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Original Research

The effect of leg length asymmetry on leg stiffness and dynamic postural stability in vertical landing

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Abstract

Background: Assessment of asymmetries in dynamic postural stability and lower extremities kinetics during landing technique are considered factors for injury prevention and achieve optimal athletic performance. Nevertheless, the relationship between these factors has not been established. This study aimed to investigate the effects of leg length asymmetry on dynamic stability and leg stiffness upon initial contact with the ground after vertical landing. **Methods**: Twenty healthy adult men landed on the ground from a height of 30 cm; we measured leg length, leg stiffness, lateral pelvic tilt angle, peak vertical force (PVF), the loading rate, dynamic postural stability index (DPSI), and the correlations among these variables. **Results**: At initial contact, the right leg was significantly longer and showed greater lateral pelvic tilt than the left leg. These characteristics increased the loading rate at the time of PVF on the right leg, which in turn affected leg stiffness and pelvic tilt. The DPSI was also decreased for the right leg compared with the left leg. In the correlation analysis, we observed strong, positive correlations and high explanatory power for PVF, the loading rate, vertical stability index, and DPSI, with $r \geq 0.822$ and $R^2 \geq 57\%$. **Conclusions**: The identified associations support the validity of the result, showing that the right leg failed in its rapid stabilization strategy. The leg length asymmetry is suspected to affect asymmetrical impact patterns, DPSI, and leg stiffness. Given the number of individuals with leg-length inequalities who play sports relying on jumping and landing patterns, reducing the rate of injury possibly incurred.

Keywords: vertical landing; leg stiffness; dynamic postural stability; asymmetry

1. Introduction

Injuries during sports activities typically occur during landing and affect the health of the patient [1]. Unilateral and bilateral landing both occur, depending on the nature of the specific sport and the athlete's objectives, but the function of both legs is assumed to be symmetrical based on biomechanical research data on landing [2].

Humans tend to favor one side of the body, and people can be observed to use one hand or foot preferentially with regard to sporting ability [3,4]. People usually demonstrate dominant or non-dominant orientation, but while approximately 90% of people are right-hand dominant, only 25%–45% show a preference for the right leg during lower limb movements [5]. This is because movements of the dominant lower limb require more brain activity than upper limb movements [6,7].

Biomechanical differences in dominant or non-dominant orientation are related to physiological, anatomic, and lower limb asymmetry, and cause considerable confusion during landing [1]. Variation in athletic performance is a natural biological phenomenon that reflects various factors, such as the mechanisms of stress distribution, response to fatigue, technical level and adaptation, and environmental interactions [2]. Thus, the variance in athletic performance that is produced on each side of the body can be affected by systematic or random factors, and there is no clear reason to control this variance. On the other hand, the upper and lower limbs of humans experience similar overall stress, and asymmetric athletic performance between sides can contribute to unilateral injury when it is chronic or accumulated over time [2].

Various factors can cause significant stress on the musculoskeletal system, but analyzing only one leg and assuming that the functions of both legs are the same lead to limitations in athletic function that can be generated by minute differences. In the long term, asymmetric landing can develop into severe fractures and sprains, or even degenerative arthritis [8]. Differences in lower limb length are major factors in acute and chronic injury of the sacroiliac joint [9], and are closely related to the development of lower back pain [10]. Specifically, excessive movements and failure to control the magnitude of the ground reaction force (GRF) in the longer leg can cause injury and secondary falls [11,12], but the function of both legs needs to be assessed simultaneously to enhance our understanding of the causes of injury and improve safety.

During landing, the length of the lower limb is altered by flexion and extension of the hip, knee, and ankle joints, as well as tilting of the trunk [13,14]. According to the movement systems theory, the composition of the sensorimotor system originates from interactions within a complex system, including the neuromuscular and musculoskeletal systems, and adjustment or restriction of the degrees of freedom [15]. In fact, numerous musculoskeletal elements, including the muscles, tendons, and ligaments, act together like a spring during walking, running, and landing. When the lower limb functions in a healthy manner, leg stiffness can be properly adjusted in diverse conditions [16,17], but it is not clear what effect differences in leg lengths have on optimal stiffness control and dynamic stability upon initial contact.

In vertical landing, it is important to simultaneously recognize asymmetry in leg length and changes in the vertical GRF, and to predict a pearson's capacity to maintain their balance while transitioning from a dynamic to a stationary position during athletic exercises. It is essential to understand the stabilization time, depending on the muscle patterns, reflexes, and reactions that are required to minimize the range of the GRF [18]. New research is required to help prevent injury and achieve optimal athletic performance by evaluating leg stiffness, which can be observed using the lower limb length and size of the GRF.

We investigated the effects of leg length asymmetry on leg stiffness and dynamic postural stability upon initial contact with the ground in participants who performed vertical landing movements. In addition, we aimed to quantify landing strategies and mechanisms through a correlation analysis of related variables for each leg, such as the peak vertical force (PVF), loading rate, leg length, and pelvic angle.

2. Materials and methods

2.1 Subjects

20 healthy adult men participated (mean ± standard deviation: 21.70 years ± 1.89 years, height: 1.77 m ± 5.58 m, weight: 82.73 kg ± 17.50 kg) in this study. They met the subject criteria for this study. Before participating in the experiment, all participants received a thorough explanation of the study's content and aims, and provided their voluntary written consent.

2.2 Experimental procedures

To examine the relationship between leg length and leg stiffness during vertical landing movements, we determined the dominant and non-dominant legs using the ball-kick and step-up tests. Without being briefed on the ball-kick test in advance, participants were instructed to kick a ball. The leg that had initial contact during the ball-kick test was identified as the dominant leg, and the leg that first stepped onto the bench was identified as the dominant one in the step-on test [19,20].

For kinematic and kinetic measurement of the main joints of the lower limbs during landing movements, a motion analysis device consisting of eight motion analysis cameras (6 Eagle and 2 Raptor-E Camera System, Motion Analysis Corporation, Santa Rosa, CA, USA), and two force plates (OR6-5-2000, Advanced Medical Technologies Inc., Watertown, MA, USA) were used. GRF data were obtained through two force plates sampling at 600 Hz. Before measuring, we installed eight cameras each to the right, left, and front of the participant, all of which were within a 7 m range. After preparing an environment in which the range of motion could be captured, we performed calibration to establish the spatial coordinates. The sampling rate of the camera was set to 120 frames/sec, and the margin of error was 0.3 mm or less. Next, we attached 19 reflective markers (15 markers were used to track the position and movement while 4 markers were used for calibration only and removed after it) that were 15 mm in diameter on the lower limbs of the participants in accordance with the Helen Hayes Markers Set, which allowed us to assess landing movements (Fig. 1).

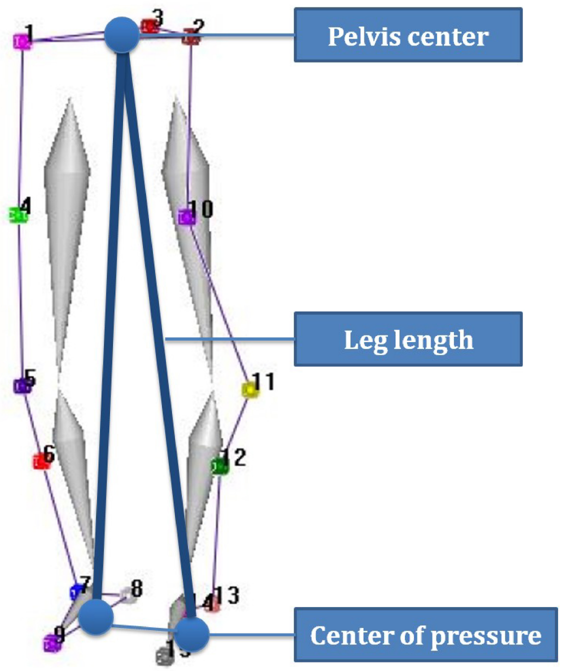


Fig. 1. Marker attachment point.

In order to simulate the landings encountered during athletic participation, subjects were asked to perform a bilateral drop landing task. To orient participants with task, each subject was asked to perform 3–5 practice trials. Once subjects were comfortable with the task, they were asked to perform 3 test trials for landing task from a height of 30 cm. Each subjects were instructed to jump with their heads up and hands in a position to touch the designated marker and place their hands on their hips as soon as they felt stable. Researchers and participants determined whether their movements were successful. We measured the mean values from three trials.

2.3 Data measurement

We calculated the angle between the center of pressure (COP) and the center of the pelvis using the angle that we observed in the frontal plane (Fig. 2).

$$\text{Tilt angle } = \tan^{-1}\left[\frac{\mathrm{Z_{cop}}\ (0) - \text{pelvis}}{\mathrm{X_{cop-pelvis}}}\right] \tag{1}$$

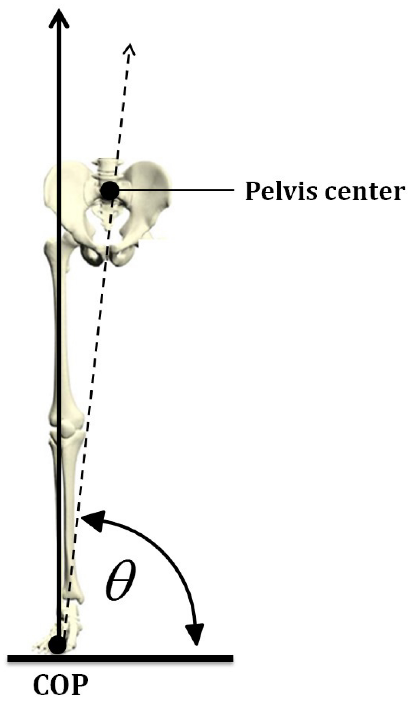


Fig. 2. Tilt angle between pelvis center and center of pressure (COP).

We calculated dimensionless leg stiffness (K_{leg}) using Silder's method [15]. For leg length, we calculated the distance from the center of the pelvis to the COP at initial contact in three dimensions, and we calculated the leg stiffness based on the time of PVF.

$$\mathrm{K_{leg}} = \frac{\mathrm{PVF}}{(l_o - l_{\min})\,/l_o{}'} \tag{2}$$

$$l_o, l_{\min} = \sqrt{(X_{\text{pelvis}} - X_{\text{COP}})^2 + (Y_{\text{pelvis}} - Y_{\text{COP}})^2 + (Z_{\text{pelvis}} - Z_{\text{COP}})^2} \quad Z_{\text{cop}} = 0 \tag{3}$$

PVF was calculated by dividing the PVF produced upon landing (N) by the participant's body weight. The resulting value was then calibrated (N/BW). l_o is the calibrated value of the rate of change in the leg length during the stance phase, and l_{min} is the minimum leg length. Leg length was measured from the center of pressure [21] to the central point of the pelvis [22] as shown in Fig. 1.

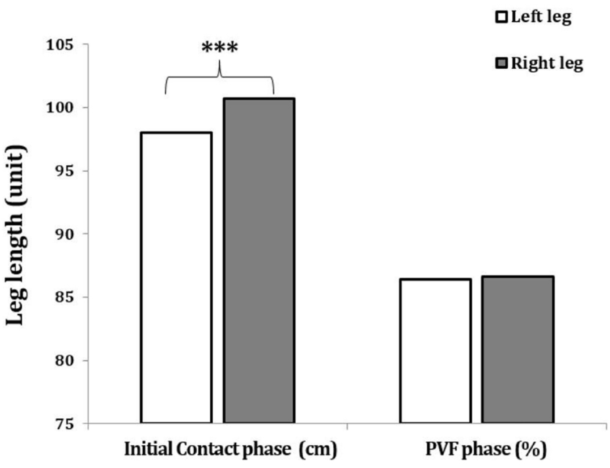
To measure the dynamic postural stability index (DPSI) during landing, we referred to the protocol of Wikstrom [18], but to determine a clear endpoint after landing, we modified the time at which we used the data points and the equation for the vertical direction. The medial-lateral stability index (MLSI) and anterior-posterior stability index (APSI) assess the fluctuations from 0 along the frontal and sagittal axes of the force plate, respectively. The vertical stability index (VSI) assesses the fluctuation from the subject's body weight to standardize the vertical GRF along the vertical axis of the force plate.

$$\begin{aligned} \text{MLSI} &= \surd[\Sigma\,(0 - \mathrm{X_{PVF}})^2 \,/\text{ number of data points }] \\ \text{APSI} &= \surd[\Sigma\,(0 - Y_{\text{PVF}})^2 \,/\text{ number of data points }] \\ \text{VSI} &= \surd[\Sigma\,(0 - Z_{\text{PVF}})^2 \,/\text{ number of data points }] \\ \text{DPSI} &= \surd[\Sigma\,(0 - \mathrm{x_{PVF}})^2 + \Sigma\,(0 - \mathrm{y_{PVF}})^2 + \Sigma\,(0 - \mathrm{z_{PVF}})^2 \\ &\quad /\text{ number of data points }] \end{aligned} \tag{4}$$

An increased DPSI value indicated less stability, and a decreased DPSI value indicated improved stability. We calculated the asymmetry index of the variables that were calculated above using Robinson's equation (1987) with the mean values of the left and right sides. An asymmetry index value of 0 corresponded to prefect symmetry, while higher values indicated increased asymmetry.

$$\mathrm{AI}(\%) = \left|\frac{\text{Right-Left}}{\frac{1}{2}(\text{ Right } + \text{ Left })}\right| \times 100 \tag{5}$$

All kinematic and kinetic data were processed using Cortex 4 (Motion Analysis Corporation, Santa Rosa, CA, USA). The body was assumed to be a linked rigid body system. The center of mass of the pelvis and each segment were calculated by assigning coordinates to the central point of body joints and were used as parametric data [22]. The two-dimensional (2D) data collected by the eight-image analysis camera were converted to three-dimensional (3D) data by the non-linear transformation method. To remove errors due to noises in data processing, Butterworth low-pass digital filtering was used for smoothing, and the



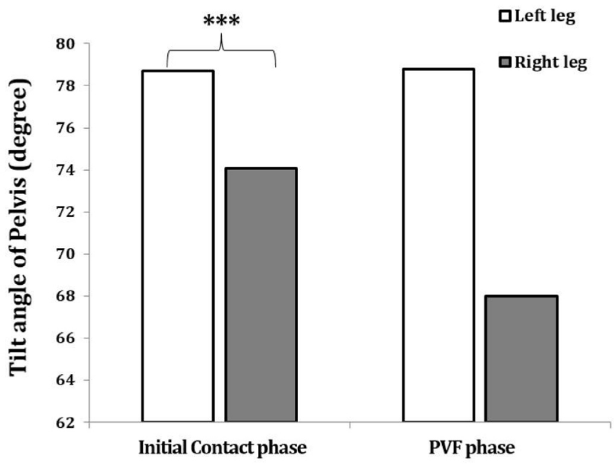


Fig. 3. Result of the leg length (left), and tilt angle of pelvis center (right) according to the landing leg. ***$p < 0.001$.

cut-off frequency was set to 10 Hz.

Using the SPSS 23.0 statistics program (IBM Corp., Armonk, NY, USA), we calculated the means and standard deviations for the kinematic and mechanical variables that were described above, and we performed paired samples *t*-tests to compare the two legs. We used Pearson's correlation coefficients to analyze the correlations between the variables of interest and the asymmetry index, and we used a statistical significance level of $p < 0.05$ for all tests.

3. Results

In the Table 1, we compared the kinematic and mechanical variables during the vertical landing movement, the leg length at initial contact with the ground was significantly longer for the right leg than for the left ($p < 0.001$) (Fig. 3). Pelvic tilt was significantly larger for the right leg than the left leg ($p < 0.001$) (Fig. 3). At the time of PVF, leg length, stiffness, and tilt showed no significant difference between the two sides, but the loading rate was statistically higher for the right leg than the left leg ($p < 0.05$) (Fig. 4). There was no significant difference in the anterior-posterior DPSI, but the right leg showed significantly increased vertical DPSI and reduced stability ($p < 0.05$) (Fig. 5).

In the correlation analysis using the asymmetry index as Table 2, we limited our analysis to results with $r \geq 0.70$ to determine clear correlations. Strong, positive correlations with high explanatory power were observed for the PVF, with $r = 0.837$ (y = 1.0597x – 23.881, $R^2 = 0.070$, $p < 0.01$) for the loading rate, $r = 0.822$ (y = 0.8417x – 25.298, $R^2 = 0.675$, $p < 0.01$) for the vertical stability index, and $r = 0.808$ (y = 0.8117x – 21.469, $R^2 = 0.653$, $p < 0.01$) for the bilateral DPSI. Strong, positive correlations were also observed for the loading rate, with $r = 0.900$ (y = 0.7276x – 7.6325, $R^2 = 0.809$, $p < 0.01$) for the vertical stability index and $r = 0.898$ (y = 0.7123x – 4.2263, $R^2 = 0.806$, $p < 0.01$) for the DPSI, and $r = 0.900$ (y = 0.9723x + 3.1016, $R^2 = 0.983$, $p < 0.01$) for the correlation between the vertical

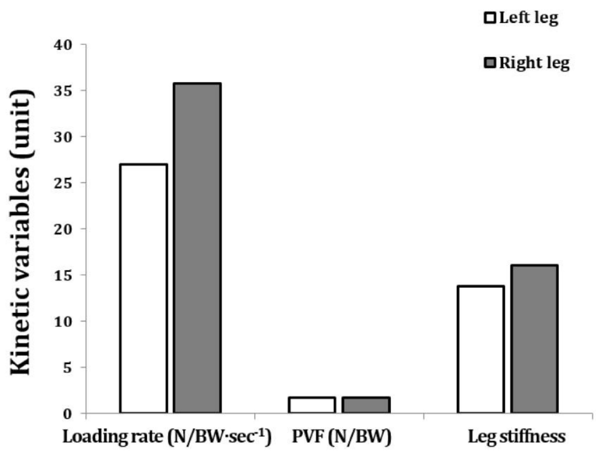


Fig. 4. Loading rate, PVF, and leg stiffness according to the landing leg.

stability index and DPSI.

4. Discussion

Diverse exercise, in which jumping and landing movements are the major components, contributes to the development and persistence of injuries [15,23,24]. Lower limb alignment differences or strength, endurance or power deficits may also play a role and explain findings [25]. Several further factors are related to injury, but in particular, asymmetrical leg length is apparent upon initial contact with the ground. However, there are no clear methods or information to successfully control impact absorption and stability in patients with asymmetric leg length.

Generally, the kinematic characteristics of asymmetry can affect musculoskeletal relationships such as the force-length relationship, and one study argued that an excessive load can be sustained by the muscle tissue of one leg when the force distribution between the two legs is altered [26,27]. In our study, when we observed the mechanical

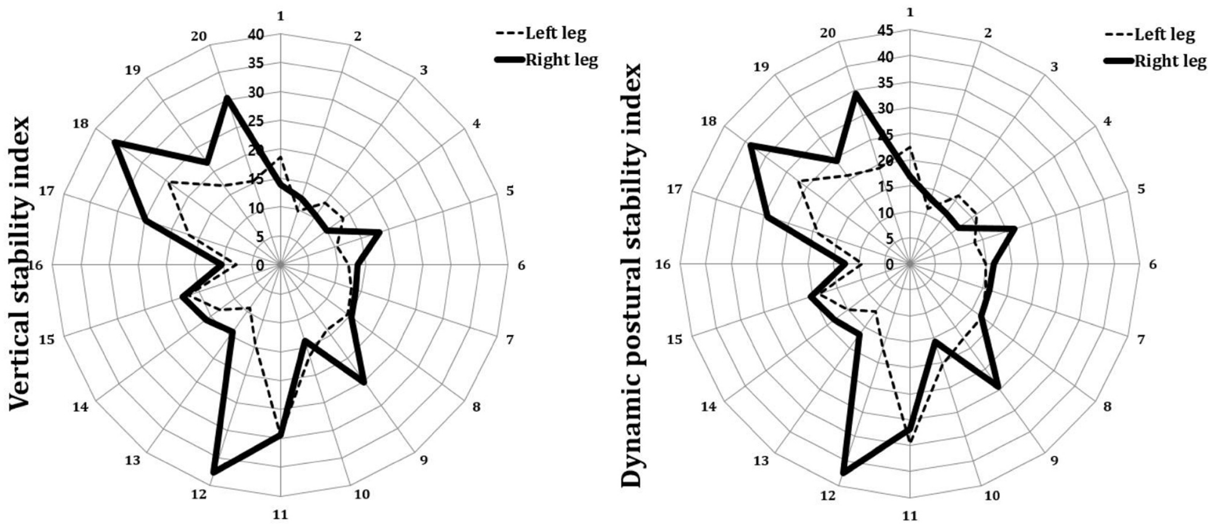


Fig. 5. Medial-lateral and anterior-posterior stability index for each subjects.

Table 1. Result of kinetic variables between bilateral leg during vertical landing.

Section	Variables	Vertical landing			t	*p*-Value
		Left leg	Right leg	*AI (%)*		
Initial contact	Lengths (cm)	98.05 ± 4.75	100.70 ± 4.95	2.51	5.732	0.001***
	Tilt angle of pelvis (deg)	78.68 ± 2.18	74.05 ± 2.07	5.98	6.702	0.001***
PVF phase during landing	Leg length (%)	86.40 ± 2.82	86.60 ± 3.88	1.80	1.758	0.095
	Leg stiffness	13.77 ± 4.63	16.10 ± 9.05	1.15	1.262	0.222
	Tilt angle of pelvis (deg)	78.77 ± 1.71	68.00 ± 34.86	3.87	1.369	0.187
	Peak vertical force (N/BW)	1.77 ± 0.39	1.71 ± 0.40	4.09	0.501	0.622
	Loading rate (N/BW/sec)	26.97 ± 9.81	35.82 ± 20.34	19.54	2.319	0.032*
DPSI	Medial-lateral	1.48 ± 0.76	1.07 ± 0.52	29.08	2.722	0.014*
	Anterior-posterior	1.45 ± 0.42	1.65 ± 0.69	14.77	1.321	0.202
	Vertical	14.98 ± 5.03	19.25 ± 8.54	21.85	2.740	0.013*
	Dynamic postural stability	17.92 ± 5.68	21.98 ± 9.21	18.14	2.346	0.030*

*$p < 0.05$; ***$p < 0.001$.

Table 2. Results of correlation (*r*).

Section	Initial contact (IC)	PVF phase during landing								
	Tilt angle	Leg length	PVF	stiffness	Tilt angle	Loading rate	MLSI	APSI	VSI	DPSI
Length (IC)	0.080	–0.542	0.314	0.035	0.418	0.071	–0.019	–0.218	–0.061	–0.075
Tilt angle (IC)		–0.062	0.195	–0.198	0.620	–0.015	0.312	–0.192	–0.106	–0.092
Leg length			0.166	–0.333	–0.430	0.239	0.223	0.368	0.322	0.344
PVF				0.413	0.008	0.837**	0.583	0.372	0.822**	0.808**
Stiffness					–0.249	0.622	0.175	0.262	0.540	0.521
Tilt angle						–0.241	–0.167	–0.511	–0.321	–0.356
Loading rate							0.534	0.525	0.900**	0.898**
MLSI								0.478	0.604	0.666
APSI									0.595	0.675
VSI										0.991**

**$p < 0.01$.

characteristics of the lower limbs at the time of PVF after the initial contact, we found that the length of the right and left legs was very similar. Each leg could absorb the PVF symmetrically. Conversely, the loading rate and leg stiffness were higher in the right leg than in the left leg. A high loading rate indicates that a large GRF cannot be distributed properly across the lower limb joints within a short time [28], and is closely related to neuromuscular problems, joint degeneration, and bone fracture [29,30]. In addition, an increased loading rate is associated with increased leg stiffness, which is consistent with the results of our study, in which leg stiffness was higher in the right leg, which showed an increased loading rate [31].

Regarding the calculation of leg stiffness in our study, there was a strong association between the change in leg length and impact. Increasing the range of motion of the lower limb joints can effectively decrease leg stiffness and the impact that is transferred from the feet to the head [32,33]. However, stiff motion of the right leg at initial contact with the ground was inevitably maintained until the time of PVF, resulting in failure to control leg stiffness. In other words, a certain amount of stiffness control is essential during exercise, and one study reported that stiffness can be controlled appropriately [34,35]. In the case of asymmetric leg length at the initial contact, similar to our study, other authors thought that leg stiffness is controlled by increased lateral pelvic tilt.

In the present study, the right leg, which was longer at the initial contact than the left, showed a worse anterior-posterior stability index, vertical stability index, and DPSI than the left leg. Athletes can reduce the risk of injury upon landing by maintaining a neutral posture and stabilizing the leg as soon as possible after landing; leg length asymmetry indicates failure of this strategy [12,36]. In particular, when we performed a correlation analysis for each leg using the asymmetry index, we observed correlations of $r \geq 0.822$ and $R^2 \geq 57\%$ for the PVF, loading rate, vertical stability index, and DPSI. These variables were calculated from the impact type or size of the impact over time, and thus are closely related to postural control and the stabilization time. Increased values for these variables indicate a relatively large increase in force components within a short time.

Maintaining balance during landing reflects successful impact absorption [18]. Because landing strategies are usually planned [37,38], temporal restrictions in an unpredictable situation can lead to high loads that can damage the spine or nerve pathways in and around the spine [39–41]. Therefore, leg length asymmetry at initial contact indicates reduced postural control in one leg to maintain stability and impact absorption. Therefore, an increased load is applied to one side of the body that increases the risk of secondary ankle sprain or slipping.

5. Conclusions

Leg length asymmetry at initial contact with the ground, which was the focus of our study, is thought to play a subtle but important role in the rate of injury. Despite increasing the lateral pelvic tilt to offset the impact pattern, we found that successful maintenance of the landing function in both legs was impossible. Therefore, if participants in sports practice maintain leg length symmetry on landing, it should enable them to successfully land in various sporting environments.

Author contributions

Conceptualization, KK and KJ; methodology, KJ and SH; software, KJ; validation, KK, KJ and SH; formal analysis, KJ and SH; investigation, KK, KJ and SH; resources, KK; data curation, KJ and SH; writing—original draft preparation, KJ and SH; writing-review and editing, KJ and SH; project administration, KK; funding acquisition, KK. All authors have read and agreed to the published version of the manuscript.

Ethics approval and consent to participate

This study was approved by the Institutional Review Board at Incheon National University (INUIRB No. 7007971-202012-002A). All participants provided written informed consent.

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Conflict of interest

The authors declare no conflict of interest.

References

[1] Niu W, Wang Y, He Y, Fan Y, Zhao Q. Kinematics, kinetics, and electromyogram of ankle during drop landing: a comparison between dominant and non-dominant limb. Human Movement Science. 2011; 30: 614–623.

[2] Schot PK, Bates BT, Dufek JS. Bilateral performance symmetry during drop landing: a kinetic analysis. Medicine and Science in Sports and Exercise. 1994; 26: 1153-1159.

[3] Carpes FP, Mota CB, Faria IE. On the bilateral asymmetry during running and cycling–a review considering leg preference. Physical Therapy in Sport. 2010;11:136–142.

[4] Serrien DJ, Ivry RB, Swinnen SP. Dynamics of hemispheric specialization and integration in the context of motor control. Nature Reviews Neuroscience. 2006; 7: 160–166.

[5] Čuk, T., Leben-Seljak, P., & Štefančič, M. Lateral asymmetry of human long bones. Variability and Evolution. 2001; 9:19–32.

[6] Kapreli E, Athanasopoulos S, Papathanasiou M, Van Hecke P,

Keleki D, Peeters R, *et al*. Lower limb sensorimotor network: issues of somatotopy and overlap. Cortex. 2007; 43: 219–232.
[7] Luft AR, Smith GV, Forrester L, Whitall J, Macko RF, Hauser T, *et al*. Comparing brain activation associated with isolated upper and lower limb movement across corresponding joints. Human Brain Mapping. 2002; 17: 131–140.
[8] Murray-Leslie CF, Lintott DJ, Wright V. The knees and ankles in sport and veteran military parachutists. Annals of the Rheumatic Diseases. 1977; 36: 327–331.
[9] Cummings G, Scholz JP, Barnes K. The effect of imposed leg length difference on pelvic bone symmetry. Spine. 1993; 18: 368–373.
[10] Fann AV. The prevalence of postural asymmetry in people with and without chronic low back pain. Archives of Physical Medicine and Rehabilitation. 2002; 83: 1736–1738.
[11] Boden BP, Dean GS, Feagin JA, Garrett WE. Mechanisms of anterior cruciate ligament injury. Orthopedics. 2000; 23: 573–578.
[12] Wikstrom EA, Powers ME, Tillman MD. Dynamic stabilization time after isokinetic and functional fatigue. Journal of Athletic Training. 2004; 39: 247.
[13] Chijimatsu M, Ishida T, Yamanaka M, Taniguchi S, Ueno R, Ikuta R, *et al*. Landing instructions focused on pelvic and trunk lateral tilt decrease the knee abduction moment during a single-leg drop vertical jump. Physical Therapy in Sport. 2020; 46: 226–233.
[14] Critchley ML, Davis DJ, Keener MM, Layer JS, Wilson MA, Zhu Q, *et al*. The effects of mid-flight whole-body and trunk rotation on landing mechanics: implications for anterior cruciate ligament injuries. Sports Biomechanics. 2020; 19: 421–437.
[15] James C. Considerations of movement variability in biomechanics research. Innovative Analyses of Human Movement. 2004: 29–62.
[16] Farley CT, Houdijk HH, Van Strien C, Louie M. Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. Journal of Applied Physiology. 1998; 85: 1044–1055.
[17] Silder A, Besier T, Delp SL. Running with a load increases leg stiffness. Journal of Biomechanics. 2015; 48: 1003–1008.
[18] Wikstrom EA, Tillman MD, Smith AN, Borsa PA. A new force-plate technology measure of dynamic postural stability: the dynamic postural stability index. Journal of Athletic Training. 2005; 40: 305.
[19] Hoffman M, Payne VG. The effects of proprioceptive ankle disk training on healthy subjects. The Journal of Orthopaedic and Sports Physical Therapy. 1995; 21: 90–93.
[20] Verhagen E, Bobbert M, Inklaar M, van Kalken M, van der Beek A, Bouter L, *et al*. The effect of a balance training programme on centre of pressure excursion in one-leg stance. Clinical Biomechanics. 2005; 20: 1094–1100.
[21] Bullimore SR, Burn JF. Consequences of forward translation of the point of force application for the mechanics of running. Journal of Theoretical Biology. 2006; 238: 211–219.
[22] Plagenhoef S, Evans FG, Abdelnour T. Anatomical data for analyzing human motion. Research Quarterly for Exercise and Sport. 1983; 54: 169–178.
[23] Konradsen L. Sensori-motor control of the uninjured and injured human ankle. Journal of Electromyography and Kinesiology. 2002; 12: 199–203.
[24] Konradsen L, Voigt M. Inversion injury biomechanics in functional ankle instability: a cadaver study of simulated gait. Scandinavian Journal of Medicine and Science in Sports. 2002; 12: 329–336.
[25] Dufek JS, Bates BT. Biomechanical factors associated with injury during landing in jump sports. Sports Medicine. 1991; 12: 326–337.
[26] Pappas E, Carpes FP. Lower extremity kinematic asymmetry in male and female athletes performing jump-landing tasks. Journal of Science and Medicine in Sport. 2012; 15: 87–92.
[27] Vagenas G, Hoshizaki B. A Multivariable Analysis of lower extremity kinematic asymmetry in running. International Journal of Sport Biomechanics. 1992; 8: 11–29.
[28] Irmischer BS, Harris C, Pfeiffer RP, Debeliso MA, Adams KJ, Shea KG. Effects of a knee ligament injury prevention exercise program on impact forces in women. Journal of Strength and Conditioning Research. 2004; 18: 703–707.
[29] Crossley K, Bennell KL, Wrigley T, Oakes BW. Ground reaction forces, bone characteristics, and tibial stress fracture in male runners. Medicine and Science in Sports and Exercise. 1993; 31: 1088–1093.
[30] Mizrahi J, Susak Z. Analysis of parameters affecting impact force attenuation during landing in human vertical free fall. Engineering in Medicine. 1982; 11: 141–147.
[31] Hennig EM, Lafortune MA. Relationships between ground reaction force and tibial bone acceleration oarameters. International Journal of Sport Biomechanics. 1991; 7: 303–309.
[32] Hewett TE, Lindenfeld TN, Riccobene JV, Noyes FR. The effect of neuromuscular training on the incidence of knee injury in female athletes. The American Journal of Sports Medicine. 1999; 27: 699–706.
[33] McMahon TA, Valiant G, Frederick EC. Groucho running. Journal of Applied Physiology. 1987; 62: 2326–2337.
[34] Farley CT, Blickhan R, Saito J, Taylor CR. Hopping frequency in humans: a test of how springs set stride frequency in bouncing gaits. Journal of Applied Physiology. 1991; 71: 2127–2132.
[35] Farley CT, González O. Leg stiffness and stride frequency in human running. Journal of Biomechanics. 1996; 29: 181–186.
[36] Ross SE, Guskiewicz KM. Examination of static and dynamic postural stability in individuals with functionally stable and unstable ankles. Clinical Journal of Sport Medicine. 2004; 14: 332–338.
[37] Baker J, Côté J, Abernethy B. Learning from the experts: practice activities of expert decision makers in sport. Research Quarterly for Exercise and Sport. 2003; 74: 342–347.
[38] Besier TF, Lloyd DG, Ackland TR. Muscle activation strategies at the knee during running and cutting maneuvers. Medicine and Science in Sports and Exercise. 2003; 35: 119–127.
[39] Borotikar BS, Newcomer R, Koppes R, McLean SG. Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. Clinical Biomechanics. 2008; 23: 81–92.
[40] McLean SG, Samorezov JE. Fatigue-induced ACL injury risk stems from a degradation in central control. Medicine and Science in Sports and Exercise. 2009; 41: 1661–1672.
[41] Proske U. Kinesthesia: the role of muscle receptors. Muscle and Nerve. 2006; 34: 545–558.