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Research article

Inter-Segmental Timing as A Contributing Factor to Knee Loading in Stop-Jump Landings

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Abstract

Landing tasks are a major contributor to knee injuries in sport. While biomechanical determinants of knee joint loading have already been studied, less attention has been given to the role of inter-segmental timing during sport-specific landings. This study aimed to examine both biomechanical and temporal predictors of knee joint loading during horizontal and vertical phases of stop-jump landings. Thirty-three male volleyball and basketball players performed stop-jump tasks while lower-limb kinematics, kinetics, patellar tendon forces, and anterior tibial shear forces were calculated. Statistical Parametric Mapping and Linear Mixed Models were used to identify phase-specific differences and predictors of knee loading. Inter-segmental timing, quantified as foot-contact delay between body segments, emerged as a significant predictor of knee joint loading across conditions. This finding extends previous work by highlighting inter-segmental timing as an additional factor influencing knee joint loading alongside established biomechanical predictors such as displacement velocity, body mass, and knee range of motion. Horizontal landing phases showed a stiffer strategy compared to vertical phases, with reduced joint range of motion (40–83%) and increased knee valgus. These differences were associated with higher anterior tibial shear (121%) and patellar tendon (156%) forces. These results may be particularly relevant in sport situations where landings are not fully planned and can be altered by external perturbations such as opponents' deceiving actions.

Key words: LCA, patellar tendinopathy, knee, biomechanics.

Introduction

Jumping and landing tasks are frequent in sports such as volleyball or basketball. These activities represent risks in terms of both non-contact traumatic (ie. ACL) and overuse (ie. patellar tendinopathy) injuries. Incidence rates per 1000 athlete-exposure of ACL injury is relatively low (from 0.091 to 0.110 among female athletes and from 0.024 to 0.027 among male athlete) (Martinez-Calderon et al., 2025) but have large consequences in terms of out-of-field time and return-to-play level (Ardern et al., 2014). Patellar tendinopathy is a common overuse injury in activities requiring repeated elastic energy storage-and-release such as landing. Prevalence can be higher than 20% in volleyball and basketball players over 18 years old (Nutarelli et al., 2023).

Even though a recent meta-analysis (Romero-Franco et al., 2020) has not been able to demonstrated that kinetic or kinematic variables of landing tasks were associated with non-contact knee injuries, it was also identified (Pedley et al., 2020) that most of the research studies focused on vertical jumps/landings (countermovement jumps, drop jumps, squat jumps) and only to a lesser extent on horizontal ones despite their prevalence in the players' practice. Furthermore, these horizontal jumps/ landings (hop test, broad jump) may not fully represent the field reality of the players both in terms of intensity and technique. Several studies (Heebner et al., 2017; Krause et al., 2022; Saadat et al., 2024) have demonstrate that factors such as the addition of a subsequent jump, the restriction of arm movement, the direction and type of movement will influence the kinematics and kinetics of the jump and landing. These observations emphasize the interest in studying tasks, which are representative of the players' practice. Such tasks may contribute to better identify players at risk and/or better understand the injuries mechanisms. Only few studies (Edwards et al., 2012) have investigated stop-jump landings, even though they represent a frequent transition from an horizontal displacement to a vertical one, which is common in sports like volleyball or basketball.

The aim of the study is first to compare the vertical and horizontal landing phases of a stop jump from a kinematic and kinetic point of view. Contrary to previous studies, the whole landing period will be investigated rather than only specific events (i.e. contact time, maximal knee flexion time…). This approach should allow to better conclude on the landing strategies and their impact on the knee. Then, the identification of the landing technique predictors of knee loading (with a specific interest to the patellar tendon and tibial anterior shear forces) will be investigated. The statistical approach chosen in the present study, based on Linear Mixed Models, should allow more robust and informative results than individual correlations, which has been more commonly used.

Methods

Thirty-three male players (21.9 ± 2.8 years old, 79.1 ± 8.1 kg, 1.87 ± 0.07 m) from basketball (47%) and volleyball (53%) participated in this study. The power of the analysis was computed using the GILMMPSE software for Linear Mixed Models (LMM). The target type I error was fixed to 0.05. We identified the landing type (horizontal vs. vertical) as a repeated measure. The estimated post-hoc power was approximately equal to 0.9.

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Participants had to be aged between 18 and 30, play volleyball or basketball at least three times a week in a training or competition context. Volleyball players had to play in the middle blocker, hitter, or opposite positions to ensure that they jumped on a regular basis. Volunteers with a history of injury or pathology in the lower limbs, spine, or neurological pathology likely to influence biomechanics, or cardiorespiratory disease were excluded from the study. Volunteers were required to refrain from unusual physical activity for 72 hours prior to the test, from taking any medication or performance-enhancing drugs that could influence their performance, and from eating for two hours prior to the session. The study complies with the Declaration of Helsinki and was approved by the institutional ethical committee. All participants provided informed consent.

Each volunteer was instrumented with 20 markers attached on the lower limbs (great trochanters, lateral and medial femoral condyles, lateral malleoli, calcaneum, first and fifth metatarsophalangeal joints, hallux) and on the pelvis (both anterior and posterior superior iliac spines). The 3D positions of the markers were measured using four Codamotion CX1 units (Charnwood Dynamics, Rothley, UK) at a sampling rate of 200 Hz. The accuracy of the system was previously evaluated (Schwartz et al., 2015). Ground reaction forces were measured separately for each foot using two force plates (Kistler™ type 9281 EA, Kistler AG, Switzerland) at a sampling rate of 1000 Hz.

After warming up during approximately 10 minutes (39 squats with increasing weight, 60 bench step ups of increasing intensity, two times 30s of skipping rope), the volunteers familiarized themselves with the stop jump task (height repetitions with gradually increased intensity). The height of a basketball placed above the force platforms was adjusted to allow the volunteer to touch it with their fingertips. The stop jump (Figure 1) consisted of a horizontal run-up of three steps (Wang, 2011), followed by a landing and a vertical push-off with both feet on the platforms in order to jump as high as possible (touch the ball). The landing had to be performed with one foot on each platform. The approach speed was left to the discretion of the volunteer. The stop jump task was repeated five times. The protocol was repeated twice, one week apart to better capture the natural variability of the volunteers' technique. The test-retest reliability of the stop jump was previously evaluated (Milner et al., 2011) demonstrating a good overall reliability. Trials were kept when the volunteers landed with one leg on each platform with no significant imbalance during the second (vertical) landing. The volunteers performed a maximum of six trials in each condition. Because some landings were not performed on the force plates, only two to three trials were kept at each session for further analysis. In total, 371 and 372 landings (including both legs) were analyzed for the horizontal and vertical jumps respectively.

The multi-body skeletal model was constrained to seven degrees of freedom (dof) for each leg (hip joint: three dof, knee: one dof ankle: three dof). Only one degree of freedom was used to model the knee because soft tissue artefacts can introduce substantial errors, making the interpretation of knee abduction/adduction angles unreliable (Stagni et al., 2005). The ankle was modelled with three degrees of freedom to be able to better capture dynamic knee valgus, which results from the contribution of the ankle, knee and hip joints (Hewett et al., 2005). The ankle centers were determined as the mid-point of the malleoli, the knee centers as the mid-point of the femoral condyles, and the hip centers using the regression equation described in (Hara et al., 2016). Inverse kinematics and dynamics were performed in Visual3D Professional software (HAS-Motion, Ontario, Canada).

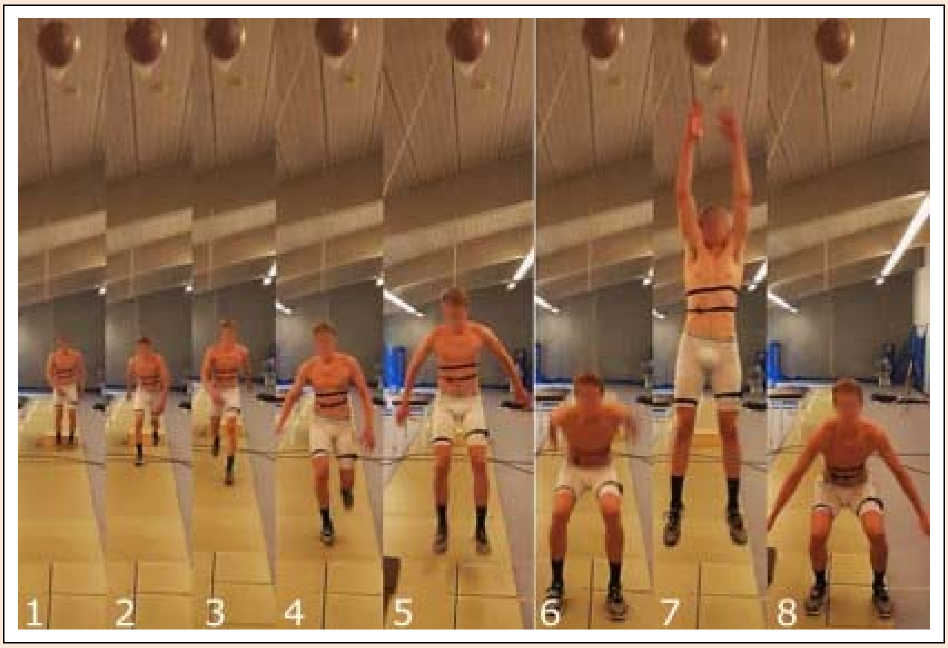

Figure 1. Illustration of the stop jump task including the horizontal approach of three steps (images 1 to 4), the horizontal jump (image 5), the horizontal landing (image 6), the vertical jump (image 7) and the vertical landing (image 8).

3D marker positions and analog force measurements were filtered through a zero-phase 4th order low-pass Butterworth filter at a cut-off frequency of 10 Hz and 20 Hz respectively (Zhou et al., 2021). Joint angles were determined using the Cardan sequence defined in (Wu et al., 2002). As the in-vivo measure of the patellar tendon strength is not feasible, a numerical approach was used based on the inverse dynamic output (sagittal knee moment) and the estimation of the moment arm as a function of the knee flexion (Herzog and Read, 1993). The patellar tendon loading rate was finally derived from the patellar tendon strength time series. The proximal tibial anterior shear force was derived from the net knee joint force expressed in the tibia coordinate system. The tibial anterior shear forces was investigated as it has previously been identified as one of the most direct loading mechanism of the ACL (Dai et al., 2012).

Initial contact was defined when the total vertical ground reaction forces exceeded 20 N. The end of the landing phase was defined at knee maximal flexion (Yu et al., 2004; Zhou et al., 2021) for both the horizontal landing phase (after the three-step run-up) and the vertical landing phase (after the maximal vertical jump). The landing was then time normalized in percent of the phase for further statistical investigations. In addition to the time series, the following biomechanical characteristics of the landing kinematics and kinetics were extracted: horizontal and vertical velocities, delay between feet landing, ankle, knee and hip amplitudes during landing, knee valgus, maximum sagittal knee moment, maximum patellar-tendon force. Knee-to-ankle separation ratio (KASR) distance was used as a proxy for knee dynamic valgus (Ortiz et al., 2016). KASR was preferred over the isolated measurement of knee abduction/adduction rotation because, as already mentioned, this approach reflects the combined contribution of the hip, knee, and ankle joints to dynamic valgus (Hewett et al., 2005) and 2) avoid relying on values that may be significantly biased by soft tissue artefacts (Stagni et al., 2005). A ratio equal to one represents an alignment of the knees and the ankles. A ratio inferior to one indicates than the knee joints are more medially positioned than the ankle joints (indicating dynamic knee valgus) (Mizner et al., 2012). The delay between feet landing was computed as the difference between the landing time of the foot of interest with respect to its counterpart. Therefore, a zero value indicates simultaneous landing of both feet, while a negative value indicates early landing of the foot of interest.

Results

The comparison of the landings of the vertical and horizontal jumps demonstrates significant differences both in terms of kinematics (Figure 2) and in terms of stress on the patellar tendon and tibial anterior shear force (Figure 3). In the sagittal plane, the horizontal landing demonstrates reduced joint amplitude at the three joints (hip: 17.1 ± 8.0° vs. 51.5° ± 21.9, $p < 0.001$; knee: 51.4 ± 9.6° vs. 64.5 ±

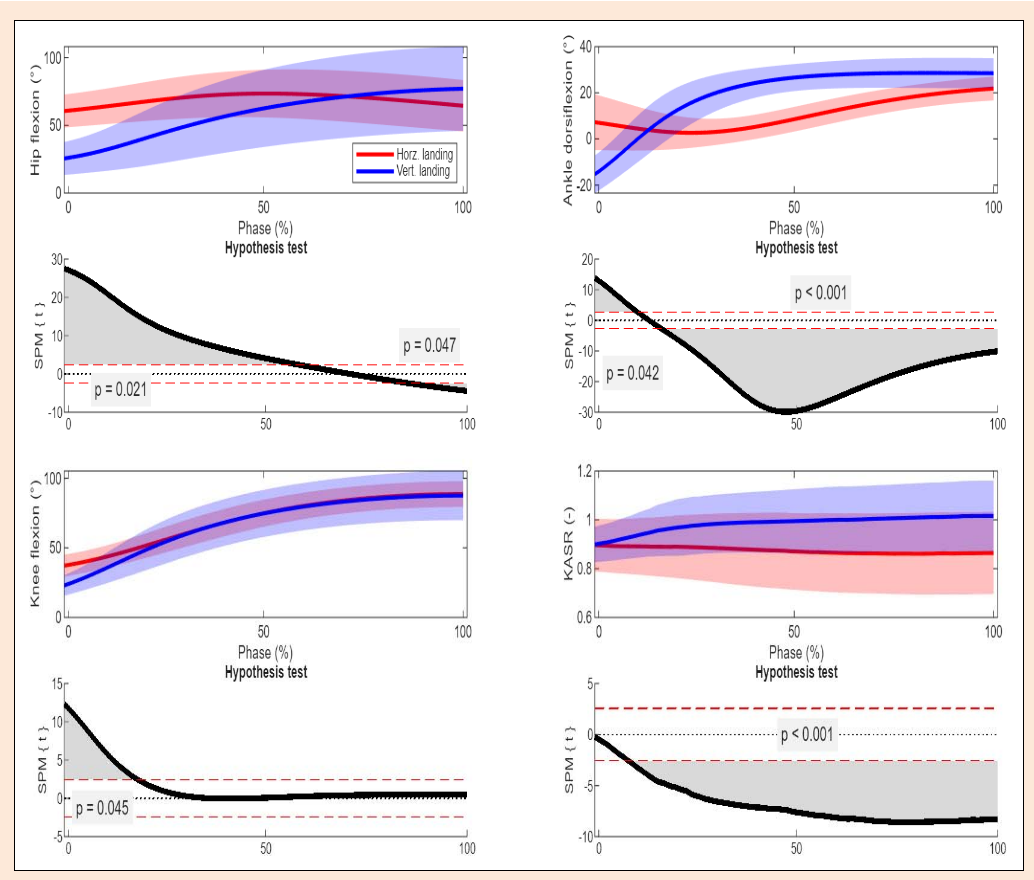

Figure 2. Lower limb (ankle, knee, hip) kinematics in the sagittal plane, knee-to-ankle separation ratio (KASR). Statistical Parametric Mapping (SPM) statistics are presented on the graphs below the time series; statistical differences are indicated by the grey shaded areas.

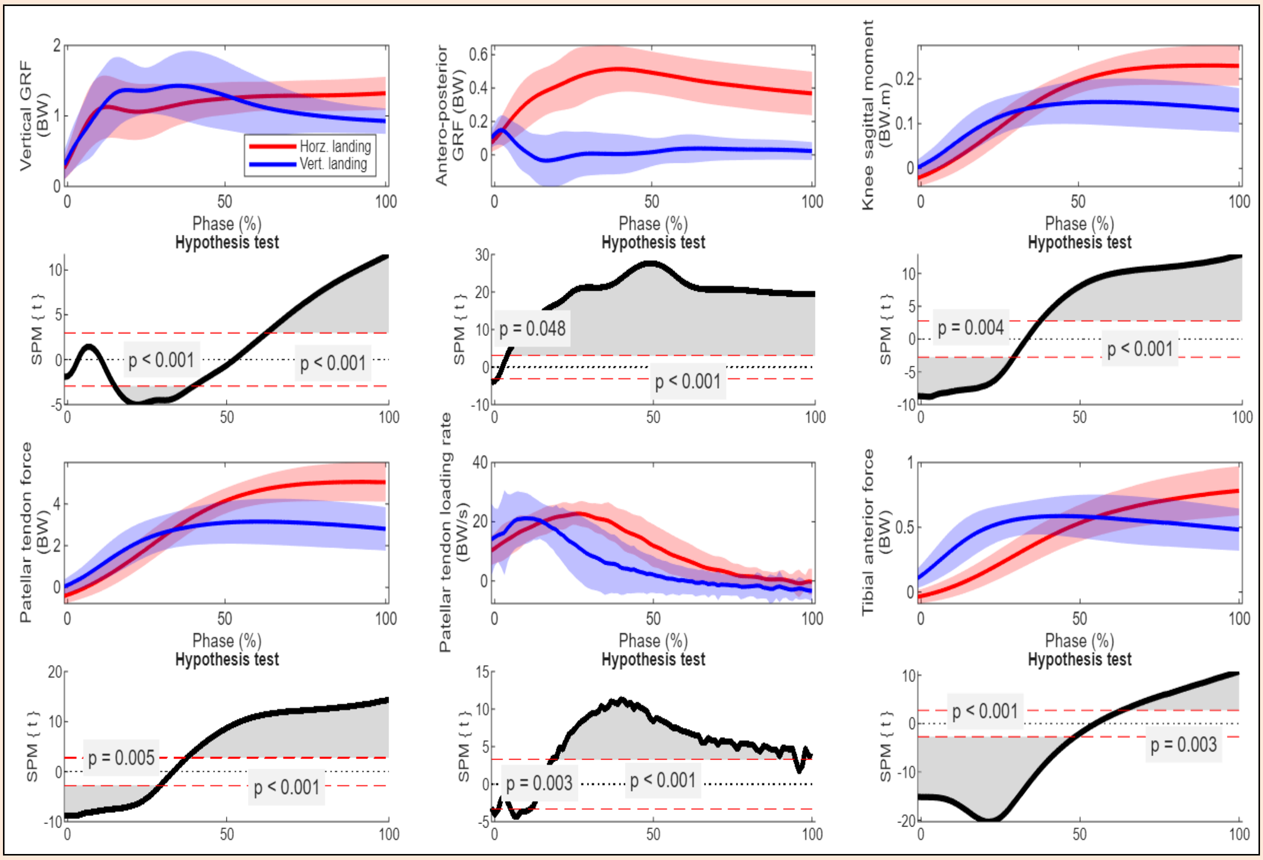

Figure 3. Vertical and anterior ground reaction forces, knee joint moment in the sagittal plane, the patellar tendon force and loading rate, and the tibial anterior shear force during a horizontal landing (red) and a vertical landing (blue). Statistical Parametric Mapping (SPM) statistics are presented on the graphs below the time series, statistical differences are indicated by the grey shaded areas.

13.9°, $p < 0.001$; ankle: 21.9 ± 7.6° vs. 44.6 ± 8.6°, $p < 0.001$) and increased valgus (derived from the knee-to-ankle separation ratio: 0.86 ± 0.17 vs. 1.02 ± 0.15, $p < 0.001$) at knee maximal flexion. During the horizontal jump landings, volunteers tend to have rear foot strike while during the vertical one, volunteers land first on their forefoot as shown by the ankle dorsiflexion negative angle (7.3 ± 12.2° vs. -15.3 ± 8.2°, $p < 0.001$). The horizontal landing leads to a peak vertical ground reaction force in the first half of the reception phase whereas the vertical landing presents increasing vertical forces during all landing duration. Larger antero-posterior ground reactions forces are observed during the horizontal landing (0.57 ± 0.17 BW vs. 0.22 ± 0.07 BW). The peak tibial anterior force of the vertical landing occurs during the first phase of the landing phase, whereas during the horizontal landing, the tibial anterior force constantly increases during the reception. Finally, if the estimated patellar tendon force is larger during a horizontal jump at its maximum (5.15 ± 0.90 BW vs. 3.55 ± 0.96 BW, $p < 0.001$), the peak loading rates remain close (24.5 ± 5.0 BW/s vs. 25.9 ± 10.6 BW/s, $p = 0.22$).

Results of the LMM model for the horizontal jump landing (conditional fit: $R^2 = 0.39$, $p < 0.001$) indicate that weight ($p < 0.001$), horizontal velocity ($p < 0.001$), landing delay ($p = 0.001$) and knee amplitude ($p < 0.001$) are significantly related to the patellar tendon maximal force during landing (Table 1). Concerning the vertical jump landing, the LMM model (conditional fit: $R^2 = 0.274$, $p < 0.001$) indicates that weight ($p < 0.001$), landing delay ($p < 0.001$), vertical velocity ($p = 0.05$), and knee ($p < 0.001$)

Table 1. Kinematic and kinetic predictors of the patellar tendon maximal force during a horizontal jump landing. Parameter Estimates (Fixed coefficients).

			95% Confidence Intervals				
Names	**Estimate**	**SE**	**Lower**	**Upper**	**df**	**t**	**p**
(Intercept)	4043.49	72.6	3900.7	4186.3	0.925	55.681	.015
Weight (kg)	260.81	41.6	179.1	342.5	366.026	6.277	<.001
Landing delay (s)	-198.75	61.2	-319.1	-78.4	364.199	-3.247	.001
Velocity horz (m/s)	282.92	49.9	184.7	381.1	298.858	5.667	<.001
Velocity vert (m/s)	7.54	50.3	-91.3	106.4	366.090	0.150	.881
Hip amplitude (°)	-10.70	53.6	-116.1	94.7	366.247	-0.200	.842
Ankle amplitude (°)	-61.43	44.6	-149.1	26.2	366.970	-1.378	.169
Knee amplitude (°)	-282.15	75.5	-430.7	-133.7	350.377	-3.736	<.001
Knee valgus (m)	-66.76	42.9	-151.1	17.6	366.323	-1.556	.121

Table 2. Kinematic and kinetic predictors of the patellar tendon maximal force during a vertical jump landing. Parameter Estimates (Fixed coefficients).

			95% Confidence Intervals				
Names	Estimate	SE	Lower	Upper	df	t	p
(Intercept)	2907.0	56.1	2796.6017	3017.4	0.917	51.781	.017
Weight (kg)	193.2	46.6	101.4931	284.9	366.206	4.142	<.001
Landing delay (s)	-348.4	48.3	-443.2549	-253.5	366.003	-7.219	<.001
Velocity horz (m/s)	53.5	54.8	-54.2522	161.2	366.245	0.976	.330
Velocity vert (m/s)	91.6	46.6	-0.0630	183.2	366.543	1.965	.050
Hip amplitude (°)	-576.8	91.4	-756.4335	-397.1	366.562	-6.312	<.001
Ankle amplitude (°)	-43.4	65.7	-172.5486	85.7	366.997	-0.662	.509
Knee amplitude (°)	403.8	86.8	233.1082	574.5	366.814	4.652	<.001
Knee valgus (m)	21.1	49.5	-76.3678	118.5	348.260	0.425	.671

and hip amplitudes ($p < 0.001$) are all significantly related to the patellar tendon maximal force (Table 2).

Results of the LMM model for the horizontal jump landing (conditional fit: $R2 = 0.26$, $p < 0.001$) indicate that weight ($p = 0.005$), landing delay ($p < 0.011$), horizontal velocity ($p < 0.001$) and ankle ($p = 0.034$) and knee amplitudes ($p < 0.001$) are all significantly related to the tibial anterior maximal shear force during landing (Table 3). Concerning the vertical jump landing, the LMM model (conditional fit: $R^2 = 0.305$, $p < 0.001$) indicates that weight ($p = 0.033$), landing delay ($p < 0.001$), hip amplitude ($p < 0.044$) and knee valgus ($p < 0.001$) are significantly related to the tibial anterior maximal shear force (Table 4).

Table 3. Kinematic and kinetic predictors of the anterior tibial maximal shear force during a horizontal jump landing. Parameter Estimates (Fixed coefficients).

			95% Confidence Intervals				
Names	Estimate	SE	Lower	Upper	df	t	p
(Intercept)	613.14	14.18	585.26	641.0	0.934	43.253	.019
Weight (kg)	21.38	7.56	6.52	36.2	366.023	2.829	.005
Landing delay (s)	-48.68	11.14	-70.58	-26.8	365.223	-4.370	<.001
Velocity horz (m/s)	46.54	9.10	28.65	64.4	314.542	5.116	<.001
Velocity vert (m/s)	11.68	9.14	-6.30	29.7	366.079	1.278	.202
Hip amplitude (°)	16.94	9.75	-2.22	36.1	366.217	1.738	.083
Ankle amplitude (°)	17.21	8.11	1.26	33.2	366.999	2.122	.034
Knee amplitude (°)	-67.62	13.75	-94.66	-40.6	355.110	-4.919	<.001
Knee valgus (m)	-2.08	7.80	-17.43	13.3	366.286	-0.267	.790

Table 4. Kinematic and kinetic predictors of the anterior tibial maximal shear force during a vertical jump landing. Parameter Estimates (Fixed coefficients).

			95% Confidence Intervals				
Names	Estimate	SE	Lower	Upper	df	t	p
(Intercept)	16.04	7.48	1.32	30.749	366.086	2.143	.033
Weight (kg)	-79.16	7.74	-94.38	-63.942	366.001	-10.227	<.001
Landing delay (s)	1.61	8.79	-15.67	18.894	366.104	0.184	.854
Velocity horz (m/s)	10.70	7.48	-4.02	25.417	366.897	1.430	.154
Velocity vert (m/s)	-29.63	14.68	-58.49	-0.765	366.894	-2.019	.044
Hip amplitude (°)	13.33	10.54	-7.40	34.053	366.673	1.264	.207
Ankle amplitude (°)	-23.22	13.93	-50.61	4.172	366.409	-1.667	.096
Knee amplitude (°)	27.59	7.98	11.89	43.288	365.577	3.456	<.001
Knee valgus (m)	16.04	7.48	1.32	30.749	366.086	2.143	.033

Discussion

Non-contact knee injuries, such as ACL injuries, often occur during deceleration or landing tasks such as the stop jump (Dai et al., 2012). The present study compared the horizontal and vertical landing phases of a stop jump to better understand the knee joint stress in a task representative of the players' practice. Our results demonstrate that the landing kinematics of the vertical and horizontal jumps present significant differences during the landing phases illustrating two different strategies. The horizontal landing could be described as more stiff with reduced hip (75%), knee (22%) and ankle (56%) range of motion in comparison with the vertical landing. The horizontal landing strategy mostly relies on the knee joint. Because of the significantly larger knee joint moment in the sagittal plane, it can be hypothesized that the quadriceps is more recruited and primarily mobilized to soften the significantly larger antero-posterior ground reaction forces. These observations are in line with known ACL loading mechanisms in non-contact ACL injuries, which include altered kinematics, larger posterior ground reaction forces, and large quadriceps muscle force (Boden and Sheehan, 2022; Yu and Garrett, 2007). The landing technique observed in our study confirms previous results (Edwards et al., 2012) and could be explained by a landing point anterior to the body center of mass, requiring already flexed hips at impact. Stiffer landing (reduced range of motion) was shown to increase peak forces at the patellar tendon (Van der Worp et al., 2014) as also observed in our study. Additionally, the heel-strike landing strategy reduce the ability of the triceps surae to absorb the landing forces and could therefore increase the risk of ACL injury (Larwa et al., 2021). Our results not only observed significantly larger (35%) tibial anterior shear forces during the horizontal landing but also a dynamic knee valgus which is maintained during all the landing phase whereas the same valgus tends to be resorbed during the vertical landing. If isolated valgus moment was not shown to be a significant contributing factor to ACL injuries, it can increase the strain applied on the ligament (Alentorn-Geli et al., 2009). Overall, our observations depict a horizontal landing technique that may solicit more the knee than the vertical landing. This phase of the stop jump would therefore be more likely to cause acute or chronic injuries. Indeed, the vertical landing strategy involves all three ankle, knee and hip joints during the landing phase, which is permit by a straighter pose and a forefoot contact at impact.

To explain the stress applied to the knee, our study identified factors common to both vertical and horizontal landings, which are the weight and the velocity of the main direction of displacement. These non-modifiable factors (in most cases) have in common to be directly related to the kinetic energy that needs to be dissipated during the landing phase. Interestingly, the velocity that best predicts both patellar tendon forces and tibial anterior shear forces is the velocity related to the main direction of the displacement (forward velocity for the horizontal landing, and downward velocity for the vertical landing). This observation clarifies previous results (Edwards et al., 2012) stating that vertical ground reaction forces should not be used to reflect patellar tendon forces during a stop jump. The role played by the direction of the motion on the knee constraints further reinforce the need to work on situations that replicate game-related situations. Our observation that the speed of a stop jump (~2.44 $m.s^{-1}$ at impact in the present study) is 20% greater than the one of a broad jump (~2.05 $m.s^{-1}$) (Boone et al., 2021) reinforces this finding. Concerning the landing technique, which represents a modifiable factor, the knee range of motion appears to be a significant contributor to the knee constraints in most conditions. Larger amplitude of knee motion during landing would contribute to reducing both the patellar tendon forces and the tibial anterior shear forces. This result reinforces previous observations (Van der Worp et al., 2014) stating that subjects with patellar tendon ultrasonographic abnormality, which are at higher risk to develop a patellar tendinopathy, present a less upright position at initial contact and therefore less range of motion available to dissipate the landing energy. The landing delay between both feet was not yet investigated in the literature. However, field realistic landings such as stop jumps can be viewed as an intermediate situation between purely analytical one leg landings and two-legs simultaneous landings. Previous studies (Wang, 2011) have demonstrated that one leg landing lead to greater ground reaction forces and tibial anterior shear force. Interestingly, our results comfort these previous investigations in this intermediate situation, which was found to significantly contribute to knee joint constraints. The contribution of this technical aspect of the landing (feet synchronization) to injury during field playing may therefore have been underestimated up to now and could be a working hypothesis to reduce the risk of knee injuries. Ankle mobility only stands out in the vertical landing model of the tibial anterior shear force. This result confirms previous results (Zhou et al., 2021), which demonstrated that during an horizontal stop jump landing, a forefoot landing strategy (rather than a rear foot landing strategy) allows to significantly reduce the energy dissipation in the knee joint towards the ankle joint. The absence of significant results in the other conditions should not necessarily be interpreted as the absence of a role of the ankle joint. Indeed, our findings may be due to consistent rear- or forefoot-landing strategies depending on the landing task, preventing the model from benefiting from enough variability. Altering the natural rear foot landing strategy towards a forefoot landing strategy should therefore be investigated in the future in a population with patellar tendon pain. However, it is important to bear in mind that changing the landing technique may have a negative effect on performance (the height of the jump) (Dai et al., 2015) and that the acceptability of any prevention or rehabilitation program must take this parameter into account.

Some limitations should be acknowledged for this study including ones related to the biomechanical model. The patellar tendon force was estimated in two-dimension despites its complex behavior (Edwards et al., 2012). Furthermore, the stress at the knee joint were estimated based on an inverse dynamic procedure only, not taking into account potential muscle co-contractions, which has been identified (Sell et al., 2007) as potential predictors of the knee stress. A recent study (Garcia et al., 2022) demonstrated that the method used in the present study could lead to higher estimations of the patellar force values even though the trends remains the same using both methodologies. The authors were, however, not able to conclude on the most realistic results, recommending further research. Finally, the study was performed on male volunteers only. The conclusion should not be generalized without care to a female population as a previous study (Chappell et al., 2002) has identified kinematic differences during stop jump landing between male and female athletes.

Conclusion

In conclusion, the results obtained in the present study comparing the horizontal and vertical landing phases of a stop jump demonstrated the existence of two distinct landing strategies. The horizontal landing was characterized by a stiffer landing (reduced joint range of motion), a rear foot landing pose as well as increased knee valgus. Overall, the horizontal landing presents characteristics that could identify it as more at risk, emphasizing the importance of investigating tasks that are closely related to the players' game reality. If non-modifiable factors such as weight and velocity of displacement were shown to influence the stress at the knee joint, modifiable factors related to the landing technique were also identified, such as the knee range of motion and the synchronization of the feet contact at impact.

Acknowledgements

The datasets generated during the current study are not publicly available but are available from the corresponding author upon reasonable request. The authors declare that they have no conflict of interest. All experimental procedures were conducted in compliance with the relevant legal and ethical standards of the country where the study was carried out. The authors declare that no Generative AI or AI-assisted technologies were used in the writing of this manuscript.

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Key points

- Inter-segmental timing appears as a significant predictor of knee joint loading during landing.
- Significant higher anterior tibial shear and patellar tendon forces occur during horizontal landing.
- Horizontal landing phases showed a stiffer landing strategy than vertical ones.

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