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

David F Graham[1,3], Luca Modenese, PhD[1], Grant Trewartha, PhD[2], Christopher P Carty, PhD[1], David G. Lloyd, PhD[1], Rod S Barrett, PhD[1]

[1]School of Allied Health Sciences and Menzies Health Institute Queensland, Griffith University, Queensland, Australia

[2]Department for Health, University of Bath, United Kingdom

[3]School of Science and Technology, University New England, New South Wales, Australia

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Corresponding Author

David F Graham

School of Science and Technology

University of New England, Armidale, NSW 2350

Email dgraha23@une.edu.au

Abstract

The purpose of this study was to use a musculoskeletal model to estimate hip and knee joint contact loads in community dwelling older adults during maximal recovery from forward loss of balance by stepping and to assess the association of contact loads with other measures of recovery kinematics. Participants (n = 106) were released from a series of increasing static forward lean angles until the maximum lean angle that each participant could recover from with a single step was identified. Peak hip and knee joint contact loads following touchdown of the stepping leg were computed using muscle force estimates obtained using static optimisation. Peak contact loads ranged from 5.1-12.3 body weights for the hip and 3.2-10.7 body weights for the knee. Peak joint contact loads were significantly correlated with the initial lean magnitude (Hip: r = 0.55; Knee: r = 0.32), as well as trunk flexion angle at foot contact (Hip: r = 0.30; Knee: r = 0.35) and step length (Hip: r = 0.54; Knee: r = 0.24). Overall findings indicated that older adults experience joint contact loads during maximal balance recovery by stepping that are 3-4 times higher than those reported for normal gait, and exceed hip contact loads previously reported to cause femoral fractures in individuals with severe osteoporosis and suboptimal neuromuscular function. Improving trunk control during recovery from forward loss of balance by stepping may decrease joint contact loads and corresponding the risk of bone and/or joint injury.

Abstract length = 242 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) and are considered unlikely to cause spontaneous hip fracture because the mechanical failure load of cadaveric femurs from older adults ranges from 5.5 to 14 body weights (Schileo et al. 2014). However Viceconti et al. (2012) demonstrated via the 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 typically experienced 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 successful recovery from an unexpected trip perturbation experienced during walking. At present however the magnitude of hip and knee 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 and knee joint contact loads experienced during balance recovery by stepping may be reduced.

The ability of older adults to recover from a large forward balance perturbation by stepping significantly predicts the risk of real world falls in the following 12 months (Carty et al. 2015) and is largely determined by an ability to resist forward trunk flexion during the stepping response (Barrett et al. 2012; Grabiner et al. 2008; Owings et al. 2001), take a suitably long recovery step (Graham et al. 2015; Karamanidis et al. 2008; Schillings et al.

2005) and 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 cocontraction of spine, hip and knee muscles during balance recovery from an equivalent balance perturbation (Graham et al. 2014) and might therefore be considered an example of suboptimal motor control that adversely affects balance recovery and simultaneously increase joint contact forces.

The purpose of this study was to use a musculoskeletal model to estimate hip and knee joint contact loads in older adults during recovery from a forward loss of balance when released from their maximum recoverable initial static forward lean angle. A secondary purpose was to assess the association of contact loads with other measures of balance recovery kinematics. We hypothesised that the magnitude of joint contact loads during balance recovery would be positively correlated with initial lean magnitude as well as variables previously reported to influence recovery performance, namely increased step length and increased trunk flexion angle at foot contact of the stepping leg.

Methods

Participants

One hundred and six community dwelling older adults aged 65 to 80 years (age: 72.0 ± 4.8 years; height: 1.67 ± 0.09 m, mass: 75.4 ± 12.5 kg) were recruited at random from the local electoral roll. Individuals previously diagnosed with neurological, metabolic, cardiopulmonary, 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 in Carty et al., (2011). Participants stood barefoot with their feet shoulder-width apart in an upright posture and were subsequently tilted forward, with their feet flat on the ground, until the required load in body weight (BW) was recorded on a load cell (S1W1kN, XTRAN, Australia) placed in series with an inextensible cable. One end of the cable was attached to a safety harness worn by the participant at the level of their sacrum and the other end was attached to an electric winch on a rigid metal frame located behind the participant. The length of the cable was adjusted until the required force on the cable was achieved. Care was taken to ensure the cable was aligned parallel with the ground and that participants kept their head, trunk and extremities aligned prior to cable release. 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. Participants were instructed to relax their muscles while leaning and to regain balance with a single step using the stepping lower limb of their choice following cable release. The instruction to attempt to recover using a single step was reiterated prior to every trial. A second cable, instrumented with a load cell (S1W1kN, XTRAN, Australia), 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 was displayed in real time on a computer monitor and was visually inspected by the investigator to ensure anticipatory actions (e.g., antero-posterior and medio-lateral weight shifting) were not evident in the period immediately prior to cable release. Following an initial trial at a 15%BW lean angle the Maximal Recoverable Lean Angle (MRLA) was determined by systematically increasing the lean magnitude by ~1%BW increments until the participant could no longer recover with a single step. For each trial, participants were classified as adopting either a single or a multiple step balance recovery strategy using previously published criteria (Carty et al. 2011) where a multiple step was identified by a) a second step of any kind by the stepping limb or progression of the non-stepping limb past the stepping foot following the initial step, b) lateral deviation of the lateral malleolus marker on the non-stepping foot by greater than 20% of body height from its position at cable release and c) if a force of greater than 20% BW was detected in the load cell attached to the ceiling restraint. For the purpose of this study only the MRLA trial was analysed. Trajectories of 51 reflective markers attached to each participant (Barrett et al. 2012) were recorded at 200 Hz using a 10 camera, 3-dimensional motion capture system (Vicon Motion Systems, LA, USA). Ground reaction forces were collected simultaneously at 1 kHz using two 900 mm x 600 mm piezoelectric force platforms (Kistler Instruments, USA). One plate was located beneath the two feet in the initial forward lean position and the second plate was located 800mm (center of plate one to center of plate two) anterior to the first plate in order to record ground reaction forces associated with touchdown of the stepping foot. Marker trajectory and ground reaction force data were filtered using a 4^{th} order, zero-lag, low-pass, Butterworth filter with a cut-off frequency of 20Hz (Bisseling et al. 2006). Specific events during the stepping phase of balance recovery were defined as follows: Cable release (CR) was identified from a 5 N drop in force measured in the horizontal restraining cable, toe off (TO) was identified from the first vertical motion greater than 2.5 mm of the great toe marker on the stepping foot (De Witt 2010) and foot contact (FC) from a force in excess of 5% of the participants body weight recorded on the anterior force plate. For the purpose of this study the length of each trial was the period from CR to 0.25 seconds after FC.

Computation of hip and knee joint contact forces

All data analysis were performed using OpenSim (version 3.2) (Delp et al. 2007) in conjunction with custom Matlab scripts (Version 2014b, The Maths Works, USA). A scalable anatomical model consisting of 17 bodies, 17 joints, 92 muscle actuators and 36 degrees-offreedom (Hamner et al. 2010) was used as the initial generic model for analysis. A wrap object was embedded in the generic model to maintain anatomically accurate erector spinae muscles moment arms during trunk flexion (Graham et al. 2014). Model scaling and inverse kinematic analysis (Lu et al. 1999) were performed by fitting the anatomical model to measured 3D marker positions with a high weighting on virtual markers 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. An evaluation of the simulations was conducted by comparing the experimentally collected muscle EMG to the corresponding muscle activation from static optimisation (see Supplementary Figure 1) in accordance with recommended best practice (Hicks et al. 2015). Passive muscle forces were also 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). 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). An example of contact loads calculated at the hip and knee joint for a representative subject is presented in Figure 1, together with ground reaction forces for both legs. The same OpenSim analysis was used to calculate joint reactions by disabling the muscles and providing the joint moments necessary to equilibrate the model through idealized torque actuators. The relative contribution of muscle forces to the total joint contact load was obtained by subtracting the joint reaction load from the total joint contact load.

Statistical Analysis

The Pearson Product Moment Correlation Coefficient was used to examine the relations between joint contact loads and initial lean angle, 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 (relative to the vertical axis). Statistical analyses were performed using SPSS (Version 22, IBM SPSS, USA). Significance was accepted for $P<0.05$.

Results

Mean peak contact loads were approximately 8 and 6 times body weight for the hip and knee respectively (Table 1). The largest peak joint contact loads experienced by an individual were 12.3 BW for the hip joint and 10.7 BW for the knee joint. Muscle forces contributed 95% of the total hip joint contact force and 80% of the total knee joint contact force respectively. Hip and knee joints joint contact loads were significantly correlated to maximal recoverable lean angle (Figure 2) as well as trunk flexion angle at foot contact and step length (Figure 3) ($p < 0.05$ for all correlations).

Discussion

This study confirmed that large peak contact loads are generated in maximal recovery from forward loss of balance by stepping, with individual peak contact loads following touchdown of the stepping leg ranging from 5.1-12.3 body weights for the hip and 3.2-10.7 body weights for the knee. In support of our hypothesis, the magnitude of hip and knee joint contact load was positively correlated with the intensity of the balance perturbation, and also with variables previously demonstrated to be associated with recovery performance, namely trunk flexion angle at touchdown of the stepping leg and recovery step length.

When released from the mean static forward lean angle of approximately 21 degrees, average peak joint contact loads were approximately 8 BW for the hip and approximately 6 BW for the knee. Average peak hip joint contact loads in the present study were therefore similar to the peak value of 8.7 BW reported for stumbling by Bergmann et al (1993), approximately 4 times higher than previously reported for slow walking on level ground (Bergmann et al. 2001) and approximately 1.5 times larger than those reported for stair descent and running at 8 km/hr (Bergmann et al. 1993). Similarly, peak knee joint contact loads were approximately 3 times higher than those reported for walking (Fregly et al. 2012) and around 1.7 times higher than those for stair descent (Kutzner et al. 2010). The relationship between joint contact load and the intensity of the balance perturbation was strongest for the hip (r = 0.55), where average peak hip loads increased by a factor of around 2 across the range of perturbation intensities examined, compared with the knee (r = 0.32), which increased around 1.5 times over the same range of perturbation intensities. In agreement with previous reports (Correa et al. 2010; Herzog et al. 2003; Winby et al. 2009) muscle force was the main determinant of the joint contact force, which in the present study accounted for 95% of total hip contact force and 80% of total knee contact force. The ability to generate large hip muscle forces and sustain large hip joint contact loads therefore appears critical for successful recovery from forward loss of balance by stepping which is consistent with the finding that lower limb muscle weakness predicts the ability of older adults to recover from forward loss of balance with a multiple compared to single steps (Carty et al. 2012a). The large peak hip joint contact loads identified in the present study are also similar to 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 suboptimal neuromuscular control and low bone mineral density.

Step length was correlated to the joint contact loads at the hip (r = 0.54) and knee (r = 0.24). A long step is important for balance recovery because it places the base of support further in front of the whole body centre of mass where GRF vector can more effectively reduce forward and downward centre of mass progression. However a large step also comes at the expense of larger joint contact forces, especially at the hip where a doubling of step length corresponds to around 50% increase in hip joint contact force. It therefore follows that participants that use short steps may do so to minimise joint loading, perhaps to maximise joint stability (Bergmann et al. 2004) or minimise joint pain, even though balance recovery is compromised.

The trunk flexion angle at foot contact was significantly correlated with hip (r = 0.30) and knee (r = 0.35) joint contact force. Excessive trunk flexion represents suboptimal motor control during balance recovery (Graham et al. 2014), and can also distinguish between older adults that require multiple versus single steps to recover from a fixed initial lean magnitude (Barrett et al. 2012; Grabiner et al. 2008; Owings et al. 2001). Given that the amount and rate of trunk flexion during balance recovery can be improved through repeated exposure to the task (Barrett et al. 2012; Carty et al. 2012c), training may be expected to reduce joint contact loads during balance recovery through the combined effect of improved trunk control and corresponding reduction in step length required to achieve dynamic stability.

A limitation of the present study was that, consistent with previous computational studies aiming to estimate hip contact loads in daily living activities (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 suboptimal neuromuscular control (Martelli et al. 2011; Modenese et al. 2013) including high levels of muscle co-contraction which would be expected to result in even higher joint contact loads. Further Bergmann et al. (2004) reported that unanticipated versus anticipated loss of balance resulted in higher versus lower contact loads during balance recovery. As participants in the present study were aware of their impending loss of balance, just not the exact timing, it is possible that the joint contact loads reported here may be smaller than for a completely unanticipated fall. In future it will be necessary to evaluate how the application of joint contact force vectors interact with the geometry and material properties of the proximal femur to more accurately determine a direct link to risk of femoral fracture.

Conclusion

Hip and knee joint contact loads in the stepping limb during recovery from forward loss of balance in older adults are 2-3 times higher than those previously reported for normal gait. Hip joint contact loads in particular were of similar magnitude to those previously reported to cause femoral fracture in individuals with a combination of suboptimal control and severe osteoporosis. Although a long recovery step is a feature of successful balance recovery, this comes at the expense of increased hip and knee contact forces. Conversely, large trunk flexion angles are a feature of poor balance recovery performance, and are also associated with larger hip and knee contact forces. Balance training that improves trunk control may therefore simultaneously improve balance recovery performance and decrease hip and knee joint loading.

Conflict of Interest

The authors declare that they have no conflicting interests.

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Figure 1
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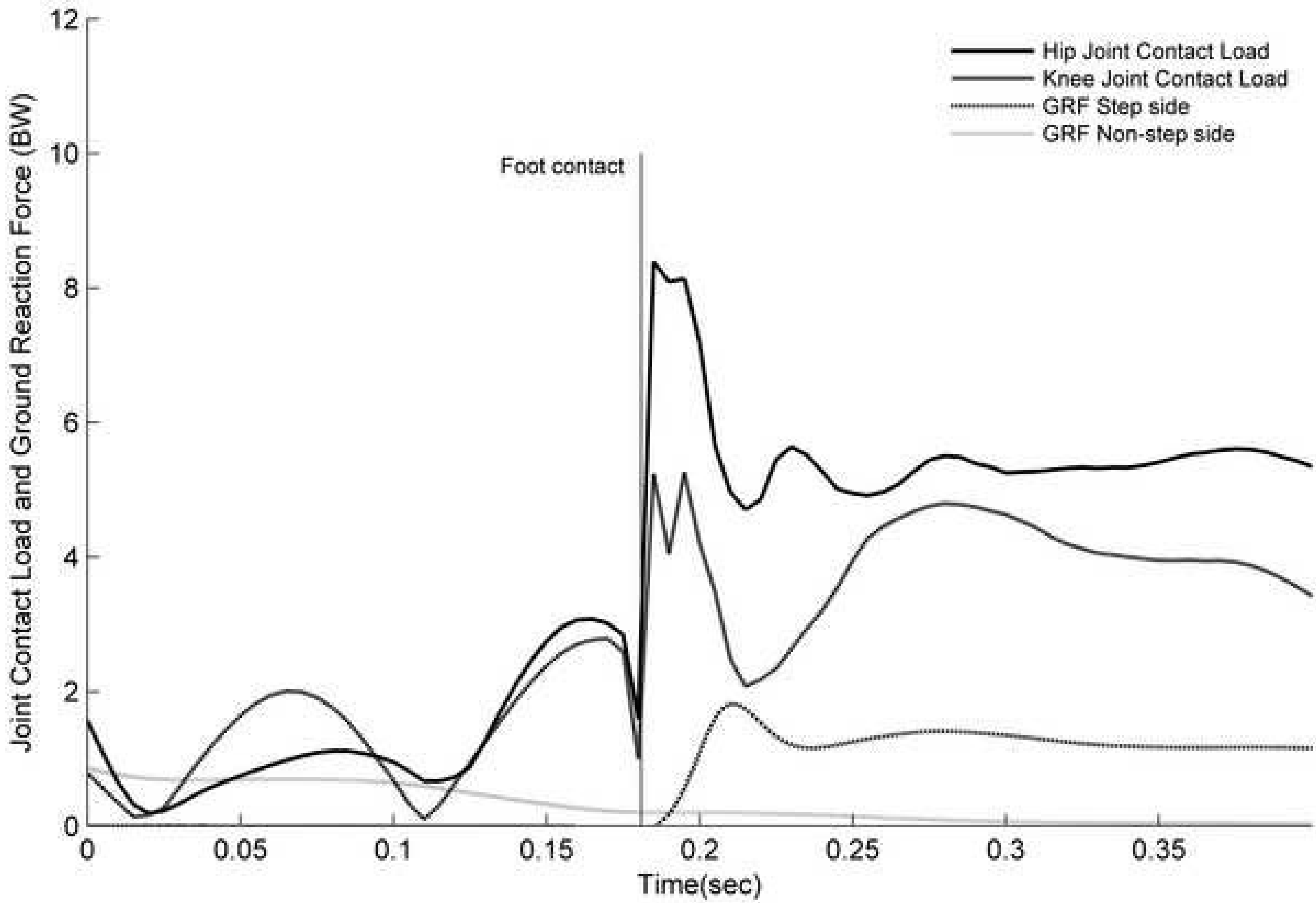

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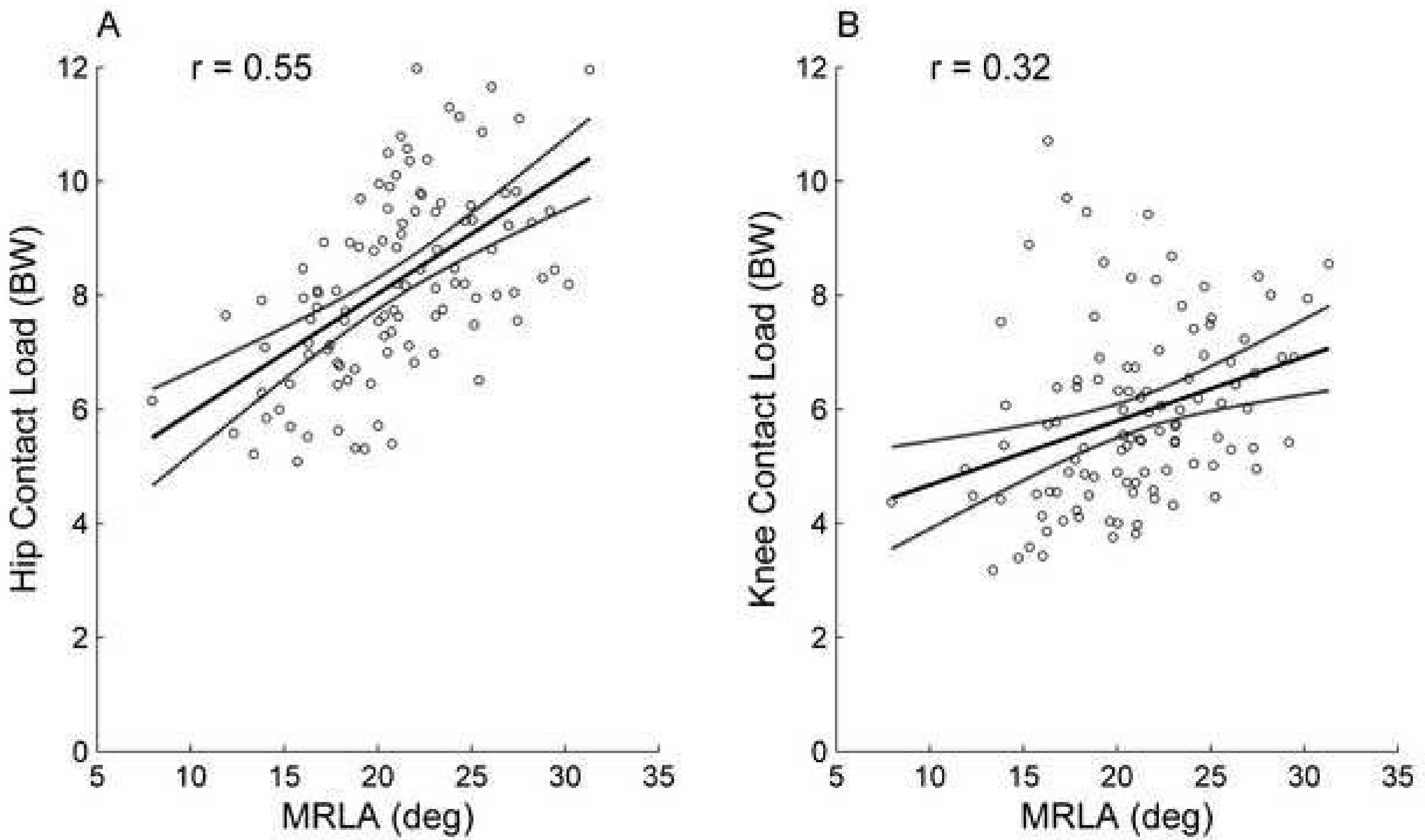

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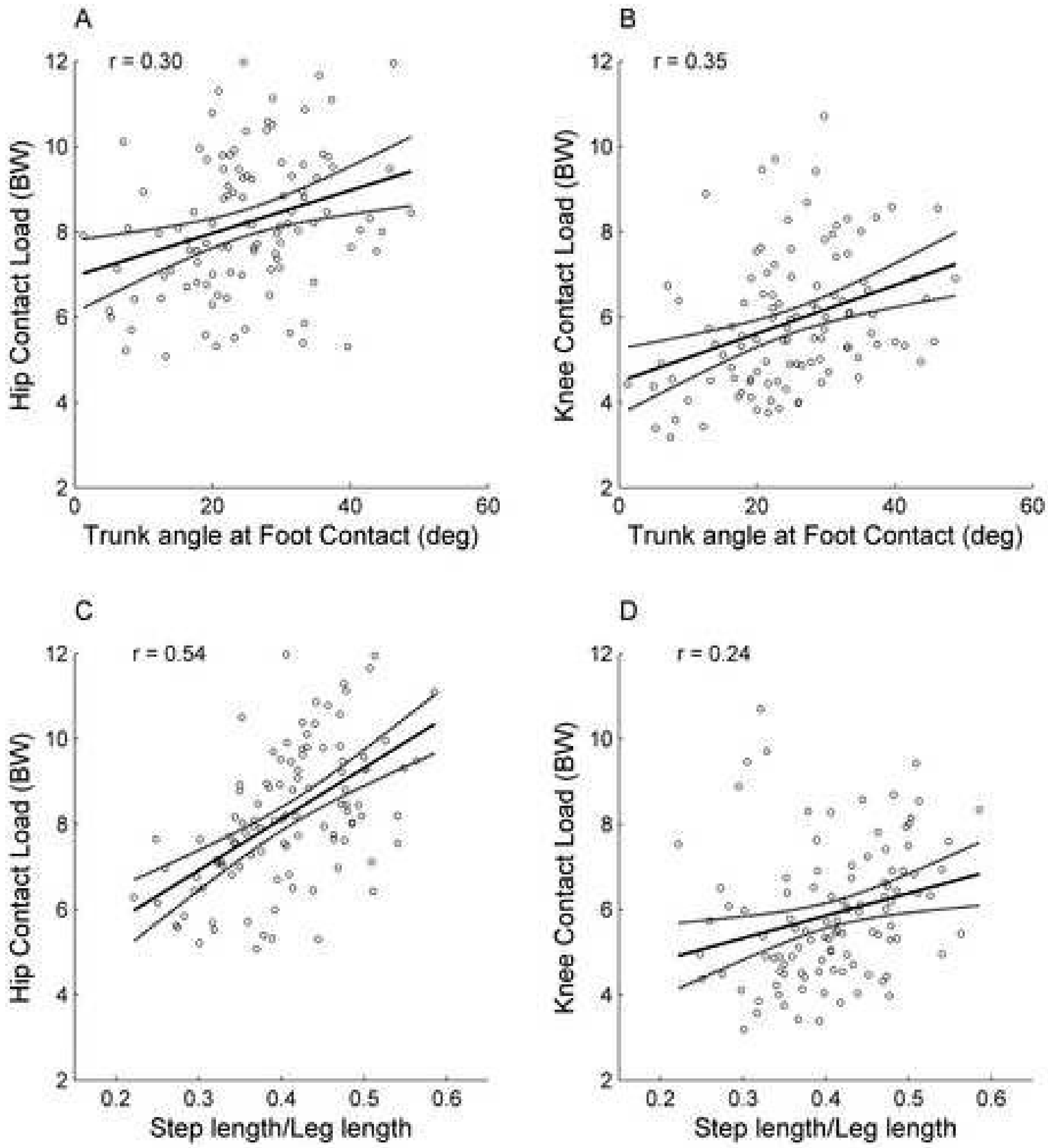

Table 1

Table 1. Summary data for balance recovery (n = 106).

	Mean ± SD	95% CI	Range
Maximum recoverable lean angle (°)	20.9 ± 4.4	20.2-21.8	7.9-31.3
Peak hip contact load (BW)	8.22 ± 1.68	7.94-8.55	5.1-12.3
Peak knee contact load (BW)	5.90 ± 1.60	5.64-6.24	3.2-10.7
Peak ground reaction force (BW)	1.83 ± 0.57	1.72-1.90	0.98-3.39
Trunk flexion angle at foot contact (°)	25.0 ± 9.9	23.3-26.9	11.3-48.8
Step length/leg length	0.41 ± 0.08	0.40-0.42	0.22-0.59

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