Optimal fall indicators for slip induced falls on a cross-slope

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Optimal Fall Indicators for Slip Induced Falls on a Cross-Slope

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Abstract: Slip induced falls are among the most common cause of major occupational injuries in the UK as well as being a major public health concern in the elderly population. This study aimed to determine the optimal fall indicators for fall detection models which could be used to reduce the detrimental consequences of falls. A total of 264 kinematic variables covering three dimensional full body model translation and rotational measures were analysed during normal walking, successful recovery from slips and falls on a cross-slope. Large effect sizes were found for three kinematic variables which were able to distinguish falls from normal walking and successful recovery. Further work should consider other types of daily living activities as results show that the optimal kinematic fall indicators can vary considerably between movement types.

Keywords: fall detection; slips; falls; kinematics; cross-slope; balance control

Practitioner's summary: Fall detection models are used to minimise the adverse consequences of slip induced falls, a major public health concern. Optimal fall indicators were derived from a comprehensive set of kinematic variables for slips on a cross-slope. Results suggest robust detection of falls is possible on a cross-slope but may be more difficult than level walking.

1. Introduction

Loss of balance leading to a fall has serious and sometimes life threatening consequences in young and older adults (Bhatt, Wening, and Pai 2006). Slips and trips were the most common cause of major occupational injuries in the UK in 2012-2013 which resulted in a loss of 1,332,000 working days (Health and Safety Executive, 2013). In the US, 15% of non-fatal injuries and 14% of fatal injuries were due to same level falling accidents (US Bureau of Labor statistics, 2012). Slipping is commonly the triggering event of occupational falls accounting for between 40 and 50% of falling incidents (Courtney et al. 2001). Ageing results in an increased susceptibility to falling (Rubenstein 2006) with falls caused by slips responsible for approximately 60% of outdoor falls among community living adults over the age of 70 (Luukinen et al. 2000). Falls and related injuries are therefore a major public health concern in the elderly population (Kannus et al. 2007). Even in the absence of injury, falls can cause fear, loss of independence and a reduced ability to function suitably in social roles (Rogers and Mille 2003; Berg et al. 1997).

The effective prevention of falls due to slips relies on an in-depth understanding of the biomechanics of slips and slip mechanics, particularly during the balance recovery phase (Hu and Qu 2013; Redfern et al. 2001; Lockhart, Woldstad, and Smith 2003). This knowledge can be used to facilitate the development of fall detection models. Fall detection can be used to minimise injury severity either by initiating timely medical assistance for fallers or by activating a fall prevention device to avoid physical impacts (Bourke, O'brien, and Lyons 2007). Previous attempts to develop fall detection models have most often been concerned with determining the optimal locations for fall detection sensors with fall indicators which were predetermined without theoretical or experimental basis (Wu 2000; Doukas et al. 2007; Nyan, Tay, and Mah 2008; Nyan, Tay, and Murugasu 2008; Bagalà et al. 2012). A recent study addressed this limitation by experimentally examining a large set of kinematic measures to determine their ability to differentiate slip-induced falls from normal walking and successful recovery (Hu and Qu 2013). Vertical kinematic measures of the upper arm, trunk and head segments were identified as the optimal variables which could detect falls during the early stance phase of gait. This study was limited to slip induced falls on a level walkway and therefore the ecological validity of these results to many activities of daily living was limited (Hsu et al., 2015).

A better understanding of slip dynamics during non-level walking conditions is important as surface incline in a direction perpendicular to the line of progression (i.e.

cross-slopes) are a regular feature of our physical environment; pavements often have intermittent or prolonged cross-slopes which may impede gait stability (Dixon and Pearsall 2010), additionally in hilly areas, inclined roads may need to be traversed. Previous research has indicated an increased likelihood of slips occurring on a cross-slope than on a flat surface, postulated to be due to the reduced normal force acting on the foot requiring a higher coefficient of friction to avoid a slip (Lawrence et al., 2015). This suggests that there may be an increased perceived risk of falling when walking along a potentially slippery cross slope. Furthermore, it has been suggested that cross-slopes may be a causative factor in falls due to the asymmetrical demands of cross-slope walking which may introduce functional balance barriers (Dixon and Pearsall, 2010). Due to the reported increased functional demands of walking on a cross-slope (Dixon and Pearsall 2010), the consequence of slipping on a cross slope could be quite considerable, potentially presenting a high risk of injury (Lawrence et al., 2015). The aim of this study was to determine the optimal fall indicators for fall detection research by experimentally examining the ability of kinematic measures to differentiate falls from non-fall activities on a cross-slope.

2. Methods

2.1 Data Collection

Participants were 14 young men aged 25.3 ± 2.9 years (mean ± SD), height 1.79 ± 0.10 m, and mass of 72.5 ± 5.6 kg. Participants were healthy and free from injury with no history of balance or musculoskeletal disorders. Written informed consent was obtained from all participants and the ethics committee of Sheffield Hallam University granted approval prior to the study.

A Polhemus Liberty system (Polhemus Inc., Colchester, VT, USA) was used to track full body kinematics using 14 sensors, sampling at 240 Hz. The data were filtered using a second order dual pass Butterworth filter with optimal cutoff frequency of 6Hz determined using residual analysis (Winter, 1980). During trials participants were fitted into a harness attached to an overhead rail in order to prevent injury from fall impacts. Walking trials were conducted on a wooden walkway which had a total length of 4.8m. The walkway had a 7° tilted cross-slope descending from left to right relative to the walking direction. The first 2.3m and last 1m section of the walkway were covered in non-slip rubber. The middle 1.5 m section was covered by removable vinyl flooring. The walkway had profile consisting of a sloped and a flat section throughout its length (figure 1); the flat section was necessary to minimise the risk of participants slipping off the edge of the walkway and reduce the chance of injury to the lower limbs.

2.2 Procedure

Sensors were placed on moulded thermoplastic and placed on body segments using a self-adhesive wrap. The location of sensor attachment sites were chosen to prevent movement restriction, minimise soft tissue movement artefact and minimise distortion to the electromagnetic field caused by metal in the harness. With the participant standing in the anatomical position, the locations of 35 anatomical landmarks were digitised using a stylus, with their position recorded in the relevant segment's sensor local coordinate system. These landmarks were used to create proximal and distal joint centre positions for each segment and define anatomical coordinate systems in accordance with International Society of Biomechanics guidelines (Wu and Cavanagh 1995). In this way a 14 segment body model was defined which consisted of the head, upper arms, forearms, lower legs, thighs, feet, pelvis, lower trunk and upper trunk. For each body segment, the corresponding segmental COM was estimated using the parameters provided by De Leva (1996). Segment anatomical coordinate systems were defined such that the x, y, z axes were predominantly medio-lateral, longitudinal and anterior-posterior respectively. Joint angles for all joints were reported about three axes; flexion-extension (F/E), abduction-adduction (A/A) and internal-external rotation (I/E).

Each participant was asked to walk at a self-selected normal walking speed along the cross-slope. Participants always walked on the sloped part of the walkway with both feet. The flat section was a safety consideration in order to eliminate the risk of participants slipping off the side of the walkway. Participants always started at the same end of the walkway and therefore the slope was always in the same direction relative to the direction of travel. Participants were instructed to look straight ahead when walking to divert their attention from the floor and preclude awareness of changes in surface. Furthermore, it was ensured that changes in the surface were not visible. For slip trials an odourless soapy solution was placed on the vinyl section of the cross-slope. Between trials, the contaminant was completely removed from the surface of the cross slope and the soles of the participant's shoes. To prevent awareness of the experimenter changing the floor surface, participants sat facing away from the cross-slope listening to music and the time between trials was kept consistent (≈ 120 seconds). Participants were not restricted as to which foot came in to contact with slippery surface first in order to negate any effect controlling for slipped leg could have had on normal gait patterns. For the analysis however, only trials where the right foot came into contact with the slippery surface first were included. Slip trials could be unambiguously categorised as falls or successful recovery from visual inspection of the video footage. Slip/fall categorisations were verified using quantitative methods previously employed to detect falls as follows; falls would be identified if the midpoint between the left and right hip joint centres (estimated by the anterior superior iliac spines (ASIS) markers) dropped below 95% of its minimum height measured during normal walking (Beschorner and Cham 2008; Hu and Qu, 2013). Otherwise, balance recovery was considered to be successful.   A total of 20 walking trials were collected for each participant. The first five were normal walking trials without the risk of a slip. Five out of the following 15 trials were randomly selected as slip trials.

2.3 Data Reduction

Of the 70 slip trials recorded, 10 were unusable and therefore excluded from the analysis; 7 trials were excluded because participants grabbed the safety harness rope during slipping. A further 3 were excluded as the participants had not taken 3 steps before coming into contact with the slippery surface, therefore failing to establish steady state gait (Muir et al., 2014).   A total of 16 trials were excluded due to the right foot not being the first to come into contact with the slippery surface. One normal walking trial was randomly selected for analysis for each participant; a total of 14 normal walking trials, 36 successful recovery and 8 falls were recorded and used for subsequent data analysis. Five out of 14 participants fell at least once (1 participant fell 3 times, 1 fell twice and 3 fell once). The remaining 9 participants were able to successfully recover every time they were slipped. Table 1 shows the walking speed and stance duration for selected trials. For slip trials (successful recovery and falls) walking speed and stance duration was calculated using a normal gait cycle before the participant reached the slippery surface. Gait events were analysed using Visual3D^{TM} software (C-Motion, Inc., Rockville, MD, USA). Heel strike (HS) and Toe Off (TO) events were identified by a kinematic method which used the anterior-posterior velocity of the foot segment relative to the pelvis (Zeni, Richards, and Higginson 2008); HS was defined as the instant of relative positive-to-negative zero crossing of the foot segment velocity, and similarly TO was the negative-to-positive zero crossing. There were no significant differences between gait speed and stance duration between normal walking, successful recovery and falls (table 1).

2.4 Dependant Variables

Dependant variables included segment translational measures, segment rotational measures and joint angular measures as used in Hu and Qu (2013). Segment translational measures included the linear velocity and acceleration of the segmental COM. Segment rotational and joint angular measures were defined by the orientation of the segment relative to the global reference frame and proximal segment, respectively. Angular velocity and acceleration were analysed. Slipped heel velocity and acceleration were also included in the analysis as previous studies had indicated that heel kinematics are important in the analysis of slip biomechanics (Redfern et al. 2001). Kinematics were calculated for each of the 14 body segments. Linear velocity $(\dot{x}_i, \dot{y}_i, \dot{z}_i)$ and linear acceleration $(\ddot{x}_i, \ddot{y}_i, \ddot{z}_i)$ were calculated from differentiated segment COM displacement data and presented along the three axes of the global reference frame. The global reference frame was a conventional three orthogonal axes system aligned such that the z axis was parallel to the walking direction, the y axis was perpendicular to the floor (vertical) and the x axis was defined as the cross product of the z and y axes. Segment rotational velocity $(\dot{\alpha}_i, \dot{\beta}_i, \dot{\gamma}_i)$ and acceleration $(\ddot{\alpha}_i, \ddot{\beta}_i, \ddot{\gamma}_i)$ were calculated using Euler angles (α, β, γ) ordered rotational measures about the Z_i , Y_i and X_i axes, respectively, following ISB standards (Wu and Cavanagh 1995).

2.5 Statistical Analysis

Three 'activities' were analysed: normal walking, successful recovery and falls. For normal walking, data were selected from a complete stance phase during a gait cycle.

For the slip trials (successful recovery and falls), data were analysed from the point at which the slipped foot contacted the slippery surface, i.e. at heel strike of the slipped foot. To account for temporal differences in gait cycles between trials and activities, the time over which data were analysed was normalised to the mean duration of the stance phase of the normal walking gait cycles using spline interpolation (MATLAB 8.2, The MathWorks Inc., Natick, MA, USA). Uni-variate analysis of covariance (ANCOVA) - with walking speed as the covariate - was performed for each 2% time interval between 0% and 30% of stance, i.e. 0%, 2%, 4%… 30%. Given the large number of dependent variables, statistical significance (α) was set to 0.01, to reduce the type I family wise error rate. *Post hoc* pairwise comparisons using the Bonferroni correction were further conducted on the dependant variables that were significantly different among activities. In addition, effect sizes quantified using Cohen's *d* were calculated for dependant variables that were able to differentiate falls from non-fall activities at three or more consecutive time points with values of 0.2, 0.5 and 0.8 correspond to 'small', 'medium' and 'large' effect sizes, respectively (Cohen 1992). All statistical analysis was carried out using Matlab™ statistics toolbox (MATLAB and Statistics Toolbox Release 2013b, The MathWorks, Inc., Natick, MA, USA).

3. Results

The ANCOVA identified 173 dependant variables that were significantly different across the three activities (p <0.01) in the early stance phase (table 2). *Post hoc* multiple comparisons identified 22 dependant variables which were significantly different between falls versus normal walking and successful recovery for at least three consecutive time points in the early stance phase (table 3). The effect sizes for the comparison of falls versus normal walking were large for $\dot{x}_{L\_Foot}$, $\dot{x}_{L\_Shank}$, $\dot{y}_{L\_Thigh}$ and

$Y\ddot{\theta}_{L\_Hip}$ (table 4). For falls versus successful recovery there were large effect sizes for $\dot{x}_{L\_Foot}$, $\dot{x}_{L\_Shank}$ and $Y\ddot{\theta}_{L\_Hip}$ at 0%, 22-28% and 16% into stance, respectively. Mean profiles of the dependant variables averaged over the entire sample with large effect sizes are shown in figure 2. Falls resulted in large differences in $\dot{x}_{L\_Shank}$ between 22% and 28% into stance and for $Y\ddot{\theta}_{L\_Hip}$ at 16% into stance. For $\dot{x}_{L\_Foot}$ there were considerable differences between falls, successful recovery and normal gait for 0-12% of stance.

4. Discussion

A total of 264 kinematic variables covering three dimensional full body model translation and rotational measures were analysed during normal walking, successful recovery and falls on a cross-slope. There were 173 kinematic variables which were significantly different ($p<0.01$) between activities in the early stance phase of gait (0-30%) and of these there were 22 variables that were able to distinguish falls from both normal walking and successful recoveries for at least three consecutive 2% time intervals. Effect sizes were calculated to identify which differences in kinematic variables were practically meaningful, i.e. large enough to be easily detected. Large effect sizes for distinguishing between normal walking and falls were found for left foot linear velocity in the medial-lateral direction ($\dot{x}_{L\_Foot}$), left shank linear velocity in the medial-lateral direction ($\dot{x}_{L\_Shank}$) and left hip angular acceleration about the Y axis ($Y\ddot{\theta}_{L\_Hip}$). Left shank medial-lateral linear velocity was significantly different between activities with a large effect size for the greatest portion of the early stance phase (22%-28%) which indicates that this may be the most robust variable for use with fall detection models.

For slips on a level walkway, Hu and Qu (2013) identified five kinematic variables which had large effect sizes for falls versus successful recovery; head vertical acceleration, upper arm vertical velocity, trunk vertical velocity, shank frontal velocity, and head frontal angular velocity. In the current study, three variables were associated with large effect sizes for fall vs. successful recovery, and of those three, two only showed large differences for a single 2% time step during stance. This suggests it may be more difficult to detect falls during slips on a cross-slope. Effect size is a measure of the size of the differences in the means between groups in relation to the between-participants variability. For successful recovery versus a fall, the pooled group standard deviation was, in general, large. The large variability might be due to the highly asymmetric nature of the slipping movement. Cross-slopes have been shown to induce substantial asymmetrical changes in gait dynamics (Dixon and Pearsall 2010; Pearsall et al. 2007; Urry 2002; De Garie and Pearsall 2000; Nicolaou and Pearsall 2002; Walsh et al. 2000). It has been shown that, during gait, sagittal plane kinematics are modified to adapt to and minimise ground height differences between the up-slope (US) and down-slope (DS) feet induced by a cross-slope (Dixon and Pearsall 2010). A more pronounced asymmetry of coronal lower extremity kinematics between the US and DS limbs is required to stay true to the forward course without slippage and avoid falling (Dixon and Pearsall 2010).

A further factor which may have added to the variability of kinematic variables was the range of balance recovery strategies employed by the participants. During successful recovery, participants could be grouped into using either a single step strategy or a cross over strategy. For the single step strategy the step was directed so as to recover lateral stability whereby the slipped foot slid laterally down the slope in order to widen the base of support (i.e. the distance between the feet). This was similar to the strategy described by Rogers and Mille (2003) for recovery of sideways falls. For the cross over strategy, the un-slipped leg crossed either in front of or behind the slipped leg requiring a more complex limb trajectory than the single step strategy (Rogers and Mille 2003). Trunk position also varied between participants, some were able to maintain a forward facing stance during the slip whilst others rotated their body inwards so that at the end of the slip the body was facing up the slope. Arm movements were also observed: arms were elevated laterally outwards and upwards, although sometimes only one arm was raised. It has been previously suggested that such arm movements aim to assist in shifting the COM anteriorly after being initially displaced posteriorly by the slip (Marigold, Bethune, and Patla 2003). Falls generally occurred as a falling backward motion either with the body facing in the direction of travel or turning to face downslope.

Previous studies have indicated that humans alter their gait patterns when walking on a known slippery surface (Moyer et al. 2006) and that prior slip experience causes more pronounced gait adaptations than awareness alone (Heiden et al. 2006). During the experiment, all slips were expected with the only difference between slip trials being that some were after prior experience of a fall. Therefore gait patterns may have been more similar between trials than comparisons of unexpected slips with expected slips. As such, there were no significant differences between gait speed and stance duration between normal walking, successful recovery and falls. Therefore multiple slip trials from individual participants were included in the analysis.

To negate any effect controlling for slipped foot could have had on normal gait patterns, participants were not restricted as to which foot came into contact with the slippery surface first. This meant that the number of trials included in the analysis was reduced and resulted in a small sample of fallers (n = 8) compared to successful recoveries (n = 14). Each fall and successful recovery was treated as independent samples and as such some participants were weighted more heavily than others.

This study was limited to young male participants and as such the optimal fall indicators and effect sizes may differ for other populations such as the over 65s, a high risk category for slip injury (Berg et al. 1997; Sterling, O'Connor, and Bonadies 2001). Young and older adults have been shown to employ different balance control strategies (Rogers and Mille 2003). Further work should include older participants.

Unlike other fall detection studies which examined instantaneous kinematic variables such as peak heel velocity which could have occurred at different times relative to gait events (Cham and Redfern 2002) this study aimed to identify kinematic measures that could differentiate falls from non-fall activities in a real time manner. Lead time refers to the time between fall detection and fall impact. A large lead time is advantageous as this indicates more time available for protective systems to activate before impact. Therefore, identification of the kinematic variables that are able to detect falls soonest is an important aspect to consider in the development of fall detection models. Of the three kinematic variables with large effect sizes, left foot linear velocity in the medial-lateral direction ($\dot{x}_{L\_Foot}$) became significantly different 0% into the stance phase with a medium effect size between 2-12% of stance. Therefore, using this parameter would result in the largest lead time.

A fall indicator must have three features to achieve good performance. First, it must be able to differentiate from non-fall activities. Second, the difference must be large enough to be detected. Third, it should detect falls quickly , i.e. the fall detector should be able to detect when a fall is going to happen in time for action to be taken (Hu and Qu 2013). Our statistical analysis method addressed each of these features and advanced on previous studies by considering gait on a cross-slope. Results should be used to inform the development of fall detection sensors which can be worn by a person at risk of falling. This study analysed body segment kinematics as these parameters are directly amenable to real time measurement using wearable, low cost hardware. For example, previous studies have used accelerometers placed on the relevant body segment(s) to measure kinematics (Wu 2000; Doukas et al. 2007; Nyan, Tay, and Mah 2008; Nyan, Tay, and Murugasu 2008; Bagalà et al. 2012). Large effect sizes were found for three kinematic variables which were able to distinguish falls from normal walking and successful recovery. These large effect sizes indicate that these differences would be large enough to be practically meaningful for use with a fall detection model. The importance of selecting appropriate fall indicators for fall detection models was emphasised by the inclusion of a fall-like activity (successful recovery after a slip) which reduced the number of suitable kinematic variables considerably. The inclusion of a fall-like activity was important as it better replicates daily living situations. Further work should consider other types of daily living activities as this study has shown that results can vary considerably between types of movement analysed.

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Table 1. Walking speed and stance duration for each activity, mean (SD).

	Normal walking		Successful recovery		Falls		*p* value
Walking speed (m/s)	1.16	(0.16)	1.02	(0.26)	1.17	(0.08)	0.114
Stance duration (ms)	710	(114)	757	(255)	747	(82)	0.188

Table 2. Summary of the results from the ANCOVA (mean ± SD)

Dependant variables	% into stance	Normal Walking	Successful recovery	Fall	P values for independent variables	P values for covariate
Segment linear acceleration (m/s^2)						
$\ddot{x}_{Head}$	4	-0.54 ± 0.75	-1.11 ± 1.08	1.20 ± 2.16	0.004	0.54
$\ddot{y}_{Head}$	22	0.63 ± 1.52	-0.33 ± 1.19	-2.92 ± 2.48	0.003	0.71
$\ddot{z}_{Head}$	18	-0.45 ± 1.17	-0.25 ± 1.58	-2.15 ± 3.77	0.006	0.97
$\ddot{x}_{L\_UpperArm}$	2	-0.44 ± 0.94	-0.70 ± 1.26	0.00 ± 1.61	0.005	0.22
$\ddot{z}_{L\_UpperArm}$	4	0.50 ± 1.24	-0.55 ± 1.55	-1.29 ± 2.45	0.005	0.97
$\ddot{x}_{R\_UpperArm}$	0	-0.69 ± 0.54	-0.78 ± 1.01	0.87 ± 2.89	0.000	0.05
$\ddot{z}_{R\_UpperArm}$	0	1.46 ± 0.95	1.32 ± 1.46	3.42 ± 2.66	0.000	0.02
$\ddot{y}_{L\_Forearm}$	28	0.31 ± 1.70	-0.11 ± 5.17	-0.67 ± 6.94	0.006	0.67
$\ddot{z}_{L\_Forearm}$	2	-2.34 ± 2.26	-4.40 ± 3.56	-4.44 ± 3.45	0.008	0.02
$\ddot{x}_{R\_Forearm}$	0	-0.49 ± 0.69	0.10 ± 2.37	1.76 ± 3.93	0.007	0.01
$\ddot{z}_{R\_Forearm}$	2	2.52 ± 1.67	3.68 ± 3.08	5.03 ± 3.91	0.003	0.00
$\ddot{x}_{UpperTrunk}$	28	-0.24 ± 1.33	-0.75 ± 1.07	-0.53 ± 1.38	0.002	0.24
$\ddot{z}_{UpperTrunk}$	4	2.17 ± 1.17	0.83 ± 1.59	1.03 ± 3.63	0.005	0.00
$\ddot{x}_{LowerTrunk}$	28	0.54 ± 0.68	0.45 ± 1.18	0.44 ± 3.25	0.000	0.16
$\ddot{y}_{LowerTrunk}$	22	0.74 ± 1.41	-0.55 ± 1.42	-3.03 ± 2.08	0.005	0.42
$\ddot{z}_{LowerTrunk}$	8	0.87 ± 0.76	0.56 ± 0.93	-0.17 ± 2.83	0.009	0.03
$\ddot{z}_{Pelvis}$	20	-1.35 ± 1.88	-1.31 ± 1.09	-0.74 ± 3.01	0.001	0.00
$\ddot{x}_{L\_Thigh}$	0	-0.18 ± 0.88	-0.02 ± 1.31	-1.34 ± 1.79	0.001	0.93

$\ddot{y}_{R\_Thigh}$	26	-0.23	±	0.86	-0.63	±	1.09	-0.30	±	1.72	0.003	0.02
$\ddot{x}_{L\_Shank}$	0	1.34	±	1.36	0.59	±	2.72	-1.90	±	3.91	0.003	0.08
$\ddot{z}_{R\_Shank}$	12	-4.86	±	3.87	-3.26	±	3.18	-2.83	±	3.26	0.004	0.00
$\ddot{x}_{L\_Foot}$	14	0.56	±	1.02	1.08	±	3.97	4.37	±	5.90	0.004	0.61
$\ddot{y}_{L\_Foot}$	0	1.93	±	2.11	1.21	±	1.29	1.59	±	0.78	0.005	0.01
$\ddot{z}_{L\_Foot}$	4	3.14	±	1.77	2.67	±	3.98	8.27	±	10.59	0.003	0.37
Segment linear velocity (m/s)												
$\dot{x}_{Head}$	12	-0.14	±	0.13	-0.22	±	0.18	0.06	±	0.32	0.009	0.77
$\dot{y}_{Head}$	10	-0.04	±	0.07	-0.02	±	0.08	-0.02	±	0.23	0.008	0.00
$\dot{z}_{Head}$	0	1.15	±	0.24	1.07	±	0.33	1.25	±	0.08	0.004	0.00
$\dot{x}_{L\_UpperArm}$	10	-0.10	±	0.12	-0.17	±	0.11	-0.02	±	0.22	0.008	0.02
$\dot{x}_{R\_UpperArm}$	0	-0.06	±	0.09	-0.10	±	0.12	0.03	±	0.22	0.007	0.31
$\dot{y}_{R\_UpperArm}$	6	-0.17	±	0.08	-0.16	±	0.10	-0.23	±	0.20	0.006	0.22
$\dot{z}_{R\_UpperArm}$	0	1.16	±	0.24	1.12	±	0.27	1.42	±	0.10	0.000	0.00
$\dot{y}_{L\_Forearm}$	14	-0.06	±	0.11	0.02	±	0.28	0.24	±	0.45	0.002	0.69
$\dot{z}_{L\_Forearm}$	20	0.79	±	0.19	0.54	±	0.35	0.64	±	0.85	0.003	0.00
$\dot{x}_{R\_Forearm}$	6	-0.04	±	0.10	-0.02	±	0.28	0.10	±	0.38	0.009	0.17
$\dot{z}_{R\_Forearm}$	4	1.15	±	0.27	1.08	±	0.31	1.46	±	0.17	0.003	0.00
$\dot{y}_{UpperTrunk}$	0	-0.17	±	0.08	-0.17	±	0.05	-0.25	±	0.09	0.005	0.03
$\dot{z}_{UpperTrunk}$	0	1.15	±	0.28	1.12	±	0.30	1.35	±	0.07	0.000	0.00
$\dot{x}_{LowerTrunk}$	12	-0.08	±	0.10	-0.14	±	0.11	0.01	±	0.21	0.008	0.04
$\dot{z}_{LowerTrunk}$	0	1.15	±	0.24	1.08	±	0.30	1.35	±	0.08	0.000	0.00
$\dot{x}_{Pelvis}$	16	-0.04	±	0.11	-0.04	±	0.09	-0.17	±	0.16	0.009	0.03
$\dot{y}_{Pelvis}$	30	0.15	±	0.11	0.05	±	0.10	-0.10	±	0.22	0.003	0.00
$\dot{z}_{Pelvis}$	0	1.16	±	0.30	1.13	±	0.29	1.34	±	0.10	0.000	0.00
$\dot{x}_{L\_Thigh}$	0	0.02	±	0.08	0.02	±	0.08	-0.09	±	0.19	0.001	0.07
$\dot{y}_{L\_Thigh}$	0	-0.15	±	0.08	-0.13	±	0.05	-0.25	±	0.14	0.000	0.29

$\dot{z}_{L\_Thigh}$	0	1.05	±	0.29	1.05	±	0.32	1.18	±	0.07	0.006	0.00
$\dot{z}_{R\_Thigh}$	0	1.27	±	0.38	1.16	±	0.33	1.40	±	0.08	0.000	0.00
$\dot{x}_{L\_Shank}$	0	0.06	±	0.07	0.01	±	0.10	-0.17	±	0.40	0.000	0.18
$\dot{z}_{L\_Shank}$	24	2.17	±	0.43	1.71	±	0.83	2.31	±	0.29	0.008	0.00
$\dot{y}_{R\_Shank}$	18	-0.03	±	0.05	-0.05	±	0.04	-0.11	±	0.08	0.009	0.74
$\dot{z}_{R\_Shank}$	0	1.22	±	0.42	1.17	±	0.28	1.36	±	0.12	0.000	0.00
$\dot{x}_{L\_Foot}$	0	0.04	±	0.08	-0.03	±	0.15	-0.32	±	0.69	0.000	0.15
$\dot{y}_{L\_Foot}$	0	0.07	±	0.11	0.07	±	0.05	0.08	±	0.09	0.005	0.00
$\dot{z}_{L\_Foot}$	0	0.25	±	0.46	0.09	±	0.10	0.02	±	0.28	0.000	0.42
$\dot{z}_{R\_Foot}$	20	0.06	±	0.08	0.26	±	0.25	0.68	±	0.52	0.003	0.30
Joint Angular Acceleration (°/s^2)												
$X\ddot{\theta}_{Neck}$	16	-10.31	±	488.40	-126.84	±	743.95	-737.80	±	992.14	0.003	0.04
$Y\ddot{\theta}_{Neck}$	28	-290.42	±	256.60	-20.61	±	503.32	-915.90	±	1400.61	0.002	0.11
$X\ddot{\theta}_{L\_Shoulder}$	6	-113.02	±	431.00	-363.15	±	851.45	-99.90	±	1567.88	0.006	0.10
$Y\ddot{\theta}_{L\_Shoulder}$	30	-326.57	±	640.02	-238.09	±	1240.68	-92.36	±	3013.27	0.000	0.55
$Y\ddot{\theta}_{R\_Shoulder}$	0	-259.05	±	486.21	-643.65	±	846.59	1406.61	±	2276.74	0.001	0.13
$X\ddot{\theta}_{L\_Elbow}$	0	-351.34	±	401.56	-441.89	±	677.14	58.33	±	905.33	0.001	0.88
$Y\ddot{\theta}_{L\_Elbow}$	30	222.40	±	613.89	484.42	±	1414.92	-536.93	±	2991.45	0.007	0.13
$Z\ddot{\theta}_{L\_Elbow}$	10	-201.17	±	210.04	-115.50	±	566.61	-1012.79	±	757.90	0.007	0.13
$X\ddot{\theta}_{R\_Elbow}$	2	-91.76	±	299.69	-24.89	±	543.07	-400.46	±	779.13	0.009	0.38
$Y\ddot{\theta}_{R\_Elbow}$	26	-115.20	±	295.05	201.76	±	897.10	618.20	±	1146.09	0.000	0.42
$Z\ddot{\theta}_{R\_Elbow}$	4	2.26	±	116.65	32.90	±	284.55	-171.82	±	303.50	0.007	0.17
$Z\ddot{\theta}_{Thorax}$	14	-21.17	±	418.48	-42.41	±	520.61	116.34	±	1727.52	0.008	0.06
$Y\ddot{\theta}_{Pelvis}$	30	-108.63	±	387.49	31.46	±	696.31	639.79	±	1123.09	0.002	0.68
$Z\ddot{\theta}_{Pelvis}$	12	101.35	±	369.86	46.65	±	418.61	-497.49	±	1179.34	0.004	0.00

$Y\ddot{\theta}_{L\_Hip}$	4	369.64	±	572.21	262.83	±	753.83	1667.92	±	2091.28	0.000	0.26
$Z\ddot{\theta}_{L\_Hip}$	6	708.42	±	369.87	441.07	±	743.64	1427.81	±	1711.23	0.000	0.89
$X\ddot{\theta}_{R\_Hip}$	22	762.23	±	782.32	412.93	±	744.06	1084.21	±	1161.22	0.006	0.01
$Y\ddot{\theta}_{R\_Hip}$	14	408.43	±	1304.84	-660.14	±	1330.02	-1705.47	±	2791.50	0.006	0.88
$Z\ddot{\theta}_{R\_Hip}$	14	-104.67	±	864.74	36.75	±	711.16	875.88	±	1769.77	0.000	0.83
$X\ddot{\theta}_{L\_Knee}$	26	-1990.77	±	1219.95	-1509.20	±	1956.67	-2999.89	±	2423.72	0.001	0.00
$Y\ddot{\theta}_{L\_Knee}$	10	283.02	±	962.11	9.16	±	1052.75	-970.14	±	2460.71	0.009	0.71
$X\ddot{\theta}_{R\_Knee}$	0	3350.62	±	2014.98	4140.08	±	1303.62	4012.39	±	1821.48	0.007	0.09
$Y\ddot{\theta}_{R\_Knee}$	26	-252.51	±	946.80	-243.16	±	807.23	-589.18	±	2145.20	0.005	0.24
$Z\ddot{\theta}_{R\_Knee}$	16	173.12	±	712.55	-65.70	±	592.69	-564.91	±	1094.98	0.009	0.56
$X\ddot{\theta}_{R\_Ankle}$	30	481.84	±	359.18	-45.46	±	649.00	217.40	±	732.04	0.008	0.00
$Z\ddot{\theta}_{R\_Ankle}$	22	-332.70	±	496.56	-100.18	±	445.32	-151.03	±	828.19	0.004	0.70
Joint Angular Velocity (°/s)												
$X\dot{\theta}_{Neck}$	22	0.99	±	19.93	0.21	±	51.25	-49.20	±	79.39	0.008	0.38
$Y\dot{\theta}_{Neck}$	0	3.65	±	13.64	10.70	±	29.92	9.75	±	50.34	0.001	0.05
$Z\dot{\theta}_{Neck}$	14	-25.71	±	30.23	-11.24	±	26.01	-16.50	±	54.21	0.005	0.15
$Z\dot{\theta}_{L\_Shoulder}$	16	-5.08	±	19.63	10.68	±	40.44	49.76	±	65.27	0.008	0.59
$Y\dot{\theta}_{R\_Shoulder}$	12	-0.01	±	25.14	-10.18	±	64.60	95.38	±	119.34	0.009	0.61
$X\dot{\theta}_{L\_Elbow}$	8	-24.93	±	34.49	4.67	±	48.82	10.64	±	91.96	0.001	0.87
$Y\dot{\theta}_{L\_Elbow}$	8	14.53	±	28.56	25.51	±	39.88	-34.20	±	81.47	0.009	0.03
$Z\dot{\theta}_{L\_Elbow}$	18	-21.78	±	14.65	-8.47	±	31.05	-58.25	±	52.31	0.003	0.08
$X\dot{\theta}_{R\_Elbow}$	8	-22.40	±	21.17	-18.30	±	31.86	-47.26	±	32.38	0.003	0.00
$Z\dot{\theta}_{Thorax}$	18	27.66	±	17.88	12.75	±	24.00	6.04	±	48.10	0.003	0.70
$Z\dot{\theta}_{Pelvis}$	18	-8.08	±	25.22	0.34	±	22.21	-20.76	±	55.76	0.003	0.02
$Y\dot{\theta}_{L\_Hip}$	0	-4.26	±	25.80	13.66	±	37.70	-70.61	±	101.20	0.000	0.03
$Z\dot{\theta}_{L\_Hip}$	0	9.81	±	19.42	-9.56	±	25.79	-43.08	±	88.24	0.000	0.00

$X\dot{\theta}_{R\_Hip}$	6	13.95	±	33.38	8.95	±	25.68	-1.07	±	56.89	0.001	0.12
$Y\dot{\theta}_{R\_Hip}$	10	32.30	±	61.90	32.98	±	50.70	75.76	±	65.95	0.003	0.01
$Z\dot{\theta}_{R\_Hip}$	16	-21.14	±	30.06	-25.29	±	30.21	8.02	±	61.49	0.001	0.32
$X\dot{\theta}_{L\_Knee}$	0	64.87	±	70.12	77.06	±	54.44	89.41	±	44.83	0.002	0.04
$X\dot{\theta}_{R\_Knee}$	4	107.17	±	58.92	84.86	±	49.98	124.61	±	63.45	0.001	0.00
$Z\dot{\theta}_{R\_Knee}$	12	1.60	±	27.38	-1.00	±	22.53	15.63	±	27.98	0.007	0.08
$X\dot{\theta}_{L\_Ankle}$	0	-18.11	±	33.31	-28.84	±	35.50	12.62	±	64.24	0.004	0.12
$X\dot{\theta}_{R\_Ankle}$	20	-59.93	±	49.34	-42.69	±	35.20	-49.97	±	64.15	0.000	0.01
Ankle Marker Linear Kinematics (m/s)												
$\ddot{A}_{L\_x}$	0	0.70	±	1.05	0.01	±	2.02	-3.26	±	6.21	0.000	0.08
$\dot{A}_{L\_x}$	0	0.02	±	0.05	-0.09	±	0.18	-0.17	±	0.37	0.000	0.25
$\dot{A}_{L\_y}$	0	0.31	±	0.28	0.31	±	0.22	0.30	±	0.26	0.009	0.00
$\dot{A}_{L\_z}$	0	0.22	±	0.51	0.10	±	0.09	0.26	±	0.23	0.000	0.84
A_{L_y}	22	-0.69	±	0.06	-0.72	±	0.09	-0.71	±	0.09	0.009	0.02
$\ddot{A}_{R\_x}$	12	-0.87	±	0.79	-2.88	±	3.41	-3.41	±	4.00	0.003	0.77
$\ddot{A}_{R\_z}$	0	-30.42	±	12.02	-34.10	±	10.32	-34.51	±	4.03	0.008	0.00
$\dot{A}_{R\_x}$	30	0.00	±	0.01	-0.37	±	0.33	-0.90	±	0.61	0.009	0.19
$\dot{A}_{R\_z}$	20	0.02	±	0.02	0.25	±	0.25	0.70	±	0.59	0.002	0.23
A_{R_z}	0	-0.45	±	0.72	0.36	±	0.40	0.21	±	0.33	0.000	0.00
Segment Rotational Acceleration (°/s^2)												
$\ddot{\alpha}_{Head}$	10	78.79	±	196.99	83.96	±	273.60	318.94	±	529.39	0.008	0.42

$\ddot{\beta}_{Head}$	14	-86.35	±	157.90	-48.18	±	147.73	-315.68	±	479.88	0.000	0.33
$\ddot{\alpha}_{L\_UpperArm}$	6	297.98	±	239.77	580.87	±	699.66	730.96	±	660.32	0.008	0.95
$\ddot{\beta}_{L\_UpperArm}$	2	381.00	±	794.70	375.47	±	674.01	-476.27	±	1340.48	0.009	0.91
$\ddot{\gamma}_{L\_UpperArm}$	10	-210.54	±	216.36	-207.06	±	560.32	-777.46	±	877.64	0.003	0.34
$\ddot{\alpha}_{R\_UpperArm}$	12	-222.62	±	179.48	-251.87	±	537.61	-735.34	±	947.02	0.005	0.26
$\ddot{\beta}_{R\_UpperArm}$	0	264.23	±	439.89	444.64	±	685.20	-1467.13	±	2148.44	0.000	0.02
$\ddot{\alpha}_{L\_Forearm}$	0	764.16	±	501.67	779.34	±	484.13	428.51	±	984.77	0.000	0.01
$\ddot{\beta}_{L\_Forearm}$	18	213.84	±	458.76	304.29	±	788.00	-672.85	±	1635.70	0.009	0.01
$\ddot{\alpha}_{R\_Forearm}$	12	-140.92	±	324.90	-276.98	±	610.44	-1208.82	±	1330.77	0.007	0.07
$\ddot{\beta}_{R\_Forearm}$	26	493.44	±	460.09	19.11	±	856.66	-407.82	±	2677.40	0.002	0.57
$\ddot{\gamma}_{UpperTrunk}$	26	117.77	±	278.88	-110.15	±	442.03	16.46	±	938.30	0.001	0.67
$\ddot{\alpha}_{LowerTrunk}$	12	-75.03	±	341.87	24.27	±	315.77	-9.57	±	640.57	0.003	0.84
$\ddot{\beta}_{LowerTrunk}$	0	-61.04	±	140.01	-69.61	±	220.95	-198.57	±	308.45	0.004	0.41
$\ddot{\gamma}_{LowerTrunk}$	2	24.89	±	88.04	-0.95	±	196.77	-158.65	±	337.94	0.005	0.35
$\ddot{\gamma}_{Pelvis}$	12	87.26	±	348.48	-35.03	±	348.51	-569.90	±	857.50	0.006	0.01
$\ddot{\alpha}_{L\_Thigh}$	0	-602.68	±	484.25	-472.02	±	445.43	-406.57	±	261.59	0.006	0.00
$\ddot{\beta}_{L\_Thigh}$	4	329.64	±	630.20	-28.23	±	726.19	1555.87	±	2505.11	0.000	0.22
$\ddot{\gamma}_{L\_Thigh}$	0	281.11	±	191.51	205.30	±	559.20	-153.41	±	853.44	0.004	0.12
$\ddot{\alpha}_{R\_Thigh}$	2	454.42	±	866.87	176.65	±	605.39	172.35	±	1337.00	0.006	0.00
$\ddot{\beta}_{R\_Thigh}$	16	-851.87	±	780.23	-770.95	±	1466.77	-1667.14	±	2545.04	0.001	0.94
$\ddot{\gamma}_{R\_Thigh}$	14	31.38	±	869.51	31.22	±	672.13	438.79	±	887.29	0.009	0.11
$\ddot{\alpha}_{L\_Shank}$	26	-2014.64	±	1036.29	-1767.12	±	1729.79	-3184.56	±	1934.78	0.004	0.00
$\ddot{\beta}_{L\_Shank}$	8	694.04	±	480.52	323.85	±	2027.10	1657.14	±	3356.47	0.004	0.49
$\ddot{\gamma}_{L\_Shank}$	4	342.96	±	348.45	148.29	±	416.74	968.74	±	1367.46	0.000	0.10
$\ddot{\alpha}_{R\_Shank}$	0	3528.04	±	1454.57	4221.41	±	1033.32	4161.21	±	994.63	0.003	0.00
$\ddot{\beta}_{R\_Shank}$	4	1679.69	±	1346.41	1197.26	±	1599.52	1877.42	±	2435.46	0.001	0.88
$\ddot{\gamma}_{R\_Shank}$	6	-447.77	±	513.53	-357.24	±	412.59	-607.13	±	651.00	0.003	0.67

$\ddot{\beta}_{L\_Foot}$	4	256.95	±	256.15	20.65	±	1162.00	1921.62	±	3788.66	0.000	0.03
$\ddot{\beta}_{R\_Foot}$	2	911.40	±	980.73	413.90	±	1436.88	2755.92	±	2886.53	0.001	0.04
$\ddot{\gamma}_{R\_Foot}$	24	-83.78	±	319.34	-105.56	±	290.15	-389.97	±	421.63	0.008	0.53
Segment Rotational Velocity (°/s)			±			±			±			
$\dot{\beta}_{Head}$	12	8.25	±	11.46	6.34	±	10.85	-5.58	±	12.36	0.009	0.10
$\dot{\beta}_{L\_UpperArm}$	4	23.45	±	31.35	28.66	±	48.01	-36.13	±	55.49	0.005	0.00
$\dot{\gamma}_{L\_UpperArm}$	8	0.28	±	15.87	1.93	±	36.96	-39.15	±	70.53	0.006	0.37
$\dot{\alpha}_{R\_UpperArm}$	18	-17.81	±	22.36	-21.16	±	44.92	-52.18	±	67.37	0.006	0.34
$\dot{\beta}_{R\_UpperArm}$	4	8.67	±	17.26	17.49	±	46.06	-69.26	±	94.22	0.004	0.20
$\dot{\alpha}_{L\_Forearm}$	12	73.58	±	49.58	31.24	±	67.33	32.64	±	101.28	0.008	0.85
$\dot{\beta}_{L\_Forearm}$	18	47.33	±	34.10	38.22	±	64.63	-19.05	±	101.15	0.008	0.19
$\dot{\alpha}_{R\_Forearm}$	26	-31.33	±	36.40	-64.24	±	61.99	-144.70	±	115.80	0.003	0.14
$\dot{\beta}_{R\_Forearm}$	10	1.53	±	22.05	29.56	±	44.16	-90.93	±	126.31	0.004	0.49
$\dot{\alpha}_{UpperTrunk}$	24	4.63	±	11.68	13.73	±	22.65	-22.62	±	48.02	0.003	0.26
$\dot{\beta}_{UpperTrunk}$	0	8.30	±	10.32	14.87	±	28.59	7.93	±	52.95	0.001	0.00
$\dot{\gamma}_{UpperTrunk}$	14	-18.46	±	19.31	-0.98	±	18.69	-17.20	±	46.72	0.007	0.35
$\dot{\beta}_{LowerTrunk}$	0	4.75	±	11.23	15.85	±	25.54	-16.98	±	41.97	0.000	0.00
$\dot{\gamma}_{LowerTrunk}$	12	15.36	±	10.92	13.20	±	19.80	-15.78	±	33.19	0.008	0.13
$\dot{\alpha}_{Pelvis}$	14	-16.44	±	26.36	0.79	±	22.04	-13.98	±	31.94	0.007	0.00
$\dot{\beta}_{Pelvis}$	0	21.27	±	15.44	20.66	±	29.25	3.67	±	69.82	0.003	0.38
$\dot{\gamma}_{Pelvis}$	14	-1.41	±	22.42	9.98	±	20.00	-34.59	±	54.28	0.004	0.36
$\dot{\alpha}_{L\_Thigh}$	18	-112.06	±	56.17	-69.87	±	55.32	-120.74	±	27.88	0.007	0.00
$\dot{\beta}_{L\_Thigh}$	0	17.01	±	25.60	34.33	±	49.44	-66.94	±	165.65	0.000	0.28
$\dot{\gamma}_{L\_Thigh}$	0	14.95	±	13.22	5.83	±	26.14	-35.56	±	96.80	0.000	0.03
$\dot{\alpha}_{R\_Thigh}$	4	-1.79	±	34.07	3.78	±	26.38	-9.69	±	45.48	0.009	0.38

$\dot{\gamma}_{R\_Thigh}$	12	-21.76	±	31.93	-15.31	±	22.70	-41.39	±	35.05	0.003	0.31
$\dot{\alpha}_{L\_Shank}$	0	93.12	±	81.54	107.26	±	37.35	116.50	±	13.39	0.000	0.00
$\dot{\beta}_{L\_Shank}$	0	7.34	±	19.98	19.99	±	60.00	-81.58	±	172.09	0.000	0.01
$\dot{\gamma}_{LShank}$	0	1.66	±	9.94	-2.83	±	26.46	-49.03	±	95.96	0.000	0.08
$\dot{\alpha}_{R\_Shank}$	6	115.92	±	38.35	112.82	±	41.23	132.55	±	35.14	0.002	0.00
$\dot{\gamma}_{R\_Shank}$	14	-16.60	±	12.58	-15.72	±	21.43	-21.81	±	21.43	0.006	0.54
$\dot{\alpha}_{L\_Foot}$	0	75.01	±	58.10	78.43	±	60.46	129.12	±	73.83	0.000	0.00
$\dot{\beta}_{L\_Foot}$	0	1.88	±	16.67	19.56	±	76.65	-100.48	±	192.83	0.000	0.01
$\dot{\gamma}_{L\_Foot}$	0	-40.71	±	28.96	-29.06	±	31.73	-58.29	±	58.60	0.000	0.04
$\dot{\gamma}_{R\_Foot}$	18	7.64	±	17.61	3.42	±	12.68	1.43	±	40.94	0.001	0.96

Notes: 'Point of % into stance' indicates the earliest moment when significant difference among 'activities' was found in the early stance phase.

Subscript R and L refer to right and left sides respectively.

Table 3. Summary of *post hoc* multiple comparisons

Dependant variable	% into stance	Normal walking vs. successful recovery		Normal walking vs. falls		Successful recovery vs. falls	
		Lower bound	Upper bound	Lower bound	Upper bound	Lower bound	Upper bound
Segment linear acceleration (m/s^2)							
$\ddot{z}_{Head}$	18	-12.64	5.67	-24.33	-0.51	-17.42	-0.45
$\ddot{y}_{R\_Thigh}$	26	-2.60	8.60	1.16	15.72	0.25	10.63
$\ddot{z}_{R\_Shank}$	12	-24.11	7.87	-44.07	-2.45	-29.96	-0.32
Segment linear velocity (m/s)							
$\dot{y}_{R\_UpperArm}$	6	-0.77	0.35	-1.49	-0.03	-1.07	-0.03
$\dot{z}_{R\_UpperArm}$	0	-0.09	1.35	0.55	2.42	0.19	1.52
$\dot{z}_{R\_UpperTrunk}$	0	0.13	1.42	0.57	2.25	0.04	1.23
$\dot{z}_{LowerTrunk}$	0	0.01	1.12	0.55	2.00	0.20	1.22
$\dot{y}_{L\_Thigh}$	0	-0.63	0.07	-1.15	-0.25	-0.74	-0.10
$\dot{x}_{L\_Shank}$	0	-0.24	1.05	-1.72	-0.04	-1.89	-0.69
$\dot{x}_{L\_Foot}$	0	-0.46	1.49	-3.05	-0.51	-3.20	-1.39
Joint angular acceleration (°/s^2)							
$Y\ddot{\theta}_{L\_Hip}$	4	-4253.48	4831.71	26.14	11845.96	1437.32	9856.54
$Z\ddot{\theta}_{L\_Hip}$	6	-3320.81	4813.97	254.92	10838.26	1030.77	8569.25
Joint angular velocity (°/s)							
$Z\dot{\theta}_{Neck}$	14	-202.04	104.88	-401.73	-2.42	-295.71	-11.28
$Z\dot{\theta}_{Pelvis}$	18	-182.11	103.97	-373.10	-0.92	-280.49	-15.38
$X\dot{\theta}_{R\_Hip}$	6	-183.80	138.52	-423.44	-4.10	-340.48	-41.78
$X\dot{\theta}_{R\_Knee}$	4	-90.87	412.85	98.22	753.55	31.50	498.29
Segment rotational acceleration (°/s^2)							
$\ddot{\beta}_{Head}$	14	-1320.76	557.71	-3211.49	-767.59	-2478.40	-737.62

$\ddot{\beta}_{R\_UpperArm}$	0	-3582.19	4937.11	-11132.85	-49.25	-10215.91	-2321.10
Segment rotational velocity (°/s)							
$\dot{\gamma}_{L\_Thigh}$	0	-125.17	212.70	-445.61	-6.03	-426.14	-113.03
$\dot{\gamma}_{L\_Shank}$	0	-196.89	119.01	-535.06	-124.08	-437.00	-144.25
$\dot{\alpha}_{R\_Shank}$	6	-66.36	276.91	49.61	496.21	8.58	326.69
$\dot{\gamma}_{L\_Foot}$	0	-279.28	48.55	-521.08	-94.57	-344.36	-40.55

Notes: Lower bound and upper bound define the 99% confident interval for the mean difference of the measure between the two activities. Significant difference is determined if the 99% confidence interval does not contain zero. The point of % into stance indicates the earliest moment when significant difference between falls and non-fall activities was found in the early stance phase. Subscript R and L refer to right and left sides respectively.

Table 4. Effect sizes for the comparisons of falls versus recovery and falls versus normal walking

Dependant variables	% into stance	Normal walking vs. successful recovery	Normal walking vs. falls	Successful recovery vs. falls
Segment linear velocity (m/s)				
$\dot{y}_{L\_Thigh}$	28	0.61*	1.11**	0.57*
$\dot{x}_{L\_Shank}$	22	0.83**	1.37**	0.80**
$\dot{x}_{L\_Foot}$	0	0.53	0.89**	0.85**
Joint angular acceleration (°/s^2)				
$Y\ddot{\theta}_{L\_Hip}$	16	0.67*	1.61**	1.04**

**indicates large effect size (0.80), * indicates medium effect size (0.5). Subscript R and L refer to right and left sides respectively.

Figure 1. Cross-slope set up; wooden walkway covered in non-slip rubber and a 1.5m section of vinyl.

Figure 2. Mean kinematic profiles for dependant variables with large effect sizes for successful recovery vs. falls averaged over the entire sample. The time during which kinematic variables were significantly different for successful recovery versus falls is highlighted in grey.