± 10	53 ± 9 ^{1,2}	4,84*
$\Delta\theta_{ANKLE(200)}$ (°)	51 ± 9	54 ± 11	54 ± 11	1,04
$\Delta\theta_{ANKLE(100)}$ (°)	57 ± 7	55 ± 6	47 ± 6 ^{1,2}	20,45***
$\Delta\theta_{KNEE(200)}$ (°)	80 ± 8	81 ± 7	70 ± 9 ^{1,2}	11,52***
$\Delta\theta_{HIP(100)}$ (°)	36 ± 7	36 ± 8	26 ± 11 ^{1,2}	4,84*
$\Delta\theta_{HIP(200)}$ (°)	67 ± 9	70 ± 10	63 ± 12 ^{1,2}	5,66*
Velocity $vy_{CM(100)}$ (ms⁻¹)	-1,746 ± 0,303	-1,810 ± 0,287	-1,516 ± 0,365 ^{1,2}	14,59***
Velocity $vy_{CG(200)}$ (ms⁻¹)	-0,947 ± 0,356	-0,777 ± 0,429 ¹	-0,458 ± 0,261 ^{1,2}	23,45***
Velocity $vx_{CM(100)}$ (ms⁻¹)	0,171 ± 0,095	0,223 ± 0,096	0,755 ± 0,334 ^{1,2}	64,07***
Velocity $vx_{CM(200)}$ (ms⁻¹)	-0,055 ± 0,104	0,081 ± 0,126	1,308 ± 0,654 ^{1,2}	103,97***

*** $p < 0,001$; ** $p < 0,01$; * $p < 0,05$

The results obtained show that the shorter vertical displacement of the CM observed in LPR is associated with a shorter angular displacement of knee joints ($\Delta\theta_{KNEE(100)}$, $\Delta\theta_{KNEE(200)}$; $p < 0.001$) and hip ($\Delta\theta_{HIP(100)}$, ($\Delta\theta_{HIP(200)}$; $p < 0.05$), as compared to the other landing modalities. Table 2 displays velocities at the end of the impact (100 ms) and absorption phase (200 ms). Significant differences were found in the vertical velocity of the CM at the end of the two phases ($vy_{CM(100)}$, $vy_{CM(200)}$; $p < 0.001$). The lowest vertical velocity occurred in LPR ($vy_{CM(100)}$, $vy_{CM(200)}$), as compared to the other two landing modalities. In contrast, horizontal velocities ($vx_{CM(100)}$ and $vx_{CM(200)}$; $p < 0.001$) were higher in LPR, which compensates the lower vertical velocity of the CM in LPR.

Table 3 contains an analysis of the contribution of each body segment until the end of the impact (100 ms) and absorption (200 ms) phase, expressed in percentages of vertical displacement of the CM. In general, data suggest that thighs and calves are the body segments that contribute the most to impact force attenuation in the three types of landing at the end of the two time periods. Inferential statistics shows significant differences in the contribution of the trunk+head, calves and thighs at the end of the two time periods. According to hypothesis tests, differences only remained significant between LPR and the other two landing modalities. Differences between LPR and the other types of landing included: a) The mean contribution of trunk+head was greater at the end of the impact phase ($p < 0.01$); this difference increased at the end of the absorption phase ($p < 0.001$); b) the contribution of the thighs was smaller at the two time periods ($p < 0.001$); and c) The contribution of the calves was greater at the two time periods ($p < 0.001$).

Table 3.- Descriptive and inferential statistics for body segment contribution in the three experimental conditions. Data are expressed as percentages of the displacement of the CM in the two time periods (100 and 200 ms after initial contact with the force plate).

Variables	Discrete (DL)	Preceding Jumping (LPJ)	Preceding Running (LPR)	F
Carms ₍₁₀₀₎ (%)	7,94 ± 2,74	8,08 ± 2,61	7,88 ± 3,12	0,05
Carms ₍₂₀₀₎ (%)	8,88 ± 3,68	9,43 ± 2,81	7,38 ± 2,82	1,33
Ctrunk+head ₍₁₀₀₎ (%)	7,70 ± 4,04	9,89 ± 5,39	11,37 ± 5,51 ^{1,2}	5,46**
Ctrunk+head ₍₂₀₀₎ (%)	13,89 ± 4,86	15,64 ± 4,52	21,82 ± 6,03 ^{1,2}	20,43***
Cthighs ₍₁₀₀₎ (%)	29,09 ± 9,60	27,74 ± 9,13	10,91 ± 8,86 ^{1,2}	45,84***
Cthighs ₍₂₀₀₎ (%)	35,31 ± 5,41	34,86 ± 5,68	20,03 ± 8,33 ^{1,2}	27,00***
Clegs ₍₁₀₀₎ (%)	33,04 ± 8,14	32,56 ± 8,49	47,00 ± 13,00 ^{1,2}	28,57***
Clegs ₍₂₀₀₎ (%)	28,21 ± 6,91	28,68 ± 5,76	41,22 ± 10,46 ^{1,2}	18,29***
Cfeet ₍₁₀₀₎ (%)	23,45 ± 7,29	23,85 ± 8,15	23,88 ± 7,68	0,06
Cfeet ₍₂₀₀₎ (%)	13,41 ± 4,46	12,94 ± 5,37	9,81 ± 5,03 ^{1,2}	6,81**

*** $p < 0,001$; ** $p < 0,01$; * $p < 0,05$

DISCUSSION

In sports, the primary function of landing is attenuating ground reaction forces during initial contact. Impact force attenuation can be evaluated based on the peak GRF during heel contact (2-PvGRF) (Cámara, Calleja-González, Martínez, & Fernández-López, 2013; Decker et al., 2003; Rojano, Rodríguez & Berral, 2013).

Force attenuation is important as it is associated with injury (Eriksen et al., 2013; Chappell, Yu, Kirkendall & Garret, 2002; McNair, Prapavessis & Callender, 2000). Surprisingly, peak vertical ground reaction forces (PvGRF-2) are better absorbed in preparatory landing, as compared to discrete landing, which only purpose is absorbing the impact.

The fact that force attenuation is lower in DL might be explained by kinematics. Thus, in DL the braking distance is reduced to obtain a similar reduction of velocity at the end of the impact phase. Nevertheless, impact absorption might be more efficient in preparatory landings as a result of the greater muscle preactivation required for initiating the subsequent muscle flexion-extension cycle (Cavagna, Dusman & Margaria, 1968; Gollhofer & Kyröläine, 1991; Komi, 1992). Additionally, LPJ and LPRA involve greater neural and reflex activity during impact absorption. This activity is increased to adapt joint angles to the post-landing movement (Sampello, 2005).

This hypothesis has been consistently confirmed for landing preceding jumping (LPJ) due to its similarity with the countermovement of vertical jumps. Yet, evidence supporting this hypothesis in the context of jumping preceding running (JBR) is less solid. A very different impact absorption strategy has been documented for jumps followed by a subsequent task. In LPR, the upper-body rotates externally about an axis that passes through the ankle joint during most of the impact phase and shifts to the medium foot to absorb the vertical impact. This strategy was documented by Ridderikhoff, Batelaan & Bobbert (1999) and Gutiérrez-Dávila, Amaro, Garrido & Rojas, (2014) for horizontal jumps. The data shown in Table 2 support this strategy, with a greater angular displacement of the ankle at the end of the impact phase.

In JBR, the center of mass migrates forward during ground contact. Also, the horizontal displacement of the CM at the end of the impact and absorption phase was greater in JBR than in the other landing modalities. This position and displacement of the CM during impact absorption reduces the effect of vertical ground reaction forces to the magnitude of the 2-PvGRF and cause a delay of the 2-PvGRF (see tables 1 and 2). In LPR, vertical forces are attenuated and the CM shifts forward to start running before impact absorption is completed. Published data on angular displacement in DL and LPJ confirm that impact absorption during the impact phase relies first on the knees and then on the ankles, which is consistent with the contributions of Decker et al. (2003). In contrast, this order is reversed in LPR, where impact is primarily absorbed by the ankle joint (see Table 2). This difference might be due to the upper-body external rotation about an axis that passes through the ankle joint while the feet is planted on the floor to attenuate vertical forces.

The data obtained on the contribution of body segments to the vertical displacement of the CM (table 3) confirm the mechanism of rotations described for the impact phase. Notably, the main contributor to the vertical displacement of the CM is the external rotation of the legs on the ankle joints. The contribution of the legs is significantly greater in JBR. In contrast, no differences

were observed related to the contribution of the feet when they rotate backwards to fall flat on the floor. This mechanism causes a greater angular displacement of the ankle joints related to the forward displacement of the legs. Also, the smaller contribution of the thighs when rotating backwards with respect to the knee joint reduces the angular displacement of the knees (see Tables 2 and 3).

The data obtained suggest that the strategies employed to attenuate vertical forces during JBR cause a forward angular acceleration of the legs with respect to an axis that passes through the ankle. This acceleration causes the thighs to flex, thereby resulting in the displacement of the CM forward. This mechanism increases strain within the anterior cruciate ligament (ACL), which role is preventing the tibia from shearing forwards in relation to the femur. Strain increases when the CM moves towards a foot to rapidly release the free leg. When forces shift to one of the feet, force momentum on knee joints during vertical force attenuation increase. A positive correlation has been documented between vertical ground reaction forces and forward acceleration of the tibia, which increases the risk for ACL injury (McNair & Marshall, 1994; Ericksen et al., 2013).

A change of direction during landing (without anticipation) is a risk factor for non-traumatic ACL injury. An unexpected change of direction before landing produces knee rotation and valgus displacement when the foot is in contact with the ground. The reason is that the foot moves backwards to attenuate the peak vertical ground reaction force (2-PvGRF). Knee rotation and valgus displacement have been associated with non-traumatic ACL injury (Cowling & Steele 2001; Hewett, et al., 2005).

CONCLUSIONS

Peak vertical ground reaction forces (2-PvGRF) are attenuated more effectively in preparatory landing (LPR and LPJ), as compared to discrete landing (DL). Yet, impact attenuation occurs by two different mechanisms. In LPJ, muscle preactivation prior to the flexion-extension cycle, added to a greater neural and reflex activity result in more effective impact absorption. Neural and reflex mechanisms are activated to modulate joint angles for the subsequent jump. In contrast, impact absorption occurs by a different kinematic mechanism in LPR.

LPR produces a forward angular acceleration of the legs. This acceleration produces thigh flexion, thereby causing the CM to shift forward. This mechanism increases strain on the anterior cruciate ligament (ACL), which role is preventing the tibia from sliding forward on the femur. Strain increases when the vertical force shifts to the rear foot to rapidly release the front foot.

This kinematic strategy combined with knee rotation and valgus displacement as a response to an unexpected change of direction before landing is a risk factor for non-traumatic ACL injury.

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