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Hip joint contact loads in older adults during recovery from forward loss of balance by stepping

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Abstract

Hip joint contact loads during activities of daily living are not generally considered high enough to cause acute bone or joint injury. However there is some evidence that hip joint loads may be higher in stumble recovery from loss of balance. A common laboratory method used to evaluate balance recovery performance involves suddenly releasing participants from various static forward lean magnitudes (perturbation intensities). Prior studies have shown that when released from the same perturbation intensity, some older adults are able to recover with a single step, whereas others require multiple steps. The main purpose of this study was to use a musculoskeletal model to determine the effect of three balance perturbation intensities and the use of single versus multiple recovery steps on hip joint contact loads during recovery from forward loss of balance in community dwelling older adults (n = 76). We also evaluated the association of peak hip contact loads with perturbation intensity, step length and trunk flexion angle at foot contact at each participant's Maximum Recoverable Lean Angle (MRLA). Peak hip joint contact loads were computed using muscle force estimates obtained using Static Optimisation and increased as lean magnitude was increased and were on average 32% higher for Single Steppers compared to Multiple Steppers. At the MRLA, peak hip contact loads ranged from 4.3-12.7 body weights and multiple linear stepwise regression further revealed that initial lean angle, step length and trunk angle at foot contact together explained 27% of the total variance in hip joint contact load. Overall findings indicated that older adults experience peak hip joint contact loads during maximal balance recovery by stepping that in some cases exceeded loads reported to cause mechanical failure of cadaver femurs. While step length and trunk flexion angle are strong predictors of step recovery performance they are at best moderate predictors of peak hip joint loading.

Abstract length = 306 words

Introduction

Contact loads in the hip joint during normal walking are reported to be in the vicinity of 2-4 times body weight (Bergmann et al. 2001; Bergmann et al. 1993). These loads are well below the mechanical failure load of 5.5 to 14 body weights reported by Schileo et al. (2014) who tested the load bearing capacity of femurs from older adults in conditions that approximated the stance phase of gait. However Viceconti et al. (2012) demonstrated via use of a musculoskeletal modelling approach that a combination of sub-optimal neuromuscular control and severe osteoporosis may make spontaneous fracture during walking feasible, and thereby explain the small proportion of femoral fractures that occur in the apparent absence of high-energy trauma that may occur due to a fall. It therefore follows that motor tasks where larger impulsive loads than those associated with gait are applied, could produce hip loads that are in the range associated with failure, perhaps even in the absence of degraded neuromuscular control and severe osteoporosis. One such motor task where high joint contact loads are experienced is the stumbling response used to recover balance from a trip perturbation. Bergmann et al. (1993) reported peak hip contact loads as high as 8.7 body weights in patients fitted with an instrumented hip replacement during a stumble recovery from an unexpected trip perturbation experienced during walking. At present however the magnitude of hip joint contact loads during maximal balance recovery by stepping, and the extent to which these forces are affected by the balance perturbation intensity and motor control strategy used during balance recovery by stepping remain unknown. Such information would inform efforts to understand the mechanical risk factors associated with femoral fracture and implant loosening and help identify ways by which hip contact loads experienced during balance recovery by stepping may be reduced.

A common method used to evaluate balance recovery performance involves suddenly releasing participants from various static forward lean magnitudes (perturbation intensities).

Carty et al. (2015) reported that older adults are significantly less likely to experience a real world fall if they are able to recover from a large forward perturbation intensity or use a single versus a multiple step recovery strategy when released from a set perturbation intensity. Recovery from a large perturbation intensity and recovery using a single recovery step are strongly associated with the ability to resist forward trunk flexion during the stepping response (Barrett et al. 2012; Grabiner et al. 2008; Owings et al. 2001), the ability to take a suitably long recovery step (Graham et al. 2015; Karamanidis et al. 2008; Schillings et al. 2005) and the ability to produce adequate hip and knee joint powers in the stepping limb (Carty et al. 2012b; Graham et al. 2015; Madigan 2006). Recovery step length, trunk angle at touchdown of the stepping limb and lower limb joint moments and powers during recovery from forward loss of balance are all reported to increase with balance perturbation intensity (Carty et al. 2012b; Madigan et al. 2005) and would therefore be expected to result in a corresponding increase in lower extremity muscle force and hence joint contact loads for larger balance perturbations. Poor trunk control in particular has been shown to result in more co-contraction of spine, hip and knee muscles during the stepping phase of balance recovery from an equivalent balance perturbation and might therefore be considered an example of inefficient coordination that adversely affects balance recovery (Graham et al. 2014). However the effect of single versus multiple step recovery on hip joint contact loads remains unknown.

The purposes of this study were to (1) determine the effect of balance perturbation intensity on peak hip contact loads during balance recovery using the single step balance recovery strategy, (2) compare the effect of single versus multiple step balance recovery strategy on peak hip contact loads during balance recovery from the same perturbation intensity, and (3) evaluate the association of peak hip contact loads with perturbation intensity, step length and trunk flexion angle at foot contact at each participant's maximum recoverable lean angle

(MRLA). We hypothesised that hip loads would be greater at higher balance perturbation intensities and during the single compared to multiple step balance recovery strategy, and that step length, MRLA, and trunk flexion angle at foot contact would be associated with peak hip contact loads.

Methods

Participants

Participants consisted of a sub-sample of one hundred and six community dwelling older adults (Age: 72.0 ± 4.8 years; Height: 1.67 ± 0.09 m, Mass: 75.4 ± 12.5 kg) from a larger prospective study (Carty et al. 2015), which were recruited at random via letters sent to 5000 residents aged 65 to 80 years that were registered on the local electoral roll. Individuals previously diagnosed with neurological, metabolic, cardio-pulmonary, musculoskeletal and/or uncorrected visual impairment were excluded. Ethics approval was obtained from the Institutional Human Research Ethics Committee and all relevant ethics guidelines including provision of informed consent were followed.

Experimental procedures

The balance recovery protocol was undertaken as reported previously by Carty et al. (2011) and is only described here briefly, a detailed description of this procedure is provided in Appendix 1. Participants were positioned in a forward lean posture with lean perturbation measured in body weights (BW) recorded on a load cell (S1W1kN, XTRAN, Australia) placed in series with an inextensible cable. The cable was attached to a safety harness at the level of their sacrum and cable length was adjusted until the required force was achieved. The cable was released at a random time interval (2-10 s) following achievement of the prescribed posture and cable force (± 1% BW) through the disengagement of an electromagnet located in-series with the cable. A second instrumented cable, which attached the safety harness to the ceiling, was used to prevent participants from contacting the ground in the event of a failed recovery. Centre of pressure location, displayed in real time on a computer monitor, was visually inspected to ensure anticipatory actions were not evident prior to cable release.

Following familiarisation, participants performed 4 trials at each of the 15% BW, 20% BW and 25% BW perturbation intensities in randomised order. For each trial, participants were classified as adopting either a single or multiple step balance recovery strategy using previously published criteria (Carty et al. 2011). Single and Multiple Steppers were then participants who exclusively recovered a single or multiple recovery steps respectively at each of the 3 perturbation intensities investigated. The MRLA was determined by systematically increasing perturbation intensity by ~1% BW increments from the last intensity recovered from with a single step until the participant could no longer recover with a single step. The final trial at which the participant was able to recover using a single step was taken to represent their MRLA. Trajectories of 51 reflective markers attached to each participant (Barrett et al. 2012) and Ground Reaction Forces under each foot were recorded simultaneously. For analytical purposes the length of each trial was the period from toe off of the stepping foot (TO) to the maximum knee joint angle made by the stepping leg following foot contact (KJM).

Computation of hip joint contact loads

Data analyses were performed using OpenSim (version 3.2) (Delp et al. 2007) in conjunction with custom Matlab scripts (Version 2014b, The Maths Works, USA). The model described by Hamner et al. (2010) including 17 bodies (head, torso, pelvis, and bilateral humerus, radius, ulna, hand, femur, tibia, foot) with 17 joints and 36 degrees of freedom (pelvis: 6, neck: 3, lumbar joints: 3, hip: 3, shoulder joints: 3, wrist: 2, elbow: 1, radioulnar: 1, knee: 1, ankle: 1) was used as the initial generic scalable model. 92 hill-type muscle actuators were used to actuate the lumbar and lower extremity joints while the arms were driven by torque actuators. The mass of the harness worn during balance recovery trials was added to the model as a component of the total mass of the participant. A wrap object was embedded in the generic model as previously reported (Graham et al. 2014) that matched erector spinae muscle moment arms during trunk flexion (Daggfeldt et al. 2003). Model Scaling and Inverse Kinematic analyses (Lu et al. 1999) were performed by fitting the anatomical model to measured 3D marker positions with a high weighting on virtual markers attached to the pelvis an those which defined the joint centre of the hip, knee and ankle. Joint centres were estimated from experimental marker trajectories: the regression equations of Harrington et al. (2007) were used for the hip joint (as suggested by Kainz et al. (2015)), while the knee and ankle joint centres were identified as the midpoints of the femoral condyles and the medial and lateral malleoli respectively. Residual Reduction Analysis (RRA) was subsequently performed to improve the dynamic consistency between measured ground reaction forces and the mass-acceleration product of the model (Delp et al. 2007). The Static Optimisation tool in OpenSim was used to calculate muscle forces using a cost function to minimise the sum of squared muscle activations within the force-length-velocity constraints of each muscle. Joint contact loads were computed using the Joint Reaction analysis available in OpenSim, which calculates contact loads through a recursive procedure equivalent to resolving the free body diagrams of the rigid bodies included in the model, starting from the most distal and moving proximally (a detailed description of the tool implementation can be found in Steele et al. (2012)).

Model evaluation

Models were evaluated according to the recommendations of Hicks et al. (2015) to ensure that possible sources of error were minimised to within recommended tolerances. Participant data were excluded from further analysis if the pelvis from the generic model was scaled in depth or width in excess of two standard deviations from the mean value of the average male or female geometry reported by Reynolds et al. (1982). Marker tracking errors, the influence of RRA on joint kinematics, trunk COM and residual forces and moments, and agreement between model activations and measured EMG activity of key muscles were then evaluated across all simulations. Hip joint contact load estimates of Multiple Steppers at the 20% BW perturbation intensity were compared to hip contact loads associated with stumbling during level walking and stumbling during stair climbing measured using an instrumented hip prostheses (Bergmann et al. 2004). Additionally, we compared the hip joint contact load estimates during the stance phase of walking for 10 older adults with the direct measurements made using an instrumented hip prosthesis (Bergmann et al. 2001) and indirect estimates from a computational modelling study of hip joint loading during gait (Giarmatzis et al. 2015).

Statistical Analysis

A repeated measures general linear model was used to assess the effect of the three perturbation intensities (15%, 20% and 25%BW) on each dependent measure (hip contact load, step length, trunk angle at foot contact). A priori contrasts were used to make comparisons between the successive perturbation intensities. A between factor general linear model was used to assess the effect of step strategy (Single Steppers versus Multiple Steppers) at the 20% BW perturbation intensity on each dependent measure. Pearson Product Moment Correlation Coefficients were used to examine the relations between hip joint contact loads experienced during the MRLA trial and the MRLA, step length normalised to participant leg length (leg length was defined as the distance between the hip and ankle joint centres) and trunk flexion angle at foot contact. These data were subsequently entered into a stepwise multiple regression model with entry and exit criteria of $p<0.05$ and $p>0.05$, respectively. All statistical analyses were performed using the Statistical Package for the Social Sciences (SPSS, Version 22, IBM, USA). Significance was accepted for $p<0.05$.

Results

Model evaluation

The evaluation of pelvic dimensions of each scaled model resulted in a reduction of the number of included participants from 106 to 76. Pelvic scaling factors for included participants were 1.10 ± 0.08, 1.18 ± 0.09 and 1.08 ± 0.08 respectively for width, depth and height. Mean pelvic width and depth of the scaled models were 0.29 ± 0.01m and 0.17 ± 0.02m respectively and were on average larger compared to the width (0.24 ± 0.04m) and depth (0.14 ± 0.03m) obtained from Reynolds et al. (1982). Data were normally distributed about the mean in both dimensions. Mean peak RMS errors for Scaling and Inverse Kinematics were 0.018 ± 0.005 m and 0.037 ± 0.028 m respectively. Mean residual pelvic forces and moments were all below 5% BW and 0.05 Nm/kg respectively (Supplementary Figure 1). Peak RMS errors between residual reduced kinematics and experimental kinematics were below 2.5° across all DOF in all simulations (Supplementary Figure 2). On average RRA modified the trunk COM location in the vertical, anterior/posterior and medial/lateral dimensions by 0.00 ± 0.04 m, 0.05 ± 0.03 m and 0.01 ± 0.03 m respectively. Qualitative agreement was also achieved between model activations and measured EMG activity of key muscles (Supplementary Figure 3).

The mean peak hip contact load for Multiple Steppers at the 20% BW perturbation intensity was within 10% of the peak load associated with stumbling during gait (Bergmann et al. 2004) and within 20% of the load associated with stumbling during stair climbing (Bergmann et al. 2004) (Figure 1a). During walking gait, early stance and late stance mean peak hip joint contact loads were 34% and 20% higher than those measured by Bergmann et al. (2001)

(Figure 1b) but were 23% and 47% BW lower than numerical estimates for young adults walking at a similar speed (Giarmatzis et al. 2015). Finally, passive muscle forces were checked for each simulation and found to be negligible (i.e. muscles tended to operate on the ascending limb and plateau region of the force-length relation).

<Insert Figure 1 about here>

Effect of balance perturbation intensity

A total of 20 participants were able to recover balance with a single step from each of 15, 20 and 25% BW perturbation intensities. Perturbation intensity had a significant main effect on normalised step length ($F = 26.7$, $p < 0.01$), trunk flexion angle at foot contact ($F = 13.2$, $p < 0.01$) and peak hip contact load ($F = 14.9$, $p < 0.01$). A priori-contrasts revealed that normalised step length, trunk flexion angle at foot contact and peak hip contact load were higher at the 20% BW compared to 15% BW condition and at the 25% BW compared to 20% BW condition (Table 1).

<Insert Table 1 about here>

Effect of step strategy

For the purpose of comparing the effects of step strategy on hip joint contact loads Single Steppers ($n = 20$) were compared to Multiple Steppers ($n = 18$) at the 20% BW perturbation intensity. Single Steppers compared to Multiple Steppers used a significantly higher normalised step length ($F = 7.3$, $p < 0.01$), trunk flexion angle at foot contact ($F = 4.2$, $p = 0.03$) and had higher peak hip contact loads ($F = 4.1$, $p = 0.01$) during recovery from the 20% BW perturbation intensity (Table 2).

<Insert Table 2 about here>

Relation between kinematic measures and hip contact loading at the maximal recoverable lean angle (MRLA)

At the mean MRLA the mean peak hip joint contact loads were approximately 9 times BW (Table 3) with the largest peak hip joint contact load experienced by an individual was 12.7 BW. Hip joint contact loads were significantly correlated to MRLA (r = 0.49) as well as trunk flexion angle at foot contact (r = 0.45) and step length (r = 0.41) ($p < 0.05$ for all correlations) (Figure 2). When all variables were entered into a stepwise multiple linear regression equation of the form $Y = A_1X_1 + A_2X_2 + A_3X_3 + A_4$, MRLA ($X_1$), normalised step length (X_2) and trunk flexion angle at foot contact (X_3) together accounted for 27% of the variance in hip contact load (Y) (SEE = 1.7). The corresponding regression coefficients (A_1–A_4) were: 0.185, 0.265, 0.153 and 2.789.

<Insert Figure 2 about here>

<Insert Table 3 about here>

Discussion

A musculoskeletal model was used in the present study to investigate the effect of perturbation intensity on peak hip joint contact loads during single-step balance recovery (i.e. same strategy-different intensity) and the effect of single versus multiple step balance recovery strategy on the peak hip joint contact loads during recovery at the same perturbation intensity (i.e. same intensity-different strategy). In support of our hypotheses, peak hip joint contact loads increased with each increase in balance perturbation intensity for older adults that were able to recover with a single step. Peak hip joint contact loads were also found to be higher for older adults that were able to recover with a single compared to multiple step balance recovery strategy when evaluated at the same perturbation intensity. Similar to previous studies step length and trunk flexion angle increased as the initial perturbation intensity was increased, and at the fixed perturbation intensity, Single Steppers took longer steps and used a more upright trunk posture than their Multiple Stepper counterparts. We also demonstrated that step length and trunk flexion angle at foot contact during maximal balance recovery performance explained additional variance in peak hip joint contact loads beyond that explained by perturbation intensity alone. Taken together these findings confirm that perturbation intensity and stepping strategy adopted are important determinants of peak hip contact loading experienced during balance recovery by stepping in older adults.

The peak hip joint contact loads during balance recovery at the 15, 20 and 25% BW perturbation intensities in the present study were 7.3 ± 1.7 BW, 8.4 ± 1.7 BW and 10.7 ± 1.0 BW. These values were respectively 3.2, 3.6 and 4.7 times higher than the peak contact load of 2.3 BW previously reported for slow walking on level ground (Bergmann et al. 2001), and 1.7, 2.0 and 2.5 times higher than the peak contact load of 4.3 BW previously reported for running at 9 km/hr (Bergmann et al. 1993). The peak hip contact load estimates from the present study were also within the range of 5.5-14 BW reported to cause mechanical failure of cadaver femurs (Schileo et al. 2014). The peak hip joint contact loads associated with the highest perturbation intensity in the present study were also in excess of the upper limit of around 9 BW reported by Martelli et al. (2011) to be feasible during walking in cases of severe neuromotor degradation, and according to Viceconti et al. (2012), capable of producing spontaneous hip fractures in the presence of severe osteoporosis of the hip and degraded neuromuscular function. Balance recovery could therefore be a motor control task that imposes risk of hip fracture in individuals, particularly following large balance perturbations in individuals with sub-optimal neuromuscular control and low bone mineral density.

Hip joint contact loads were on average 32% higher for older adults that were able to recover from the 20% BW perturbation intensity using a single step (8.4 ± 1.7 BW) compared to multiple step (6.5 ± 1.1 BW) recovery strategy, and were therefore slightly lower in the Multiple Stepper group compared to the peak hip contact load of 8.7 BW reported for stumbling by Bergmann et al (1993). Previous studies have suggested that a multiple step recovery is associated with an increased risk of experiencing a real world fall (Carty et al. 2015; Hilliard et al. 2008; Mille et al. 2013) and reflects underlying lower limb muscle weakness (Carty et al. 2012a) and concomitant lower limb muscle inhibition during balance recovery (Cronin et al. 2013). However the findings presented here may also suggest that older adults could also adopt a multiple step strategy, in part to protect the hip against large peak contact loads during balance recovery.

Peak hip contact loads ranging from 4.3 to 12.7 BW were generated during maximal recovery from forward loss of balance by stepping. While 24% of the variance in peak hip contact load following touchdown of the stepping leg was explained by perturbation intensity alone, a further 3% was explained by the addition of step length and trunk angle at foot contact to the regression model. Although step length and trunk angle at foot contact are strong predictors of balance recovery performance (Grabiner et al. 2008; Graham et al. 2015; Karamanidis et al. 2008; Schillings et al. 2005), they appear at best moderate predictors of hip joint contact load. The relatively low amount of total variance in hip joint contact load explained in the multiple regression model further reinforces the importance of subject-specific dynamic simulations, such as that used in the present study, for studying joint loading.

The results of this study should be considered with the following limitations in mind. First, hip joint contact loads have previously been shown to be sensitive to errors in pelvic scaling, which strongly influence the location of the hip joint centre location (Lenaerts et al. 2009; Martelli et al. 2015). Efforts were made in the present study to minimise these errors by excluding participants where pelvic scaling factors were large relative to pelvic geometries reported in the literature (Reynolds et al. 1982). The main reason for high scale factors in our excluded participants was associated with difficulties in representing pelvic geometry with skin mounted markers due to high centralised adiposity. While any remaining errors in pelvic scaling were unlikely to affect the mean hip contact loads reported here, a degree of caution is nevertheless warranted when interpreting values at the upper and lower bounds of the hip contact load distribution. Errors associated with Scaling, Inverse Kinematics and RRA were kept within recommended tolerances (Hicks et al. 2015) and residual pelvic forces and moments were also low. Second, consistent with computational studies aiming to estimate hip contact loads in activities of daily living (Giarmatzis et al. 2015; Modenese et al. 2012; Modenese et al. 2011), muscle forces were estimated using Static Optimisation with a cost function that minimised muscle activation squared (Crowninshield et al. 1981). Joint contact loads reported here are therefore unlikely to reflect sub-optimal neuromuscular control (Martelli et al. 2011; Modenese et al. 2013) including high levels of muscle co-contraction. While surface EMG from key muscles and modelled muscle activations were qualitatively similar, EMG amplitudes tended to be higher than the corresponding muscle activations immediately following foot contact, which likely reflects the inability of Static Optimisation to predict high levels of muscle co-contraction. Additionally, a rigid tendon was assumed within the Static Optimisation algorithm used in the present study. It has been demonstrated within the context of a Hill-type muscle model that model force estimates, particularly for muscles with long compliant tendons, can be sensitive to this assumption (Millard et al. 2013). The influence of the rigid tendon assumption within the current study is unknown and therefore requires further investigation. Third, surface EMG data from only one muscle that crossed the hip (Medial Hamstring) was collected and so the ability to compare measured and modelled hip muscle activations was limited. Fourth, direct validation of model predicted hip contact loads was not possible in the present study however model hip contact load predictions were found to be in relative agreement with hip joint loads measured using an instrumented hip prosthesis during a stumbling task (Bergmann et al. 2004) and walking (Bergmann et al. 2001) as well as hip contact loads during gait estimated using methods similar to those reported here (Giarmatzis et al. 2015). Finally, in future it will be of benefit to evaluate how the application of joint contact loads interact with the geometry and material properties of the proximal femur to more accurately determine the risk of femoral fracture during balance recovery by stepping.

Conclusion

Hip contact loads increased as a function of perturbation intensity and were higher during single versus multiple step recovery from the same perturbation intensity. The magnitude of peak hip joint loads during maximal recovery efforts experienced by some individuals exceeded the loads required to cause mechanical failure of older cadaver femurs. Single step balance recovery from large postural perturbations may therefore present a risk of fracture in some individuals, most notably those with severe osteoporosis. While step length and trunk flexion angle are strong predictors of step recovery performance, they are at best moderate predictors of peak hip joint loading during maximal recovery from forward loss of balance with a single step.

Conflict of Interest

The authors declare that they have no conflicting interests.

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Figure Captions

Figure 1 (A) Comparison of hip joint contact loads during balance recovery from the present study with similar data from Bergmann et al. (2004). (B) Comparison of hip joint contact loads from ten healthy older adults during the stance phase of walking at 1.00 ± 0.01 m.s^{-1} from the present study with similar data from Bergmann et al. (2001) from 4 older adults walking at 1.09 ± 0.01 m.s^{-1} recorded using an instrumented prosthesis.

Figure 2. Scatterplots showing the relationships between peak hip joint contact load during the MRLA trial and (A) the maximum recoverable lean angle (MRLA), (B) trunk angle at foot contact and (C) step length/leg length. The regression line for each variable is plotted as a solid line accompanied by a dashed line representing the 95% confidence limits.

Supplementary Figure 1. Pelvic residual forces, moments and reserve actuator moments for a representative participant during balance recovery from toe off (TO) to the maximum knee joint flexion angle following foot contact (KJM).

Supplementary Figure 2. Comparative pelvic segment angles and lower limb joint angles from the stepping side leg of a representative participant during balance recovery from toe off (TO) to the maximum knee joint flexion angle following foot contact (KJM) for Inverse Kinematics and the Residual Reduction Analysis.

Supplementary Figure 3. Simulated muscle activations and EMG for key lower limb muscles across all 76 participants at the maximal recoverable lean angle from toe off of the stepping foot (TO) to knee joint maximum (KJM) following foot contact. Surface EMG activity was recorded using bipolar surface electrodes (Duo-trode, Myotronics Inc., Australia) positioned along muscle fibre direction at an inter-electrode distance of 2 cm. Data were collected telemetrically (Aurion ZeroWire; Milano, Italy) from 5 muscles of each leg: vastus medialis, biceps femoris, semitendinosus, gastrocnemius, and soleus at 1 kHz. Raw EMG signals were root mean square integrated and lowpass filtered at 10 Hz. EMG is normalised to the maximum amplitude measured during recovery and is presented in grey representing ± 1SD of the overall mean. Mean model activations are represented by the bold black line with dashed lines indicating ± 1SD.

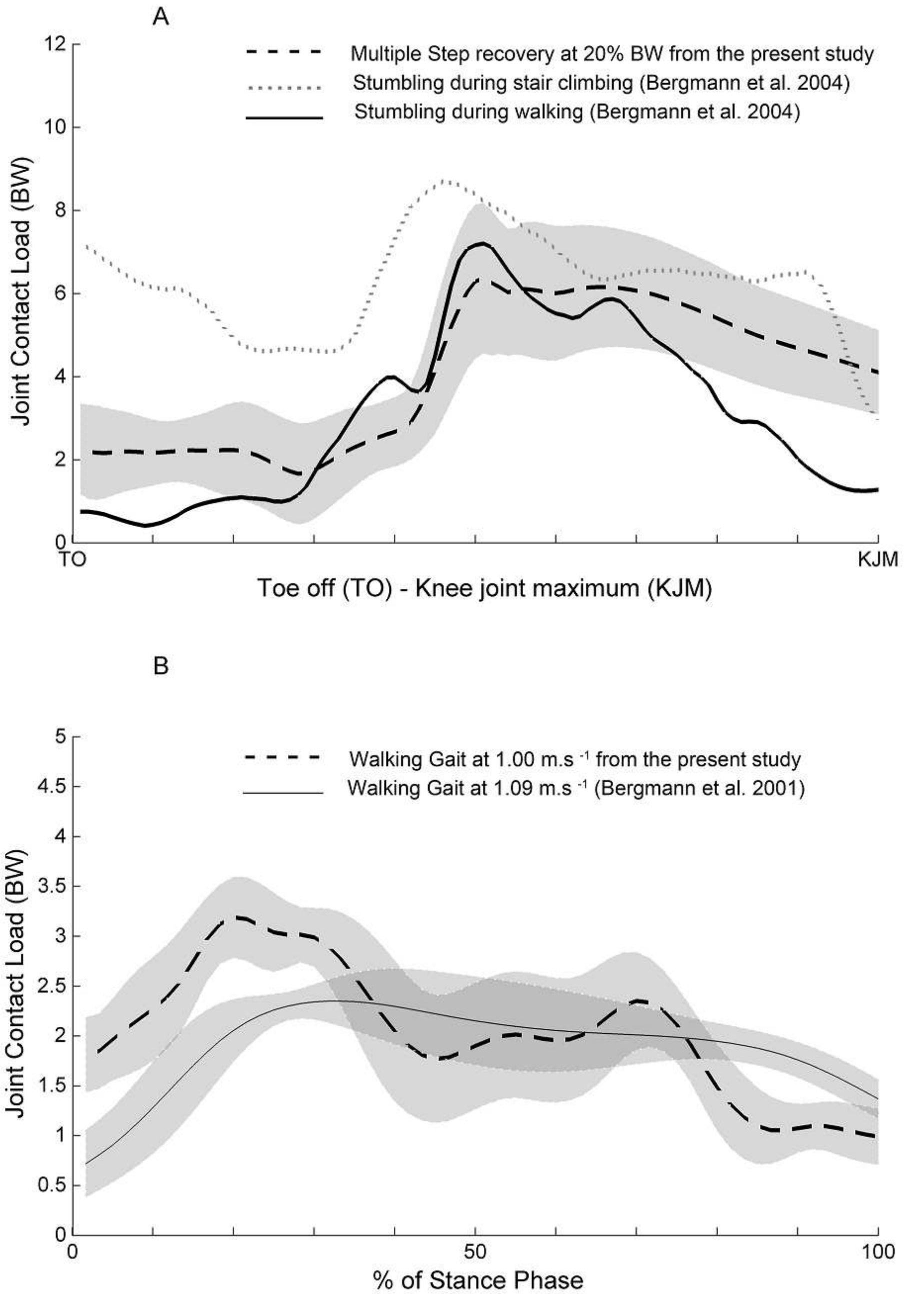

A
Multiple Step recovery at 20% BW from the present study
Stumbling during stair climbing (Bergmann et al. 2004)
Stumbling during walking (Bergmann et al. 2004)
Joint Contact Load (BW)
TO
KJM
Toe off (TO) - Knee joint maximum (KJM)
B
Walking Gait at 1.00 m.s -1 from the present study
Walking Gait at 1.09 m.s -1 (Bergmann et al. 2001)
Joint Contact Load (BW)
% of Stance Phase

Figure 1

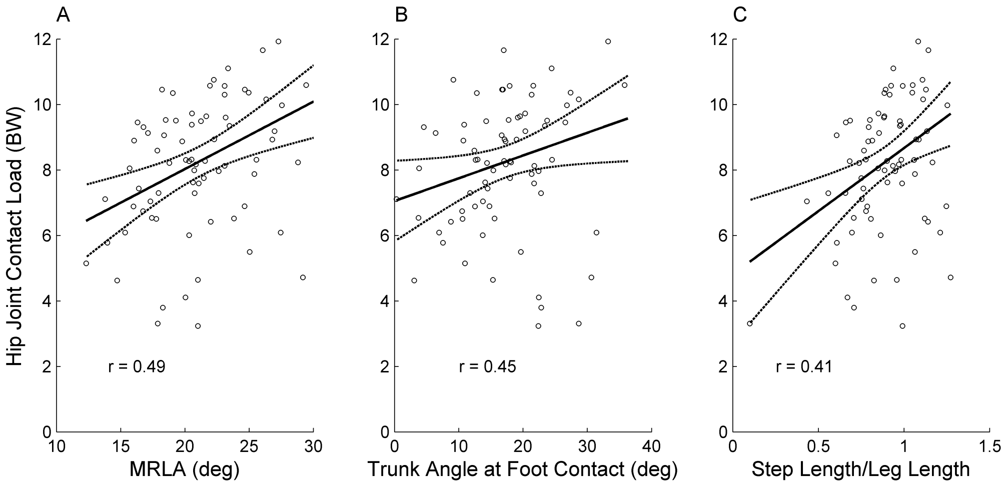
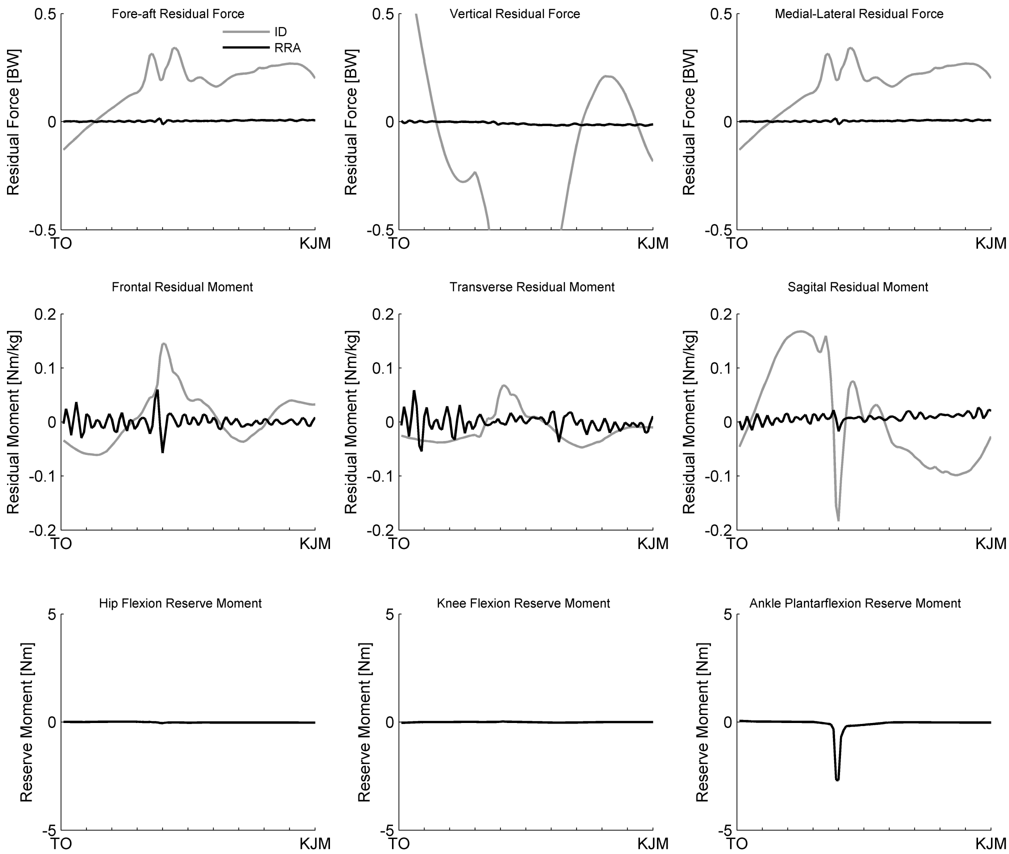
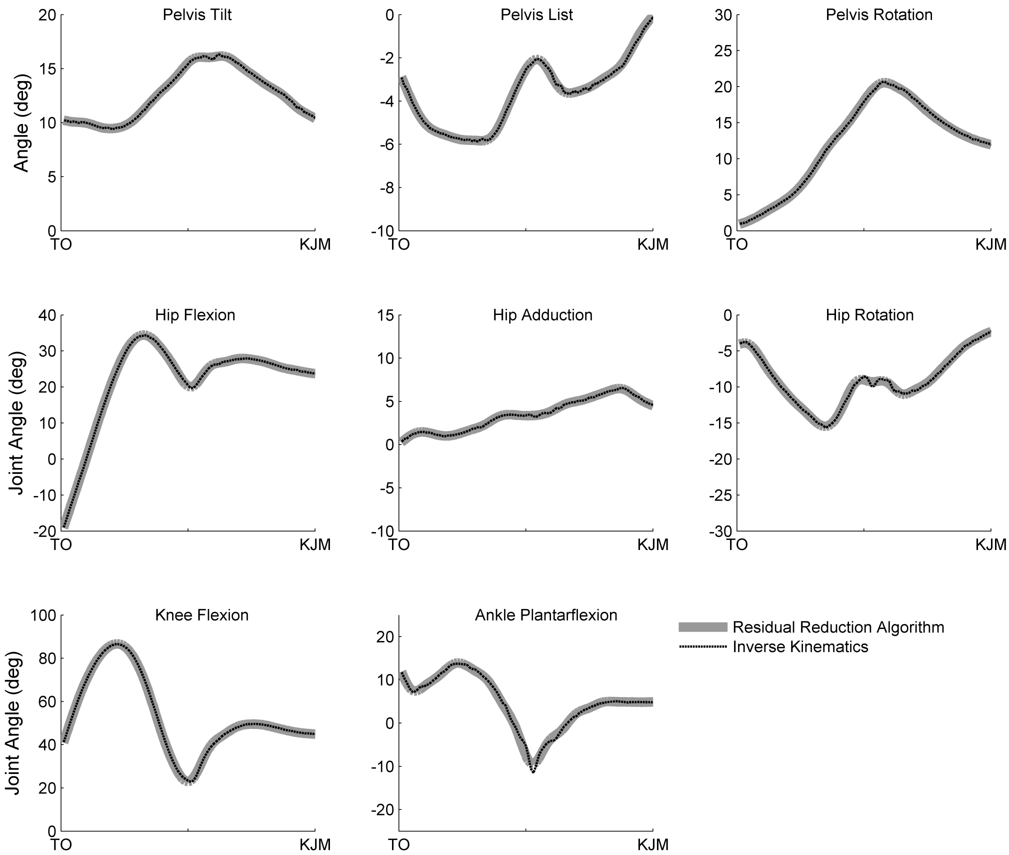

A
B
C
Hip Joint Contact Load (BW)
r = 0.49
r = 0.45
r = 0.41
MRLA (deg)
Trunk Angle at Foot Contact (deg)
Step Length/Leg Length

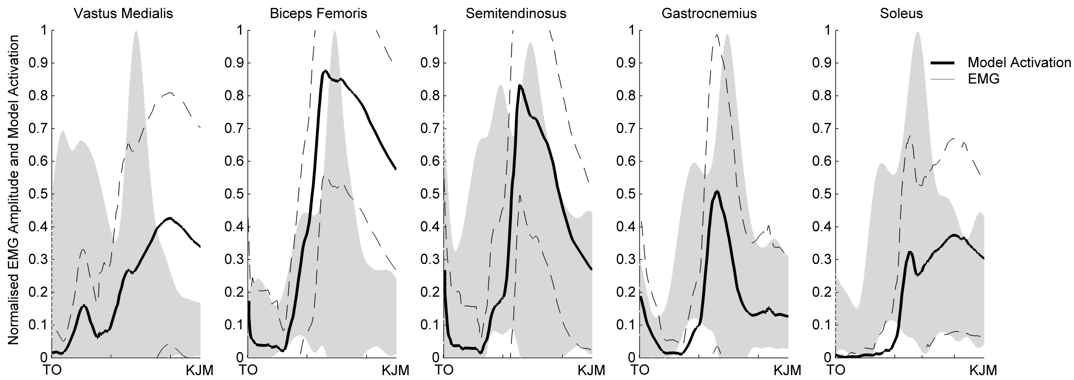
Figure 2



Supplementary Figure 1



Supplementary Figure 2



Supplementary Figure 3.