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BIOMECHANICAL DEMANDS OF THE 2-STEP TRANSITIONAL GAIT CYCLES LINKING LEVEL GAIT AND STAIR DESCENT GAIT IN OLDER WOMEN

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ABSTRACT

Stair descent is an inherently complex form of locomotion posing a high falls risk for older adults, specifically when negotiating the transitional gait cycles linking level gait and descent. The aim of this study was to enhance our understanding of the biomechanical demands by comparing the demands of these transitions. Lower limb kinematics and kinetics of the 2-step transitions linking level and descent gait at the top (level-to-descent) and the bottom (descent-to-level) of the staircase were quantified in 36 older women with no falls history. Despite undergoing the same vertical displacement (2-steps), the following significant ($p<.05$) differences were observed during the top transition compared to the bottom transition: reduced step velocity; reduced hip extension and increased ankle dorsiflexion (late stance/pre-swing); reduced ground reaction forces, larger knee extensor moments and powers (absorption; mid-stance); reduced ankle plantarflexor moments (early and late stance) and increased ankle powers (mid-stance). Top transition biomechanics were similar to those reported previously for continuous descent. Kinetic differences at the knee and ankle signify the contrasting and prominent functions of controlled lowering during the top transition and forward continuance during the bottom transition. The varying musculoskeletal demands encountered during each functional sub-task should be addressed in falls prevention programmes with elderly populations where the greatest clinical impact may be achieved. Knee extensor eccentric power through flexion exercises would facilitate a smooth transition at the top and improving ankle plantarflexion strength during single and double limb stance activities would ease the transition into level gait following continuous descent.

INTRODUCTION

Descending stairs is a common task that permits functional ambulation between different levels. The knee extensors and ankle plantarflexors play an important role in stair descent biomechanics (McFadyen and Winter, 1988; Samuel et al., 2011) by dissipating mechanical energy and enabling forward progression, respectively (Cluff and Robertson, 2011). Considerable eccentric control of the knee and ankle musculature is required to resist the downward influence of gravity as the body undergoes repetitive free fall from one step to the next. Stair locomotion presents a considerable falls risk with early work indicating that 14% of all falls occur on stairs (Cohen et al., 1986) and 75% of all stair-related falls occur during descent compared to ascent in older adults (Masud and Morris, 2001). An important element in designing effective falls prevention programmes requires a comprehensive biomechanical understanding of task demand.

Studies have frequently analysed gait cycles that are initiated and terminated on independent steps while participants negotiate the stairs using a step-over-step, reciprocal gait pattern representative of continuous descent (McFadyen and Winter, 1988; Christina and Cavanagh, 2002; Hamel et al., 2005; Sheehan and Gottschall, 2011). During continuous descent, older adults operate within a higher proportion of their maximal dynamometer-derived capacity for both knee moments (old *vs.* young; 42% *vs.* 30%) and ankle dorsiflexion angle (107% *vs.* 91%) (Reeves et al., 2008). Further work has confirmed that mechanical demands at the knee are greater than at the hip with older adults using on average 100%, and in some cases 150% of available capacity (Samuel et al., 2011). Functional demands at the hip were on average ~20% of available isometric hip strength for both the flexor and extensor muscles (Samuel et al., 2011). Demands exceeding 100% of capacity may reflect the age-related differences in voluntary drive to activate muscles during selected testing protocols and variation in the protocols utilised (i.e., contraction type, chosen angular position/ velocity) which makes direct comparisons challenging. Whilst it is well known that continuous descent poses heightened mechanical demands for older adults, the kinematic and kinetic demands of the transitions linking level and continuous descent gait are less well understood.

One study investigating the influence of step location (comparison between continuous descent in the top and mid-stair region) upon ground reaction forces (GRF) during descent found altered GRF in both young and old (Christina and Cavanagh, 2002). Interestingly, an interaction effect was observed (step location*age) such that loading rates were larger as participants progressed down the staircase and this was more apparent for older adults. In support of this, Lee & Chou, (2007) showed that both young and older adults completed the bottom transition more quickly compared to continuous descent. Moreover, the same study indicated that unlike the young, older adults were unable to reduce their centre of mass (COM) sway angles from continuous descent to the bottom transition which the authors suggested may represent a reduced ability to stabilise during this transition (Lee and Chou, 2007). Given the likely increased severity of injury that would result from a fall from the top compared to the bottom of the staircase, and the progressive change in demands thought to occur throughout descent, analysis of lower limb mechanics during both transitions is vital to provide a thorough understanding of task demand and falls risk.

To the best of the authors' knowledge, only one early study directly compared the top and bottom transitions in young adults. This work revealed that whilst lower limb joints operate within a similar range of motion (ROM) during both transitions, differing kinematic profiles were observed (Andriacchi et al., 1980). Moreover, increased external hip and knee flexor moments and earlier onset of knee extensor muscle activity were noted for the top transition, albeit these differences were not evaluated statistically (Andriacchi et al., 1980) and require confirmation. Redirecting the COM from one level to another requires a prescribed change in lower limb mechanics modulated by changes in both step height and depth in response to staircases of varying design. These movement alterations require a superior level of postural and motor control facilitating appropriate multi-segment co-ordination. The biomechanical requirements to complete both transitional phases are likely to differ from one another as has been demonstrated for stair ascent (Alcock et al., 2014a) and when comparing 1-step transitions with continuous stair gait (Sheehan and Gottschall, 2011). Identifying the biomechanical demands of these transitions would guide evidence-based recommendations for targeted exercises, especially in high-falls risk groups, and encourage safer stair locomotion. This could have greatest impact for older women due to their increased falls occurrence and amplified falls risk associated with stair locomotion (Blake et al., 1988; Campbell et al., 1989; Gine-Garriga et al., 2009).

Therefore, the aim of this study was to compare the lower limb mechanics involved in the 2-step transition from the top and bottom of the staircase in older women with no falls history. It was hypothesised that functional differences would exist between the transitions particularly during stance, with the top transition necessitating greater controlled lowering and presenting demands similar to that of continuous descent (i.e., greater eccentric control of the knee extensors in terminal stance) and the bottom transition stance phase closely representing level gait (i.e., greater concentric knee power generation mid-stance, and larger ankle plantarflexor moments).

METHODS

PARTICIPANTS

Thirty-six female participants gave written informed consent to take part in this study which received National Health Service ethical approval (08-H1305-91). Participants were recruited through the local community and were pre-screened to exclude cardiovascular, musculoskeletal or neurological complaints, visual or cognitive deficits, polypharmacy or a history of falls. Group mean[SD] characteristics were: age 71.7[7]years, range 61-83 years; height 162.8[6.6]cm; mass 70.7[12.7]kg. This study was embedded within a larger project that quantified biomechanical profiles of older women completing daily activities (Alcock et al., 2013; 2014a; 2014b)

PROTOCOL

3D kinematics of the 2-step transition from the top and the bottom of the stairs were recorded using 14 ProReflex infrared cameras sampling at 100Hz (Qualisys, Sweden). Spherical reflective markers (14mm) were placed upon the participants' lower limbs bilaterally according to a six degrees-of-freedom marker system (Cappozzo et al., 1995). A custom-built staircase was utilised (step height: 20cm, depth: 25cm, width: 80cm, top landing depth: 80cm) as described previously (Alcock et al., 2014a). Orthogonal GRFs were measured using two 400x600mm piezoelectric force platforms (model 9286AA, Kistler, Winterthur, Switzerland) sampling at 500Hz. One platform was mounted within the first step and measured forces from the 2-step transition from the top of the staircase; while one ground-mounted platform recorded forces from the 2-step transition at the bottom of the staircase (Figure 1). Analogue data were converted through a 64-bit analogue-to-digital board and recorded synchronously with kinematic data. Participants were instructed to begin each trial from the back of the top landing and completed either one or two gait cycles on the landing before descending the stairs completing a total of 8-10 descent trials. Participants were asked to continue walking beyond the bottom of the staircase (approximately 4 metres) at their self-selected pace.

All participants used a reciprocal stepping pattern naturally and without prompt, and no participant used the handrails. During descent, and on the 3-step staircase used in this study (Figure 1), the lead limb initially descended from the top landing to step 2 (1-step top transition). The trail limb then descended two steps from the top landing to step 1 (2-step top transition). The next step of the lead limb was from step 2 to the ground (2-step bottom transition). The trail limb then descended from step 1 onto the ground (1-step bottom transition). It is noteworthy that, depending on the number of steps within a given staircase, the lead/trail limb functions will alter during the bottom transition. This study is specifically focused on comparing the 2-step transitions from the top and bottom of the staircase rather than the 1-step transitions due to the larger vertical displacement involved and consequently larger ROM required.

VARIABLES

Extracted temporal-spatial variables included velocity (m/s), cycle time (s) and stance phase duration (%). Peak lower limb joint angles and ROM were calculated during each of the 2-step transitions and joint angles were time-normalised to 100% gait cycle. The gait cycle was standardised to begin with toe-off, thus presenting the swing phase first followed by the stance phase, to facilitate comparisons with previous studies' (McFadyen and Winter, 1988; Karamanidis and Arampatzis, 2010; Sheehan and Gottschall, 2011). Foot contact and toe-off events were identified from the vertical displacement of the forefoot markers relative to the staircase structure and corroborated with GRF data when available. Peak medial (Fx1), lateral (Fx2), posterior (Fy1) and anterior (Fy2) GRF values were quantified. In addition, peak vertical forces during early (Fz1) and late stance (Fz3), the minimum force mid-stance (Fz2), and load and decay rates were analysed. GRF data were normalised to body mass and time-normalised to 100% stance. Inverse dynamics were used to calculate lower limb sagittal joint moments and powers and were time-normalised to 100% gait cycle. Body mass-normalised peak joint powers were determined according to the specific bursts defined by McFadyen & Winter (1988). To ensure that kinetic differences observed were not influenced by alternative force plate mounting structures, fast Fourier analysis was performed on the force plate in each of the settings used (concrete pit and wooden inset in the staircase structure). This analysis revealed that kinetic data were not confounded as a result of force platform mounting structure (Chesters et al., 2013) and results are presented in the *supplementary material*.

DATA ANALYSIS

A static calibration trial was collected prior to the movement trials to define segment lengths and identify lower limb joint centres. The hip joint centres were derived from the CODA pelvis which was constructed in Visual 3D (Bell et al., 1989; Bell et al., 1990). The knee and ankle joint centres were defined as the midpoint between the markers defining the lateral and medial aspects (i.e. femoral epicondyles and malleolus of the fibula and tibia, respectively) of two articulating segments (i.e. thigh and shank, respectively). Marker trajectories were identified and labelled in Qualisys Track Manager (v.2.7, Qualisys, Sweden), then exported to Visual 3D (v.3.90.7, C-Motion, Germantown, MD, USA) for subsequent analysis. Kinematic data were interpolated over a maximum gap of ten frames using a cubic spline algorithm and an X-Y-Z Cardan sequence defined the order of rotations according to the right hand rule about the segment coordinate axes (x: flexion/extension, y: abduction/adduction and z: longitudinal rotation). Kinematic and kinetic data were filtered using a low-pass Butterworth filter with cut-off frequencies of 6Hz and 25Hz, respectively (Siegel et al., 1996) and all data were averaged across the completed trials.

STATISTICAL ANALYSIS

Paired samples t-tests were conducted to analyse the biomechanical differences between the top *vs.* bottom transitional gait cycles. Paired comparisons were split into three groups: temporal-spatial, kinematic, and kinetic indices. A family-wise Hommel correction was used to manage the Type I error associated with multiple comparisons (Hommel, 1988; Falk, 1989). Two-tailed significance was reported as the direction of the group differences was not known. Where statistical differences were found, effect sizes (Cohen's d) were calculated to verify these differences. Significance was set at $p \leq .05$.

RESULTS

A significantly faster velocity, shorter cycle time and stance phase duration were observed for the bottom transition compared to the top ($p \leq .0018$; d=4.6-10.2, Table 1).

Significant kinematic differences were observed between the two transitions at the hip and ankle (Table 1 and Figure 2). The limb completing the top transition demonstrated increased peak ankle dorsiflexion (late stance) and ankle ROM ($p = .0064$) compared to the limb executing the bottom transition. Peak hip extension (late stance) and ankle plantarflexion (late swing/ early stance), were significantly greater during the bottom transition compared to the top ($p = .0064$, d=9.3 and 3.7, respectively).

Several GRF parameters (Fy1, Fz1, load and decay rates) were found to be statistically greater for the bottom transition compared to the top transition (Table 2 and Figure 3). The limb completing the bottom transition generated significantly greater ankle plantarflexor moments during early and late stance compared to the top transition ($p = .0095$). All statistically significant differences were confirmed by moderate Cohen's d effect sizes ranging from 6.7-20.7. The largest difference was the peak knee extensor moment (late stance) which was reduced during the bottom transition compared to the top transition (d=20.7, p=.009).

The limb completing the bottom transition generated significantly greater knee power mid-stance (p=.0095, Table 3 and Figure 4). Knee power absorption (late stance) and ankle power absorption (mid-stance) were significantly reduced during the bottom transition compared to the top. Differences in the peak ankle power generation (mid-stance) were reduced during the bottom transition and were non-significant post-corrective procedures.

DISCUSSION

This study investigated the differences between the 2-step transitions from the top and bottom of the staircase during stair descent. Despite both gait cycles undertaking a 2-step cycle, distinct biomechanical differences and contrasting functional demands were observed. In agreement with our hypothesis, the top transition was characterised by controlled lowering (represented by a larger knee extensor moment and eccentric extensor control), similar to continuous descent (McFadyen and Winter, 1988). In comparison, the bottom transition was completed more quickly with larger GRFs and plantarflexor moments indicating a greater requirement for forward propulsion into level gait.

Demands of descent transitions compared to level gait

Level gait mechanics for the same cohort have been reported previously (Alcock et al., 2013). Both stair transitions were completed more slowly, with an increased cycle time and reduced stance phase duration compared to level gait. Knee ROM was considerably greater during both transitions (~90°) than during level gait (~60°) as was ankle ROM due to greater dorsiflexion (~two-fold increase) and plantarflexion (~four-fold increase). Both the knee extensor moment and knee power absorption burst were largest during the top transition compared to level gait and the bottom transition. Increased ankle power generation was observed during both transitions compared with level gait.

Comparison between top and bottom transitions

The two descent transitions were distinguished by peak hip extension angles during late stance (top=9.2° flexion *vs.* bottom=2.3° extension, p=.0064) such that the hip never fully extended during the top transition. Moreover, the participants in the current study demonstrated more hip extension compared to the findings presented in Samuel et al. (2011) (20° flexion). Similar magnitudes of hip flexion were noted for the top transition in the present study and the continuous cycle reported in Reeves et al. (2008a). This suggests that the stance phase of the top transition in the present study (which was completed on the staircase) exhibited similar mechanics to that observed during continuous descent gait. Variations in hip extension profiles during the top transition between the present study and that of Samuel et al. (2011) may be attributed to varying staircase dimensions (height x depth: 20x25cm vs. 18.5x28cm for the current *vs.* Samuel et al. (2011) study, respectively). The large magnitude of hip extension observed during the bottom transition acts to facilitate the increase in step length of the ipsilateral limb onto level ground thus conforming more closely to the level gait mechanics of forward propulsion. This is in contrast to the top transition, whereby step length is dictated by the proceeding staircase dimensions. Therefore chosen step length beyond the staircase was not restricted by the impending step depth and increasing step length beyond the staircase inherently necessitates increased hip extension.

Kinematic differences further distinguishing between the two transitions included a significantly reduced dorsiflexion angle (late stance) and greater plantarflexion angle (late swing/early stance), which resulted in reduced ROM during the bottom transition. Greater plantarflexion upon contact increases functional leg length and thus facilitates appropriate foot placement whilst requiring less pelvic movement in the frontal plane (i.e., pelvic obliquity - not analysed in the present study). The most marked difference between the two transitions was the peak dorsiflexion angle which was largest during the top transition. Maximising ankle dorsiflexion may strategically increase the base of support (BOS) during the top transition, and thus dynamic stability, as it allows a larger area of the foot surface to remain in contact with the ground for longer (Lark et al., 2003). This strategy was observed in the current study during the top transition and may indicate an intention to maximise dynamic stability when eccentric demands at the knee are high. Consequently, improving locomotor stability when descending from the top of the stairs may be achieved by enhancing ankle ROM particularly within the dorsiflexion range. Moreover, concurrent use of the handrails would further enhance dynamic stability during this demanding task, thus helping to alleviate fall risk concerns in high risk groups.

The limb executing the top transition displayed many GRF parameters of reduced magnitude (Table 2) compared to the bottom transition. These alterations may be attributed to the increased velocity observed during the bottom transition as demonstrated previously following continuous, rhythmic descent (Lee and Chou, 2007). In addition, it is conceivable that locomotor confidence may increase as a person descends, as the severity of potential fall-related injuries may reduce closer to the bottom of the stairs. This effect may be even more pronounced on a conventional staircase comprising a greater number of steps during which online motor programmes may be fine-tuned and automated (Schmidt, 1975) according to standardised staircase dimensions. Further work is required to determine whether these discrete transitional forms of locomotion may impact falls risk due to temporal-spatial disparities, varying staircase designs, and the presence of a fear of falling which should be monitored in future studies.

A limitation of the current study was the use of only a 3-step staircase and the lack of reciprocal, continuous descent gait cycles separating one transition from another. It may be expected that a longer staircase comprising a greater number of consecutive steps (a minimum of four steps is required to permit analysis of a single continuous cycle and top and bottom transitions) would likely result in greater momentum generated at the bottom of the stairs. However differences between transitions were still detected with the present 3-step staircase and may be further amplified when ambulating at faster velocities and thus with greater momentum. Future work may incorporate a longer top landing and explore the chosen foot placement strategies adopted in the approach to stair descent, in conjunction with both 1-step and 2-step transitional biomechanics, to provide greater detail about this potentially hazardous transitional phase. Integrating COM and BOS calculations would help to determine whether older adults strategically choose foot placement to optimise global stability during transitional phases.

Participants self-selected their lead limb for each trial to represent their habitual descent biomechanics/ patterns most accurately. Lower limb mechanics were considered symmetrical during level gait for the same cohort and as such it was not expected that limb preference due to asymmetry would have influenced the data presented (Alcock et al., 2013). However, it would be interesting to understand whether participants with large between-limb strength differences and asymmetry (i.e., due to disease, disuse or trauma) elect to use the preferred limb for a particular transition given the varying demands exposed in this study. Enhancing our understanding of transitional mechanics should be extended to comparisons with young individuals, fallers and those with compromised balance to further understand transitional demands. Moreover, it is important to consider the adjacent steps to each of these transitions (i.e. 1-step transition or continuous stair gait of the contralateral limb) given the influence on the bilateral coordination of temporal-spatial, kinematic and kinetic indices. Finally it is noteworthy to highlight the variety of methods used to define a continuous *vs.* transitional gait cycle and the gait event (foot contact/ toe-off) that is used to define the beginning of the gait cycle (stance/ swing). It is critical that clear definitions and consistent terminology are established for stair phase gait mechanics to facilitate appropriate comparisons. We propose that a continuous gait cycle is defined as one that is initiated *and* terminated on an independent step, not including that of the floor level and thus all other gait cycles would be classified as transitional.

This study is the first to identify the functional biomechanical demands of transitions between level and stair descent gait in older women. Some preliminary recommendations for stair decent rehabilitation may be made for maintaining strength and joint ROM and evaluating these parameters in exercise-based interventions with other older adult populations (fallers, individuals with balance impairments etc.) should be the focus of future work. Exercise recommendations may include incorporating the considerable eccentric control required from the knee extensors (power absorption, late stance), concentric and eccentric control from the plantarflexors (power absorption and generation mid-stance) and greater magnitudes of ankle dorsiflexion and ROM required during the top transition. In contrast, increased concentric knee power generation (mid-stance) and ankle plantarflexor moments (early and late stance) were observed during the bottom transition and improving ankle plantarflexion strength during single and double limb stance activities would ease the transition into level gait following continuous descent. Finally, reduced hip and ankle joint mobility, particularly for joint extension, may restrict the propulsion away from the stairs and consequently inhibit initiation of level gait and limit step length beyond the stairs.

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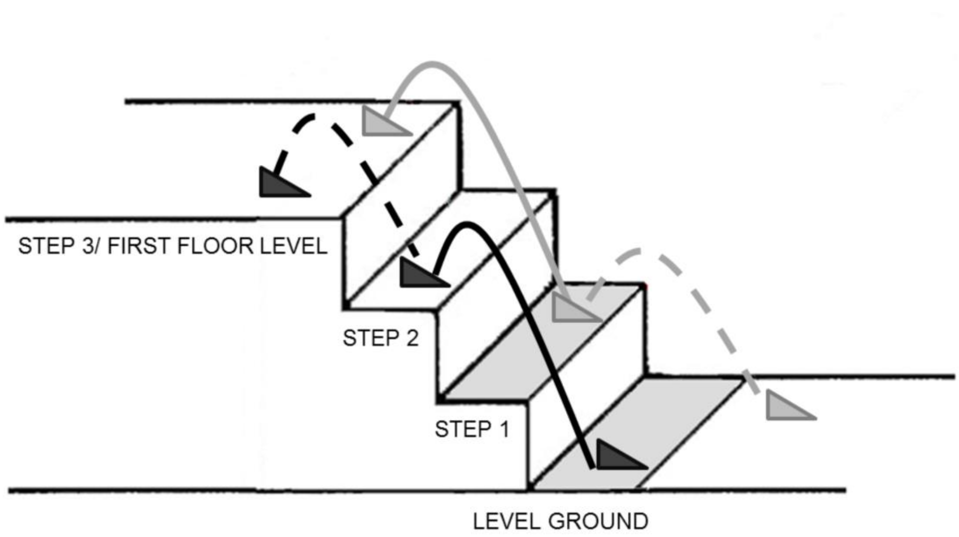

FIGURE 1 Schematic demonstrating the lead (black line) and trail (grey line) limb gait cycles during stair descent

The dashed lines represent the 1-step transitional gait cycles of the lead and trail limbs, while the solid lines represent the 2-step transitional gait cycles that were selected for further analysis. The grey shaded steps denote the positioning of force plates for kinetic data acquisition of the lead (ground) and trail (step 1) limbs. Both gait cycles studied were initiated and terminated by toe-off and data are presented firstly by swing, followed by stance.

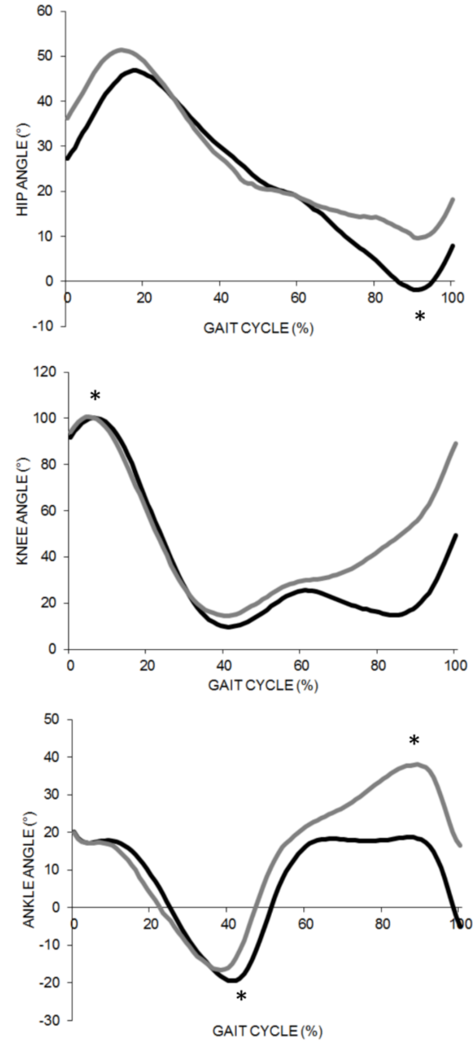

FIGURE 2 Ensemble average and time-normalised sagittal plane joint angle profiles (degrees) of the limb completing the level-to-descent gait transition (grey line, top floor level to step 1) and the limb completing the descent-to-level gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. Negative [+] values indicate extension and plantarflexion for the hip and ankle angles, respectively.

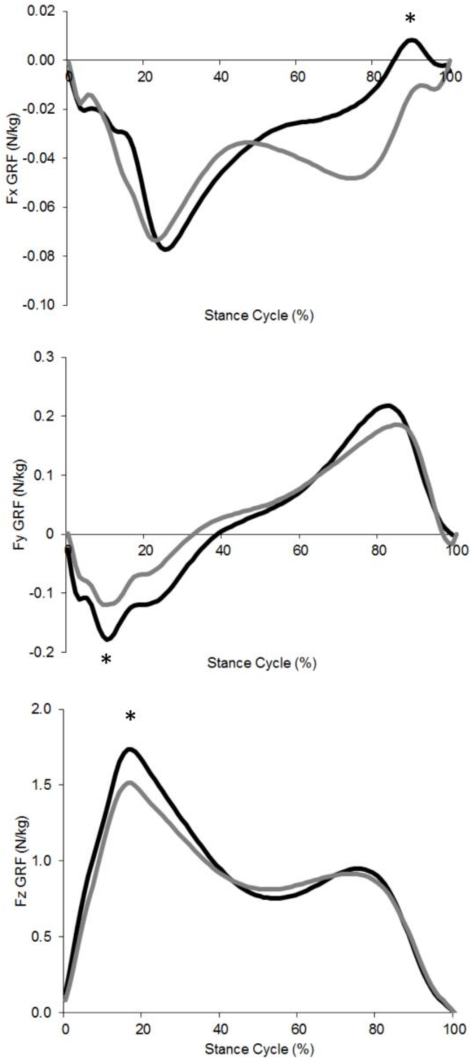

FIGURE 3 Ensemble average and time-normalised orthogonal ground reaction forces (N/kg) of the limb completing the gait-to-descent transition (grey line, top floor level to step 1) and the limb completing the descent-to-gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. Negative [-] ground reaction force values indicate lateral (Fx2) and posterior (Fy2) force components.

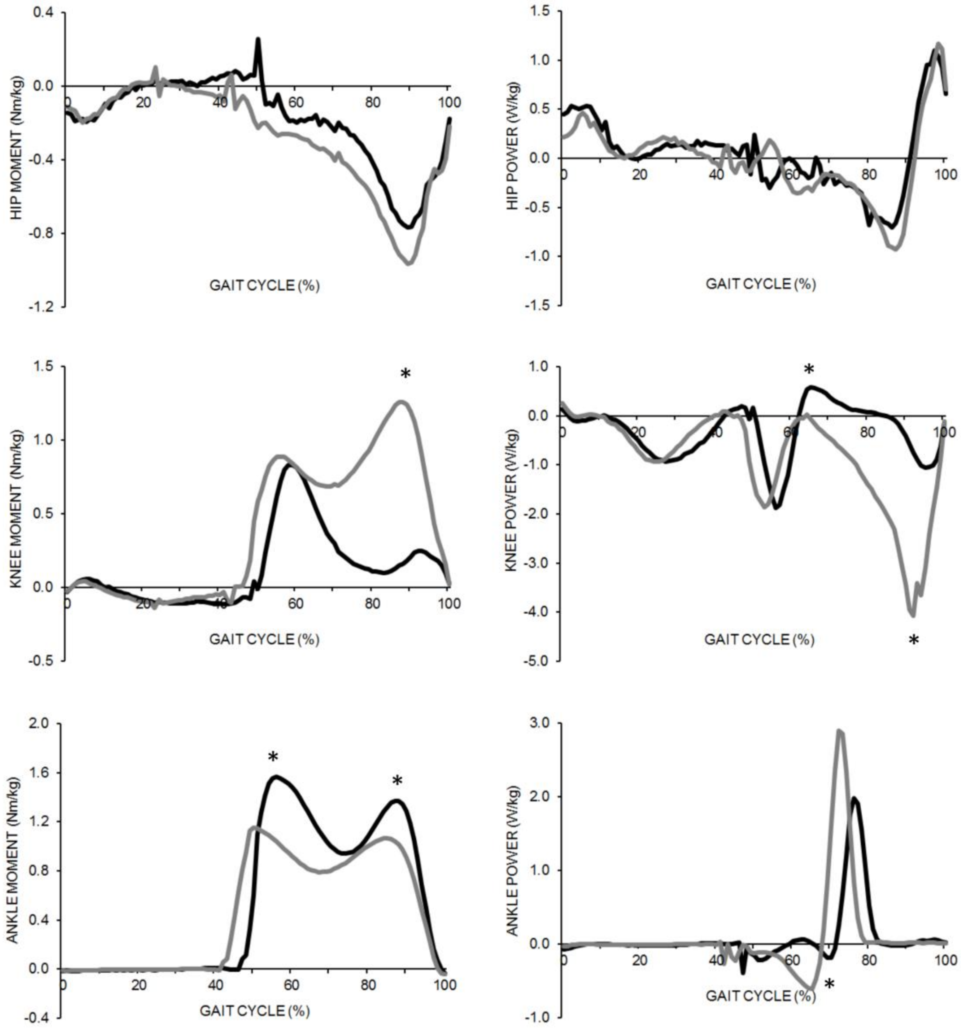

FIGURE 4 Ensemble average and time-normalised **s**agittal hip, knee and ankle joint moments (Nm/kg) and joint power profiles (W/kg) of the limb completing the gait-to-descent transition (grey line, top floor level to step1) and the limb completing the descent-to-gait transition (black line, step 2 to level ground)

* indicates significant between-limb differences (p≤.05) post corrective procedures. At the hip and knee, a positive [+] value indicates an extensor moment; at the ankle, a positive [+] value indicates a plantarflexor moment. Positive [+] powers denote concentric power generation and negative [-] powers denote eccentric power absorption at the respective joints.

TABLE 1 – Mean [SD] temporal-spatial and peak joint kinematics and ROM (degrees) parameters of the limb completing the top transition (top floor level to step 1) and the limb completing the bottom transition (step 2 to level ground)

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	*t*	SIG.	CORRECTED SIG.	COHEN'S d
TEMPORAL-SPATIAL							
GAIT SPEED (m/s)	0.64 [0.1]	0.84 [0.2]	0.17 : 0.21	16.7	.001	.0018	9.4
CYCLE TIME (s)	1.36 [0.3]	1.22 [0.2]	-5.91 : -2.91	-6.0	.001	.0018	4.6
STANCE (%)	57.7 [3.6]	53.3 [3.6]	-0.18 : -0.11	-8.4	.001	.0018	10.2
JOINT KINEMATICS (degrees)							
HIP FLEXION (Early swing)	53.3 [7.8]	46.8 [10.0]	-8.73 : -0.52	-2.3	.029	.1128	
HIP EXTENSION (Late stance)	9.2 [11.4]	-2.3 [9.3]	-14.15 : -8.78	-8.7	.001	.0064	9.3
HIP ROM	44.4 [8.2]	50.1 [6.9]	1.02 : 8.48	2.6	.014	.0713	
KNEE FLEXION (Early swing)	103.1 [7.2]	100.5 [9.4]	-5.00 : -0.25	-2.2	.031	.1128	
KNEE ROM	91.0 [5.4]	92.3 [7.3]	-1.49 : 3.95	0.9	.362	1.000	
ANKLE DORSIFLEXION (Early swing)	18.8 [8.3]	20.7 [7.6]	-0.88 : 4.79	1.4	.170	.541	
ANKLE PLANTARFLEXION (Late swing/ Early stance)	-18.3 [5.8]	-21.0 [6.6]	-3.99 : -1.38	-4.2	.001	.0064	3.7
ANKLE DORSIFLEXION (Late stance)	39.4 [7.8]	22.6 [4.9]	-19.38 : -14.30	-13.5	.001	.0064	22.0
ANKLE ROM	57.7 [6.1]	45.1 [5.7]	-14.92 : 10.21	-10.8	.001	.0064	17.7

ROM denotes range of motion. Shaded areas indicate significant between-limb differences. At the hip and ankle joints, a negative value [-] indicates hyper[extension] and plantarflexion, respectively.

TABLE 2 - Mean [SD] ground reaction forces (GRFs) and peak internal joint moments (Nm/kg) of the limb completing the top transition (top floor level to step 1) and the limb completing the bottom transition (step 2 to level ground)

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	*t*	SIG.	CORRECTED SIG.	COHEN'S d
			GROUND REACTION FORCES (*N/Kg)*				
MEDIAL FX1 GRF	0.01 [0.02]	0.02 [0.02]	0.00 : 0.02	2.474	.020	.1562	
LATERAL FX2 GRF	-0.08 [0.02]	-0.08 [0.03]	-0.01 :0.01	-0.294	.771	1.0000	
POSTERIOR FY1 GRF	-0.13 [0.03]	-0.18 [0.03]	-0.06 : -0.03	-5.836	.001	.0095	10.2
ANTERIOR FY2 GRF	0.21 [0.05]	0.21 [0.04]	-0.02 : -0.02	0.147	.884	1.0000	
VERTICAL FZ1 GRF	1.53 [0.19]	1.76 [0.22]	0.16 : 0.29	7.412	.001	.0095	9.1
VERTICAL FZ2 GRF	0.80 [0.09]	0.77 [0.10]	-0.08 : 0.00	-1.843	.076	.5021	
VERTICAL FZ3 GRF	0.94 [0.10]	0.97 [0.09]	-0.01 : 0.08	1.635	.114	.6994	
LOAD RATE [*N/kg/s]*	12.6 [3.9]	16.4 [4.5]	2.42 : 5.08	5.813	.001	.0095	7.5
DECAY RATE [*N/kg/s]*	4.6 [1.2]	6.2 [1.3]	0.98 : 2.11	5.600	.001	.0095	9.1
			JOINT MOMENTS (*Nm/Kg)*				
HIP FLEXOR MOMENT (Late stance)	-1.05 [0.5]	-0.88 [0.3]	-0.38 : 0.16	1.420	.168	.9018	
KNEE EXTENSOR MOMENT (Early stance)	0.93 [0.5]	0.85 [0.4]	-0.22 : 0.06	-1.152	.259	1.0000	
KNEE EXTENSOR MOMENT (Late stance)	1.23 [0.5]	0.31 [0.1]	-1.11 : -0.73	-9.903	.001	.0095	20.7
ANKLE PLANTARFLEXOR MOMENT (Early stance)	1.21 [0.3]	1.50 [0.4]	-0.38 : -0.19	6.330	.001	.0095	6.7
ANKLE PLANTARFLEXOR MOMENT (Late stance)	1.13 [0.1]	1.36 [0.2]	-0.32 : 0.13	4.860	.001	.0095	9.1

Shaded areas indicate significant between-limb differences, Negative [-] ground reaction force values indicate lateral (Fx2) and posterior (Fy2) force components. At the hip and knee, positive [+] values indicate extensor moments and at the ankle joint, positive [+] values indicate a plantarflexor moment

TABLE 3 - Mean [SD] peak joint powers (W/kg) of the limb completing the top transition (top floor level to step 1) and the limb completing the bottom transition (step 2 to level ground)

VARIABLE	TOP TRANSITION	BOTTOM TRANSITION	95% CONFIDENCE INTERVAL (Lower : Upper)	*t*	SIG.	CORRECTED SIG.	COHEN'S d
			JOINT POWERS [W/kg]				
HIP POWER GEN (Early swing)	0.59 [0.48]	0.67 [0.33]	0.03 : 1.14	0.865	.396	1.0000	
HIP POWER GEN (Late stance)	1.27 [0.61]	1.35 [0.88]	-0.29 : 0.44	0.428	.672	1.0000	
KNEE POWER ABS (Mid-swing)	-0.75 [0.29]	-0.75 [0.30]	-0.11 : 0.12	0.122	.904	1.0000	
KNEE POWER ABS (Early stance)	-2.10 [1.22]	-2.10 [1.48]	-0.59 : 0.60	0.029	.977	1.0000	
KNEE POWER GEN (Mid-stance)	0.34 [0.54]	0.75 [0.40]	0.13 : 0.70	2.955	.006	.0095	5.8
KNEE POWER ABS (Late stance)	-3.91 [1.41]	-1.24 [0.32]	2.12 : 3.23	9.911	.001	.0095	18.3
ANKLE POWER ABS (Early stance)	-1.02 [1.46]	-1.31 [2.33]	-2.81 : -1.29	-1.570	.126	.7215	
ANKLE POWER ABS (Mid-stance)	-0.78 [0.57]	-0.36 [0.48]	0.52 : 1.43	4.642	.001	.0095	6.1
ANKLE POWER GEN (Mid-stance)	3.52 [1.42]	3.09 [1.24]	0.42 : 0.89	-2.090	.046	.3292	

Shaded areas indicate significant between-limb differences. GEN denotes generation and ABS denotes absorption.

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JBiomech - Supplementary Material

Introduction

Musculoskeletal modelling of human movement requires the capture of accurate and valid kinetic data. Instrumented staircases such as the one in the present study are often unique in design, but permit kinetic data collection via force platforms embedded into metal or wooden staircases (Nadeau et al., 2003; Mian et al., 2007; Reeves et al., 2008), independent step structures (McFadyen and Winter, 1988), or concrete supports (Hamel et al., 2005), and those consisting of a structure placed on top of existing floor-mounted platforms (Lark et al., 2003). However, staircase design may introduce error when comparing stairway-derived forces with ground-mounted force platforms due to the material properties of the mounting structure.

Studies utilising instrumented staircases composed of wooden steps supported by metal frames (Chapdelaine et al., 2005; Della Croce and Bonato, 2007) have reported reductions in the natural frequency from staircases placed upon existing ground-mounted platforms when compared to stair-mounted platforms (Della Croce and Bonato, 2007). Conversely Chapdelaine *et al.* (2005) were unable to detect a natural frequency in the vertical direction due to a small oscillation impulse amplitude. Whilst alterations in the natural frequency have been shown to not impede upon the low frequencies typically associated with foot contact during gait and stair locomotion (Antonsson and Mann, 1985; Chapdelaine et al., 2005), it is not clear if the experimental set-up used in the present study provides robust kinetic data. Many studies employing the use of instrumented stairways or walkways have neglected to quantify the spectral power lost due to force plate mounting or define the signal filter introduced. Custom built experimental staircases are often constructed from wood (Lark et al., 2003; Nadeau et al., 2003; Vanicek et al., 2010; Alcock et al., 2014) conforming to building regulation dimensions with three steps (Andriacchi et al., 1980; Lu and Lu, 2006; Mian et al., 2007; Beaulieu et al., 2008; Vanicek et al., 2010; Alcock et al., 2014). Therefore, to validate such designs this supplementary material presents an evaluation of the power lost and signal filter introduced by the 3-step custom-built staircase utilised in the current study and others published previously (Vanicek et al., 2010; Alcock et al., 2014).

Methods

Staircase design and kinetic data acquisition

Dimensions and structure of the custom-built wooden staircase and associated force plate mounting are presented in Figure S.1. The 3-step staircase was comprised of two independent sections allowing a platform to be embedded in the first step with a 10mm gap around the platform edge.

Vertical ground reaction forces (GRF) were collected from a piezoelectric force platform (model 9286AA, Kistler, Winterthur, Switzerland) sampling at 500Hz through a 64-bit analogue-to-digital board. A 3kg medicine ball was released by hand from a 1-metre height (measured by a stadiometer) onto the platform and allowed to bounce once, two such trials were completed. This process was performed with the force platform embedded into: (1) a floor-mounted level concrete pit (FP_{GROUND}); and (2) the first step of a wooden 3-step stairway (FP_{STEP}).

Spectral analysis

Spectral analysis (SA) of the vertical GRF from each trial was performed by FFT between 0-250 Hz in 2048 bins at a resolution of 0.122 Hz using Matlab (R2008a, Mathworks, Natick, MA). Mean power spectrums were produced. 50SA (median frequency) and 95SA, defined as the spectral frequency at which 50% and 95% of the power fell below; and total energy (TE) of each spectrum were calculated. Additionally, the transfer function for FP_{STEP} with respect to FP_{GROUND} was calculated between 0Hz and 18Hz. SA performed on a previously recorded vertical GRF trace recorded during gait analysis (Male, age=27yrs, height=1.84m, mass=78.1kg, gait speed=1.12m/s) defined this frequency range as containing 99.95% of spectral power during foot strike. This transformation also allowed the volume of spectral power lost (%) during gait due to the transfer function to be calculated.

Statistical analysis

Independent samples t-tests were performed on 50SA, 95SA, and TE for each condition using SPSS (v18.0, SPSS Inc., Chicago, IL). Homogeneity of variance was assessed using Levene's statistic and equal variances were assumed. Statistical significance was set at $p \leq 0.05$.

Results

Impulse from the ball-drop contained energy across a wide range of frequencies (see Figure S.2). 50SA was reported as 16.58 Hz for FP_{GROUND}, indicating most of this energy was at low frequencies. Power spectrum for FP_{STEP} deviated from FP_{GROUND} at ~10 Hz. Significant differences were observed for 50SA and TE. The mean 50SA (M = 16.58, SD = 0.92, CI = 15.44:17.71) and mean 95SA (M = 52.47, SD = 3.02, CI = 48.72:56.21) for FP_{STEP} were significantly different to the mean 50SA (M = 20.39, SD = 0.46, 95%CI = 19.81:20.96) and mean 95SA (M = 52.78, SD = 1.16, 95%CI = 51.34:54.23) for FP_{GROUND}. Similarly, when considering TE, FP_{STEP} (M = 13344.78, SD = 872.42, 95%CI = 12261.52:14428.04), was significantly different to FP_{GROUND} (M=17107.46, SD=578.83, 95%CI=15967.93:17939.99).

The calculated transfer function indicated that limited signal filtering occurred and only at the highest frequencies for FP_{STEP} (Figure S.3). When considered with respect to the power spectrum during foot contact, the total loss of power was found to be 2.2% for FP_{STEP} (Figure S.3).

Discussion

This analysis has shown that kinetic data sampled by force platforms embedded in instrumented wooden stairways were altered at high frequencies. However, in the frequency range of interest to gait (0-18 Hz) and in this study, the differences were considered minimal. Significant differences observed in 95SA and TE reflected energy across the complete frequency spectrum. This suggests modifications in high frequency platform response may have occurred when mounted in the wooden step. 50SA was changed significantly; this may have indicated some alteration of the low frequency force response relevant to gait in this condition. However, the analysis of transfer functions suggests only a small portion of TE was lost when the platform was staircase-mounted. This loss was considered negligible in comparison to other errors introduced during in motion capture (Chiari et al., 2005).

Whilst other studies have investigated the acquisition of kinetic data from instrumented stairways, those studies applied impulses of low magnitude (Della Croce and Bonato, 2007) (0.1kg from 1-metre height), and only considered natural frequencies. The experiment presented in this supplementary material considered a much larger impulse and is the first to quantify the energy lost due to staircase design. It was noted the drop-ball procedure provided energy from a wide range of frequencies and TE from ball-drop was ~2 times that produced by a foot strike during gait. Activities such as stair climbing and faster walking produce greater energy, thus, the impulse selected was of suitable size to assess force platform performance in a gait laboratory. As the largest component of the GRF vector, the vertical GRF was analysed due to its considerable influence on kinetic computations. Furthermore, the vertical GRF is thought to be the most consistent during gait, as the medio-lateral and anterior-posterior forces can vary substantially, and was therefore appropriate to represent analysis of force platform performance.

In conclusion, this analysis found that negligible power was lost when mounting a force plate into a 3-step wooden staircase structure and may alleviate concerns that the kinetic differences highlighted between the transitional steps at the top and bottom of the staircase may have been filtered substantially as a result of staircase mounting. Moreover, this methodology may be repeated in gait laboratories using custom-built staircases made of alternative materials and comprising of more steps and force platforms.

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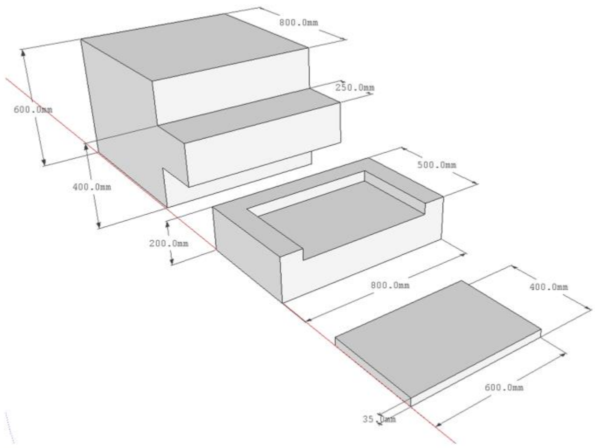

Figure S.1 – Geometric drawing of the 3-step custom built staircase components depicting the main structure comprising steps 2 and 3 (far left), integrated first step housing the force plate (centre; FP_{STEP}) and the force plate (right; FP_{GROUND})

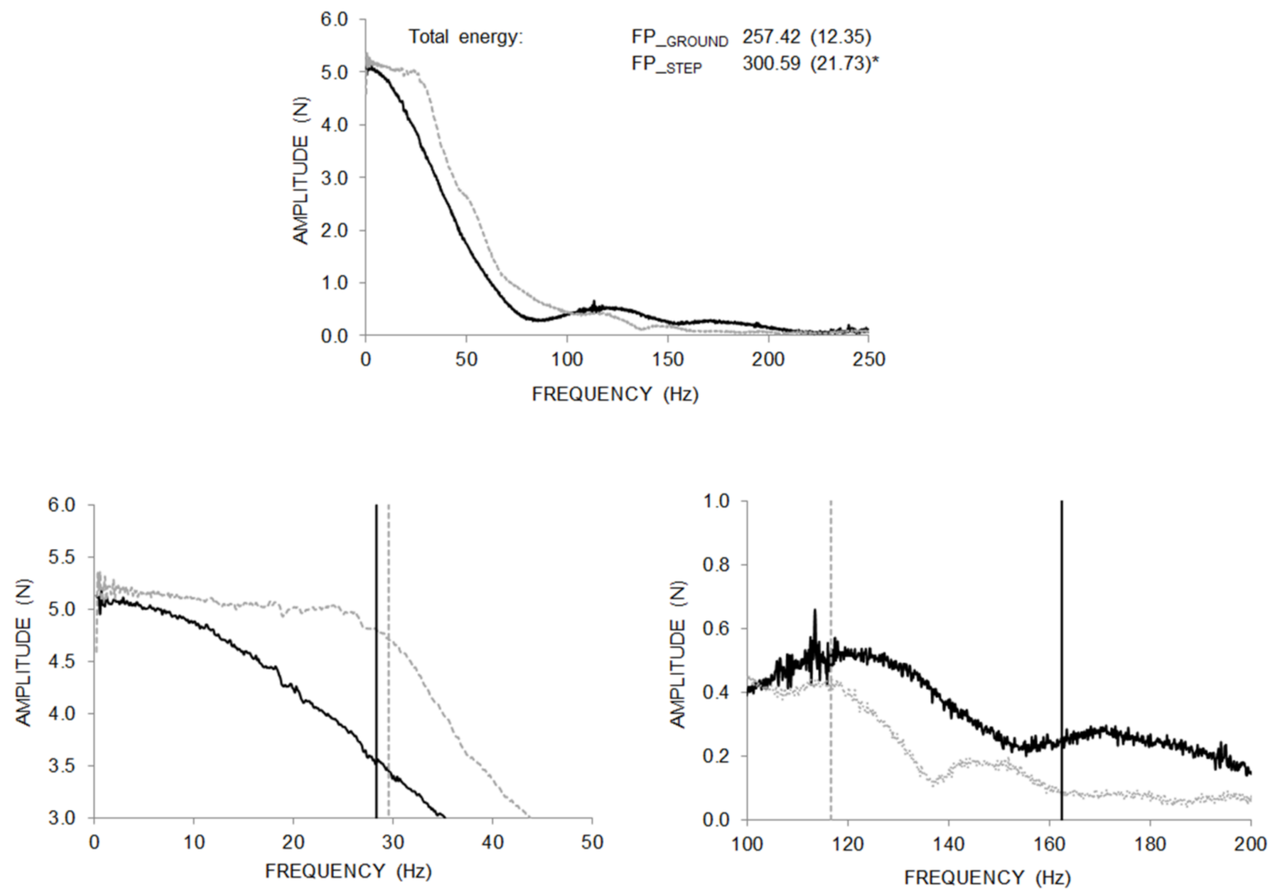

Figure S.2 – Power spectrum of the ball drops displayed on a normal scale (top), and magnified scale depicting the mean 50SA (bottom left) and mean 95SA (bottom right) for FP_{GROUND} (black solid line) and FP_{STEP} (grey dashed line)

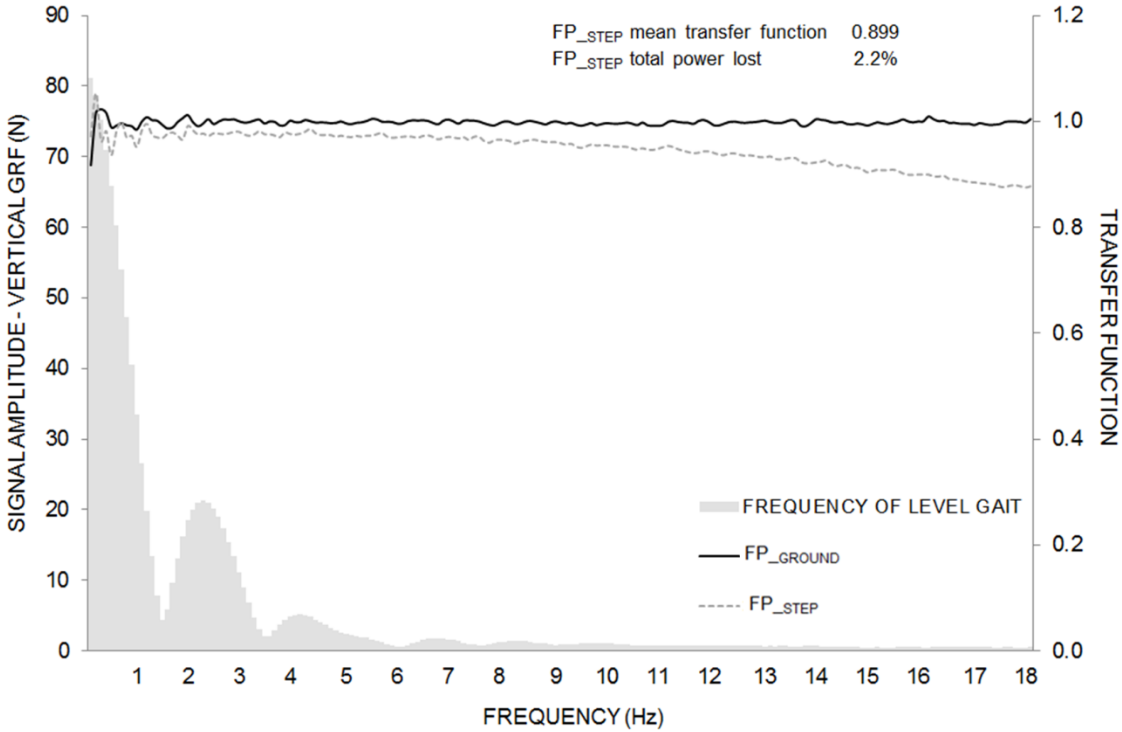

Figure S.3 – Power spectrum of an example foot contact during level gait (shaded grey) up to 95% power, and transfer functions for FP_{GROUND} (black solid line) and FP_{STEP} (blue dashed line)