

Article

Sex and Limb Dominance Differences in Postural Control Performance of Young Adults: A Third-Order Polynomial Decay Approach

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Abstract

This study systematically evaluated postural control performance on dominant and non-dominant sides in young adults of different sexes using a Third-Order Polynomial Decay fitting method combined with time-domain stability features. A total of 126 participants (66 males, 60 females) performed single-leg landing tasks, during which ground reaction forces (GRF) and center of pressure (COP) data were collected using a Kistler 3D force platform and Bioware acquisition system. Dynamic stability times in the anterior–posterior, medial–lateral, and vertical directions were calculated with polynomial fitting, and additional time-domain measures were used to assess static and dynamic stability. Results showed that on the non-dominant side, participants exhibited significantly longer dynamic stability times ($p = 0.015$), greater root mean square distance ($p = 0.005$), and longer total sway path ($p = 0.005$) in the anterior–posterior direction compared with the dominant side. Significant sex differences were also found in vertical stability index ($p = 0.044$), dynamic stability index ($p = 0.047$), total sway path ($p < 0.001$), anterior–posterior sway path ($p = 0.001$), and medial–lateral sway path ($p < 0.001$). In conclusion, the dominant limb demonstrated superior stability, males showed better static control, and females displayed greater dynamic stability, underscoring the importance of targeted non-dominant training and sex-specific balance strategies.

Keywords: postural control; limb dominance; sex differences; single-leg landing; third-order polynomial decay



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1. Introduction

Postural control stability generally refers to an individual's ability to maintain their body posture and keep the center of mass (COM) within the base of support during movement [1]. This ability is typically divided into dynamic postural stability and static postural stability. Dynamic postural stability emphasizes the capacity to maintain or regain balance when the body is in motion or subjected to external perturbations, while static postural stability focuses on maintaining the COM and body posture under stationary conditions [2].

The human body relies on multisensory inputs—such as visual, vestibular, and proprioceptive information—to perceive its spatial position and posture. It then uses rapid neuromuscular responses to adjust muscle tension and joint angles in order to maintain balance. Good postural stability not only enhances athletic performance but also plays a crucial role in preventing sports injuries. In contrast, poor postural control may increase the risk of falls and musculoskeletal injuries [3,4]. Therefore, scientific training aimed at improving postural stability can effectively reduce the likelihood of such injuries.

After anterior cruciate ligament (ACL) injury, the dynamic postural stability of both the injured and the contralateral healthy limb can be affected. This impairment is primarily associated with neuromuscular control deficits, including insufficient preparatory and reactive muscle activity, as well as post-injury central nervous system reorganization [5,6]. Individuals with unilateral ACL injury exhibit prolonged time to stabilization (TTS) during single-leg landing tasks, indicating compromised dynamic postural stability. Furthermore, the dominance (or non-dominance) of the injured limb may further influence bilateral stability, highlighting the significant role of lateralization in motor control [7]. To obtain a more objective and reliable assessment of dynamic postural stability, the present study employed a third-order polynomial decay fitting method to extract TTS time-domain features. This approach enables precise identification of the time point at which an individual achieves dynamic stabilization by fitting the entire ground reaction force (GRF) signal [8].

Among the various factors affecting postural stability, sex and limb dominance are particularly prominent. Differences between males and females in anatomical structure, muscle strength, bone density, and neuromuscular control mechanisms [9] may contribute to distinct postural control and stability performance. In addition, although human anatomical structures are generally symmetrical, functional asymmetry often exists in physiological performance, a phenomenon referred to as lateralization. Prolonged functional asymmetry may result in differences in limb morphology, joint range of motion, muscle strength, and postural stability [10].

Although previous studies have examined the independent effects of sex and limb dominance, their combined influence on dynamic and static postural stability remains unclear and debated. In this study, a third-order polynomial decay fitting method was applied to systematically explore these factors. Importantly, our goal was not only to verify known differences but also to reveal novel, task-specific distinctions between dynamic and static stability and to quantify the magnitude of these effects. The results offer refined, data-driven insights to guide the development of individualized exercise and rehabilitation programs.

2. Methods

A total of 126 young adults (66 males and 60 females; Table 1) were recruited for this study from Yanshan University. The sample size was determined a priori using G*Power (Version 3.1.9.7; Heinrich Heine University Düsseldorf, Düsseldorf, Germany) based on a two-way ANOVA (sex $\times$ limb dominance) design, with a medium effect size ($f = 0.25$), significance level $\alpha = 0.05$, desired statistical power = 0.95, two groups (male/female), two measurements (dominant/non-dominant), correlation among repeated measures = 0.5, and nonsphericity correction $\epsilon = 1$, which yielded a minimum required sample size of 54 participants. To improve representativeness and account for potential dropouts, 126 participants were ultimately recruited.

Table 1. Basic characteristics of participants.

Sex	N	Age (y)	Height (cm)	Weight (kg)	BMI(kg/m ²)
Male	66	20.26 ± 1.42	180.23 ± 4.23	72.03 ± 7.62	22.17 ± 2.19
Female	60	20.30 ± 2.71	166.22 ± 5.60	57.81 ± 7.50	20.90 ± 2.16
Total	126	20.28 ± 2.12	173.56 ± 8.57	65.26 ± 10.37	21.56 ± 2.26

The dominant limb was determined using a ball-kicking test [11], where the leg that kicked the ball the farthest was identified as the dominant side. The ball-kicking test has been validated in previous studies as a reliable approach for identifying lower-limb dominance [12,13]. Prior to data collection, all participants were fully informed of the study procedures, objectives, and requirements, and signed written informed consent forms. None of the participants engaged in strenuous physical activity within 24 h before testing, and Participants were screened based on self-reported health history, including musculoskeletal injury history, neurological conditions, and visual impairments, to ensure compliance with the inclusion criteria. In addition, the vestibular system was evaluated through pre-test observation and familiarization trials, where participants were asked to perform basic balance and single-leg stance tasks; only those who could complete these tasks without signs of dizziness, imbalance, or abnormal postural responses were included in the study. This study protocol was reviewed and approved by the Ethics Committee of Qinhuangdao First Hospital (Approval No.: 2025K-136-01).

2.1. Test Protocol

Postural Stability Test (Figure 1): Participants were instructed to place their hands on their hips and stand on a 30 cm-high wooden box positioned 5 cm from a force platform. At the beginning of the test, the non-testing leg was lifted, and participants leaned forward to initiate a natural fall, landing on the testing leg. After landing, they were required to maintain balance for 20 s until an auditory cue signaled the end of the trial [14]. During the test, participants were instructed to fix their gaze on a target on a blank wall in front of them. All tests were performed barefoot to eliminate the influence of footwear on postural stability. Trials were repeated if the non-supporting leg touched the ground or if the participant used their arms to regain balance. All participants completed a warm-up and one familiarization trial on each leg prior to formal testing. Each participant completed at least three valid trials per leg, with a 1-min rest between trials to reduce fatigue-related errors [15].



Figure 1. Setup of the single-leg landing postural stability test. (a) Front view; (b) Side view. X, Y, and Z indicate the force platform's anterior-posterior (AP), medial-lateral (ML), and vertical (V) directions, respectively.

The choice of experimental conditions followed previous studies on dynamic postural stability. A 30 cm-high box has been commonly used to induce sufficient vertical displacement without increasing injury risk, while barefoot testing and standardized gaze fixation were applied to minimize external influences on postural control. Similarly, the 20-s trial duration and the requirement of three valid repetitions per leg were adopted to ensure both the reliability and reproducibility of the measurements [14,15].

2.2. Data Acquisition and Processing

A Kistler 3D force platform (sampling rate: 1000 Hz) and the Bioware data acquisition system (Kistler Instrumente AG, Winterthur, Switzerland) [16] were used to collect GRF and center of pressure (COP) data during the dynamic postural stability tests. The first 5 s after landing were defined as the dynamic phase, and the period after 5 s was defined as the static phase [17]. The collected data were low-pass filtered at 50 Hz using Bioware software [18,19]. All variables related to GRF and COP were computed in MATLAB (R2020a, MathWorks, Natick, MA, USA) [20] to evaluate both dynamic and static balance abilities.

2.3. Main Outcome Measures

2.3.1. Time to Stabilization (TTS)

TTS was calculated using the Third-Order Polynomial Decay fitting method [8], as described in Equation (1). To avoid distortion due to the sharp increase in GRF during the first 100 milliseconds after landing, this period was excluded from the fitting process (Figure 2).

$$f(x) = a \cdot t^{-3} + b \cdot t^{-2} + c \cdot t^{-1} + d \quad (1)$$

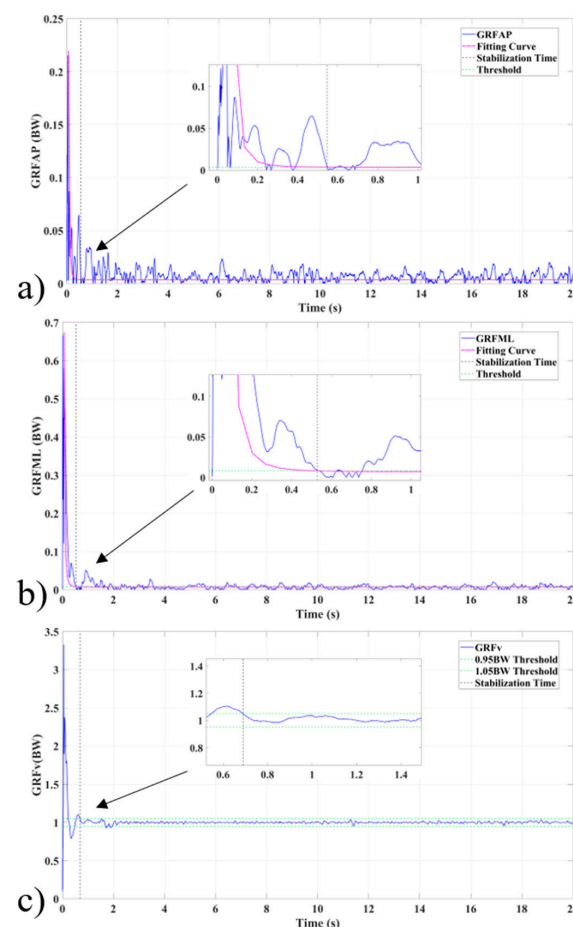


Figure 2. Example of TTS calculation for a single trial of one participant. (a) Anterior–posterior (AP) direction; (b) Medial–lateral (ML) direction; (c) Vertical (V) direction.

TTS in the anteroposterior direction (TTS_{AP}): defined as the time point within 0–12 s post-landing when the rectified AP GRF falls below 1.80% of BW. see Equation (2).

$$f(x) = t^{-3} - 30t^{-2} + 30t^{-1} - 0.01 \quad (2)$$

TTS in the mediolateral direction (TTS_{ML}): defined as the point when the rectified ML GRF falls below 1.23% BW (Figure 1). see Equation (3).

$$f(x) = t^{-3} - 10t^{-2} + 10t^{-1} - 0.01 \quad (3)$$

TTS in the vertical direction (TTS_V): defined as the earliest time when vertical GRF remains within 95–105% BW for at least 100 ms.

2.3.2. Dynamic Postural Stability Index (DPSI)

DPSI reflects overall postural stability by combining GRF variability across AP, ML, and vertical directions. A lower DPSI indicates better dynamic stability [21]. To account for differences in participant body weight, GRF data were normalized. The DPSI was computed over a 3-s window starting from initial ground contact: see Equation (4).

$$DPSI = \sqrt{\frac{\sum (0 - GRFx)^2 + \sum (0 - GRFy)^2 + \sum (Body\ Weight - GRFz)^2}{Number\ of\ data\ point}} \div Body\ Weight \quad (4)$$

AP Stability Index (APSI): reflects AP-direction stability; see Equation (5).

$$APSI = \sqrt{\frac{\sum (0 - GRFx)^2}{Number\ of\ data\ point}} \div Body\ Weight \quad (5)$$

ML Stability Index (MLSI): reflects ML-direction stability; see Equation (6).

$$MLSI = \sqrt{\frac{\sum (0 - GRFy)^2}{Number\ of\ data\ point}} \div Body\ Weight \quad (6)$$

Vertical Stability Index (VSI): reflects vertical stability; see Equation (7).

$$VSI = \sqrt{\frac{\sum (Body\ Weight - GRFz)^2}{Number\ of\ data\ point}} \div Body\ Weight \quad (7)$$

2.3.3. Root Mean Square Distance (RDIST)

RDIST quantifies the dispersion of AP and ML displacements from the mean COP location during the 10–15 s window post-landing; see Equation (8). It is commonly used to measure the spatial variability of COP [22]:

$$RDIST = \sqrt{\frac{1}{n} \sum_{i=1}^n \|x_i - \bar{x}\|^2} \quad (8)$$

where:

- x_i —AP position of the COP at sample i .
- $\bar{x}$ —Mean AP COP position.
- n —Total number of samples.

$RDIST_{AP}$: Root mean square (RMS) of displacement in AP direction; see Equation (9).

$$RDIST_{AP} = \sqrt{\frac{1}{n} \sum_{i=1}^n (x_{i,AP} - \bar{x}_{AP})^2} \quad (9)$$

$RDIST_{ML}$: RMS of displacement in ML direction; see Equation (10).

$$RDIST_{ML} = \sqrt{\frac{1}{n} \sum_{i=1}^n (x_{i,ML} - \bar{x}_{ML})^2} \quad (10)$$

2.3.4. Total Excursions (TOTEX)

TOTEX refers to the total length of the COP trajectory over time, calculated from 10 to 15 s post-landing; see Equation (11) [23]:

$$TOTEX = TOTEX_{AP} + TOTEX_{ML} \quad (11)$$

$TOTEX_{AP}$: total AP sway path; see Equation (12).

$$TOTEX_{AP} = \sum_{n=1}^{N-1} |AP[n+1] - AP[n]| \quad (12)$$

$TOTEX_{ML}$: total ML sway path; see Equation (13).

$$TOTEX_{ML} = \sum_{n=1}^{N-1} |ML[n+1] - ML[n]| \quad (13)$$

2.3.5. 95% Confidence Ellipse Area (AREA-CE)

The AREA-CE indicates the area encompassing approximately 95% of the COP data points, reflecting both dispersion and spatial concentration. It was computed during the 10–15 s period post-landing; see Equation (14) [24].

$$AREA - CE = \Pi(AP_{max} - AP_{min})(ML_{max} - ML_{min}) \quad (14)$$

2.4. Statistical Analysis

A two-way ANOVA was conducted to examine the main and interaction effects of sex and limb dominance on both dynamic and static postural stability indicators. Normality of the data was assessed using the Shapiro–Wilk test, and homogeneity of variance was evaluated with Levene’s test in SPSS (Version 23.0; IBM, Armonk, NY, USA). If assumptions of normality were met, two-way ANOVA was applied. To control for type I error due to multiple comparisons, Bonferroni correction was applied for all post-hoc pairwise comparisons. All results are reported as mean $\pm$ standard deviation ($M \pm SD$), with significance set at $p < 0.05$. Data visualization was performed using Origin 2021.

3. Results

As illustrated in Figures 3 and 4, results from the two-way ANOVA indicated that there were no significant interaction effects between sex and limb dominance across all measured variables ($p > 0.05$).

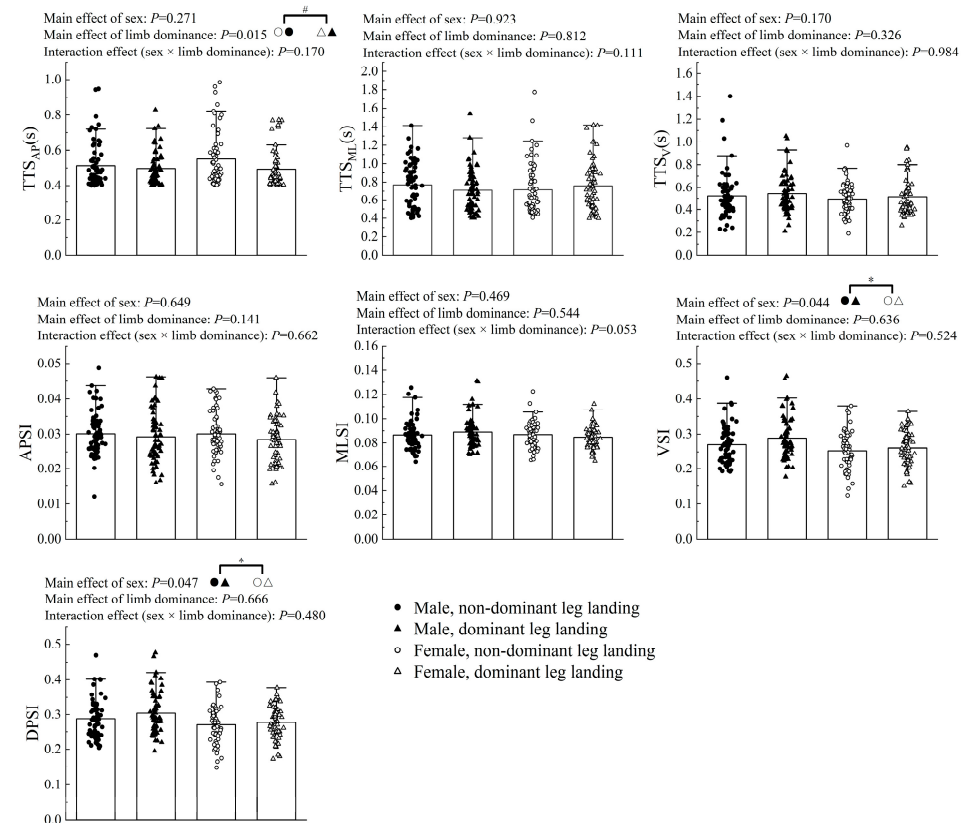


Figure 3. Comparison of dynamic postural stability between sexes and limb dominance. Note: * indicates a significant difference between males and females; # indicates a significant difference between dominant and non-dominant sides.

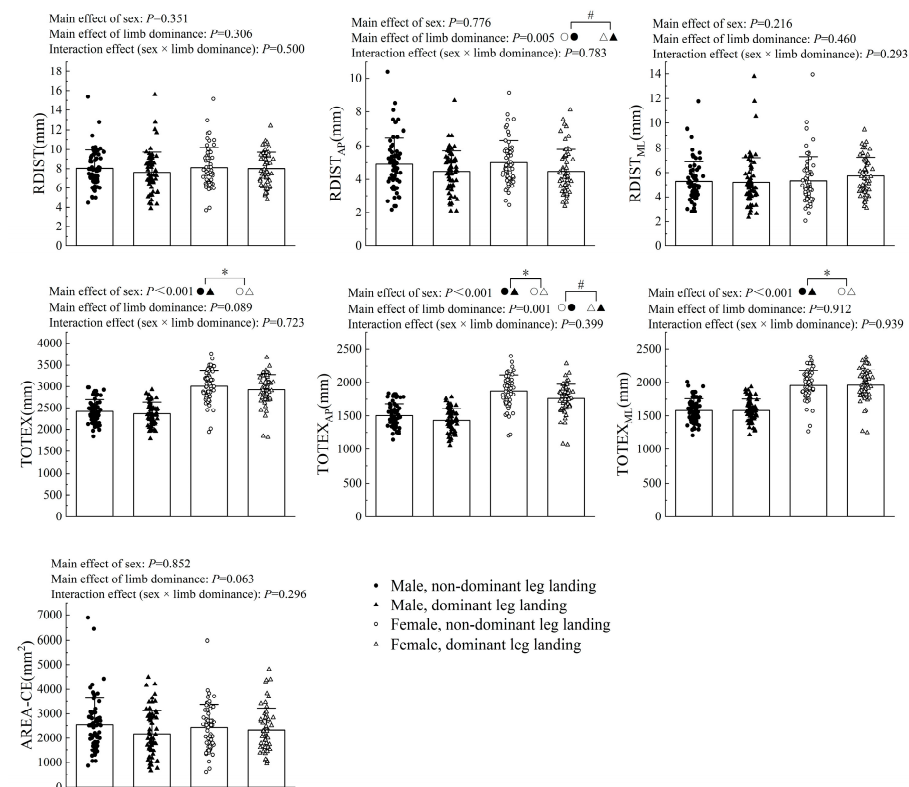


Figure 4. Comparison of static postural stability between sexes and limb dominance. Note: * indicates a significant difference between males and females; # indicates a significant difference between dominant and non-dominant sides.

3.1. Dynamic Postural Stability Results

The two-way ANOVA results (see Figure 3) showed that, regardless of sex, the TTS_{AP} on the non-dominant side after landing (0.53 s vs. 0.49 s, $p = 0.015$, $F = 5.954$, partial $\eta^2 = 0.023$) was significantly greater than that on the dominant side. Irrespective of limb dominance, males exhibited significantly greater VSI (0.28 vs. 0.26, $p = 0.044$, $F = 4.103$, partial $\eta^2 = 0.016$) and DPSI (0.29 vs. 0.27, $p = 0.047$, $F = 3.982$, partial $\eta^2 = 0.016$) compared to females.

3.2. Static Postural Stability Results

Results of the two-way ANOVA for static postural stability (see Figure 4) showed that, regardless of sex, the $RDIST_{AP}$ on the non-dominant side (4.95 mm vs. 4.46 mm, $p = 0.005$, $F = 7.892$, partial $\eta^2 = 0.031$) and the $TOTEX_{AP}$ (1678.73 mm vs. 1590.76 mm, $p < 0.001$, $F = 188.490$, partial $\eta^2 = 0.432$) were significantly greater than those on the dominant side. Regardless of limb dominance, males demonstrated significantly lower values in total sway path compared to females, including $TOTEX$ (2395.40 mm vs. 2966.24 mm, $p < 0.001$, $F = 215.546$, partial $\eta^2 = 0.465$), $TOTEX_{AP}$ (1469.39 mm vs. 1816.64 mm, $p = 0.001$, $F = 12.379$, partial $\eta^2 = 0.048$), and $TOTEX_{ML}$ (1580.18 mm vs. 1959.16 mm, $p < 0.001$, $F = 225.726$, partial $\eta^2 = 0.476$).

4. Discussion

A key finding of this study is the dissociation between dynamic and static postural stability across sexes and between dominant and non-dominant limbs. Females demonstrated superior dynamic stability, whereas males showed better static stability, with the dominant limb exhibiting superior dynamic and static postural control compared with the non-dominant limb under standardized assessment. This pattern highlights that postural control strategies are highly task- and limb-dependent, with substantial effect sizes indicating meaningful practical implications.

4.1. Sex and Limb Dominance Differences in Dynamic Postural Stability

In this study, a Kistler three-dimensional force platform was used to quantify the TTS and DPSI after single-leg landing. These indicators have been validated to possess good reliability, validity, and sensitivity [25,26], and are widely used for assessing dynamic postural stability after single-leg landings. The current findings revealed that TTS_{AP} on the non-dominant side was significantly greater than that on the dominant side, indicating that the non-dominant limb required more time to stabilize in the anteroposterior direction after landing. According to Takaya et al. [27], a longer TTS_{AP} may represent a physiological protective adaptation mechanism, allowing for greater degrees of freedom in the sagittal plane, thereby granting the body more time and space to adjust posture and reduce the risk of injury due to imbalance.

Moreover, this study found that, compared with females, males exhibited significantly higher VSI and DPSI values after single-leg landing, indicating that females performed better in both horizontal-plane dynamic stability and overall dynamic stability. Similar findings were reported by Dallinga et al. [28], who found that males had higher VSI and DPSI than females in forward, diagonal, and lateral single-leg landings. Wikstrom et al. [29] also observed greater VSI and DPSI in males following maximum vertical jump landings, consistent with the present results.

It is worth noting that some previous studies reported findings that are not fully consistent with ours. For example, Wikstrom [26] evaluated dynamic postural stability after single-leg landings in healthy adults and found that females exhibited superior vertical dynamic stability compared with males, while no sex differences were observed

in the anteroposterior direction. This differs from our results, which showed that males demonstrated better static postural stability, whereas females performed better in dynamic stability. The discrepancy may be primarily due to differences in assessment methods. For instance, Wikstrom et al. focused only on vertical dynamic stability, whereas our study evaluated anteroposterior and mediolateral directions as well as static and dynamic stability. Participant characteristics and experimental design may also have some effect but are likely less influential. These inconsistencies suggest that sex differences in postural control are influenced by multiple factors and highlight the need for further research to clarify the underlying mechanisms.

Dynamic balance refers to the ability of the body to maintain or quickly regain stability while in motion or under external force. Enhancing dynamic balance during athletic training and rehabilitation can help reduce injury risk and improve performance. The observed lower dynamic balance ability in males may be attributed to differences in landing strategies [30,31]. Studies have shown [32] that females tend to adopt softer and more controlled landing strategies, which help reduce vertical ground reaction forces and subsequently lower injury risk. Conversely, males are more likely to employ forceful and direct “hard” landings, which, while potentially advantageous in certain athletic contexts, may lead to increased vertical ground reaction forces and a higher risk of injury. Additionally, females tend to have a relatively lower center of gravity, and when the base of support remains unchanged, a lower center of mass improves stability. Therefore, females’ superior dynamic balance performance observed in this study may be attributed to such biomechanical advantages. They should be considered as possible mechanisms rather than definitive conclusions.

4.2. Sex and Limb Dominance Differences in Static Postural Stability

Current research on static balance control primarily relies on COP parameters. The COP parameters used in this study quantitatively reflect postural stability and are widely adopted in evaluating static balance ability [33–35]. COP sway is negatively correlated with static postural stability [36]. This study found that $RDIST_{AP}$ and $TOTEX_{AP}$ were significantly greater on the non-dominant side than on the dominant side. These differences, mainly observed in the anteroposterior direction, are attributed to the regulation of ankle plantarflexion and dorsiflexion torques [37]. It has been suggested that the non-dominant ankle possesses relatively weaker muscular strength and cannot generate sufficient torque to counteract postural instability [38], resulting in greater COP sway in the anteroposterior direction during postural stability testing. Further analysis indicated that the non-dominant limb adopts a different strategy to maintain static posture by increasing forward body inclination, which may help lower the center of gravity and enhance single-leg stance stability. At the same time, the non-dominant side tended to make quicker backward adjustments when attempting to regain balance, contributing to greater $RDIST_{AP}$ and $TOTEX_{AP}$.

Additionally, the findings revealed that males exhibited significantly lower $TOTEX$, $TOTEX_{AP}$, and $TOTEX_{ML}$ than females following landing, indicating better static postural stability in males. It is noteworthy that while males performed better in static stability, females outperformed males in dynamic postural stability. These results align with previous research [39] and further confirm sex differences in postural control. This implies that both sex and limb dominance should be considered during balance training, with individualized training strategies adopted to enhance postural stability, reduce injury risk, and improve performance.

In summary, comparisons of dynamic and static postural stability following single-leg landings revealed significant differences by sex and limb dominance. The non-dominant

side exhibited longer stabilization times, larger root mean square distances, and greater total excursion in the anteroposterior direction, whereas the dominant side demonstrated superior balance control. In terms of sex differences, males showed better static postural stability, while females performed better in dynamic stability, likely due to their use of more controlled and softer landing strategies that reduce vertical ground reaction forces and injury risk. These findings suggest that training and rehabilitation programs should account for both limb dominance and sex differences, using targeted interventions to enhance postural stability, minimize injury risk, and improve overall athletic performance.

4.3. Practical Applications of Polynomial Fitting Method

Furthermore, the polynomial fitting method used to calculate TTS and quantify postural stability in this study may have practical applications in both clinical and sports settings. By providing a standardized and sensitive approach to evaluate dynamic and static balance, this method can assist clinicians and coaches in identifying individuals with balance deficits, monitoring training progress, and tailoring rehabilitation programs. For example, targeted interventions could be designed to improve limb-specific stabilization times or compensate for sex-related differences in dynamic balance. The utility of such approaches is supported by recent studies highlighting the integration of biomechanical assessments with functional and sensory-motor training to enhance health and athletic performance [40,41].

Importantly, the quantitative TTS and DPSI indices obtained from polynomial fitting could also be incorporated into biomechanical or musculoskeletal modeling studies. Integrating these indices into forward-dynamics simulations or predictive models may allow researchers to explore sex- and limb-specific postural control mechanisms, refine analytical frameworks, and predict individual responses to interventions. This approach bridges descriptive data analysis and predictive modeling, enhancing the methodological impact of the study.

4.4. Limitations

A limitation of the present study is that the landing strategy of participants, including the part of the foot contacting the force platform, trunk forward lean, and knee flexion angle, was not strictly controlled or measured. These factors can influence postural stability outcomes, as different landing patterns may result in variations in ground reaction forces and center of pressure trajectories. Therefore, individual differences in natural landing strategies could have contributed to variability in the stability measures. Future studies should consider standardizing or recording these biomechanical parameters to further reduce variability and clarify their effects on postural control.

5. Conclusions

The analysis revealed that, compared with the non-dominant limb, the dominant side exhibited superior performance in stabilization time, anteroposterior root mean square distance, and total excursion following single-leg landings, indicating better dynamic and static postural control in the dominant limb among young adults. Furthermore, males demonstrated significantly better static postural stability than females, whereas females showed superior dynamic postural performance. These results suggest that incorporating balance training for the non-dominant limb can markedly improve overall postural stability in young individuals. Simultaneously, both sex and limb dominance should be considered during dynamic and static balance training, with targeted interventions implemented to improve training efficiency and optimize athletic performance.

Moreover, the polynomial fitting method applied in this study could be extended in future research to incorporate advanced movement analysis techniques, such as computer vision-based motion capture and markerless tracking systems [42,43]. These approaches may allow real-time monitoring of postural stability, facilitate remote or large-scale assessments, and complement traditional force platform measurements, enhancing both clinical and sports applications.

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Conflicts of Interest: The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Abbreviations

The following abbreviations are used in this manuscript:

COM	Center of mass
ACL	Anterior cruciate ligament
TTS	Time to stabilization
GRF	Ground reaction force
COP	Center of pressure
DPSI	Dynamic postural stability index
RDIST	Root mean square distance
TOTEX	Total excursions
AREA-CE	95% Confidence ellipse area

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