

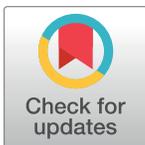
RESEARCH ARTICLE

Limits in reliability of leg-spring and joint stiffness measures during single-leg hopping within a sled-based system

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OPEN ACCESS

Citation: Diggin D, Anderson R, Harrison AJ (2019) Limits in reliability of leg-spring and joint stiffness measures during single-leg hopping within a sled-based system. PLoS ONE 14(12): e0225664. <https://doi.org/10.1371/journal.pone.0225664>

Editor: Andreas Kramer, Universitat Konstanz, GERMANY

Received: June 18, 2019

Accepted: November 8, 2019

Published: December 5, 2019

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Data Availability Statement: The data underlying the results presented in the study is publicly available at: <https://figshare.com/> (DOI: [10.6084/m9.figshare.10308539](https://doi.org/10.6084/m9.figshare.10308539)).

Funding: The authors received no specific funding for this work.

Competing interests: The authors have declared that no competing interests exist.

Abstract

Research examining the reliability of stiffness measures during hopping has shown strong consistency in leg-spring stiffness (k_{leg}), but high variability in joint stiffness (k_{joint}) measures. Sled-based systems (SBS) reduce movement degrees-of-freedom and are used to examine stretch-shortening cycle (SSC) function under controlled conditions. The aim of this study was to examine the reliability of k_{leg} and k_{joint} during single-leg hopping within an SBS. Kinematic and kinetic data were collected on four occasions (Day_1, Day_2, Day_3 and Day_3_{Offset}). Participants completed two trials of single-leg hopping at different frequencies (1.5, 2.2 and 3.0 Hz) while attached to an inclined-SBS. Stiffness was determined using models of leg-spring (k_{leg}) and torsional (k_{joint}) stiffness. Statistical analysis identified absolute and relative measures of reliability. Results showed moderate reliability for k_{leg} at 1.5 Hz between inter-day testing bouts, and weak consistency at 2.2 and 3.0 Hz. Examination of intra-day comparisons showed weak agreement for repeated measures of k_{leg} at 1.5 and 2.2 Hz, but moderate agreement at 3.0 Hz. Limits in k_{leg} reliability were accompanied by weak-to-moderate agreement in k_{joint} measures across inter- and intra-day testing bouts. Results showed limits in the reliability of stiffness measures relative to previous reports on overground hopping. Lack of consistency in k_{leg} and k_{joint} may be due to the novelty of hopping within the current inclined-SBS. Constraints imposed on the hopping task resulting from SBS design (e.g. additional chair mass, restricting upper body movement) may have also influenced limits in k_{leg} and k_{joint} reliability. Researchers should consider these findings when employing inclined-SBS of a similar design to examine SSC function.

Introduction

Stretch-shortening cycle (SSC) tasks are characterised by eccentric lengthening immediately followed by concentric shortening [1]. These actions are typical of human movement (e.g. walking, running, jumping and throwing) and serve to enhance concentric force output and

movement efficiency [1,2]. Knowledge of mechanisms controlling SSC function are therefore relevant to general health and athletic performance. SSC function has subsequently been examined during walking [3], running [4,5], jumping [4], under fatigued conditions [6–10] and before and after training [11,12].

It is understood that the efficiency of SSC task performance is regulated by a combination of elastic and neural mechanisms [1]. Of importance to elastic mechanisms of SSC performance is the property stiffness (k). Stiffness represents the resistance of a body to deformation and reflects the elastic nature of the body [13,14]. In humans, greater stiffness is found to enhance the operational effectiveness of the SSC [15–17], yet higher levels of stiffness also contribute to musculoskeletal injury [18]. When examining stiffness characteristics during SSC tasks, researchers often model the elastic nature of the entire leg as a linear-spring (i.e. k_{leg}) [9,19–22]. Torsional stiffness models are also used to clarify the contribution of the stiffness of each joint (k_{joint}) to k_{leg} [7].

The reliability of k_{leg} and k_{joint} measures has been examined during SSC tasks [22,23]. During double- and single-leg hopping, Joseph *et al.* [23] and Diggin *et al.* [22] showed good agreement ($ICC > 0.80$) for repeated measures of k_{leg} when hopping frequency is controlled. In contrast, measurement consistency declined ($ICC = 0.20–0.86$) during running and hopping at self-selected frequencies [23]. Furthermore, k_{joint} measures have consistently demonstrated questionable reliability during natural hopping and running tasks. Joseph *et al.* [23] showed poor reliability for all k_{joint} variables measured ($ICC = 0.51–0.92$; $CV > 20\%$). Diggin and colleagues [22] showed strong consistency for repeated measures of k_{ankle} , but weak-to-moderate agreement for k_{knee} and k_{hip} ($ICC = 0.57–0.93$) during single-leg hopping. It must be noted that in efficient SSC tasks (e.g. hopping), participants must coordinate the body's musculoskeletal structures (e.g. segments, muscles, joints) in an integrated fashion to achieve the movement goal. These requirements produce inherent variability in movement patterns which can impact movement consistency and measurement reliability [24]. It is possible that limits in k_{leg} and k_{joint} reliability previously reported, were caused by extraneous movements (esp. upper body) that impacted landing consistency.

Researchers have used sled-based systems (SBS) to offset issues associated with extraneous upper body movements [8,19,25,26]. While construction varies between laboratories, these systems often include rails positioned at angles below vertical orientation (e.g. $0^{\circ}–30^{\circ}$ to horizontal) [12,27]. A force plate mounted at the end of the system provides kinetic information of participants' performance [6,25,28]. Participants perform SSC tasks (e.g. drop / rebound jumps, hopping) while attached to a chair / platform (sled) that slides up and down the inclined rails. SBS regulate levels of mechanical load experienced during SSC tasks by (i) altering rail angle [27]; (ii) employing a winch system to control initial eccentric loading [19,29]; and (iii) reducing movement degrees-of-freedom [10,30]. For lower body SBS, performance of SSC tasks while attached to the chair limits upper body movement and allows researchers to control impact velocities and eccentric loading [8,19]. Furthermore, the restricted path of the sledge chair might reduce task constraints and facilitate consistent landing patterns which can impact measurement reliability.

The reliability of kinematic and kinetic variables measured during SSC task performance within an SBS has been examined. Debenham and colleagues [10] reported strong agreement ($ICC = 0.87$) for ankle joint range-of-motion recorded during single-leg hopping tasks within an SBS. During drop (DJ) and rebound jump (RBJ) tasks, Flanagan and Harrison [19] reported strong consistency (Cronbach's $\alpha > 0.95$) for temporal and kinetic variables tested (e.g. flight time, $F_{z\ max}$, reactive strength index). Similarly, during a simulated hopping task within an SBS, Furlong and Harrison [29] reported strong consistency (Cronbach's $\alpha > 0.90$) for temporal (CT, flight time) and kinetic (e.g. $F_{z\ max}$, peak rate of force development) variables.

Flanagan and Harrison [19] reported strong reliability (Cronbach's $\alpha > 0.95$) for repeated measures of k_{leg} during jumping tasks (Cronbach's $\alpha > 0.95$) within an SBS.

Given the purpose of the SBS is to reduce extraneous upper body movement to encourage landing and measurement consistency, it is possible that stiffness measures (esp. k_{joint}) achieved during hopping within an SBS might demonstrate strong reliability. To date, the consistency of k_{leg} and k_{joint} measures recorded during hopping within an SBS has not been examined. The aim of this study was to examine the reliability of stiffness measures during hopping within an inclined-SBS. It was hypothesized that restricting extraneous upper body movement would facilitate strong consistency for k_{leg} and k_{joint} variables during hopping tasks.

Materials and methods

Participant characteristics

Thirty-two students (16 males and 16 females; mean \pm SD age 21.3 ± 2.9 years; stature 1.70 ± 0.08 m; mass 69.9 ± 10.0 kg) volunteered to participate. Volunteers were recruited via e-mail circulated to the campus community. All participants were physically active as members of a university sports club, attending bi-weekly practices and represented the university at varsity level. Participants were injury free for at least six months prior to testing as determined via health screening questionnaire. Procedures were approved by the research ethics committee of the Faculty of Education and Health Science at the University of Limerick, and all participants provided written informed consent. To limit the effects of fatigue on stiffness measurement, participants were instructed to abstain from vigorous physical activity 24 hours before testing bouts.

Experimental procedures

Participants completed a familiarization session one week prior to initial testing to acquaint them with procedures (Day_0) [31]. Following this, participants underwent testing on four occasions. The first (Day_1), second (Day_2) and third (Day_3) testing bouts took place at the same time of day, spaced three to seven days apart [23,32]. On the final test day, participants completed procedures twice; at their typically scheduled time (Day_3) and six hours prior to or following their scheduled time (Day_3_{Offset}) [22].

For each test, participants wore dark, tight-fitting clothing. Retro-reflective markers (14 mm) were placed at six anatomical locations on participants' right side (acromion process, greater trochanter, lateral epicondyle of the femur, lateral malleolus, calcaneus and fifth metatarsal). Marker placement was conducted by the same tester throughout for consistency. During all test bouts (Day_0 –Day_3), participants completed two 10 second trials of single-leg hopping at each of 1.5, 2.2 and 3.0 Hz in time with a digital metronome (TempoPerfect Metronome, NCH Software, Greenwood Village, CO, USA) while secured within an SBS (Fig 1). The design of the SBS was described previously [12,19]. As previous research suggests little effect of leg dominance on k_{leg} or k_{joint} [23], participants were instructed to land as close as possible to the force plate center (AMTI OR6-5; AMTI, Watertown, MA, USA) in time with the metronome, using their right leg and natural hopping technique. Trials were accepted for analysis if participants hopped within $\pm 5\%$ of the target frequency [33,34]. The order of hopping trials was randomly assigned, and participants received 60 seconds recovery between each trial to limit fatigue effects. An analogue triggering device was used to initiate 3D kinematic and kinetic data acquisition simultaneously. Kinematic data were recorded using three MAC Eagle cameras (MotionAnalysis Corporation, Santa Rosa CA, USA) operating at 200 Hz. Kinetic data were recorded at 1 kHz over the 10 second duration.



Fig 1. Image showing the experimental set-up for hopping trials within the sled-based system.

<https://doi.org/10.1371/journal.pone.0225664.g001>

Data processing

Analysis of data revealed an inconsistent delay in the initiation of kinematic and kinetic recordings. Whittlesey and Robertson [35] noted that kinematic and kinetic input signals must be temporally aligned to facilitate valid inverse dynamics analysis. Work in our laboratory has shown that maximum rate of change in fifth metatarsal marker acceleration profiles (i.e. peak jerk), coincide with ground contact to within $3.83 (\pm 1.06)$ ms of force plate criterion measures during overground hopping. Fifth metatarsal vertical coordinate data were subsequently differentiated to jerk [36]. The time of peak jerk was determined, and kinematic data were subsequently aligned with the time at which the vertical force increased above 5 N.

Marker trajectories were digitized using Cortex motion analysis software (version 2.1; MotionAnalysis Corporation, Santa Rosa, CA, USA). 3D kinematic and kinetic data were subsequently exported and analyzed using customized MS Excel macros. Recorded kinematic and kinetic data were concurrently filtered using a fourth-order Butterworth Low-Pass digital filter with an optimal cut-off of 11 Hz determined via residual analysis [36]. 3D coordinate data were subsequently interpolated to 1 kHz using a cubic spline. Filtered kinematic and kinetic data were used to calculate resultant joint moments occurring about ankle (M_{ankle}), knee (M_{knee}) and hip (M_{hip}) joints throughout the ground contact phase of each trial using inverse dynamics [7,23,37]. Segment inertia and mass characteristics were determined using the standards of Dempster [38]. Having calculated resultant moments, average torsional stiffness of the ankle (k_{ankle}), knee (k_{knee}) and hip (k_{hip}) were calculated as a ratio of changes in joint moment (ΔM_{joint}) and angle ($\Delta \theta_{\text{joint}}$) for respective lower limb joints [7,23].

In addition, k_{leg} was recorded throughout the ground contact phase of all hopping trials using the spring-mass model [21]. Thus k_{leg} was calculated as the ratio of $F_{z \text{ max}}$ and maximum leg compression (ΔLeg_L) occurring during ground contact as measured from video records [39], where leg length represented the distance between the greater trochanter and the center-of-force [40]. In all analyzed trials, the temporal occurrence of discrete events of $F_{z \text{ max}}$ and ΔLeg_L coincided to within 10% of the hop period.

Statistical analysis

The mean of 5 consecutive hops occurring within the middle of each 10-second trial was calculated for dependent variables and compared within and between experimental bouts. Differences between the means recorded for each experimental bout were established for each hopping frequency (HopFreq) using repeated measures ANOVA (parametric variables) or Friedman's test (non-parametric variables). Reliability analysis employed both 'absolute' and 'relative' measures. A two-way random effects intra-class correlation coefficient (ICC) model examined the relative agreement between parametric and log-transformed dependent variables, while measurement consistency was established from effect size (ES) and ICC 95% confidence interval results [31,41,42]. Variables were considered strong when $ICC \geq 0.90$ [43], the lower-bound level of the ICC 95% confidence interval ($ICC_{Lower} > 0.80$ [31] and Cohen's $ES < 0.50$ (moderate effect) [44]. Moderate reliability occurred if ICC ranged from 0.80 to 0.89, $ICC_{Lower} \geq 0.70$ and $ES < 0.50$, while 'weak' reliability occurred if the previous criteria were not met. Within-participant variation in dependent variables were also examined using typical error (TE) [41], however these statistics were not included in reliability indexing. All statistical analysis was conducted using SPSS (version 23.0, IBM Corporation, Armonk, NY, USA).

Results

Results for control variables of contact time (CT; $F_{(1.3, 3.9)} = 2.14$; $p > 0.05$), flight time (FT; $F_{(1.7, 5.0)} = 2.84$; $p > 0.05$) and HopFreq ($F_{(1.8, 5.3)} = 0.56$; $p > 0.05$; Table 1) revealed no significant difference between means for these variables across inter-day testing bouts. Reliability analysis showed strong agreement for variables of CT and FT ($ICC = 0.95\text{--}0.97$; $ICC_{Lower} = 0.81\text{--}0.92$; $ES = 0.01\text{--}0.09$) at 1.5 and 3.0 Hz with moderate reliability evident at 2.2 Hz ($ICC = 0.94$; $ICC_{Lower} = 0.79\text{--}0.92$; $ES = 0.10$). Across intra-day testing bouts, analysis revealed no significant difference in CT ($F_{(2, 18)} = 0.10$; $p > 0.05$), FT ($F_{(2, 18)} = 0.29$; $p > 0.05$) and HopFreq ($F_{(1.6, 43.6)} = 1.24$; $p > 0.05$) when measured six-hours apart. The high ICC's (0.92–0.98), moderate-to-high ICC_{Lower} coefficients (0.82–0.96) and low ES (0.04–0.13), suggest 'strong' intra-day agreement for variables of FT and CT. Despite moderate-to-strong reliability indices for FT and CT variables, HopFreq exhibited weak consistency throughout.

Friedman's test revealed no significant change in k_{leg} ($\chi^2(5) = 0.68\text{--}10.23$; $p > 0.05$) across Day_1, Day_2 and Day_3 testing bouts at any HopFreq. Despite exhibiting little change between days, reliability indices showed k_{leg} exhibited weak-to-moderate consistency ($ICC = 0.71\text{--}0.89$; $ICC_{Lower} = 0.29\text{--}0.73$; $ES = 0.08\text{--}0.26$) between inter-day bouts across all HopFreq (Table 2). Examination of kinematic and kinetic variables (Table 3) shows limits in k_{leg} measurement agreement originated from limits in $F_{z\ max}$ and ΔLeg_L measurement consistency. Limits in k_{leg} measurement agreement, were accompanied by weak consistency in k_{knee} and k_{hip} between Day_1, Day_2 and Day_3 across all HopFreq ($ICC = 0.62\text{--}0.84$; $ICC_{Lower} = 0.02\text{--}0.61$; $ES = 0.10\text{--}0.70$). Only k_{ankle} exhibited moderate-to-strong consistency at slower HopFreq ($ICC = 0.91\text{--}0.94$; $ICC_{Lower} = 0.76\text{--}0.83$; $ES = 0.17\text{--}0.18$). Examination of kinematic and kinetic variables (Table 3) shows that despite limits to k_{knee} and k_{hip} at 1.5 Hz, participants exhibited weak-to-strong consistency for joint kinetic and kinematic variables. At 2.2 and 3.0 Hz, limits in k_{knee} and k_{hip} resulted from weak-to-moderate agreement for joint kinetics and kinematics. Data show that strong consistency in k_{ankle} measures at 1.5 Hz was accompanied by strong consistency in $M_{ankle\ max}$ and $\Delta\theta_{ankle}$. Limits in k_{ankle} measurement agreement at 2.2 and 3.0 Hz were accompanied by varied consistency in ankle kinematic and kinetic measures. Joint kinetics and kinematics showed no change ($p > 0.05$) between inter-day test bouts at any hopping frequency.

Table 1. Descriptive and reliability statistics (mean ± s) for control variables of CT, FT, and HopFreq recorded between inter- and intra-day testing bouts during hopping within a SBS.

Variables	Inter-day Comparisons								
	Mean (± s)			ICC	ICC 95% CI		TE	ES	Index
	Day_1	Day_2	Day_3		Lower	Upper			
1.5 Hz									
CT (s)	0.529 (0.063)	0.539 (0.052)	0.524 (0.060)	0.966	0.918	0.990	0.016	0.058	Strong
FT (s)	0.136 (0.064)	0.127 (0.054)	0.142 (0.060)	0.965	0.918	0.990	0.015	0.070	Strong
HopFreq (Hz)	1.50 (0.02)	1.50 (0.01)	1.50 (0.02)	0.681	0.468	0.830	0.02	0.12	Weak
2.2 Hz									
CT (s)	0.370 (0.029)	0.377 (0.025)	0.367 (0.020)	0.942	0.808	0.993	0.011	0.090	Strong
FT (s)	0.083 (0.027)	0.077 (0.026)	0.087 (0.021)	0.938	0.794	0.993	0.011	0.096	Moderate
HopFreq (Hz)	2.21 (0.03)	2.21 (0.02)	2.21 (0.02)	0.732	0.543	0.860	0.02	0.03	Weak
3.0 Hz									
CT (s)	0.274 (0.013)	0.274 (0.013)	0.274 (0.013)	0.949	0.872	0.987	0.006	0.006	Strong
FT (s)	0.059 (0.014)	0.059 (0.013)	0.060 (0.012)	0.948	0.868	0.987	0.006	0.032	Strong
HopFreq (Hz)	3.01 (0.04)	3.00 (0.03)	3.00 (0.03)	0.721	0.541	0.847	0.03	0.07	Weak
Variables	Intra-day Comparisons								
	Mean (± s)			ICC	ICC 95% CI		TE	ES	Index
	Day_3	Day_3 _{Offset}	Lower		Upper				
1.5 Hz									
CT (s)		0.524 (0.060)	0.516 (0.073)	0.970	0.935	0.988	0.014	0.080	Strong
FT (s)		0.142 (0.060)	0.151 (0.074)	0.969	0.935	0.988	0.011	0.086	Strong
HopFreq (Hz)		1.50 (0.02)	1.51 (0.02)	0.773	0.606	0.880	0.02	0.28	Weak
2.2 Hz									
CT (s)		0.367 (0.021)	0.360 (0.029)	0.981	0.959	0.993	0.007	0.083	Strong
FT (s)		0.087 (0.021)	0.093 (0.029)	0.982	0.961	0.994	0.007	0.077	Strong
HopFreq (Hz)		2.21 (0.02)	2.20 (0.03)	0.740	0.544	0.865	0.02	0.08	Weak
3.0 Hz									
CT (s)		0.274 (0.013)	0.275 (0.013)	0.921	0.827	0.970	0.006	0.131	Strong
FT (s)		0.06 (0.012)	0.059 (0.011)	0.916	0.818	0.969	0.007	0.044	Strong
HopFreq (Hz)		3.00 (0.03)	2.99 (0.04)	0.736	0.542	0.861	0.03	0.31	Weak

Abbreviations: CT = contact time; FT = flight time; HopFreq = hopping frequency; ICC = intraclass correlation coefficient; 95% CI = 95% confidence interval; TE = typical error; ES = effect size; Index = reliability index resulting from absolute and relative reliability statistics. Note. Central statistics represent the mean (± s) of five consecutive hops recorded for two trials across Day_1, Day_2, Day_3, and Day_3_{Offset} experimental bouts.

<https://doi.org/10.1371/journal.pone.0225664.t001>

Examination of intra-day testing comparisons (Table 2) showed no change in k_{leg} between Day_3 and Day_3_{Offset} ($\chi^2(3) = 0.60-3.19; p > 0.05$). Despite this, k_{leg} measures exhibited weak consistency at 1.5 and 2.2 Hz HopFreq (ICC = 0.59–0.86; ICC_{Lower} = 0.09–0.70; ES = 0.11–0.35) and moderate consistency at 3.0 Hz (ICC = 0.87; ICC_{Lower} = 0.71; ES = 0.17). Limits in k_{leg} reliability originated from strong and weak consistency in $F_{z\ max}$ and ΔLeg_L respectively at all HopFreq. Repeated measure ANOVA results showed no change in k_{joint} variables between Day_3 and Day_3_{Offset} ($F(1.1, 22.6) = 0.17-1.07; p > 0.05$). Despite this, k_{knee} and k_{hip} exhibited weak-to-moderate consistency between Day_3 and Day_3_{Offset} (ICC = 0.29–0.89; ICC_{Lower} = 0.00–0.72; ES = 0.02–0.50). In contrast, k_{ankle} exhibited strong consistency at 3.0 Hz but weak consistency at other hopping frequencies. Examination of kinematic and kinetic data (Table 4) showed limits in k_{joint} reliability originated from largely weak-to-moderate consistency in joint kinematic and kinetic inputs between Day_3 and Day_3_{Offset}. Limits in

Table 2. Descriptive and reliability statistics (mean ± s) for variables of k_{leg} and k_{joint} recorded between inter- and intra-day testing bouts during hopping within a SBS.

Variables	Inter-day Comparisons							TE	ES	Index
	Mean (± s)			ICC	ICC 95% CI					
	Day_1	Day_2	Day_3		Lower	Upper				
1.5 Hz										
k_{leg} (kN.m ⁻¹)	4.85 (0.82)	4.70 (1.04)	5.00 (1.80)	0.887	0.731	0.968	0.57	0.08	Moderate	
k_{ankle} (Nm.deg ⁻¹)	2.27 (0.69)	2.22 (0.69)	2.45 (0.64)	0.936	0.833	0.985	0.32	0.18	Strong	
k_{knee} (Nm.deg ⁻¹)	1.14 (1.28)	1.65 (2.29)	0.94 (0.72)	0.789	0.485	0.940	7.44	0.10	Weak	
k_{hip} (Nm.deg ⁻¹)	14.70 (14.18)	11.61 (5.16)	10.49 (5.17)	0.838	0.611	0.954	18.48	0.34	Weak	
2.2 Hz										
k_{leg} (kN.m ⁻¹)	7.69 (1.25)	8.19 (1.63)	8.22 (1.29)	0.770	0.430	0.940	0.9	0.26	Weak	
k_{ankle} (Nm.deg ⁻¹)	4.06 (1.57)	3.41 (0.93)	3.75 (1.08)	0.909	0.762	0.979	2.20	0.17	Moderate	
k_{knee} (Nm.deg ⁻¹)	24.9 (31.56)	15.21 (12.57)	7.17 (6.22)	0.742	0.015	0.982	8.21	0.70	Weak	
k_{hip} (Nm.deg ⁻¹)	109.77 (154.56)	100.68 (133.98)	121.02 (171.15)	0.623	0.376	0.794	17.37	0.05	Weak	
3.0 Hz										
k_{leg} (kN.m ⁻¹)	11.5 (3.30)	11.01 (2.30)	12.79 (7.20)	0.713	0.288	0.925	5.11	0.20	Weak	
k_{ankle} (Nm.deg ⁻¹)	5.12 (1.81)	4.53 (1.64)	5.04 (1.96)	0.884	0.695	0.973	1.69	0.03	Weak	
k_{knee} (Nm.deg ⁻¹)	28.47 (27.18)	41.72 (74.19)	61.05 (82.54)	0.747	0.584	0.861	13.93	0.35	Weak	
k_{hip} (Nm.deg ⁻¹)	47.84 (73.03)	20.57 (23.85)	34.73 (59.67)	0.625	0.382	0.795	17.30	0.17	Weak	
Variables	Intra-day Comparisons							TE	ES	Index
	Mean (± s)		ICC	ICC 95% CI						
	Day_3	Day_3 _{Offset}		Lower	Upper					
1.5 Hz										
k_{leg} (kN.m ⁻¹)		5.00 (1.80)	5.00 (1.33)	0.858	0.699	0.944	1.05	0.11	Weak	
k_{ankle} (Nm.deg ⁻¹)		2.45 (0.64)	2.61 (0.86)	0.833	0.635	0.937	0.76	0.13	Weak	
k_{knee} (Nm.deg ⁻¹)		0.94 (0.72)	8.99 (8.98)	0.386	0.000	0.859	8.98	0.02	Weak	
k_{hip} (Nm.deg ⁻¹)		10.49 (5.17)	17.93 (11.75)	0.864	0.694	0.951	10.85	0.18	Weak	
2.2 Hz										
k_{leg} (kN.m ⁻¹)		8.22 (1.29)	11.76 (14.30)	0.594	0.091	0.854	1.68	0.35	Weak	
k_{ankle} (Nm.deg ⁻¹)		3.75 (1.08)	5.39 (6.29)	0.847	0.642	0.948	1.13	0.25	Weak	
k_{knee} (Nm.deg ⁻¹)		7.17 (6.22)	12.70 (12.48)	0.683	0.000	0.941	11.60	0.15	Weak	
k_{hip} (Nm.deg ⁻¹)		121.02 (171.15)	24.71 (19.08)	0.888	0.417	0.997	16.18	0.10	Weak	
3.0 Hz										
k_{leg} (kN.m ⁻¹)		12.79 (7.20)	12.56 (5.25)	0.865	0.705	0.949	3.90	0.17	Moderate	
k_{ankle} (Nm.deg ⁻¹)		5.04 (1.96)	6.70 (4.14)	0.915	0.814	0.968	2.26	0.14	Strong	
k_{knee} (Nm.deg ⁻¹)		61.05 (82.54)	6.76 (5.11)	0.287	0.000	0.747	7.95	0.50	Weak	
k_{hip} (Nm.deg ⁻¹)		34.73 (59.67)	12.43 (11.61)	0.885	0.720	0.963	10.10	0.06	Moderate	

Abbreviations: k_{leg} = leg-spring stiffness; k_{ankle} , k_{knee} , k_{hip} = stiffness variables for ankle, knee, and hip joints, respectively; ICC = intraclass correlation coefficient; 95% CI = 95% confidence interval; TE = typical error; ES = effect size; Index = reliability index resulting from absolute and relative reliability statistics. Note. Central statistics represent the mean (± s) of 5 consecutive hops recorded for 2 trials across Day_1, Day_2, Day_3, and Day_3_{Offset} experimental bouts.

<https://doi.org/10.1371/journal.pone.0225664.t002>

consistency in kinematic and kinetic inputs occurred despite no change in these data between Day_3 and Day_3_{Offset} testing bouts ($\chi^2(3) = 0.98-3.86; p > 0.05$).

Discussion

SBS are used to examine SSC function under controlled conditions. Sled design regulates eccentric loading and limits extraneous movement to improve landing and measurement

Table 3. Descriptive (mean ± s) and reliability statistics for kinetic and kinematic variables recorded between inter-day testing bouts during hopping within a SBS.

Variables	Inter-day Comparisons						TE	ES	Index
	Mean (± s)			ICC	ICC 95 % CI				
	Day 1	Day 2	Day_3		Lower	Upper			
1.5 Hz									
F _{z max} (N)	435 (109)	423 (93)	446 (107)	0.976	0.943	0.993	23	0	Strong
M _{ankle max} (Nm)	53.27 (14.83)	52.83 (14.38)	54.72 (14.13)	0.936	0.846	0.981	6.11	0.07	Strong
M _{knee max} (Nm)	5.05 (10.89)	8.50 (7.76)	6.04 (9.57)	0.665	0.167	0.905	8.59	0.07	Weak
M _{hip max} (Nm)	75.35 (39.48)	84.88 (37.26)	94.19 (51.26)	0.953	0.888	0.986	14.69	0.29	Strong
Δθ _{ankle} (°)	24.62 (5.67)	25.06 (5.66)	24.78 (5.47)	0.964	0.908	0.991	2.20	0.02	Strong
Δθ _{knee} (°)	15.09 (6.92)	13.95 (6.46)	15.05 (5.93)	0.937	0.850	0.982	1.90	0.00	Strong
Δθ _{hip} (°)	5.95 (3.20)	5.47 (2.95)	6.42 (3.05)	0.960	0.900	0.989	1.00	0.10	Strong
ΔLeg _L (m)	0.05 (0.05)	0.07 (0.04)	0.08 (0.05)	0.889	0.731	0.968	0.01	0.41	Moderate
2.2 Hz									
F _{z max} (N)	439 (82)	425 (77)	456 (66)	0.913	0.783	0.977	21	0	Moderate
M _{ankle max} (Nm)	53.77 (10.33)	50.53 (10.94)	54.11 (10.58)	0.847	0.589	0.965	6.01	0.02	Weak
M _{knee max} (Nm)	7.77 (5.04)	8.68 (7.37)	9.70 (8.68)	0.725	0.299	0.929	4.05	0.18	Weak
M _{hip max} (Nm)	75.40 (29.76)	67.25 (30.11)	89.95 (90.95)	0.842	0.585	0.963	46.03	0.19	Weak
Δθ _{ankle} (°)	14.63 (3.03)	15.01 (3.03)	15.08 (2.41)	0.943	0.857	0.985	1.40	0.11	Strong
Δθ _{knee} (°)	2.66 (1.25)	3.02 (1.32)	3.18 (1.30)	0.682	0.194	0.917	1.00	0.27	Weak
Δθ _{hip} (°)	1.93 (1.11)	1.51 (0.88)	1.84 (0.99)	0.827	0.567	0.955	0.70	0.06	Weak
ΔLeg _L (m)	0.04 (0.03)	0.04 (0.03)	0.04 (0.03)	0.721	0.331	0.926	0.01	0.15	Weak
3.0 Hz									
F _{z max} (N)	446 (54)	445 (63)	458 (47)	0.967	0.917	0.991	16	0	Strong
M _{ankle max} (Nm)	53.39 (9.61)	54.40 (11.15)	60.71 (19.71)	0.922	0.807	0.980	4.70	0.36	Strong
M _{knee max} (Nm)	12.88 (11.33)	8.91 (11.13)	8.47 (12.46)	0.873	0.688	0.966	7.57	0.25	Weak
M _{hip max} (Nm)	49.92 (34.54)	51.79 (26.49)	71.57 (47.38)	0.883	0.712	0.969	15.89	0.40	Moderate
Δθ _{ankle} (°)	10.93 (2.61)	12.49 (3.38)	11.87 (3.80)	0.901	0.755	0.974	2.00	0.19	Moderate
Δθ _{knee} (°)	2.85 (1.04)	3.04 (1.82)	2.42 (2.11)	0.519	0.000	0.876	1.30	0.18	Weak
Δθ _{hip} (°)	4.27 (1.81)	4.27 (2.03)	4.03 (1.58)	0.728	0.308	0.930	1.70	0.09	Weak
ΔLeg _L (m)	0.02 (0.02)	0.03 (0.02)	0.03 (0.02)	0.785	0.451	0.945	0.01	0.34	Weak

Abbreviations: ICC = intraclass correlation coefficient; 95% CI = 95% confidence interval; TE = typical error; ES = effect size; Index = reliability index resulting from absolute and relative reliability statistics; F_{z max} = maximum ground reaction force; M_{ankle max}, M_{knee max}, and M_{hip max} = maximum resultant moments recorded for ankle, knee, and hip joints, respectively; Δθ_{ankle}, Δθ_{knee}, and Δθ_{hip} = relative angular displacement at ankle, knee, and hip joints, respectively; ΔLeg_L = change in leg length. Note. Central statistics represent the mean (± s) of 5 consecutive hops recorded for 2 trials across Day_1, Day_2, and Day_3 experimental bouts.

<https://doi.org/10.1371/journal.pone.0225664.t003>

consistency. During overground hopping and running, k_{leg} has demonstrated good reliability [22,23]. In contrast, k_{joint} measures have demonstrated limited consistency. The reliability of k_{leg} and k_{joint} while hopping within an SBS had not been examined. This study provides a comprehensive analysis of the inter- and intra-day reliability of k_{leg} and k_{joint} achieved during hopping within an SBS.

Similar to the findings of Hobara and colleagues [21] k_{leg} and k_{joint} increased with increases in HopFreq in the present study. Values of 13.9–28.9 kN.m⁻¹ have been reported for k_{leg} during natural hopping conditions on different surfaces and at different frequencies [22,23,30,45,46]. In addition, values of 6.9–12.0 Nm.deg⁻¹ have been reported for k_{ankle} during natural hopping [45]. In the present study, stiffness values are lower than reported during natural hopping [22,23]. This can be explained by the orientation of the frame of the SBS (i.e. 30° to the horizontal) which reduces (half) the effect of gravitational acceleration [12,19]. Therefore, lower

Table 4. Descriptive (mean ± s) and reliability statistics for kinetic and kinematic variables recorded between intra-day testing bouts during hopping within a SBS.

Variables	Intra-day Comparisons						TE	ES	Index
	Mean (± s)		ICC	ICC 95% CI					
	Day_3	Day_3 _{Offset}		Lower	Upper				
1.5 Hz									
F _{z max} (N)	446 (107)	467 (125)	0.970	0.936	0.988	19	0	Strong	
M _{ankle max} (Nm)	54.72 (14.13)	57.56 (18.48)	0.902	0.790	0.962	9.19	0.09	Moderate	
M _{knee max} (Nm)	6.04 (9.57)	8.37 (14.59)	0.651	0.110	0.896	10.83	0.20	Weak	
M _{hip max} (Nm)	94.19 (51.26)	84.64 (53.07)	0.852	0.679	0.944	19.04	0.05	Weak	
Δθ _{ankle} (°)	24.78 (5.47)	23.1 (5.0)	0.907	0.803	0.964	2.80	0.11	Strong	
Δθ _{knee} (°)	15.05 (5.93)	3.1 (6.8)	0.895	0.774	0.959	4.20	0.02	Moderate	
Δθ _{hip} (°)	6.42 (3.05)	5.2 (3.5)	0.953	0.899	0.981	0.90	0.06	Strong	
ΔLeg _L (m)	0.08 (0.05)	0.09 (0.02)	0.857	0.696	0.944	0.01	0.13	Weak	
2.2 Hz									
F _{z max} (N)	456 (66)	483 (83)	0.991	0.980	0.997	14	0	Strong	
M _{ankle max} (Nm)	54.11 (10.58)	71.58 (51.69)	0.716	0.372	0.897	35.27	0.28	Weak	
M _{knee max} (Nm)	9.70 (8.68)	10.93 (9.36)	0.752	0.414	0.919	7.36	0.10	Weak	
M _{hip max} (Nm)	89.95 (90.95)	86.71 (71.85)	0.894	0.757	0.964	38.91	0.04	Moderate	
Δθ _{ankle} (°)	15.08 (2.41)	14.7 (5.0)	0.848	0.653	0.946	1.90	0.15	Weak	
Δθ _{knee} (°)	3.18 (1.30)	-2.3 (2.5)	0.885	0.744	0.959	1.40	0.23	Moderate	
Δθ _{hip} (°)	1.84 (0.99)	-1.9 (1.9)	0.833	0.621	0.940	0.80	0.23	Weak	
ΔLeg _L (m)	0.04 (0.03)	0.06 (0.01)	0.643	0.171	0.878	0.01	0.23	Weak	
3.0 Hz									
F _{z max} (N)	458 (47)	461 (54)	0.982	0.959	0.993	13	0	Strong	
M _{ankle max} (Nm)	60.71 (19.71)	62.57 (16.98)	0.613	0.187	0.851	6.29	0.37	Weak	
M _{knee max} (Nm)	8.47 (12.46)	10.59 (15.30)	0.804	0.519	0.937	9.30	0.22	Weak	
M _{hip max} (Nm)	71.57 (47.38)	65.13 (49.60)	0.844	0.651	0.944	20.20	0.02	Weak	
Δθ _{ankle} (°)	11.87 (3.80)	9.50 (8.10)	0.975	0.946	0.991	5.30	0.19	Strong	
Δθ _{knee} (°)	2.42 (2.11)	-3.60 (2.50)	0.534	0.000	0.828	1.10	0.25	Weak	
Δθ _{hip} (°)	4.03 (1.58)	-5.10 (2.50)	0.799	0.563	0.924	2.00	0.34	Weak	
ΔLeg _L (m)	0.03 (0.02)	0.04 (0.01)	0.749	0.435	0.913	0.01	0.01	Weak	

Abbreviations: ICC = intraclass correlation coefficient; 95% CI = 95% confidence interval; TE = typical error; ES = effect size; Index = reliability index resulting from absolute and relative reliability statistics; F_{z max} = maximum ground reaction force; M_{ankle max}, M_{knee max}, and M_{hip max} = maximum resultant moments recorded for ankle, knee, and hip joints, respectively; Δθ_{ankle}, Δθ_{knee}, and Δθ_{hip} = relative angular displacement at ankle, knee, and hip joints, respectively; ΔLeg_L = change in leg length. Note. Central statistics represent the mean (± s) of 5 consecutive hops recorded for 2 trials across Day_3 and Day_3_{Offset} experimental bouts.

<https://doi.org/10.1371/journal.pone.0225664.t004>

ground reaction forces (i.e. F_{z max}) are expected in the present study relative to published reports on overground hopping. This is supported by the work of Harrison *et al.* [25] and Flanagan and Harrison [19] who reported lower values for k_{leg} during DJ and RBJ tasks (3.4–8.5 kN.m⁻¹) performed within the current SBS, relative to natural conditions. The stiffness data recorded during the present study are in line with previous data reported for SSC tasks performed within an inclined-SBS.

Hobara *et al.* [21] showed increases in k_{leg} with increases in HopFreq. Thus, regulation of HopFreq between testing bouts was important to reliability analysis. In the current study HopFreq displayed weak-to-moderate consistency across all trials despite CT and FT variables showing strong agreement at all HopFreq between Day_3 and Day_3_{Offset}. Inter-day comparisons showed strong consistency for CT throughout. In contrast, FT exhibited strong agreement at 1.5 and 3.0 Hz but moderate consistency at 2.2 Hz. Hopkins and colleagues [41]

suggested low ICC's occur in situations where between participant variation is low for a given variable. In the present study, HopFreq was regulated to within 5% of the target frequency. Thus, low levels of variability may have produced spurious reliability indices in HopFreq and in FT measures during hopping at 2.2 Hz. The low TE (0.01–0.02) and ES (≤ 0.31) for CT, FT and HopFreq support this. Thus, HopFreq was controlled adequately throughout.

Despite regulation of HopFreq in the present study, k_{leg} exhibited moderate reliability at 1.5 Hz and weak reliability at 2.2 and 3.0 Hz (ICC = 0.59–0.89) between inter-day test bouts. These reliability indexes were accompanied by strong consistency for k_{ankle} at 1.5 Hz only. Remaining k_{joint} variables demonstrated weak-to-moderate agreement between Day_1, Day_2 and Day_3 testing bouts. Between Day_3 and Day_3_{Offset}, k_{leg} exhibited moderate reliability at 3.0 Hz but weak consistency at the other frequencies. The moderate index for k_{leg} at 3.0 Hz was accompanied by strong consistency for k_{ankle} . Variables of k_{knee} and k_{hip} however, exhibited weak-to-moderate consistency throughout. Examination of kinematic and kinetic inputs (Tables 3 and 4) showed that limits in k_{leg} reliability occurred due to limits in ΔLeg_L reliability, while $F_{z\ max}$ exhibited moderate-to-strong consistency. The present data also showed that limits in k_{joint} measurement agreement resulted from inconsistencies in $M_{joint\ max}$ and $\Delta\theta_{joint}$ at the highest hopping frequencies (2.2 and 3.0 Hz).

While the present study did not include a control condition, previous work in our lab examined the reliability of k_{leg} and k_{joint} measures in overground hopping [22]. This study employed the same methodology and used physically active participants consistent with the current work. Our work showed strong agreement (ICC = 0.95–0.98) for k_{leg} and k_{ankle} measures at different hopping frequencies [22]. In contrast, k_{knee} and k_{hip} exhibited weak-to-moderate agreement. Given our previous findings, that many stiffness variables showed weak-to-moderate agreement in the present study is surprising. Particularly since the primary function of the current inclined-SBS is to limit extraneous upper body movement to enhance landing and measurement consistency during SSC tasks [8,19].

It is understood that in biological measurement, observed scores are composed of a true score and additional error [42]. Weir [42] suggests that errors in measurement data result from instrumentation errors associated with equipment used, in addition to participant / tester error, modelling error and biological variability. In the context of the current work, instrumentation errors arise from equipment (force plate and video) use and protocol adherence between testing bouts. It is important to note that participants were instructed refrain from physical activity prior to data collection. Furthermore, force plate and video equipment were calibrated as per the manufacturer's guidelines prior to each data collection period. In addition, equipment set-up and protocols used were consistent with published works showing strong reliability for k_{leg} and k_{ankle} during overground hopping [22]. Consequently, deteriorations in measurement consistency relative to published reports are not due to differences in testing protocols between studies.

Of additional concern to the reliability of k_{leg} and k_{joint} scores is error derived from the use of mathematical models. Whittlesey and Hammill [47] noted that models are sensitive to the number of model components. Thus, small errors present early in the modelling process will propagate by the end of the simulation. This suggests that modelling errors will increase as calculations progress from distal to proximal joints. As k_{joint} is the ratio of changes in M_{joint} and θ_{joint} , it is clear that errors in M_{joint} (and k_{knee} and k_{hip}) measures will increase as inverse dynamics analysis progresses from distal to proximal joints. It is important to consider that while these errors will influence k_{joint} consistency, they are unlikely to influence k_{leg} in the present work. In addition, as modelling procedures employed in the current work are consistent with previous works [22], it is unlikely that differences in measurement consistency between the current and previous works result from modelling errors.

Considering biological variability, our previous work on overground hopping noted the existence of k_{leg} as an attractor state [22] where participants are drawn to the manipulation of leg compression (seen by variable k_{joint} or ΔLeg_L) to maintain a consistent k_{leg} under variable landing conditions [9,45]. These works also showed that stiffness variables were accompanied by mostly strong consistency for $F_{z\ max}$ and ΔLeg_L variables. In the present study, it must be considered that while the reliability of $F_{z\ max}$ was strong, the reliability of ΔLeg_L was weak-to-moderate across all testing bouts. As a result, the k_{leg} exhibited weak-to-moderate consistency throughout. It is therefore difficult to establish a consistent outcome variable (i.e. attractor state) that participants are trying to regulate. Furthermore, considering limits in the reliability of kinematic and kinetic variables measured, and the limited pattern to the nature of this reliability, we feel that our data does not support the notion that limits in reliability are due to kinematic variability to achieve consistent outcomes.

It is important to consider the novelty of the hopping task and its impact on the present data. Our previous work showed that familiarizing participants with testing procedures prior to data collection enhanced measurement reliability relative to earlier reports [23]. In the present study however, participants completed a familiarization session prior to data collection. Thus, acquainting participants with procedures did not benefit the consistency of hopping dynamics within the current SBS. One reason for this is likely the design of the current SBS. The design of the current SBS requires participants to adopt a flexed torso position while secured to it. This novel hopping position would alter the functional length of lower limb musculature compared to upright hopping, which will have impacted participants' ability to achieve consistent hopping dynamics. In addition, the mass of the chair of the current SBS was 19.6 kg. The added mass may have made it difficult for participants to achieve consistency in their hopping patterns. This is partly supported by the work of Kramer et al. [48]. The authors demonstrated that participants achieved almost natural reactive jumps following this four-week period (i.e. 12 sessions). While the instructions regarding landing dynamics differed from the current study, the design of the lightweight sled (5 kg) allowed participants to adopt natural movement patterns [27,48]. Although the authors did not report reliability indices for variables measured, the changes seen (38% increase) following four weeks of training (i.e. familiarizing) suggest the performance of SSC tasks within an SBS to be a novel task requiring extensive participant familiarization. The practicalities of including multiple familiarization sessions prior to data collection must also be considered.

Considering the lack of reliability evident in kinematics, kinetics and stiffness variables measured, it is appropriate to accept the null hypothesis. Limiting extraneous upper body movement by using an SBS during single-leg hopping did not improve the reliability of k_{leg} and k_{joint} measures relative to previous reports on overground hopping. In fact, use of the current SBS reduced consistency of stiffness measures relative to previous reports on natural conditions.

Conclusions

To summarize, the present data showed weak-to-moderate reliability for all stiffness measures during hopping using the current SBS. This was due largely to inconsistencies in leg compression (ΔLeg_L). Thus, reducing extraneous upper body movement using the current SBS did not encourage consistent stiffness regulation. In fact, hopping within the current inclined-SBS had adverse effects on measurement consistency relative to published reports on overground hopping. Limits in the consistency of stiffness measures may be due to constraints imposed by restricting upper body movement, which introduced novelty to the single-leg hopping task. A single familiarization session was therefore inadequate to achieve consistent landing

mechanics. The influence of sled design on the current findings must also be considered, with chair mass and orientation likely making it difficult for participants to achieve consistent hopping patterns. Researchers should consider these findings and endeavour to construct a light-weight system with a chair orientation that replicates the functional length of the muscle-tendon units driving the SSC task being analysed.

Author Contributions

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Methodology: David Diggin, Ross Anderson, Andrew J. Harrison.

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