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published in
Gait and Posture
2019

DOI (link to publisher)
10.1016/j.gaitpost.2019.06.015

document version
Publisher's PDF, also known as Version of record

document license
Article 25fa Dutch Copyright Act

Gait & Posture 73 (2019) 80–85

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

The assessment of single-leg drop jump landing performance by means of ground reaction forces: A methodological study

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ARTICLE INFO

Keywords:
Drop jump landing
Dynamic testing
Postural stability
Balance
Parameters

ABSTRACT

Background: Time to stabilization (TTS) and dynamic postural stability index (DPSI) are outcome measures based on ground reaction force (GRF) that are often used to quantify dynamic postural stability performance following a drop jump landing. However, their interrelations, as well as the overlap with other dynamic measures and static single-leg postural sway, are unknown.

Research question: What is the relation among TTS and DPSI, how are they related to impact forces and dynamic postural sway, and how are all these dynamic measures related to static postural sway?

Methods: A sample of 190 elite soccer players performed four single-leg drop jump landings. TTS in three directions (vertical, anteroposterior, and mediolateral), and DPSI were intercorrelated (Pearson's r), and related to impact forces and the magnitude of horizontal GRF (HGRF) from 0.4 to 2.4 s and 3.0–5.0 s following landing. All these measures were also correlated to HGRF in the static phase (i.e., 5.3–11.7 s).

Results: The TTS measures were significantly interrelated (r = 0.28-0.53), but were not significantly correlated to DPSI. TTS was more strongly related to HGRF0.4–2.4 s (r = 0.54-0.75) than to HGRF3.0–5.0 s (r = 0.32-0.54) or impact forces (r = -0.28-0.36). Vertical TTS was not significantly related to impact forces. The DPSI was most strongly related to the vertical peak force (r = 0.85), and was not significantly related to HGRF of the dynamic periods. Furthermore, TTS and dynamic HGRF were significantly related to static HGRF (r = 0.34-0.80), while DPSI and impact forces were not.

Significance: TTS and DPSI do not represent similar aspects of single-leg jump landing performance. The ability to stabilize posture seems to be represented by TTS and dynamic postural sway, which partly overlaps with static postural sway. In contrast, DPSI and vertical peak force mainly reflect the kinetic energy absorption during impact. The findings can help to better understand the meaning of the outcome measures, and to translate results to rehabilitation or prevention programs.

1. Introduction

Balance tests with outcome measures based on ground reaction forces (GRF) have been successfully applied to assess sensorimotor control [1–4]. However, there is an increasing interest in dynamic tasks to assess sensorimotor performance, of which a jump landing stabilization task is the most commonly used [5–7]. The complexity and sport specificity of the jump landing is thought to yield better sensitivity in athletes [5]. On the other hand, the task execution is more complex compared to static balance tests. For instance, the jump landing strategy is likely affected by feedforward components of motor control [8,9], which are not present in the feedback driven sensorimotor control during static balance tasks [10]. Furthermore, dependent on the jump direction, the impact of the landing yields high vertical and anteroposterior GRF values [11], which may vary between attempts.

The most commonly applied dynamic postural stability outcome measures are the 'time to stabilization' (TTS) and the 'dynamic postural stability index' (DPSI) [6,9]. The TTS aims to reflect the time it takes for an individual to stabilize following landing to a stable state (i.e., a normal single-leg stance) [12,13], or to minimize body sway as quickly

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https://doi.org/10.1016/j.gaitpost.2019.06.015
Received 15 March 2018; Received in revised form 10 April 2019; Accepted 20 June 2019
0966-6362/

as possible [14]. Several calculation methods have been proposed, which vary with respect to the directions of GRF (i.e., vertical (V), anteroposterior (AP), and mediolateral (ML)), additional processing of the GRF signal (e.g., rectifying, fitting procedures, and sequential average processing), and the definition of the stable state (stability threshold) [11,15–17]. It is likely that these differences in calculation methods of TTS explain previous contradictory findings, even within studies [12,14]. To overcome some flaws in the calculation of TTS, original calculation methods have been modified over time [17]. For instance, a commonly applied calculation method used each participant's static single-leg GRF range-of-variation as its reference to define the stable state and calculate TTS [16,18]. As a consequence, a participant with poorer static single-leg balance would have a higher threshold or reference, paradoxically resulting in a shorter TTS [12,17]. To avoid this problem, most TTS calculation methods in V direction now use a fixed but normalized to body weight reference [15,19], and this has been recently validated and recommended for AP and ML directions as well [12].

As part of the improvements to overcome the 'stability paradox' of previous TTS methods, the dynamic postural stability index (DPSI) has been introduced as well [13]. The DPSI combines the mean squared deviations of GRF from 0 to 3 s following landing in V, AP, and ML directions into one comprehensive measure [13]. Both TTS and DPSI have been claimed to provide information about the ability to regulate rapid center of mass (COM) accelerations following a single-leg landing. Therefore, both measures are often grouped together when interpreted [9,12,13,20], though previous studies did not show a significant relation between TTS and DPSI [21,22]. It is currently unknown whether the improved TTS measures and DPSI provide related information on drop jump landing performance.

Most jump tasks consists of a jump movement in vertical and anterior direction, and the impact forces in these directions are common outcome measures [6,9]. As these high impact GRF specifically yield information about the kinetic energy absorption [23,24], they are usually considered separately from dynamic postural stability measures [6,9]. As the calculation of DPSI includes the impact forces, DPSI holds information about both kinetic energy absorption and the subsequent postural sway up to 3 s post landing. This could be important since kinetic energy absorption and postural stabilization may be independent aspects of jump landing performance. In a recent study, it was shown that the magnitude of the horizontal GRF (HGRF) during 0.4–2.4 s following a single-leg drop jump landing holds different information compared to the time periods of 0-0.4 s and 3.0–5.0 s [23]. The time period of 0-0.4 s was likely affected by the high impact force in anteroposterior direction. Between 0.4 and 2.4 s the drop in COM accelerations was still rather large, between 3.0 and 5.0 s the drop was small, and after 5 s following landing the single-leg stance was considered as a static balance [23,25]. To date, the interrelations among improved TTS, DPSI, peak forces, dynamic HGRF, and static HGRF are largely unknown. Such knowledge would provide insight about the characteristics of jump landing performance that can be measured with a force plate. This may help to adequately compare studies and to translate results into underlying functional impairments. In turn, this would allow the evaluation of targeted interventions to improve these functional impairments.

We aimed to analyze the relation between the most commonly applied dynamic postural stability outcome measures (i.e., TTS-V, TTS-AP, TTS-ML, and DPSI) that are used to determine performance following a drop jump landing. Even more so, we wanted to examine how these outcome measures relate to peak impact forces (in V, AP and ML directions) and horizontal forces during dynamic time periods (i.e., 0.4–2.4 s and 3.0–5.0 s). Finally, we related all the dynamic measures to the horizontal forces of the static period as well. We hypothesized that TTS and horizontal forces during dynamic time periods are highly interrelated. Given the expected high forces of impact in vertical and anteroposterior directions, and the amplification of peaks in a root-mean-square construct, we also hypothesized that DPSI is mainly related to these peak impact forces.

Table 1
Player characteristics.

	11-12 yr (n = 35)	13-14 yr (n = 46)	15-16 yr (n = 41)	17-18 yr (n = 44)	19-33 yr (n = 24)
Age (yr)	11.8 (0.6)	13.9 (0.5)	15.8 (0.5)	17.7 (0.7)	23.2 (3.2)
Height (m)	1.52 (0.08)	1.66 (0.09)	1.76 (0.08)	1.80 (0.06)	1.82 (0.06)
Mass (kg)	41.2 (6.4)	53.9 (9.3)	65.0 (10.1)	73.5 (7.6)	77.0 (7.5)

Presented as mean (SD); n: number of participants.

2. Methods

2.1. Participants

Data were retrieved from the database of AFC Ajax, a professional soccer club. In total 190 players were measured, 143 at the start of season 2012/2013 and 47 at the start of season 2013/2014 (Table 1). Participant or parental consent (depending on the age) was collected. The local review board granted ethical approval.

2.2. Procedures

The players were asked to hop from an aerobic step of 20 cm height, which was placed 5 cm posterior to the force plate. Players took off by means of a small drop jump with two feet, landed on the testing leg at the center of the force plate, and stabilized as quickly as possible. They kept all movement to a minimum standing on the testing leg for 15 s, while keeping their hands on the iliac crest [23]. Players were instructed to look at a visual target 4 m ahead on a blank wall. Before actual testing commenced, all players completed their warm-up, as accustomed before a training session, and performed one practice trial per leg. Two tests were administered on both legs for 143 players and three tests on both legs for 47 players due to a change of protocol during the study. The left leg was designated as the initial testing leg. All trials were performed barefoot to avoid stability assistance from a shoe [26]. A trial was discarded and repeated if a player touched the floor with the other leg or if arm movement was used to regain balance.

2.3. Data processing

The GRF were recorded at 1000 samples/s, using a 40 x 60 cm AMTI force plate (type BP400600HF, Advanced Medical Technologies Inc., Watertown, MA, USA). A custom MATLAB (The Mathworks Inc., version R2018b, Natick, RI, USA) program was written for data processing. Data were cropped from time of contact (vertical GRF > 10 N) to 12 s after contact and were low pass filtered at 12 Hz with a bidirectional second order Butterworth filter [3,23]. The GRF were normalized to body weight (BW) by means of GRF (N) / BW (N), and expressed as percentage (%BW).

2.4. Dynamic stability measures

Dynamic stability measures were carefully selected from previous studies, incorporating calculation modifications and recommendations for improvement made over time [11–13,15]:

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  - TTS-V; time to stabilization in vertical direction; the first instant that the vertical GRF signal stayed within the thresholds of 95–105% BW for at least 1.0 s within the time period of 0–12 s following impact [15,27].
  - TTS-AP; time to stabilization in anteroposterior direction; the

instant that a third order polynomial decay fit (see below) of the rectified GRF signal in anteroposterior direction crossed the threshold of 1.80% BW within the time period of 0–12 s following impact [12].
- TTS-ML; time to stabilization in mediolateral direction; the instant that a third order polynomial decay fit (see below) of the rectified GRF signal in mediolateral direction crossed the threshold of 1.23% BW within the time period of 0–12 s following impact [12].
- DPSI; dynamic postural stability index; the root mean square value of the magnitude of the resultant (Euclidian sum of the V, AP, and ML components) GRF signal from 0 to 3 s following impact. Note that in V direction 100% BW was subtracted from the GRF signal before calculation [13].

The decay function to fit the rectified GRF signal in AP direction was computed in Matlab according to Wright et al. [12]: $f(t) = a{\cdot}t^{-3} + b{\cdot}t^{-2} + c{\cdot}t^{-1} + d{\cdot}t^{0}$ (initial guess of a = 1, b = -30, c = 30, d = -0.01; 'f' represents force value on the y-axis in %BW; 't' represents sample number on the x-axis (1000 samples/s)). The first 100 ms after ground contact were not taken into account, as the steep increase of the force would bias the fit in the wrong direction. In optimizing the fit, the error between the fitted curve and the rectified GRF signal in AP direction from 0.1 to 4 s was weighted twice, the error from 4 to 12 s was weighted once for as long as the signal was above 0.3 times the limit of 1.80%BW, otherwise the error was not taken into account. A similar approach was applied to the rectified GRF signal in ML direction (initial guess of a = 1, b = -10, c = 10, d = -0.01), with the proviso that the signal before the highest peak within the first 0.4 s was not taken into account and that the reference limit was set at 1.23%BW [12].

2.5. Landing and postural sway parameters

Peak forces in vertical (peakV) and anteroposterior direction (peakAP) provided information about the impact of the landing [6]. Close inspection of the patterns of GRF in ML direction for the first 0.4 s showed, in contrast to V and AP directions, inconsistent patterns and very low values for the peak force (Fig. 1). To overcome this lack of a consistent peak force in ML direction, we calculated the root-mean-square of the GRF in the ML direction for 0-0.4 s (GRFML0.4).

The magnitude of the horizontal GRF was considered a proxy for center of mass sway and the extent of corrective motor actions [1,3,28,29]. The mean length of the horizontal GRF vector (HGRF), calculated as the Euclidean norm of the AP and ML components, has shown high correlations with other balance measures, such as COP speed [23,25,28]. Previous research has shown that, following a jump-landing task of youth elite soccer players, averaging HGRF within time periods of 0.4–2.4 s, 3.0–5.0 s, 5.3–8.3 s and 8.7–11.7 s resulted in the most optimal data reduction of the total HGRF time series of 0–12 s, with minimal loss of variance between participants [23]. Fig. 2 illustrates the magnitude of HGRF following landing, by means of the averaged values of the four representative time periods [23]. In accordance with previous studies [23,25], a steady state was reached after 5 s, therefore the HGRF0.4–2.4 s and HGRF3.0–5.0 s were used to represent dynamic periods [23], while HGRF5.3–11.7 s was used as a proxy for static single-leg stance performance [25].

2.6. Data analyses

For each participant, the outcome values were averaged over all trials for each outcome measure. If data were not normally distributed, we adjusted the distribution by means of Box-Cox transformation with increments of the power by 0.5 [30]. Therefore, TTS-V, TTS-ML, and HGRF were transformed by means of inversion (data $^{-1}$). Table 2 shows the outcome values and distributions for each of the outcome measures. The inversion of the data resulted in a reverse rank of the players, which was corrected during the process of z-score calculations. All

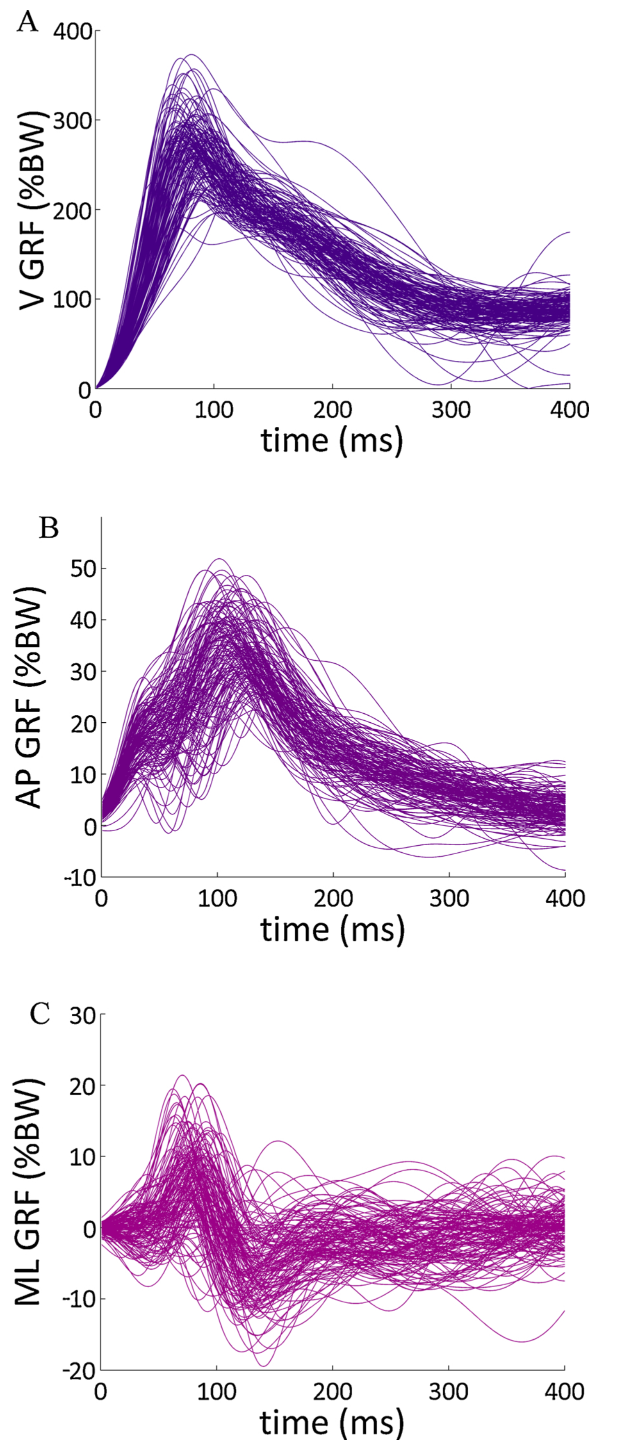

Fig. 1. The GRF (%BW) of all trials during 0 to 0.4 s after contact for vertical direction (A), anteroposterior direction (B), and mediolateral direction (C).

outcome values were standardized by calculating z-scores:

$$z\text{-}score = \frac{outcome - mean}{SD}$$

2.7. Statistical analyses

The associations across the dynamic stability outcome measures (i.e., TTS-V, TTS-AP, TTS-ML, and DPSI) were assessed by means of Pearson's correlation coefficients. The associations between on the one hand TTS and DPSI, and on the other hand the impact forces and

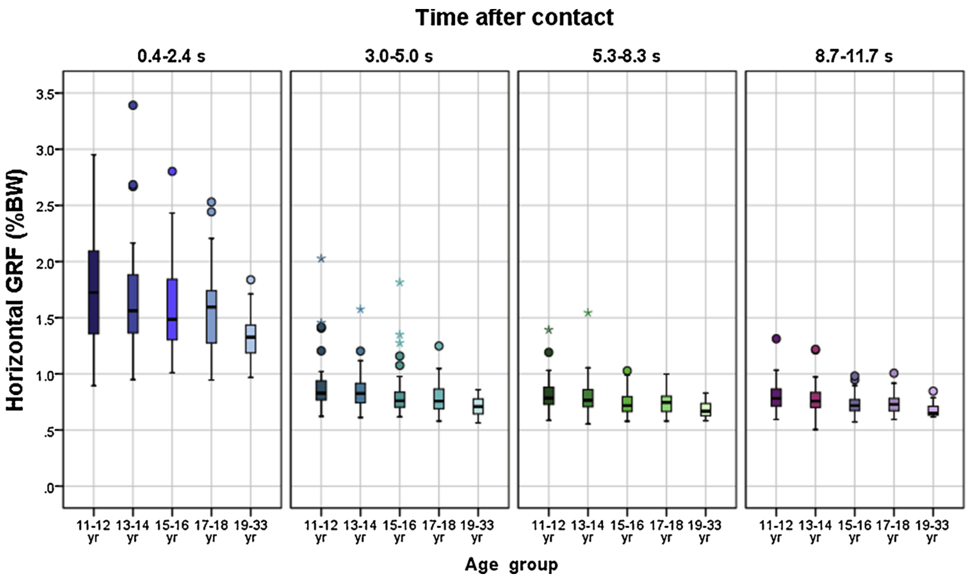

Fig. 2. Box-plot of the horizontal GRF (%BW)) following landing, illustrated by means of the averaged value of four time periods for each age group. The box represent the 25 to 75th percentile, or the interquartile distance. The line within the box is the median. The range represent the lowest or highest value within 1.5 times the interquartile distance to the box. Circle symbols represent values between 1.5–3.0 times the interquartile distance to the box ('weak' outliers), while the star symbols represent values more than 3.0 times the interquartile distance to the box ('extreme' outliers).

postural sway during the dynamic time periods (HGRF0.4–2.4 s and HGRF3.0–5.0 s) were assessed as well. Finally, the correlations between all the dynamic measures and static single-leg performance (based on HGRF > 5 s) were calculated. To interpret the strength of the correlations, for absolute values of r, 0-0.19 is regarded as very weak, 0.2-0.39 as weak, 0.40-0.59 as moderate, 0.6-0.79 as strong and 0.8–1 as very strong. Given the multiple testing, statistical significance was set at a p-value below 0.01.

3. Results

The TTS-V, TTS-AP, and TTS-ML were significantly positively interrelated with a weak to moderate strength (r = 0.26-0.53), but none of the TTS measures was significantly correlated to DPSI (Table 3A). For TTS measures, the strongest correlation was between TTS-V and HGRF0.4–2.4 s (r = 0.75). All TTS measures were stronger related to HGRF0.4–2.4 s (r = 0.54-0.75) than to HGRF3.0–5.0 s (r = 0.32-0.54) and impact forces (r = -0.28-0.36). The TTS-V was not significantly related to the impact forces. In contrast, DPSI was not significantly related to HGRF in the dynamic time periods, but was very strongly correlated to peakV (r = 0.85) (Table 3B).

Out of all the dynamic measures, the TTS measures and HGRF in dynamic time periods were significantly related to static balance, ranging from weak to very strong (r = 0.34-0.80), while impact forces and DPSI were not significantly related to static balance (Table 3C).

Table 3A
Correlations among TTS and DPSI.

	TTS-V	TTS-AP	TTS-ML	DPSI
TTS-V	x	0.43	0.53	ns
TTS-AP	0.43	x	0.28	ns
TTS-ML	0.53	0.28	x	ns
DPSI	ns	ns	ns	x

Values represent Pearson's correlation coefficient; ns: non-significant (p ≥ 0.01).
Time to stabilization in vertical (TTS-V), anteroposterior (TTS-AP), and mediolateral (TTS-ML) direction; DPSI: dynamic postural stability index. All outcome values were standardized to z-scores before analyses.

4. Discussion

The present findings demonstrate a distinction between on the one hand TTS, and postural sway during dynamic and static time periods, and on the other hand DPSI, peakV and peakAP. It seems safe to assume

Table 2
Descriptive results of the outcome measures.

	mean	SD	min	max	skewness	kurtosis	SW-test (p-value)	transformation
Dynamic stability								
TTS-V	1.05	0.49	0.41	3.43	1.7	3.8	< 0.0001	inversion
TTS-AP	0.90	0.20	0.49	1.49	0.4	0.1	0.037	x
TTS-ML	1.04	0.33	0.50	2.36	1.2	1.9	< 0.0001	inversion
DPSI	28.2	3.7	18.7	42.6	0.4	1.3	0.008	x
Impact forces								
peakV	270	26	190	346	−0.1	0.5	0.505	x
peakAP	35.5	4.7	24.4	47.5	0.4	−0.1	0.047	x
MLGRF 0-0.4 s	4.44	0.86	2.44	7.13	0.4	0.4	0.031	x
Dynamic balance								
HGRF 0.4-2.4 s	1.61	0.43	0.94	3.39	1.3	2.1	< 0.0001	inversion
HGRF 3.0-5.0 s	0.81	0.17	0.56	1.58	1.8	4.4	< 0.0001	inversion
Static balance								
HGRF 5.3-11.7 s	0.75	0.11	0.53	1.38	1.5	4.9	< 0.0001	inversion

TTS: time to stabilization in vertical (TTS-V), anteroposterior (TTS-AP), and mediolateral (TTS-ML) direction (s); DPSI: dynamic postural stability index; peak: peak GRF in vertical (peakV) and anteroposterior (peakAP) direction (%BW); MLGRF0-0.4 s: root-mean-square of GRF around zero during 0-0.4 s post landing (%BW); HGRF: mean rectified horizontal ground reaction force during different time periods post landing (%BW); SW-test: Shapiro-Wilk normality test. Transformation: type of transformation applied to the data to achieve a normal distribution of outcome values for further analyses.

Table 3B
TTS and DPSI related to impact forces and dynamic HGRF.

	Impact forces			Dynamic balance	
	peakV	peakAP	MLGRF (0-0.4 s)	HGRF (0.4-2.4 s)	HGRF (3.0-5.0 s)
TTS-V	ns	ns	ns	0.75	0.48
TTS-AP	−0.26	−0.28	ns	0.54	0.32
TTS-ML	ns	ns	0.32	0.68	0.54
DPSI	0.85	0.41	0.23	ns	ns

Values represent Pearson's correlation coefficient; ns: non-significant ($p \geq 0.01$).
peakV: peak GRF in vertical direction; peakAP: peak GRF in anteroposterior direction; MLGRF0-0.4 s: root-mean-square of GRF around zero during 0-0.4 s post landing; HGRF: mean rectified horizontal ground reaction force during 0.4–2.4 s and 3.0–5.0 s post landing; All outcome values were standardized to z-scores before analyses.

Table 3C
Dynamic measures related to static balance.

	Static balance
TTS-V	0.47
TTS-AP	0.34
TTS-ML	0.54
DPSI	ns
peakV	ns
peakAP	ns
MLGRF (0-0.4 s)	ns
HGRF (0.4-2.4 s)	0.50
HGRF (3.0-5.0 s)	0.80

Values represent Pearson's correlation coefficients; ns: non-significant ($p \geq 0.01$). Static balance is represented by HGRF 5.3–11.7 s. All outcome values were standardized to z-scores before analyses.

that the current TTS calculation methods and postural sway during 0.4–2.4 s reflect the ability to minimize body motion following landing. In contrast, DPSI, peakV and peakAP reflect the fast COM deceleration during impact of the landing (< 0.4 s), which appeared to be unrelated to subsequent motor actions used to further minimize body motion.

The ability to minimize body motion after landing reflected in TTS and postural sway during 0.4–2.4 s appeared to be partly associated with static balance performance (Table 3C). Probably, both skills rely on similar aspects of sensorimotor control. However, the weak to moderate strength of the relation also indicates the existence of other sources of variation. A jump-landing task likely creates higher demands on the sensorimotor control, including feedforward motor control components, and will have more variable performance outcomes due to increased complexity, or variations in the perturbation itself. Future studies may reveal whether the drop jump landing performance uncovers underlying sensorimotor control issues to a higher degree than static balance performance. In accordance with previous studies [31,32], the DPSI was not related to static balance performance. Instead, DPSI likely reflects a combination of jump height, forward propulsion of the drop jump and the stiffness of the body during impact, which are contributors to the fast COM decelerations during impact. Possibly, this also explains previous findings that balance training did not improve DPSI outcomes [20,33]. However, both landing kinetics and balance performance have shown to be associated with injuries [6,34]. As such, the DPSI could be relevant to consider for injury risk and rehabilitation management, but its additive value to peak impact forces has to be confirmed.

Despite the fact that TTS-V and HGRF 0.4–2.4 s rely on different dimensions (time versus magnitude of force, respectively) and different directions of force (vertical versus horizontal, respectively), their relation was strong. Furthermore, the range of the TTS outcome values (0.4–3.3 s) was in line with previous findings in elite soccer players [7], and are in accordance with previous and current observations that the largest decline in corrective motor actions occurred within 0.4–2.4 s (Fig. 2) [23,25]. These observations further strengthen the validity of TTS and HGRF0.4–2.4 s to measure the ability to minimize body motion after landing. In contrast to TTS-V, TTS-AP and TTS-ML were based on a 'best fit' calculation of the GRF. Therefore, for AP and ML direction, the GRF < 0.4 s could have affected the fitted signal of the subsequent 0.4–2.4 s time period. Perhaps this explains the significant relations of TTS-AP and TTS-ML with the impact forces.

Limitations of the present study include a low number of trials per participant (twice for each leg for the majority of participants), which was chosen to limit the burden on participants and coaches. Additionally, while the present results are most likely generalizable to other jump landing protocols, a jump in another direction will significantly change any of the direction specific parameters [26]. Furthermore, differences in TTS calculation methods can have important effects on TTS outcome values [35]. Therefore, the present findings might not be generalizable to other TTS calculation methods. We have been consistent with previous recommendations about the TTS calculation methods [12,17] and selected the most appropriate TTS calculations, but still there might exist better calculation methods.

In conclusion, the three TTS measures, and postural sway during dynamic and static time periods provided a significant overlap of information on the sensorimotor control of elite soccer players following a drop jump landing. On the other hand, the DPSI was very strongly related to peak vertical impact force. This indicates that the DPSI does not provide information about the ability to stabilize posture following landing, but reflects the kinetic energy absorption during impact.

Acknowledgements

The authors would like to thank Dr Max Feltham for his useful comments whilst preparing this manuscript.

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