5-1-2015

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Repository Citation

Nordin, Andrew D., "Investigating single-leg landing strategies and movement control across changes in task demands" (2015). *UNLV Theses, Dissertations, Professional Papers, and Capstones*. 2825.
http://dx.doi.org/10.34917/9419965

INVESTIGATING SINGLE-LEG LANDING STRATEGIES AND MOVEMENT CONTROL ACROSS CHANGES IN TASK DEMANDS

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A dissertation submitted in partial fulfillment of the requirements for the

Doctor of Philosophy – Kinesiology

Department of Kinesiology & Nutrition Sciences
School of Allied Health Sciences
The Graduate College

University of Nevada, Las Vegas
May 2015

We recommend the dissertation prepared under our supervision by

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entitled

Investigating Single-Leg Landing Strategies and Movement Control Across Changes in Task Demands

is approved in partial fulfillment of the requirements for the degree of

Doctor of Philosophy - Kinesiology
Department of Kinesiology and Nutrition Sciences

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ABSTRACT

Variability is an intrinsic characteristic of human movement, with hypothesized connections to neuromotor functioning and mechanisms of injury. The purpose of this study was to evaluate changes in movement variability among kinematic, kinetic, and electromyographic (EMG) variables following mechanical task demand manipulations during single-leg drop landings. Biomechanical outcome variables included 3 kinematic (sagittal, hip, knee, and ankle angles), 4 kinetic (sagittal hip, knee, ankle moments and vertical ground reaction force; GRFz), and 5 EMG variables (gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior muscles). Mechanical task demands were altered using load and landing height manipulations, computed as percentages of participant anthropometrics (bodyweight: BW, BW+12.5%, BW+25%, and height: H12.5% and H25%, respectively). Fewer emergent strategies were identified under greater mechanical task demands, defined using the load accommodation strategies model, alongside decreased movement variability, assessed using principal component analysis (PCA). Joint-specific biomechanical adjustments were identified, highlighting mechanisms for the observed load accommodation strategies and changes in movement variability. An increasingly upright landing posture was observed under greater mechanical task demands, decreasing effective landing height and reducing landing impulse. Alterations in movement variability were interpreted in the context of the available functional degrees of freedom at each lower extremity joint, aligning with physiological predictions and theories from motor control. The holistic approach taken in this investigation provided a more complete understanding of mechanisms contributing to changes in movement variability and factors that may underlie landing injuries.

ACKNOWLEDGMENTS

I would like to thank the University of Nevada, Las Vegas (UNLV), School of Allied Health Sciences, and the Department of Kinesiology and Nutrition Sciences for funding contributions during my doctoral studies. Additionally, sponsorship funding from the UNLV Graduate and Professional Student Association (GPSA) provided financial assistance for conference travel and professional development. Further, I must acknowledge the UNLV Graduate College for significant funding assistance through the UNLV Foundation President's Graduate Research Fellowship, as well as the Nevada System of Higher Education for the 2015 Regents' Scholar Award.

Thank you to my student colleagues in the Sports Injury Research Center for their assistance in data collections, and special recognition for the student researchers who completed the Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program (2012 & 2013).

I have tremendous appreciation for the time and feedback provided by dissertation committee members, Dr. John Mercer, Dr. Gabriele Wulf, and Dr. Sheniz Moonie, each of whom impacted my academic success and the development of my doctoral dissertation. Additional recognition is required for the previous work of Dr. Roger James and Dr. Barry Bates, who provided invaluable feedback and research direction during manuscript preparation and informal conversations. The tremendous support of Dr. Janet Dufek cannot be overstated and I am hopeful that I can provide adequate representation as the first of her UNLV doctoral students in the next phase of my academic career.

Finally, I would like to acknowledge the constant support of my family and Ms. Rebecca Pugh throughout my educational process.

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CHAPTER 1

Introduction

Variability is an intrinsic characteristic of human movement. Even the most skilled performers are unable to precisely replicate a movement pattern given the seemingly infinite number of body segment configurations and movement options available to the performer (Davids, Glazier, Araujo, & Bartlett, 2003; James & Bates, 1997; James, Bates, & Dufek, 2003; Lohse, Jones, Healy, & Sherwood, 2013). Sources and functions of trial-to-trial variations during task repetitions have been investigated from a number of perspectives and across a variety of tasks. As a result, human movement variability examinations require interdisciplinary considerations from areas including neuroscience, motor control, physiology, anatomy, and biomechanics. Within each of these areas, a functional role of variability has been acknowledged, providing a means for adaptation and system flexibility (Barrett, Noordegraaf, & Morrison, 2008; James, Dufek, & Bates, 2000; Stergiou & Decker, 2011; Stergiou, Harbourne, & Cavanaugh, 2006). Variability has therefore been considered in the context of neuromotor functioning, with hypothesized connections to understanding mechanisms of injury (Bartlett, Wheat, & Robins, 2007; Harris & Wolpert, 1998; Stergiou & Decker, 2011). Landing from a jump or an elevated surface provides a commonly performed movement with a high risk of injury (Dufek & Bates, 1992; Zhang, Bates, & Dufek, 2000), leading to investigations examining associations between movement variability and injury risk (James et al. 2000). Theoretical links between movement variability and injury have been proposed, operationalizing functional limits that characterize acute and overuse injuries (James, in Stergiou, 2004, p. 29-62). Collectively, these perspectives were used in this investigation, assessing movement variability during single-leg landings under contrasting task demands.

Sources of movement variability have been viewed from mechanical and neuromuscular perspectives, attributing variation to motor redundancy, as outlined by the degrees of freedom (DOF) problem where the number of available segmental configurations exceeds those required to complete a movement (Bernstein, 1967, p. 161-164; Davids et al., 2003; Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999; Todorov, 2006; Turvey, 1990). Along these lines, movement coordination is simplified by way of synergistic associations among effectors (muscles, joints, or limbs), providing a reduced subset of control units described using functional DOF (James & Bates, 1997; Li, 2006). Explanations have been proposed in various motor control theories for the selection of a motor pattern, explaining sources and changes in variability across changes in task demands, highlighting the functional role of variability in human movement. Contrasting sources of variation have been attributed to both deterministic and stochastic processes, varying in structure, as well as magnitude and direction (Davids et al., 2003; Diedrichsen, Shadmehr, & Ivry, 2009; Lohse et al.2013; Todorov, 2006). The anisotropic nature of variability, varying by direction, is outlined conceptually in the Uncontrolled Manifold Hypothesis (UCM), where variation is differentiated between redundant and goal-directed dimensions, represented by abstract orthogonal axes (Diedrichsen et al., 2009; Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999; Todorov, 2006). The UCM outlines testable hypotheses that can be statistically assessed using multivariate procedures examining the correlation structure among variables during task repetitions. For these reasons, changes in movement variability under contrasting task demands should be considered simultaneously from multiple sources, providing insight into movement control that may be beneficial in understanding injury mechanisms.

Physiological considerations have been provided in research, relating electrical muscle activity to neuromuscular control. Minimal intervention during movement control has been identified in motor control theories, linking patterns of muscle activation to movement efficiency as outlined in the equilibrium-point hypothesis, Optimal Control Theory (OCT), and the Constrained Action Hypothesis (CAH; Diedrichsen et al., 2009; Feldman, in Latash, 2010; Lohse et al., 2013; Todorov, 2006; Scholz & Schoner, 1999; Scott, 2012; Wulf, McNevin, & Shea, 2001). Given the focus of the proposed investigation on landing, special considerations must be given to the role of muscular contraction on electrical muscle activity. The eccentric loads placed on lower extremity muscles during landing are the result of muscle lengthening during active tension development, leading to increased risk of muscle injury (Bishop, Trimble, Bauer & Kaminski, 2000; Fang, Siemionow, Sahgal, Xiong & Yue, 2004). Patterns of electrical activity during eccentric contractions contrast those observed in concentric contractions (muscle shortening under active tension), where additional muscle force occurs without associated increases in total electrical activity (Bishop et al., 2000; Guilhem, Cornu & Guevel, 2010; James, Dufek, & Bates, 2006; Mianfang & Li, 2010). Changes in muscle activation during eccentric loading in landing have been attributed to pre-activation, recruitment of faster motor units, inhibitory mechanisms, and changes in neural inputs from the brain (Fang et al., 2004; James et al., 2006; Mianfang & Li, 2010). Outcomes from the mechanical and neuromuscular considerations for movement control therefore provide more complete explanations for the available motor responses under contrasting conditions.

Consideration for the morphological, biomechanical, and environmental constraints that shape movement outcomes provide context for the emergence of movement strategies, defined by mechanical and neuromuscular adjustments in the solution of a motor task (Bates, 1996; James & Bates, 1997; James et al., 2003). The availability of distinct neuro-musculoskeletal solutions to task changes have been characterized by movement strategies. As well, manipulations to system mass during landing and gait have demonstrated a range of available responses (Caster & Bates, 1995; James et al., 2003; James, Atkins, Dufek, & Bates, 2014). Classifications of biomechanical response strategies have been examined through changes in the ground reaction force (GRF) following modifications to the applied stressor (external force, mass, energy, or momentum). Changes in the GRF with changes in an applied stressor provide five available outcomes defined by the load accommodation strategies model: Newtonian (expected increase with increased stressor), positive biomechanical (increase lesser than predicted), fully accommodating (no change relative to baseline), negative biomechanical (decrease relative to baseline), and super-Newtonian (exceeding that predicted by the added stressor; James et al., 2003; James et al., 2014). The multiple available movement strategies are considered in line with the concepts of functional variability and functional DOF under contrasting task demands.

The load accommodation strategies model provides a means of assessing movement variability, where lesser variability is considered representative of fewer emergent strategies (Caster & Bates, 1995; James & Bates, 1997; James et al., 2000;

James, in Stergiou, 2004, p. 29-62). In addition, the outlined model provides classifications for movement outcomes, allowing variability responses to be more concretely linked with performer strategies. Associating emergent strategies with functional variability may therefore be relevant to movement control, providing potentially useful interpretations for understanding injury mechanisms. Anecdotal associations between variability and injury have been previously highlighted, where variability is considered essential in musculoskeletal health, influencing stressor magnitudes, frequency, and rates of adaptation (James, in Stergiou, 2004, p. 29-62). Functional variability limits have therefore been proposed, with excessive variability linked to mechanisms of acute injury and decreased variability linked to overuse injury (James, in Stergiou, 2004, p. 29-62). Concrete associations between variability and injury remain tenuous, however, assessing movement control characteristics through changes in task demands may provide insight into logical associations between variability and injury.

Quantification and expression of variability provides flexibility to the researcher. Although a variety of methods exist to compute movement variability, greater inclusion of variables and time series analyses are considered essential advances in understanding movement control (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Deluzio, Harrison, Coffey, & Caldwell, in Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2014; Donoghue, Harrison, Coffey, & Hayes, 2008). Principal component analysis (PCA) has gained increasingly widespread use in biomechanical investigations as a mean of reducing multivariate datasets into independent sources of variation (Brandon, Graham, Almosnino, Sadler, Stevenson, & Deluzio, 2013; Cohen, 2014; Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Federolf, Boyer, & Andriacchi, 2013; Kipp &

Palmieri-Smith, 2012; Molenaar, Wang, & Newell, 2013; Richter, McGuinness, O'Connor, & Moran, 2014; Richter, O'Connor, Marshall, & Moran, 2014, Robbins, Astephen Wilson, Rutherford, & Hubley-Kozey, 2013). PCA has been used as a means of data filtering, statistically reducing relevant information from multi-dimensional signals using the underlying correlation structure among variables (Brandon et al., 2013; Daffertshofer et al., 2004; Donoghue et al., 2008; Federolf et al., 2013; Richter et al., 2014b). Given the outlined description of movement coordination and synergies, it can be understood that the use of PCA, reducing datasets into functional units, provides interpretations aligning with concepts from motor control, including functional DOF and the UCM (Latash, 2010; Li, 2006; Lohse et al., 2013; Scholz & Schoner, 1999). Collectively, PCA has been successfully applied to datasets including variables from kinematic, kinetic, and EMG sources, demonstrating utility among a number of tasks (Daffertshofer et al., 2004; Kipp et al., 2012; Kipp et al., 2014; Li, 2006; Molenaar et al., 2013; Richter et al., 2014b). For these reasons, greater inclusion of variables contributing to movement control may be beneficial in providing insight into factors relating to injury. Additionally, applications of PCA to biomechanical time series data allows changes to be identified across a movement phase, adding greatly to the understanding of entire movement patterns beyond univariate examinations.

The outlined theoretical basis underlying this research highlights associations among movement control, performer strategies, movement variability, and potential injury mechanisms in landing. A more holistic approach was used in examining movement control in the support limb, during single-leg landings from an elevated surface, following changes in load and landing height. PCA provided the means for associating biomechanical and physiological measures with motor control theories. These hypotheses were tested statistically, examining changes in movement variability following mechanical task demand manipulations. The outlined research sought to further the understanding of lower extremity movement regulation, synthesizing interdisciplinary perspectives and making connections to potential injury mechanisms in landing.

Purpose

The purpose of this study was to evaluate changes in movement variability among kinematic, kinetic, and electromyographic (EMG) variables following mechanical task demand manipulations during single-leg drop landings. Biomechanical outcome variables included 3 kinematic (sagittal hip, knee, and ankle angles), 4 kinetic (sagittal hip, knee, ankle moments and vertical ground reaction force: GRF), and 5 EMG variables (gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior). Mechanical task demands were altered using load and landing height manipulations, computed as percentages of participant anthropometrics (bodyweight: BW, BW+12.5%, BW+25%, and height: H12.5% and H25%, respectively). Load accommodation strategies were characterized using impulse ratios relative to baseline (lowest task demands: BW•H12.5%) and 95% confidence intervals (CI) surrounding mechanically predicted impulse ratios. Collective movement variability among participants was assessed across the landing phase using principal component analysis (PCA). Follow-up analyses were carried out separately for each outcome variable, identifying changes in support limb movement patterns following mechanical task demand manipulations. PC loading vectors accessed sources of variation in the dataset, with inferential testing of PC scores identifying movement pattern changes. The number of extracted PCs and explained variance (EV) in each condition were used in interpreting movement control in each lower extremity joint.

It was hypothesized that fewer emergent strategies would be observed with greater mechanical task demands, leading to decreased movement variability. Specifically, greater load and landing height were anticipated to result in more predictable response strategies (increasingly positive biomechanical) with decreased movement variability expressed by greater explained variance among fewer extracted PCs. Statistical differences were expected among conditions for PC scores explaining the greatest proportion of the variance in each landing condition. Additionally, greater PC scores were anticipated to occur with greater mechanical task demands. Follow up analyses, for each outcome variable (kinematic, kinetic, and EMG), were expected to show greater EV among fewer PCs in proximal joints, relative to distal joints, with the number of PCs decreasing with greater load and landing height. PC loading vectors were predicted to show earlier increases in the landing phase at distal joints, with later increases at proximal joints. Joint specific strategies were expected from PC score differences among conditions. The outlined hypotheses were interpreted relative to functional variability surrounding movement repetitions, making associations between movement control strategies and potential injury mechanisms in single-leg drop landings.

Limitations

1. Subject-specific normalization of load and landing height manipulations were considered an appropriate means of reducing between-subject variability. Normalization of load and landing height to participant anthropometrics may not, however, correspond directly to levels of neuromuscular control or strength measures. Limitations were therefore acknowledged in interpreting variability responses to the outlined task manipulations.
2. Alterations in mechanical task demands were considered to occur within a range that would elicit a change in movement strategies and movement variability. The upper limits of the outlined changes may, however, have failed to reach levels necessary to evoke meaningful changes in movement control. Importantly, consideration was given to minimizing risk of participant injury during data collection.
3. The ecological relevance of the laboratory based experimental procedure may limit generalizations to performance settings, including landing from a jump.
4. Three-dimensional kinematic and kinetic measurement limitations included the use of a rigid, linked segment model in depicting human movement. The underlying assumptions are acknowledged, but these measurements are common in biomechanics.
5. Measurement of surface electromyography (EMG) presented limitations, including the interpretation of EMG data as a measure of motor unit action potentials and muscle force. EMG measures were acknowledged to be

representative of the summation of muscle activity, requiring special consideration during eccentric activities including landing.

6. PCA is a linear statistical procedure that may limit interpretations when nonlinear relationships exist among variables. As well, the temporal and magnitude normalization required during analysis may have removed useful sources of variation in the dataset. Application of PCA to biomechanical data allowed inclusion of multiple trials per participant, but involved the assessment of grouped data, which may limit the ability to differentiate between-subject from within-subject variability. Despite these limitations, PCA provided a valuable data reduction tool.
7. Associations between movement variability and injury were addressed, though injury was not specifically measured, nor were injured participants examined. Investigating variability changes under contrasting mechanical task demands were, however, considered important in understanding potential mechanisms of injury.
8. A convenience sample was used in this investigation potentially limiting generalizations to the population from which the participants were drawn.

Delimitations

1. Analysis was delimited to single-leg landings.
2. The landing phase was defined from ground contact to the time vertical center of mass (COM) velocity reached zero, ignoring preparatory movements and those occurring as participant returned to standing.

3. Lower extremity kinematics and kinetics were assessed in the support limb across the landing phase, focusing solely on sagittal plane movements at each joint and the vertical ground reaction force, despite the collection of three-dimensional data.
4. Electromyographic (EMG) variables were measured in the gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibalis anterior muscles. Other muscles may be of interest in the control of movement during landing.
5. Mechanical task demands were altered using load and landing height manipulations computed as percentages of participant anthropometrics (BW, BW+12.5%, BW+25% and H12.5%, H25%, respectively). More expansive ranges and contrasting modifications of mechanical task demands may have exposed contrasting outcomes.
6. Load manipulations were carried out through the application of loads to the anterior and posterior aspects of the trunk. Other sites of application may have lead to contrasting outcomes.
7. Male and female participants were recruited, though no considerations were given to gender comparisons in the statistical analyses.

Definitions

Single-leg landing: a drop landing performed by stepping out from an elevated surface and landing on the ground with the support of a single-leg.

Kinetics: forces causing movement.

Kinematics: observable movement outcomes without consideration for the cause (force).

Surface electromyography (EMG): measurement of electrical muscle activity via electrodes placed on the skin overlaying muscle.

Joint moment: angular or rotational force acting about an axis of rotation at each joint.

Sagittal: plane of motion spanning the vertical and anterior-posterior space that divides the body down the midline. Rotation in the sagittal plane occurs about the medial-lateral axis of the body.

Variability: trial-to-trial fluctuations of performance outcomes or movement patterns.

Principal Component Analysis (PCA): a multivariate data reduction technique used in assessing the correlation structure of a matrix. PCA extracts independent (orthogonal) sources of variation from a dataset, termed: principal components (PCs).

PC loading vector: a Principal Component (PC) loading vector, or eigenvector, is used in depicting the pattern of change that is captured by each PC across a time series.

PC score: Principal Component Scores represent a linear combination between original variables and PC loading vector coefficients, which can be used in hypothesis testing.

Eigenvalue: a matrix multiplier computationally used in performing a linear transformation to a matrix; used in PCA during the calculation of explained variance for each PC.

Constraints: morphological (anatomy), biomechanical (gravity and friction), and environmental (organization in time and space) factors shaping movement.

Strategy: a neuro-musculoskeletal solution of a motor task, uniquely selected by an individual under the presence of a given set of constraints.

Super-Newtonian Response: an increase in external force exceeding that predicted by an increased stressor.

Newtonian response: an expected linear increase in external force with an increased stressor.

Positive biomechanical: an increase in external force lesser than predicted by an increase in stressor.

Fully accommodating: an absence of change in the external force with an increase in stressor.

Negative biomechanical: a decrease in the external force despite an increase in the applied stressor**.**

Degrees of Freedom (DOF): the independent coordinates required to describe the configuration of a system, including three orthogonal axes along which rotation and translation can occur.

Functional DOF: movement options, or the reduced subset of control units arising due to neural and mechanical synergies.

Synergy: systematic correlations among effectors (muscles, joints, or limbs), as a result of mechanical, muscle, and neural linkages.

Coordination: organization of the body in time and space during task execution, effectively accomplished using synergistic associations among effectors.

Uncontrolled Manifold Hypothesis (UCM): defining redundant ($\text{UCM}_{||}$) and goal-directed dimensions ($\text{UCM}_{\perp}$) along which variability is allowed to accumulate, or is more tightly controlled.

Optimal Control Theory (OCT): outlining cost functions that underlie movement control. Effort and movement variability are minimized in finding an optimal solution to a motor problem by way of control rules.

Constrained Action Hypothesis (CAH): conscious control that interferes with automatic processes normally regulating movement.

Eccentric muscular contraction: increases in muscle length with active tension development.

Overuse injury: repeated loadings of sufficient magnitude and frequency that outpace physiological adaptations.

Acute injury: a single traumatic loading event where the failure threshold of the tissue is exceeded.

Gluteus maximus: largest superficial gluteal muscle, located on the posterior aspect of the hip, acting as a hip extensor and external rotator of the thigh.

Vastus medialis: muscle on the medial aspect of the anterior thigh, acting as a knee extensor.

Biceps femoris (long head): muscle on the posterior aspect of the thigh acting as a knee flexor and hip extensor.

Medial gastrocnemius: muscle on the medial aspect of the posterior leg acting as an ankle plantar flexor.

Tibalis anterior muscles: muscle on the lateral aspect of the anterior leg, acting as an ankle dorsiflexor and invertor.

CHAPTER 2

Review of Literature

Human movement variability

Human movement is inherently variable, leading to examinations of the sources and function of movement variability in a variety of settings and among a number of tasks (Barrett et al., 2008; Bartlett et al., 2007; James et al., 2000; Li, Haddad, & Hamill, 2005; Newell, Challis, & Morrison, 2000; Stergiou & Decker, 2011; Stergiou et al., 2006). Human movement investigations are often viewed as a performer under neural control, operating much like a computer with a central processor, consisting of inputs and outputs that are in communication via transmitted signals. Following this line of reasoning, outcomes are directed by transmitted signals, consisting of both signal and noise (Hamill, van Emmerik, & Heiderscheit, Li, 1999; Harris & Wolpert, 1998; Newell et al., 2000; Shi & Buneo, 2012; Stergiou & Decker, 2011; Stergiou et al., 2006). From this perspective, functional outcomes are dependent on the transmission of the signal with minimal noise, resulting in repeatable, stable processes (Harris & Wolpert, 1998; Newell et al., 2000; Shi & Buneo, 2012). Understanding that excessively noisy signals may result in inconsistent outcomes, it seems logical to assume that noise should be minimized, and ideally eliminated for optimal performance (Harris & Wolpert, 1998; Newell et al., 2000; Shi & Buneo, 2012; Stergiou & Decker, 2011; Stergiou et al., 2006). Assessing humans as biological computers neglects the ability of the system to adapt or change as a result of exploration both within the surrounding environment, but also within neural pathways of the body. In this sense, biological noise may be both unavoidable and essential to optimal functioning, providing the means for adaptation

(Barrett et al., 2008; James et al., 2000; Stergiou et al., 2006). The emergent school of thought views variation in performance as operational and necessary, affording the performer flexibility and the ability to adapt to stressors (Barrett et al., 2008; James et al., 2000; Li et al., 2005; Stergiou & Decker, 2011; Stergiou et al., 2006). Within this context, variability can be viewed in relation to injury, where the introduction of small changes in performance avoids repeated stress to identical tissues, more broadly distributing loads and therefore avoiding the likelihood of overuse injuries (Barrett et al., 2008; Bartlett et al., 2007; Brown, Bowser, & Simpson, 2012; James et al., 2000; James, in Stergiou, 2004, p. 29-62). As a result, this review aims to investigate the current understanding of movement variability assessments, exploring the sources of variability and making applications to movement control in landing, with connections to mechanisms of injury.

Variability examinations have been carried out on tasks in both the upper and lower extremities, ranging from simple tasks including finger oscillations to complex skills including throwing, landing, and gait (Barrett et al., 2008; Bartlett et al., 2007; Brown et al., 2012; Churchland, Afshar, & Shenoy, 2006; DiBerardino, Polk, Rosengren, Spencer-Smith, & Hsaio-Wechsler, 2010; Federolf, Tecante, & Nigg, 2012; Hamill et al., 1999; James et al., 2000; Li et al., 2005; Newell et al., 2000). Although considerable breadth is apparent in the movement variability literature, the focus of this investigation was on lower extremity functioning, with particular attention to landing. Landing provides a movement that is common in sport and has been implicated as a mechanism of injury in single and double leg conditions, following jumping, and during drop landings from an elevated platform (James et al., 2000; Padua, DiStefano, Marshall, Beutler, de la Motte, & DiStefano, 2011; Schmitz, Kulas, Perrin, Rieman, & Shultz, 2007; Scholes, McDonald, & Anthony, 2012; Torry, et al., 2011; Wikstrom, Tillman, Schneker, & Borsa, 2008). As well, landing affords the researcher experimental control, easily adjusting task demands under experimentally controlled settings. Better understanding of movement function during landing provides an avenue of research that has been extensively explored in terms of the biomechanical characteristics of movement and in terms of changes in variability from changes in task demands. Despite this, additional work can be done in this area, focusing on injury mechanisms and implications of variability changes during performance, making connections between the levels of motor control and observed movement outcomes.

A complete assessment of variability encompassing control mechanisms of human movement synthesizes information from disciplines including neuroscience, motor control, physiology, anatomy, and biomechanics. As a result, this review is far from a comprehensive examination of variability in the context of human movement, but serves as a summary of the state of the science regarding the link between variability and injury in the lower extremity in the context of landing. Additional considerations are directed toward variability measures and the implications of research design.

Theoretical foundations of movement variability in motor control

Measures of variability in human movement have been used as a window into the functioning of the neuromotor system (Barrett et al., 2008; Bartlett et al., 2007; Harris & Wolpert, 1998; James et al., 2000; Li et al., 2005; Newell et al., 2000; Scholes et al., 2012; Stergiou & Decker, 2011; Stergiou et al., 2006). It has been suggested that performance variability provides flexibility and the ability to adapt to changes in task demands or changes in the environment (Barrett et al., 2008; James et al., 2000; Li et al., 2005; Stergiou & Decker, 2011; Stergiou et al., 2006). The concept that movement emerges from constraints imposed by morphology (anatomy), biomechanics (gravity and friction), and the environment (organization in the external world in time and space) has been drawn from motor control (Bernstein 1967, p. 161-164; Higgins, 1985; James & Bates, 1997). Variability examinations have evolved from the view that variation is an inconvenient and pervasive element of human movement, to the result of complications from a complex neural system, and finally to the acknowledgement of variability as an essential component of movement (Barrett et al., 2008; James et al., 2000; Li et al., 2005; Stergiou et al., 2006). Each of these developments has been associated with specific motor control theories that will be reviewed in the following sections.

Initial investigations into movement control considered variability surrounding task repetitions as a product of redundancy (Bernstein 1967, p. 161-164; Davids et al., 2003; Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999; Todorov, 2006; Turvey, 1990). Here, redundancy is defined by the numerous mechanical degrees of freedom (DOF) associated with the human anatomical structure, which exceed the minimum number required to successfully perform a task (Bernstein 1967, p. 161-164; Davids et al., 2003). Mechanical definitions of biomechanical DOF indicate the number of independent coordinates required to uniquely describe the configuration of a system, including three orthogonal axes along which rotation and translation can occur (Davids et al., 2003; Turvey, 1990). For a theoretical point, six mechanical DOF can be characterized and applied to configurations in space.

Applied to the human structure, the biomechanical system has been suggested to consist of approximately 100 mechanical DOF, mechanically linked providing anatomical constraints at each joint (Turvey, 1990). It can be understood that organization of the numerous DOF requires reduction into a simplified subset of control units during movement regulation (Latash, 2010; Lohse et al., 2013; Turvey, 1990). Dimension reduction is accomplished through neuromuscular 'freezing', while the actual number of mechanical DOF does not change (Latash, 2010). Along these lines, a reduced subset of control units has been described in research using functional DOF (James & Bates, 1997; Li, 2006). Experimentally, functional DOF have been examined from neural and mechanical perspectives using synergies (Chvatal & Ting, 2012; Diedrichsen et al., 2009; Todorov, 2006; Turvey, 1990).

Movement control synergies

Mathematically, synergies are defined by systematic correlations among effectors (muscles, joints, or limbs), which characterize coordinated movement (Diedrichsen et al., 2009; Latash, 2010). Conceptually, the mechanical and muscle linkages, as well as the underlying neural organization describe a synergy (Diedrichsen et al., 2009; Latash, 2010; Turvey, 1990). Modular control of associated units (muscle groups) have been experimentally identified in balance, gait, and single-leg drop landing, describing not only the correlation structure among movement outcomes (Diedrichsen et al., 2009), but also the neural activation patterns controlling movement (Chvatal & Ting, 2012; Kipp et al., 2014). Synergistic associations among variables, including kinematic, kinetic, and electromyographic (EMG) signals during movement regulation expose a reduced dimensionality, attributed to functional neural hierarchies fundamental to coordination (Chvatal & Ting, 2012; Diedrichsen et al., 2009; Todorov, 2006). Coordination and synergies are described similarly in research, though coordination has been considered characteristic of the organization of the body in time and space during task execution, rather than by the internal constraints of the nervous system (Diedrichsen et al., 2009).

Functional variability

Under the outlined theoretical framework, it can be understood that redundancy leads to variability. During the control of movement, variability has been attributed to contrasting sources, each of which are considered in motor control theories. Generalized Motor Program Theory considers movement variability as arising from noise, or error, in selecting appropriate motor programs required for completing a desired task (Dufek, Bates, Stergiou, & James, 1995; Stergiou & Decker, 2011). Variability is thought to decrease as prediction error is minimized due to task-specific practice, leading to increased movement accuracy and efficiency (Harris & Wolpert, 1998; Stergiou & Decker, 2011). In contrast, Dynamical Systems Theory views biological systems as self-organizing within environmental, biomechanical, and morphological constraints, contradicting views of variability as noise and highlighting the functional role of variability in movement repetitions (Barrett et al., 2008; Bartlett et al., 2007; Brown et al., 2012; Davids et al., 2003; Hamill, van Emmerik, Heiderscheit, & Li, 1999; Li et al., 2005; Lohse et al., 2013; Newell et al., 2000; Stergiou & Decker, 2011). From this perspective, sources of variability in human movement have been attributed to both chaotic (deterministic) and stochastic (random) processes, allowing explorative and adaptive behavior (Davids et al., 2003). Trial-to-trial movement variations are considered inherent and functional characteristics of successful performance, rather than indications of movement dysfunction (Barrett et al., 2008; Bartlett et al., 2007; Brown et al., 2012; Davids et al., 2003; Hamill et al., 1999; Li et al., 2005; Lohse et al., 2013; Newell et al., 2000; Stergiou & Decker, 2011).

Commonalities among motor control theories

Expounding upon the functional role of variation in movement repetitions, a variety of theories and hypotheses exist for outlining the purpose and cause of variability in human movement. Optimal Control Theory (OCT), and the updated Optimal Feedback Control Theory, suggests that behavior is optimized using cost functions, which aim at minimizing effort and movement variability (Diedrichsen et al., 2009; Lohse et al., 2013; Scott, 2012; Todorov, 2006). Within this framework, an optimal solution is defined as a control rule, operated by motor commands that distribute work across multiple effectors, thus minimizing the cost function (Diedrichsen et al., 2009; Lohse et al., 2013; Scott, 2012; Todorov, 2006). Despite the minimization of some task-dependent cost function, including smoothness (jerk), accuracy (trajectory errors), muscle activity, or variability, the inherent and functional role of variability is acknowledged in OCT (Todorov, 2006). Rather than defining minimal variability as optimal during movement repetitions, OCT acknowledges the existence of anisotropic variability patterns, varying by direction (Diedrichsen et al., 2009; Lohse et al., 2013; Todorov, 2006). The abstract directionality of movement variation is expressed in the Uncontrolled Manifold Hypothesis (UCM) and is further explained by the minimal intervention principle (Diedrichsen et al., 2009; Lohse et al., 2013; Todorov, 2006).

Within the UCM, variation is suggested to accumulate on task irrelevant (redundant) dimensions, while being corrected on goal-directed dimensions (Diedrichsen et al., 2009; Lohse et al., 2013; Todorov, 2006). Minimal intervention has been experimentally linked with the constrained action hypothesis (CAH), where conscious control has been suggested to interfere with automatic processes, decreasing automaticity and functional variability (Kal, van der Kamp, & Houdijk, 2013; Lohse et al., 2013; Wulf et al., 2001). The concepts of automaticity and functional variability are therefore considered related, where automaticity specifically refers to the control of movement with limited conscious interference of processes that normally regulate movement (Wulf et al., 2001).

Previous research has examined automaticity in the context of movement effectiveness and movement efficiency with an eye toward movement control (Chiviacowsky, Wulf, & Wally, 2010; McNevin, Shea, & Wulf, 2003; Totsika & Wulf, 2003; Wulf, Höβ, & Prinz, 1998; Wulf et al., 2001). Generally, movement effectiveness has been assessed using measures of accuracy, consistency, and balance; while movement efficiency has been assessed using measures of muscle activation (EMG), force production, cardiovascular response, oxygen consumption, movement speed, and endurance. Evidence of improved automaticity has also been examined in the context of movement effectiveness via functional variability, movement fluidity, and regularity (Kal et al., 2013; Lohse et al., 2013). Functional variability has been examined in a variety of settings using a number of analytical techniques, including linear multivariate matrix techniques (matrix determinant and PCA; Lohse et al., 2013; Li, 2006), as well as nonlinear time series analyses (Kal et al., 2013; Stergiou et al., 2006). Collectively, these measures have been used in formulating and testing the CAH (Kal et al., 2013; Lohse et al., 2013; Wulf et al., 2001). Automaticity in research has been additionally considered in the context of increased variability along redundant dimensions, with decreased variability along goal-relevant dimensions, demonstrating agreement with OCT and UCM (Lohse et al., 2013). The outlined motor control theories each consider functional roles for variability surrounding movement repetitions through common mechanisms. Additional work uniting the outlined concepts is therefore worthwhile.

Motor control applications

Applications of OCT and the UCM in research have provided a means of accounting for sources of variability in task repetitions, with the latter used in forming testable hypotheses. A method of describing the structure of movement variability is through correlations among effectors (Diedrichsen et al., 2009). Returning to the discussion on synergies and movement coordination, it can be understood that correlation, or covariance, provide the basis for describing these concepts. From this perspective, the UCM and the associated analytical techniques provide insight into movement synergies across task repetitions (Diedrichsen et al., 2009; Latash, 2010). Defining redundant and goal-directed dimensions can be accomplished using the UCM, $UCM_{\perp}$ (perpendicular) and $UCM_{||}$ (parallel), respectively (Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999). Conceptually, the UCM is an abstract multidimensional space of possible outcomes (Lohse et al., 2013; Scholz & Schoner, 1999), where $UCM_{||}$ can be considered representative of variation that does not affect controlled variables, and can therefore be considered uncontrolled (Scholz & Schoner, 1999). Variability along the $\mathrm{UCM}_{\perp}$ is therefore predicted to be less than that along the $\mathrm{UCM}_{||}$, which can be tested statistically (Scholz & Schoner, 1999). Specifically, statistical methods examining the correlation structure among biomechanical variables across task repetitions have shown findings in line with the UCM, including Principal Component Analysis (PCA; Lohse et al., 2013; Scholz & Schoner, 1999; Federolf et al., 2013).

The utility and methodology surrounding PCA will be discussed in a later section, however, the premise of examining point-by-point correlations among repeated trials of time series variables allows the extraction of unique set of functions relating to modes of variation (Deluzio et al., in Robertson et al., 2014). In addition to the alignment of this statistical analysis with concepts outlined by UCM, the extraction of a limited number of independent principal components from a larger dataset provides interpretations common with movement synergies and functional degrees of freedom (Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Li, 2006). For these reasons, this research used PCA as a means of accessing sources of variability extracted from biomechanical time series variables collected during experimental manipulations of task demands during landing. The outlined methods were used in testing hypotheses related to movement control under a range of mechanical task demands, examining changes in variability among kinematic, kinetic, and electromyographic sources.

Considerations for the physiological processes underlying motor control are justifiably required for a complete understanding of movement regulation. Physiological underpinnings, however, are often considered in isolation from biomechanical examinations and theories from motor control. Nikolai Bernstein, the Russian physiologist, outlined the DOF problem in organizing and coordinating movement, identifying the necessity for simplified control through dimension reduction (Bernstein, 1967, p. 161-164; Turvey, 1990). Separation of the DOF problem from the underlying physiology may, however, limit interpretations in the context of the human system. The equilibrium-point hypothesis, borrowed from motor control, incorporates mechanical and physiological principles that have been suggested to align with concepts outlined in the UCM (Feldman, in Latash, 2010). Specifically, the equilibrium-point hypothesis suggests that movement emerges under the laws of physics (with consideration for the previously outlined constraints), aiming to reach a state of minimum potential energy by way of minimal muscle activation (Latash, 2010). Collectively, the equilibrium point refers to a combination of muscle length and force that minimizes electrical activity in response to the tonic stretch reflex, guiding limb through movement (Todorov, 2006; Latash, 2010). From this perspective, associations linking motor control and physiological processes are outlined, exposing common foundations. As a result, physiological considerations for the control of movement will be discussed in greater detail. The proposed research, though biomechanically driven, seeks to gain insight into the collective control of landing from neuromuscular and mechanical perspectives, necessitating interdisciplinary perspectives.

Physiological considerations for movement variability

In the context of the proposed task, the aim of the following sections is to assess the current understanding of the mechanisms that control muscular contractions during landing. As a result, particular attention will be paid to the neural control of eccentric muscular contractions, which occur during landing, with contrasts to concentric contractions. It was anticipated that highlighting factors controlling muscle tension under eccentric conditions would shed light into human movement control, emphasizing sources of movement variability and allowing inferences to be made with respect to potential injury mechanisms in landing movements.

Eccentric muscular contractions are a necessity of landing, requiring deceleration of the system mass (Bishop et al., 2000; Hedayatpour & Falla, 2012). During eccentric contractions, increases in muscle length occur while concurrently developing tension, increasing the risk of muscle strains and myotendinous injury (Bishop et al., 2000; Fang et al., 2004). Under eccentric conditions, muscles are required to absorb energy, which may be dissipated as heat or may be temporarily stored as elastic potential energy that may be recovered under short latencies and used in a subsequent concentric phase, via the stretch-shortening cycle (SSC; Bishop et al., 2000; Guilhem et al., 2010; Mianfang & Li, 2010).

Contrasts between muscular contraction types, including eccentric (muscle lengthening under active tension), concentric (muscle shortening under active tension), and isometric (tension development without changes in muscle length) have been explored in a number of settings, involving distinct muscles, and under contrasting conditions (Bishop et al., 2000; Fang et al., 2004; Guilhem et al., 2010; Hedayatpour & Falla, 2012). Additionally, comparisons have been made between muscular contraction control with respect to joint torques (angular force), electrical muscle activity (electromyography; EMG), as well as brain activity via EEG (electroencephalography) and fMRI (functional magnetic resonance imaging; Fang et al., 2004; Finucane, Rafeei, Kues, Lamb & Mayhew, 1998; James et al., 2000; Komi, Linnamo, Silventoinen & Sillnpaa, 2000; Olsson, Hedlund, Sojka, Lundstrom & Lundstrom, 2012). The outlined approaches differ with respect to the level of examination, including movement outcomes, local muscular control, and upper level (cortical) control. Given the biomechanical nature of the proposed research, a bottom-up approach will be considered, examining the role of local muscle activity on movement outcomes before discussing cortical control mechanisms. Finally, associations between movement control and injury during eccentric loading will be discussed with task specific applications to landing.

Local eccentric muscular control

Examinations exploring changes in muscle force and joint torque have been carried out under concentric and eccentric conditions, quantifying changes at different joint angles, muscle lengths, and loading rates (Liping, Wakeling, & Ferguson-Pell, 2011). The length-tension relationship dictates the ability of the muscle to generate force, where the number of cross-bridges determines force-generating capacity of muscle following shortening or lengthening of the sarcomere outside of optimal actin-myosin overlap (Liping et al., 2011). It can be understood that changes in muscle length occur as a result of changes in joint angle, therefore joint torques have been explored across joint angles. Under eccentric conditions, the active development of tension in the muscle while concurrently stretching the tissue leads to greater overall muscle force, and subsequent joint torque compared to isometric and concentric conditions (Komi et al., 2000).

During eccentric contractions muscle and tendon become less flexible, developing tension from passive elastic structures, allowing the muscle to carry heavier loads despite lengthening beyond optimal cross-bridge formation conditions (Guilhem et al., 2010). It should be noted, however, that during human movement maximal eccentric torque is usually less than reported from in vitro modeling, and may not exceed peak isometric force (Bishop et al., 2000; Linnamo, Bottas & Komi, 2000). Physiological and neural control mechanisms offer explanations for this phenomenon. Specifically, neural inhibition from feedback via joint receptors, free nerve endings in muscle, pain receptors, and Golgi tendon organs have been suggested to maintain muscle tension within limits that avoid injury (Westing, Cresswell & Thorstensson, 1991).

From a neural perspective, Golgi tendon organs (GTO) have been associated with type Ib afferent neurons, which have been shown to demonstrate increased activity as a result of increases in muscle tension, working via a disynaptic inhibitory pathway (Bishop et al., 2000; Westing et al., 1991). These safety mechanisms act as tension modulators, inhibiting muscle activation during high loading conditions (Bishop et al., 2000; Westing et al., 1991). As a result, the measurement of electrical muscle activity offers a window into neural functioning during muscular control that summarizes the outcome of these controlling factors. Although total muscle activity from EMG does not directly show inhibitory mechanisms controlling movement, synthesis from the body of literature on neural control can be applied to the interpretation of subsequent electrical activity during movement.

Neuromuscular examinations of skeletal muscle contractions typically employ measures of electrical activity using EMG via needles embedded in muscle tissue or surface skin electrodes (Finucane et al., 1998). From EMG analysis, it has been shown that under concentric conditions increases in muscle forces are largely influenced by motor unit (MU) recruitment (Liping et al., 2011). In contrast, eccentric torque is primarily controlled by changes in MU firing rate (Liping et al., 2011). Additionally, motor unit recruitment follows the size principle under loading conditions, where low-threshold motor units (MUs) are recruited prior to high-threshold MUs (Guilhem et al., 2010; Komi et al., 2000). The increased loads in eccentric contractions dictate that larger, high-threshold, MUs are preferentially recruited, generating greater twitch force from the discharge of larger MUs that innervate a greater number of muscle fibers, better distributing mechanical stress (Guilhem et al., 2010; Linnamo, Moritani, Nicol & Komi, 2003). For this reason, type II fibers are preferentially recruited during eccentric contractions, explaining the increase in discharge frequency, as opposed to the number of recruited MUs (Guilhem et al., 2010; Komi et al., 2000; Linnamo et al., 2003).

Overall, measurements of electrical muscular activity during eccentric versus concentric contraction show that eccentric contraction demonstrates lower levels of total (integrated) and mean EMG during contraction (Bishop et al., 2000). Additionally, eccentric muscular work has been shown to produce additional force without a concomitant increase in ATP and oxygen consumption, revealing decreased neural and metabolic cost for muscle (Guilhem et al., 2010). Although total and mean EMG activity are decidedly lower during eccentric contractions, peaks in electrical activity are in fact higher during eccentric work, which is the proposed result of muscle spindles facilitating motoneurons to illicit greater activation, or tension development, as a result of changes in muscle length (Bishop et al., 2000).

In opposition to Golgi tendon organs (GTO), muscle spindles have been suggested to be responsible for the instantaneous peaks in EMG activity during eccentric contractions, where the rapid muscle length changes increase neural activity more quickly than GTO induced inhibition (Bishop et al., 2000). As a result, examinations of peak to mean EMG ratio during eccentric contractions show increased electrical muscle activity, potentially demonstrating contrasting control mechanisms associated with eccentric and concentric muscle actions (Bishop et al., 2000). For this reason, peak to mean EMG ratio has been suggested to provide an indication of muscle activation variability (Bishop et al., 2000). Observed decreases, however, in joint torque variability during eccentric contractions diverged from findings of higher peak to mean EMG ratio (Bishop et al., 2000). It would be assumed that the trend of these measures would converge if measuring the same construct. This concept therefore deserves attention in future research. Particular attention should be paid to the analysis techniques and the manner in which torque variability is assessed. In the literature, torque variability has been shown to both increase and decrease when comparing eccentric versus concentric contractions (Bishop et al., 2000; Fang et al., 2004). The dissociation between these findings seems to be attributed to the manner in which variability is defined and computed.

Although several spinal level mechanisms have been proposed in regulating neural adjustments to muscle activation during contraction, including activation via increased muscle spindle sensitivity during repeated loading via muscle afferents that are sensitive to inflammation substrates and increases in pressure and temperature, neural control should extend to examinations of the motor command center: the brain (Bottas, Miettunen, Komi, & Linnamo, 2009). Inferences drawn from examinations of EMG activity and brain activation patterns during eccentric, concentric, and isometric contractions suggest that skeletal muscle is activated differently in each of these conditions (Fang et al., 2004). In particular, understanding the greater force producing demands of eccentric contractions, it has been revealed that tasks with a higher degree of difficulty are carried out using increased levels of brain activation (Fang et al., 2004). For these reasons, further explorations into the controlling mechanisms underlying muscle contraction have been carried out at the level of the brain.

Cortical eccentric muscular control

The motor system is controlled by the central nervous system, involving complex interactions between central command and peripheral reflexes, controlled at the muscle and spinal level (Fang et al., 2004; Houk, 2010; Olsson et al., 2012). Contention exists with respect to neural control of movement. From a neuroscientific perspective the degree of complexity is much greater at higher levels of control (the brain), compared to lower levels of control (muscle and spine), which dictate movement outcomes (Houk, 2010). From a behavioral motor control perspective, complexity is greater at the lower levels of control, while the brain is thought to modulate movement via prescribed neural controllers (Houk, 2010). In each case, performance emerges from neural computations following the laws of physics, where performance is derived from muscle activity (Houk, 2010).

The neuromuscular system synthesizes central motor commands into muscle forces that allow interaction with the environment. Interestingly, the convergence between the conflicting schools of thought comes at the level of the stretch reflex, which is controlled at the lower levels of the CNS in direct response to the viscoelastic properties of muscle tissue (Houk, 2010). Examinations of EMG activity allow inferences to be made with respect to neural control of movement, though this serves as the final stage of motor control without specific insight into causation. Motor control, modulated via brain functions, is most concretely understood through investigations of brain activity (Fang et al., 2004; Guilhem et al.2010; Olsson et al., 2012).

Electrical brain activity has been explored during eccentric and concentric muscular contractions using EEG (Fang et al., 2004; Guilhem et al., 2010). Additionally, functional magnetic resonance imaging (fMRI) has been used in monitoring specific brain regions that are activated during imagined maximum resistance training, in both concentric and eccentric conditions (Olsson et al., 2012). Although the external validity of the inferences drawn from imagined movements can be questioned, these explorations provide initial insight into higher-level control of human movement, under the limitations of current measurement techniques.

Investigations of electrical brain activity from EEG have demonstrated that cortical electrical potential is greater, and occurs earlier, when controlling submaximal eccentric versus concentric contractions (Fang et al., 2004; Guilhem et al., 2010; Olsson et al., 2012). Prolonged preparation time and increased activation amplitude seem to suggest that the CNS must devote greater control to movements with higher risk of injury, and a higher degree of movement difficulty, implying that control strategies differ between eccentric and concentric movements (Fang et al., 2004; Guilhem et al., 2010). Cortical activity has been implicated in planning and execution of motor activities (Fang et al., 2004). In addition to temporal and amplitude differences, specific areas of brain activity have also been shown to increase when controlling eccentric versus concentric contractions (Fang et al., 2004). These results provide direct evidence of changes in higher-level control of movement under contrasting conditions. Such examinations, however, have typically involved only forearm flexors due to the higher degree of experimental control (Fang et al., 2004; Guilhem et al., 2010). Future examinations should be carried out under applied settings, including running, jumping, or landing tasks.

Concerns with generalizability from examinations of electrical brain activity using EEG are evident, though this measurement technique affords flexibility in conducting measurements under dynamic conditions. In contrast, fMRI measurements must occur under static, or relatively static conditions, where the sampling rate of the acquired data becomes a limitation when compared to EEG measurements (Olsson et al., 2012). Investigations of brain activation using EEG do not, however, provide direct insight into activation of specific brain structures. Rather, electrical activity measured over a number of external locations on the skull is used in providing evidence of electrical activity in the underlying brain tissue (Fang et al., 2004). Functional magnetic resonance imaging (fMRI) provides real-time objective assessment of changes in brain activity within specific brain regions associated with known processes (Olsson et al., 2012). Greater understanding with respect to movement regulation within the brain can be understood, though measurements of brain activity have been restricted to imagined eccentric and concentric contractions (Olsson et al., 2012).

Exploration into specific brain regions involved in controlling muscular contractions under eccentric and concentric conditions identify that pre-frontal cortex activity is greater during eccentric movements (Olsson et al., 2012). In contrast, the motor and pre-motor cortexes show greater recruitment during concentric movements (Olsson et al., 2012). From this, it has been suggested that lower total EMG activity during eccentric contractions may be partly explained by the reduction of activity in these structures (Olsson et al., 2012). Overall, brain regions associated with eccentric contraction control include the pre-frontal cortex, the pre-motor cortex, the inferior frontal lobe and the cerebellum, all of which demonstrate significant differences in terms of total and peak activity when compared to concentric contractions (Olsson et al., 2012).

The pre-frontal cortex is implicated in controlling cognition, including high-level cognitive tasks such as working memory and episodic memory (Olsson et al., 2012). Similar to EEG explorations, this suggests that eccentric contractions appear to be more cognitively demanding, where the movement must be controlled without overloading the musculotendinous complex, resulting in injury (Fang et al., 2004; Olsson et al., 2012). Furthermore, contention exists with respect to the role of Purkinje cells within the cerebellum, which have been proposed to modulate force. The emerging evidence suggests that the cerebellum is recruited to maximize force production during concentric contractions, while the pre-frontal cortex is responsible for modulating force commands during eccentric movements (Olsson et al., 2012).

Neural inhibition at the muscle and spinal level has been discussed, including the role of the GTO. Brain activity reveals that during eccentric contractions there may also be a reduction in activity at the cortical level, particularly in the motor and pre-motor cortex (Olsson et al., 2012). Decreased muscular activity may therefore arise from decreased brain activity in the motor cortex during eccentric contractions, while inhibitory signals have been primarily focused at the spinal level. The overall conclusions that can be drawn from examinations of brain activity using EEG and fMRI show that there are different neural systems underlying eccentric and concentric control.

From a practical standpoint, it may be worthwhile to examine EMG activity while simultaneously identifying activation in brain structures using fMRI. Methodologically this is not possible due to the electrical interference between EMG equipment and the magnetic field used in fMRI (Olsson et al., 2012). For this reason, the inferences that are drawn from research involving neural control of muscular contractions are somewhat limited by the available measurement techniques. Attention to experimental design and developments in measurement techniques will allow strides to made in future investigations along this research path.

Injury and fatigue considerations in eccentric contraction control

Differences between the mechanical demands of eccentric versus concentric contractions on muscle tissue are well documented in the literature (Bottas et al., 2010; Bottas, Nicol, Komi & Linnamo, 2009; Guilhem et al., 2010; Hedayatpour & Falla, 2012; Koh & Herzog, 1998; Linnamo et al., 2000). The high tensile demands of eccentric contractions on muscle fibers induce muscle fiber damage, causing deterioration of the cytoskeleton and local inflammation responsible for Delayed Onset Muscle Soreness (DOMS; Guilhem et al., 2010; Hedayatpour & Falla, 2012; Bottas et al., 2010). Specific structural damages occur at the cellular level, resulting in a disorganization of sarcomeres via Z-line disruption, dilation of the transverse tubule system and fragmentation of the sarcoplasmic reticulum (Guilhem et al., 2010). These structural damages can be partially explained by the preferential recruitment of fast motor units (MUs), which innervate a greater number of muscle fibers and cause more forceful contractions that are associated with tissue damage (Bottas et al., 2009; Guilhem et al., 2010).

The preferential recruitment of fast twitch fibers (particularly type IIb/x) and the heterogeneous morphology and architecture of muscle fibers has been associated with disproportionate changes in muscle activity and load distribution on joints (Hedayatpour & Falla, 2012). As a result, non-uniform adaptations occur to muscle fibers depending on their location within skeletal muscle (Hedayatpour & Falla, 2012). Eccentric muscular contractions have been proposed to elicit changes in serial sarcomere number within muscle fibers, though disagreement is evident in the literature (Guilhem et al., 2010; Koh & Herzog, 1998). At best, an increase in the number in sarcomeres within a muscle fiber is minimal, while eccentric training has demonstrated increases in muscle mass and pennation angle, allowing greater force production (Guilhem et al., 2010; Koh & Herzog, 1998).

Repeated eccentric repetitions or training sessions lead to adaptations, including increased active and passive stiffness of the musculotendinous system, which may increase the susceptibility to injury under lengthening conditions observed during eccentric activity (Guilhem et al., 2010). Adaptations have been documented via performance changes following a ‘repeated bout’ effect, where muscle damage, loss of strength, and DOMS effects have been shown to decrease in subsequent eccentric training sessions (Guilhem et al., 2010). Changes in performance as a result of repeated eccentric contractions have been explored in terms of structural changes, as well as the influence of subsequent neural changes. Increased passive stiffness has been associated with muscle soreness and sensitization of small muscle afferents, resulting in perceived pain (Bottas et al., 2010). The influence of pain on performance has been directed toward antagonist muscle activity during voluntary movements, where fatigue-induced decrements of antagonist muscle proprioception may result in altered muscle activation patterns; specifically, painful muscles demonstrate increased antagonist muscle activity (Bottas et al., 2010; Bottas et al., 2009).

Eccentric induced muscle damage has been attributed to changes in agonist neural activity via reduced gamma co-activation, which overrides increased muscle spindle stretch response, decreasing overall electrical muscle activity (Bottas et al., 2010). Decreases in agonist muscle activity following eccentric fatigue have also been explained from muscle mechano-nociceptors, detecting pain and inhibiting muscle activation through pre-synaptic inhibition of spinal level afferents, changing c-motoneuron and muscle spindle afferent excitability (Bottas et al., 2010; Bottas et al., 2009). Neural activation of muscle tissue is also influenced by the accumulation of metabolites during contraction. Preferential recruitment of Type II fibers, working under anaerobic (non-oxidative) conditions results in the accumulation of local metabolites, reducing extra-cellular pH, increasing K^+ permeability in the muscle fiber from ATP/Ca^{2+} dependent K^+ channels, increasing excitation threshold and decreasing muscle fiber excitability (Hedayatpour & Falla, 2012). As is apparent from the examination of changes in neural activity following repeated eccentric repetitions and training sessions, a complex interaction between metabolic, structural, and anatomical factors occurs within active muscle. Overall, increased antagonist activity and decreased agonist/synergist activity have been observed under eccentric fatigue induced conditions. Understanding control mechanism underlying changes in EMG activity from fatigue provides considerations for changes in motor control under eccentric conditions.

Eccentric neural control applications

Physiological considerations for movement control highlight neural control mechanisms, with specific attention directed toward eccentric muscular contractions associated with landing. Typical investigations of neural activity are carried out via EMG measurements of the active muscles involved in controlling movement (Bishop et al., 2000; Bottas et al., 2009; Fang et al., 2004; Guilhem et al., 2010; Mianfang & Li, 2010; Yeadon, King, Forrester, Caldwell & Pain, 2010). Isolated examinations of eccentric contractions in muscles surrounding the knee and elbow joints have been explored in the literature through the use isokinetic dynamometry (constant movement velocity), which allows a high degree of experimental control (Bottas et al., 2009; Fang et al., 2004; Guilhem et al., 2010). Much has been gained from these examinations in terms of the mechanisms controlling eccentric contractions, however, the aim is to apply this knowledge to more naturally occurring human movements, including landing.

Examination of eccentric contraction control during landing seeks to better understand mechanisms controlling movement, and the high incidence of injury during these movements (Fang et al., 2004; James et al., 2000; Mianfang & Li, 2010; Yeadon et al., 2010). Additionally, neural control mechanisms are thought to be the source of the variability in movement outcome measures. As outlined, movement variability has been examined as a means of evaluating movement function and adaptation, where the ability of the motor system to vary, or broadly distribute, internal loads is thought to reduce the risk of injury and increase adaptation to a wider array of stimuli (Bartlett et al., 2007; James et al., 2000). Muscle proprioception has been investigated under eccentric conditions, where proprioception has been suggested to play an important role in optimizing motor control, though intensive eccentric contraction has been shown to impair the sense of position, which has implications on neural activity and force production in the muscle (Bottas et al., 2009).

The influence of landing height has been examined with respect to EMG activity in active muscles controlling movement, as well as the inclusion of a horizontal approach run (James et al., 2006; Mianfang & Li, 2010). From this, it has been shown that average EMG activity is largely dependent on the level of pre-activation in active muscles. Muscle pre-activation has been shown to increase average EMG activity during the eccentric contraction phase of landing (James et al., 2006; Mianfang & Li, 2010). Increased average EMG activity during the eccentric phase may allow force to be distributed across a greater number of muscle fibers, rather than showing the larger peaks in EMG activity from preferential recruitment of larger, faster MUs, leading to greater risk of injury (Guilhem et al., 2010). Overall, increases in drop height have not demonstrated concomitant increases in muscle activity, likely as a result of neuromuscular inhibition, the mechanisms of which have been discussed previously (James et al., 2006; Mianfang & Li, 2010). In addition, co-activation of agonist and antagonist muscles prior to landing has been shown to be a necessary precursor to successful landings from increased landing heights (Yeadon et al., 2010). Examining the influence of muscle pre-activation prior to landing may also be worthwhile in understanding subsequent movement variability measures upon landing.

Although the eccentric phase of landing can be understood from the concepts discussed previously, landing typically involves an eccentric phase followed by a concentric phase in rapid succession (Fang et al., 2004; Mianfang & Li, 2010). The goal of eccentric movements prior to concentric is to make use of the stretch-shortening cycle (SSC), where rapid stretching during the eccentric phase evokes a burst in spindle afferent input providing increased electrical activity during the subsequent concentric phase, under short latencies (Bishop et al., 2000; Mianfang & Li, 2010). The time between the eccentric and concentric phase is critical in making use of the stretch reflexes, which has demonstrated a synergistic relationship with pre-activation of muscle prior to landing (Mianfang & Li, 2010). From this perspective, understanding eccentric control mechanisms is important in understanding the contributions to concentric performance, particularly in movements requiring maximum velocity or power. The body of evidence examining eccentric control and the contrasts between eccentric and concentric contractions provide connections with, and explanations of, the sources of movement variability during landing. This provides an avenue for future research.

The body of literature examining eccentric muscular contraction control mechanisms demonstrates considerable depth and breadth, covering investigations of great experimental control during isolated movements, imagined movements, as well as in applied settings. Connections must be made from disciplines including neuroscience, motor control, physiology, and biomechanics in comprehensively covering this topic. Although the previous sections by no means provide an all-inclusive investigation of the literature exploring muscle physiology, distinct areas of overlap have been highlighted. Additionally, gaps in the current understanding of movement regulation and limitations in experimental procedures have been discussed. The aim of this research was to synthesize the understanding of neural control mechanisms during eccentric contractions, with particular attention to associations among variability surrounding neuromuscular input, movement kinematics, and kinetics. Electromyography offers a window into the neural activity controlling movement, though this information still provides a summary of the complex interactions between higher (brain) and lower (muscle and spinal) levels of control. It is hoped that future research, following the current investigation, will bridge the gap between movement control and potential injury mechanisms, examining the role of human movement variability from interdisciplinary perspectives.

Movement variability and performance outcomes

A consequence of the seemingly infinite number of body segment arrangements (DOF) available to a performer is the ability to select from a large pool of motor solutions in carrying out a movement (Caster & Bates, 1995; James & Bates, 1997; James et al., 2003). The inherent variability surrounding task repetitions has been discussed, however, means of describing and classifying observed outcomes requires attention. Performance of a motor task necessitates the implementation of a movement pattern under the influence of the constraints previously outlined. Selection of an appropriate movement pattern is of particular interest in understanding movement regulation under altered task demands. Previous research has used experimental manipulations of external loads as a means of implementing predictable changes to external forces (Caster & Bates, 1995; James et al., 2003; James et al., 2014). The proposed model for characterizing observed outcomes following manipulations to system mass has undergone modifications, though the premise remains the same (Caster & Bates, 1995; James et al., 2003; James et al., 2014). Mechanical and neuromuscular adjustments during movement allow external forces to be altered, as expressed through performer strategies. Subsequently, a movement strategy has been operationally defined as a neuro-musculoskeletal solution of a motor task, uniquely selected by an individual under the presence of a given set of morphological, biomechanical, and the environmental constraints (James & Bates, 1997; James et al., 2003). The model for classifying biomechanical response strategies will therefore be outlined, making associations to human movement variability examinations, as well as interpretations for understanding potential injury mechanisms in landing.

Movement variability and performer strategies

Performer strategies have been examined in the context of both landing and gait (Caster & Bates, 1995; James et al., 2003; James et al., 2014). Due to the focus of this research, landing remains the task under consideration. Initial research investigating alterations in ground reaction force (GRF) magnitudes under altered task demands was carried out via landing height manipulations, identifying trends toward increased vertical forces with increased landing height (Dufek & Bates, 1990). Segment configurations at ground contact, however, have highlighted individual performer strategies, with attention directed toward the role of the knee joint in modulating landing stiffness (Dufek & Bates, 1990). The ability of landing stiffness, as well as foot position at contact, to modify the observed patterns and magnitudes of force application expose emergent strategies within and between performers (Dufek & Bates, 1990; Dufek & Bates, 1992). The explicit, or implicit, selection of a movement strategy has therefore been associated with modifications to the structure of the motor program controlling the movement (Dufek & Bates, 1990). From these initial observations, follow-up analyses have classified landing strategies into mechanical and neuromuscular responses as evidenced through changes in

GRF and lower extremity EMG activity (Caster & Bates, 1995). A neuromuscular strategy was defined as a decrease or absence of change in impact force with the addition of mass, through a modification of the baseline neuromuscular activation pattern (Caster & Bates, 1995). Conversely, a mechanical strategy was defined as an increase in force with the addition of mass, while lacking a change in neuromuscular activity (Caster & Bates, 1995). Furthering these assertions, James, Bates, and Dufek (2003) outlined a more comprehensive model explaining impact force accommodation in response to increases in an applied stressor. The model was experimentally tested under mass manipulations, though the expected outcomes extend to stressors including landing height, technique, fatigue, postural variation, and previous injuries (James et al., 2003).

Within the continuum of available movement strategies, an individual may completely accommodate or completely ignore an applied stressor (James et al., 2003). The predicted outcomes therefore include Newtonian, biomechanical (negative and positive), and neuromuscular responses, where the level of neuromuscular accommodation determines the type of biomechanical response (Figure 1; James et al., 2014).

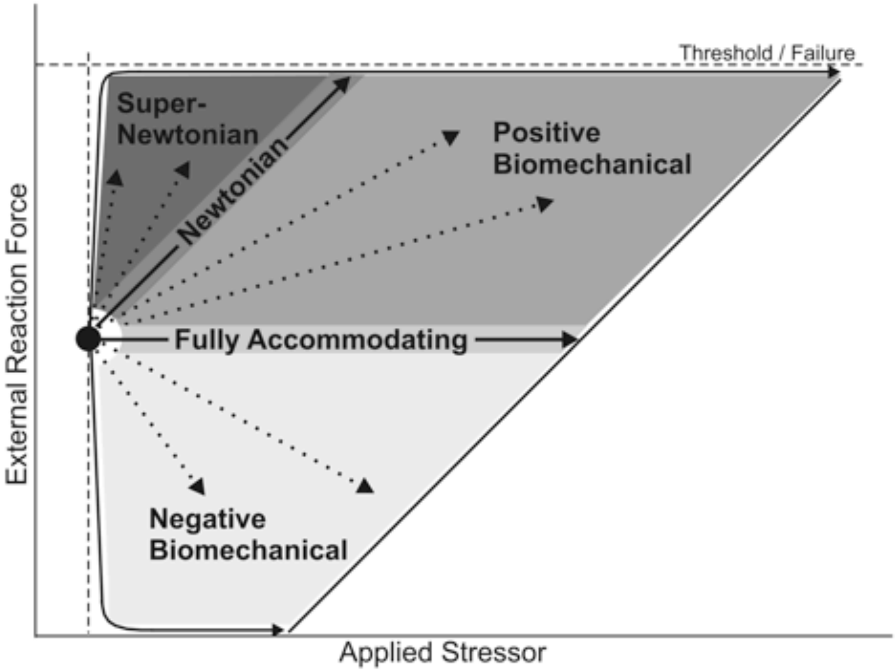


Figure 1: Load accommodation strategies model. (James et al., 2014)

The most straightforward classification is a Newtonian response, where impact forces increase linearly with the applied stressor, as outlined by Newton's 2nd Law (F = ma; James et al., 2003). Although the increase in mass can be regarded as influencing system energy (gravitational potential; PE = mgh), which provides momentum prior to ground collision (p = mv), in each case a linear relationship exists between mass and the predicted force (F = ma; James et al., 2003). The same logic can be extended to increases in landing height, where an increase in height causes a linear increase in system energy (gravitational potential; PE = mgh), producing a predictable increase in impact force. As a result, the Newtonian response would be that predicted relative to a baseline condition, with the addition of mass or height causing a predictable increase in impact force (James et al., 2003). The prediction relative to a baseline condition therefore implies that during a Newtonian response the same motor program is used, maintaining the same movement pattern and providing no neuromuscular accommodation (Caster & Bates, 1995; James et al., 2003). Positive and negative biomechanical responses therefore refer to the rates of increase (or decrease) of the impact force relative to the applied stressor (James et al., 2003). A positive biomechanical response describes a positive increase in impact force at a rate less than that of the applied stressor, indicating partial neuromuscular accommodation (Caster & Bates, 1995; James et al., 2003). A negative biomechanical response shows a decrease in impact force despite an increase in the applied stressor (James et al., 2003). Separating positive and negative biomechanical responses is the complete absence of change in the external impact force with an increase in the stressor, termed a fully accommodating response (James et al., 2003). Further refinement of the theoretical model also predicts a Super-Newtonian response, where the impact force exceeds that predicted by the Newtonian increase in mass (James et al., 2014).

The aim of this research was not to exhaustively capture the range of responses during landing height and mass manipulations, but was instead to examine the interrelatedness between movement variability alterations and emergent movement strategies when undergoing modifications to load and landing height, each altering system energy. These concepts are considered representative of similar underlying constructs, where fewer available options (functional DOF) under increasing task demands may have implications for injury susceptibility (James et al., 2000; James in Stergiou, 2004, p. 29-62). Lesser variability may be considered representative of fewer emergent strategies, with the response strategy classification having potential implications for understanding injury mechanisms (Caster & Bates, 1995; James & Bates, 1997; James et al., 2000; James in Stergiou, 2004, p. 29-62). In particular, accommodation to stressors that mitigate external forces requires neuromuscular input, with mechanical and physiological limitations bounding the outlined model (James et al., 2003). Understanding links between performer strategies and movement variability should therefore be considered in the context of injury.

Movement variability, performer strategies, and injury

The concepts of movement variability and performer strategies are unavoidably linked. Understandably, the selection of contrasting strategies across movement repetitions results in trial-to-trial variability. Explicitly, differential responses to the same treatment (stressor) are used in defining movement strategies (James & Bates, 1997).

The available biomechanical and functional DOF have been attributed to sources of between and within-subject variability (James & Bates, 1997), though the links between changes in performer strategies alongside changes in trial-to-trial variability can be more clearly connected. Movement strategy selection has been examined experimentally under altered task demands, with the previously outlined model for biomechanical response strategies providing a framework for understanding responses to applied stressors (James et al., 2003). Caster and Bates (1995) speculated that neuromuscular response strategies might be the result of perceived danger to the system with the applied stressor, further explaining, however, that the definition of a neuromuscular response might not be limited to a protective response. In either case, accommodation occurs via adjustments to the movement pattern, necessitating a greater neuromuscular response as a means of accommodating the increased load, even if only partially (Caster & Bates, 1995).

Imposing constraints on the biomechanical system through neuromuscular activity conceivably limits the emergent strategies, but still theoretically includes each of the outlined load accommodation strategies. As the stressor increases, however, the ability of the neuromuscular system to achieve negative biomechanical or fully accommodating response is plausibly exceeded. Within the tolerable limits, beyond the ability to fully accommodate and before failure (injury), positive biomechanical responses would be predicted, limiting the observed movement outcomes and decreasing movement variability as expressed through a narrowing variability bandwidth. This testable hypothesis outlines a scenario in landing where at the upper limits of the presented task demands (added load and landing height), participants are selectively (implicitly or explicitly) constrained to positive biomechanical strategies within a limited variability range. Understandably, as mechanical task demands increase (external forces), decreased variability results in repetitive loading, which may have implications for injury mechanisms (James et al., 2000).

Associations between variability and injury have remained largely anecdotal, however, the role of variability in musculoskeletal health has been logically considered in the context of factors influencing stressor magnitudes, frequency, and rates of adaptation (James, in Stergiou, 2004, p. 29-62). Functional variability limits have proposed hypotheses implicating acute and overuse injury mechanisms (James, in Stergiou, 2004, p. 29-62). Specifically, failure of the musculoskeletal system has been attributed to acute injury as a result of a single traumatic loading event where the failure threshold of the tissue is exceeded, or where repeated loadings of sufficient magnitude and frequency outpace physiological adaptations (James, in Stergiou, 2004, p. 29-62). Decreased variability is considered characteristic of the latter, where the accumulation of trauma over time outlines the variability and overuse injury hypothesis (Figure 2; James et al., 2000; James, in Stergiou, 2004, p. 29-62).

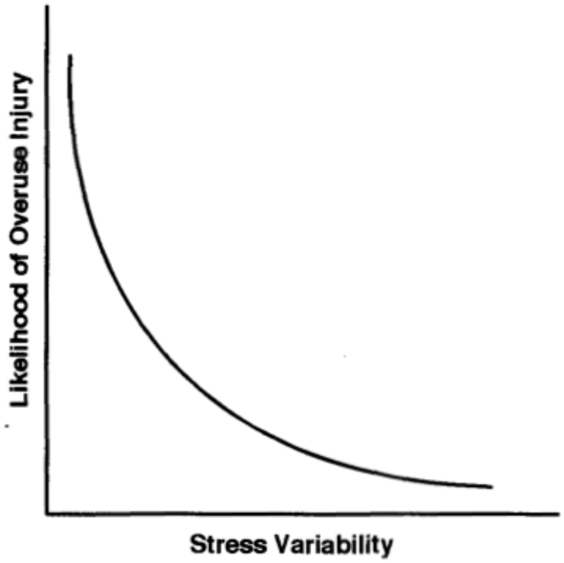


Figure 2: Theoretical relationship between stress variability and overuse injury.

(James, in Stergiou, 2004, p. 37)

Broader distribution of loads among tissues, including location, direction, magnitude, frequency and time may therefore limit the deleterious effects of repeated loadings, allowing time for adaptation (James et al., 2000; James in Stergiou, 2004, p. 29-62). Necessity for the rates of remodeling to exceed the rates of damage can be logically understood, allowing resorption of damaged tissues and deposition of healthy tissues (James in Stergiou, 2004, p. 29-62). Although the proposed research does not aim to account for internal loads or tissue level changes, making associations between models of overuse injury and trial-to-trial variability may provide useful interpretations for the hypothesized alterations in movement variability across changes in task demands.

Movement variability considerations in landing

Landing from a jump or an elevated surface has been identified as a prevalent mechanism of injury. Investigations of isolated biomechanical variables have been extensively covered in the literature, exploring potential injury mechanisms during landing. Variables under consideration have included ground reaction forces, electrical muscle activity patterns, joint moments and ranges of motion, shear and compressive forces, as well as corresponding rates of change and energy absorption/dissipation at the hip, knee, and ankle joints (Liebermann & Goodman, 2007; McNitt-Gray, 1993; Mills, Pain, & Yeadon, 2009a; Torry et al., 2011; Yeow, Lee, & Goh, 2009; Zhang et al., 2000). Changes among these variables have been explored in both real-world examinations and simulations, under single and double leg conditions, investigating landing asymmetries, preexisting injury/pathology, feedback, landing height and direction, surface and footwear characteristics, limb dominance, muscular strength ratios, lower limb stiffness, shock attenuation, and fatigue (Ali, Robertson, & Rouhi, 2014; Coventry, O’Connor, Hart, Earl & Ebersole, 2006; Decker, Torry, Wyland, Sterett & Steadmann, 2003; Devita & Skelly, 1992; Dufek & Bates, 1992; Gehring, Melnyk & Gollhofer, 2009; Hagins, Pappas, Kremenic, Orishimo, & Rundle, 2007; Iida, Kanehisa, Inaba, & Nakazawa, 2011; James et al., 2006; James, Scheuermann, & Smith, 2010; Kellis & Kouvelioti, 2009; Liebermann & Goodman, 2007; McNitt-Gray, 1993; Mills et al., 2009a; Mills, Pain, & Yeadon, 2009b; Milner, Westlake, & Tate, 2011; Niu, Wang, He, Fan, & Zhao, 2011; Peng, Kernozek, & Song, 2011; Schmitz et al., 2007; Schot, Bates, & Dufek, 1994; Shultz, Schmitz, Tritsch, & Montgomery, 2012; Wikstrom et al., 2008). Despite this work, a comprehensive understanding of the mechanisms controlling and predicting injury has yet to be reached. Additionally, rates of injury have shown little change as a result of this work.

Motivations for continued investigations into links between variability and injury include the high rates of injury, with attention directed to lower extremity joints. In particular, the anterior cruciate ligament (ACL) of the knee has been of interest in landing studies due to the high incidence of injury. In the United States, more than 1 in 3000 individuals per year experience ACL injury, with as many as 70% of ACL injuries arising from non-contact mechanisms, totaling approximately $3 billion in associated medical care (Decker et al., 2003; Delahunt et al., 2012; Padua et al., 2011; Torry et al., 2011). In terms of specific populations, a combined 88% of basketball and soccer injuries identified the mechanism of injury as occurring during deceleration type movements, where 41% of these injuries were reported to have occurred in landing (Decker et al., 2003). Additionally, 48% of injured female basketball players identified the mechanism of injury as due to landing, with 40% of elite volleyball players experiencing knee problems during their competitive careers (Zhang et al., 2000). Gender comparisons have been a major focus in the literature due to the higher incidence of knee injuries among female participants in sport, specifically at the ACL (Decker et al., 2003; Delahunt et al., 2012; Gehring et al., 2009; Hashemi et al., 2011; Schmitz et al., 2007; Torry et al., 2011). Gender differences among kinematics, kinetics, and muscle control strategies at the knee joint have been demonstrated during landing, where a more erect landing posture has been suggested to increase knee extensor moments, predisposing females to knee injuries (Decker et al., 2003; Gehring et al., 2009; Torry et al., 2011). Unfortunately, connections between knee extensor moments and shear forces have not demonstrated correspondingly high in vivo tibial translations, implicated in ACL injury, suggesting that more work can be done in understanding gender specific injury mechanisms (Torry et al., 2011).

In addition to examinations of specific populations, the extent and distribution of damage to cartilage from repetitive landing, leading to ACL failure, is poorly understood. Unfortunately, as many as 83% of ACL injured individuals will experience osteoarthritis later in life, as well as a high incidence of re-injury, or injury of the contralateral limb (Padua et al., 2011; Yeow, Cheong, Lee, & Goh, 2009). The ACL serves to provide mechanical stability to the knee joint, resisting tibial translation, and providing sensorimotor control and proprioception (Delahunt et al., 2012). Epidemiological examinations of injuries to the lower extremity have identified the knee as the most commonly injured site, associating the cause of injury to joint laxity, including specific deficits in quadriceps strength, quadriceps to hamstring ratio imbalances, as well as neuromuscular factors such as impaired proprioception and co-activation of muscles surrounding the knee joint (Dufek & Bates, 1992; Gehring et al., 2009; Hashemi et al., 2011). Additional factors include knee joint position during landing, including application of force when the knee is near full extension, as well as the surface interaction angle between the tibia and femur (Hashemi et al., 2011). Overall, the high rates of injury, specifically at the knee, as a result of deceleration and landing suggests that greater understanding of movement control is warranted, though examination of individual biomechanical variables may be of lesser importance than patterns of coordination during movement. Examinations of human movement variability offer promise in providing greater understanding of movement control, allowing inferences to be made regarding movement function and subsequent likelihood of injury.

Experimental evidence for changes in movement variability with changes in task demands have been shown in research, highlighting contrasting variability responses to changes in landing height between healthy and injury prone groups (James et al., 2000). Positive linear increases between joint kinetics and kinetic variability have been shown in research, though divergent rates of increase have also been observed (James et al., 2000). Equivocal findings among variables deserve additional attention in research, where the selected variability metric may provide contrasting trends and interpretations. A number of avenues exist to quantify variability, with considerable advances having been made in contemporary research. The following section summarizes the methods of variability expression used in previous research, outlining distinctions, advantages, and limitations of the methods used in this investigation.

Movement variability measurement methods

Quantifying and interpreting movement variability offers considerable freedom to the researcher, depending on the nature of the examination, as well as the type of movement that is being considered. The selection of an appropriate variability metric has implications on the interpretation of movement variability, particularly when comparing the results from contrasting studies (DiBerardino et al., 2010; O'Dwyer et al., 2009). Traditional examinations of movement variability have expressed variability using linear measures including standard deviation, providing a measure of centrality or deviation from the mean, expressed in the same units of the original variable (Stergiou & Decker, 2011). Standard deviation provides a sense of the absolute magnitude of changes in variability, and allows assessment across time, but comparisons between contrasting variables is limited (Stergiou & Decker, 2011). Additionally, measures of centrality view deviations from the mean as error, assuming that variations between repeated tasks are random and independent of the preceding and following repetitions (Stergiou & Decker, 2011; Stergiou et al., 2006). Movement variability investigations, however, have shown distinctions from noise, oftentimes demonstrating fractal structure (DiBerardino et al., 2010; Li et al., 2005).

Coefficient of variation is commonly used in the literature, providing a measure of standard deviation normalized to the mean of the scores in the distribution, expressed as a percentage (Brown et al., 2012; O'Dwyer et al., 2009). The use of coefficient of variation provides an easily interpretable measure of variability, but is sensitive to outlying or extreme data points, particularly for mean values near zero, and cannot be readily interpreted if the data has undergone statistical transformation (Brown et al., 2012; O’Dwyer et al., 2009). Additionally, it has been suggested that traditional measures using coefficient of variation are composed of variability from offset values and waveform variability (O’Dwyer et al., 2009; Bartlett, Bussey, & Flyger, 2006). Here, offset variability is influenced by motor performance repeatability and measurement error, while waveform variability is more directly representative of performance repeatability (O’Dwyer et al., 2009; Bartlett et al., 2006). Removal of offset, or variability derived from measurement error has been suggested as a means of improving linear variability measures (O’Dwyer et al., 2009). Overall, linear measures of variability have effectively been used in the literature as a means of quantifying magnitude variation in a time series (O’Dwyer et al., 2009; Stergiou & Decker, 2011; Stergiou et al., 2006).

In contrast, non-linear measures of variability including approximate entropy, sample entropy, correlation dimension, detrended fluctuation analysis, and largest Lyapunov exponent have been shown to be valuable in the expression of variability for cyclic movements (DiBerardino et al., 2010; Federolf et al., 2012; Kal et al., 2013; Stergiou & Decker, 2011; Stergiou et al., 2006). These non-linear measures of performance variability quantify the temporal structure of variability in a time series, which has been interpreted in the context of movement fluidity and regularity (Federolf et al., 2012; Kal et al., 2013; Stergiou & Decker, 2011; Stergiou et al., 2006). The largest Lyapunov exponent has been used throughout the literature as a means of drawing conclusions about the temporal structure of trajectories in a time series, quantifying the rate of change of a waveform shape over time, where smaller values represent decreased variability or greater predictability (Federolf et al., 2012; Stergiou & Decker, 2011; Stergiou et al., 2006). Nonlinear methods of variability analysis are therefore differentiated from solely examining variability among magnitude changes in biomechanical variables. Along these lines, phase-portrait representations of segment motion during gait provide dynamic depiction of motion via plots of position versus velocity, or angle-angle plots (DiBerardino et al., 2010). Phase-portraits offer the benefit of representing continuous gait cycles of individual body segments, allowing Elliptical Fourier Analysis of the phase-portrait, quantifying the statistical area and path length over multiple gait cycles (DiBerardino et al., 2010). Unfortunately, phase-portraits remove temporal information from examinations of variability, limiting subsequent inferences regarding the spatial and temporal origins of movement variability (Wagner, Pfesterschmied, Klous, von Duvillard, & Muller, 2012). Overall, non-linear methods of analysis show promise for cyclic movements, though limitations and weaknesses have been identified with respect to noise sensitivity (Federolf et al., 2012). Given the focus of on landing, additional means of analysis have been considered.

Principal Component Analysis

Principal component analysis (PCA) has shown utility in contemporary research as a means of reducing multivariate datasets into a smaller subset of independent sources of variation (Brandon et al., 2013; Cohen, 2014; Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Molenaar et al, 2013; Robbins et al., 2013; Richter et al., 2014b). Typically, biomechanical analyses extract a limited number of data points from a time series, overlooking patterns of change and potentially neglecting useful information (Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Donoghue et al., 2008; Richter et al., 2014b). As a result, relevant features of a movement may be overlooked. PCA has been considered an unbiased (statistically driven) means of reducing redundant information from a multi-dimensional signal, where similarities among signals may be detected using covariance (or correlation), allowing more comprehensive and efficient analyses (Brandon et al., 2013; Daffertshofer et al., 2004; Donoghue et al., 2008; Federolf et al., 2013; Richter et al., 2014b).

The purpose of the discussion surrounding the use of PCA in biomechanical investigations was not to provide an exhaustive and quantitative outline of the mathematical procedures underlying PCA, but to provide a general framework, leading into associated interpretations and limitations. Briefly, PCA involves iteratively extracting principal components (PCs) from a covariance matrix (Cohen, 2014; Donoghue et al., 2008; Deluzio et al., in Robertson et al., 2014). The use of PCA in biomechanical research has analyzed data in the form of a time series, with each trial normalized to a fixed number of data points (i.e. n = 101: 0-100% movement phase; Deluzio et al., in Robertson et al., 2014). A matrix consisting of time series trials is therefore created, inputting a trial in each row, with each column representing a point in time. A covariance matrix is then computed (n x n) and transformed, aligning PCs with directions of variation in the dataset, providing eigenvectors and eigenvalues that are used for subsequent interpretation (Deluzio et al., in Robertson et al., 2014). Eigenvalues express the proportion of explained variance in each PC, while eigenvectors (PC loading vectors) depict the pattern of variation that is captured by each PC across the time series (Deluzio et al., in Robertson et al., 2014). The ranking of PC eigenvalues corresponds with the proportion of explained variance in the dataset therefore a large proportion of the variability is accounted for in a small number of PCs (Cohen, 2014; Donoghue et al., 2008; Deluzio et al., in Robertson et al., 2014; Richter et al., 2014a). PCs are typically retained by some criterion value of explained variance or eigenvalue (i.e., greater than 90% explained variance or eigenvalue > 1.0; Brandon et al., 2013; Cohen, 2014; Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Field, 2009; Kipp, Redden, Sabick, & Harris, 2012; Robbins et al., 2013). PC scores can also be computed, representing linear combinations between the original variables and PC loading vector coefficients (Deluzio et al., in Robertson et al., 2014). PC scores have been used during hypothesis testing, examining differences between populations or experimental conditions through conventional parametric statistical tests (Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013).

Extending beyond the technical aspects surrounding PCA, its use has been shown to be effective among a number of data sources, including kinematics, kinetics, EEG, and EMG (Daffertshofer et al., 2004; Li, 2006; Kipp et al., 2014; Kipp et al., 2012; Molenaar et al., 2013). Collectively, the reduced subset of parameters (PCs) extracted from PCA has been considered representative of concepts in line with theories from motor control (Li, 2006; Todorov, 2006; Lohse et al., 2013). Specifically, PCA has been suggested to provide insight into the control of the many available DOF that are compressed along a solution space that is controlled by a smaller subset of units, often described using functional DOF and synergies (Li, 2006). Given the mathematical definition of a synergy as a systematic correlation among effectors, it can be understood that that underlying basis of PCA, using correlation or covariance (Kipp et al., 2014; Kipp et al., 2012; Wang et al., 2013), provides interpretations for the control of movement (Daffertshofer et al., 2004; Federolf et al., 2013).

Collapsing a multivariate dataset into a smaller subset of functional units from the underlying correlation structure provides information pertinent to movement coordination, synergies, and functional variability. Motor variability is more widely recognized as containing both random noise as well as regularities that are considered functional (Dafertshofer et al., 2004; Lohse et al., 2013). The use of PCA as a means of filtering signals into stochastic and deterministic components has been suggested from dynamical systems theory perspectives, potentially highlighting sources of variation that are more tightly controlled and those that are free to vary (Daffertshofer et al., 2004; Davids et al., 2003; Federolf et al., 2013). Distinctions between sources of variability are considered in the context of a number of motor control theories that have been previously outlined. Associations between the outcomes of PCA and motor control theories, include OCT and UCM, as well as minimal intervention and CAH. Specifically, greater explained variance among fewer PCs has been considered reflective of fewer controlled units during movement regulation (Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999). The subset of PCs expressing a large proportion of the variance in a dataset has been associated with distinctions between the controlled UCM$_{\perp}$ and the uncontrolled UCM$_{||}$. Presented in a less abstract manner, greater explained variance among a smaller number of PCs may demonstrate tighter regulation of a movement variable, or set of variables, during trial-to-trial repetitions. Additionally, PC loading vectors plotted across the movement phase allow identification of the contribution from each PC to the observed movement pattern (Deluzio et al., in Robertson et al., 2014). Further, PC scores allow differences to be assessed across conditions or populations, where PC scores are expressed relative to the mean waveform; high PC scores greater than the mean and low

PC scores lesser than the mean (Deluzio et al., in Robertson et al., 2014). Collectively, multiple approaches are available in the use of PCA, including the simultaneous inclusion of multiple variables (i.e. kinematics, kinetics, EMG), detecting differences between conditions or populations, as well as independent analysis for specific sources of variation within a variable. These approaches were combined in this research, examining various levels of control in the lower extremity

Limitations are acknowledged in the use of PCA in research. As with any statistical procedure, assumptions must be met prior to its use. The assumptions surrounding PCA are generally outlined in the context of Pearson product moment correlations due to the underlying foundations in correlation and covariance (Lund & Lund, 2014; SAS Institute Inc., 1989). The assumption of sampling adequacy is generally overcome in biomechanical analyses with the inclusion of multiple trials, participants, and variables. Application of PCA to biomechanical time series data ensures that each variable is continuous, with significant outliers unlikely from any individual data point. PCA assumes the data under consideration is suitable for reduction (Lund & Lund, 2014), which falls in line with the concept of functional degrees of freedom drawn from motor control (Li, 2006). Finally, the assumed underlying linearity associated with PCA may limit its applications as well as conclusions drawn from research using PCA (Li, 2006; Lohse et al., 2013; Molenaar et al., 2013). Non-linear equivalents to PCA may provide applications in future research (Molenaar et al., 2013). Although the application of PCA to biomechanical data is unconventional, its use generally conforms to the underlying assumptions (Deluzio et al., in Robertson et al., 2014). Additional limitations of PCA on human movement data include the temporal and magnitude normalization procedures that are required prior to analysis (Federolf et al., 2013). Given the applications of PCA to datasets requiring reduction, this technique has been incorporated into contemporary biomechanics research. Additionally, extensions beyond PCA are already emerging in the biomechanics literature. Due to some of the outlined limitations, alternative methods of analysis have been proposed.

Functional data analysis (FDA) views an entire sequence of measurements as a single function, rather than as discrete data points in time (Coffey, Harrison, Donoghue, & Hayes, 2011; Deluzio et al., in Robertson et al., 2014). FDA therefore acknowledges the dependence among adjacent data points, which has been cited as an improvement in the analysis of human movement data (Coffey et al., 2011; Donoghue et al., 2008). Additionally, frequency domain PCA has been used during balance assessments, with the added benefits of evaluating PCs across a given frequency range rather than as an average number of PCs for the total signal (Molenaar et al., 2013; Wang et al., 2013). Limitations of PCA are acknowledged, as well as alternative methods of analysis for human movement data. The methods used in this investigation, however, were considered worthwhile in contributing to the examination of movement variability in the context of landing.

Movement variability and research design

Biomechanical investigations typically seek to identify differences between groups or experimental conditions by examining changes in a given variable at a specific point in time, or a change in the movement pattern. Sources of variability in the data confound the detection of systematic differences, which has largely been attributed to between-subject variability, or individual differences (Bates, 1996; Bates et al., in Stergiou et al., 2004; Federolf et al.2013; James & Bates, 1997). A number of remedies exist to control for variability issues in research, including normalization of the data to subject-specific anthropometrics (i.e. height, mass, weight), the inclusion of multiple trials per participant or condition, and the inclusion of large sample sizes (Bates, 1996; Bates et al., in Stergiou et al., 2004; Federolf et al., 2013; James & Bates, 1997). Each remedy may demonstrate utility under different circumstances, but in some instances may not provide a useful solution for identifying relevant characteristics of biomechanical changes or differences. Investigations concerning research design, methods of controlling sources of variability, and the subsequent ability to draw conclusions from a dataset have been conducted in a number of disciplines, exploring contrasting variables, and statistical analysis techniques. The purpose of this section was to identify the perceived benefits and drawbacks of using PCA in human movement research.

The overwhelming emphasis of scientific research is to find patterns within data that can be generalized, or applied to the population from which the sample was drawn (Barnett, Heneman, & Libin, 2012; Dufek et al., 1995). A downfall of this approach is the tendency to neglect individual performances, instead drawing conclusions from the average of the sample, which oftentimes does not reflect the characteristics or performance of any of the sampled individuals (Dufek et al., 1995; Scholes et al., 2012; Stergiou & Scott, 2005). Specifically, collapsing individual performance measures into group statistics can in some cases mask the individual response strategies, potentially removing the ability to draw appropriate conclusions from the data, or entirely missing a meaningful effect within the data (Dufek et al., 1995; Scholes et al., 2012; Stergiou &

Scott, 2005). Given that movement variability is considered representative of individual function, movement variability analyses may provide important information regarding injury susceptibility and movement coordination at the level of the individual (Barrett et al., 2008; James et al., 2000; Li et al., 2005; Stergiou & Decker, 2011; Stergiou et al., 2006). The use of PCA in the literature is typically applied to a group model, accounting for sources of movement variability among pooled biomechanical waveforms from multiple subjects and trials (Brandon et al., 2013; Deluzio et al., in Robertson et al., 2014). A limitation of this approach is the aggregation of data among participants (Bates, 1996; Bates et al., in Stergiou et al., 2004; James & Bates, 1997), specifically limiting the ability to differentiate between-subject from within-subject variability. The use of the group model in PCA cannot be overlooked as a potential drawback, particularly in understanding mechanisms of injury that are likely subject-specific. Despite this limitation, the goal of this research was to assess sources and patterns of variation among trial-to-trial repetitions under contrasting experimental conditions, adjusting mechanical task demands through subject-specific load and landing height manipulations.

Adjusting mechanical task demands to participant anthropometrics is one means of compensation for between-subject variability. The ability of grouped data analysis to obscure phenomena at the level of the individual must nevertheless be acknowledged. Normalization plays a key role in data entry prior to PCA, normalizing waveform magnitudes (subtracting the mean and dividing by the standard deviation for each data point in the time series) and length (temporal normalization to a fixed number of data points across trials; Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013). These processes have been regarded as both strengths and limitations in research, controlling for sources of variation, but removing potentially useful sources of information (Federolf et al., 2013). The reliance on normalization and the inclusion of data points across the entire movement phase under consideration provide the basis for the advancement of the use of PCA in biomechanical investigations, providing sensitivity in detecting subtle differences in movement patterns (Federolf et al., 2013; Federolf et al., 2012). Despite the use of PCA on grouped data, the benefits of PCA have been demonstrated in small sample sizes, allowing systematic differences to be detected among populations and conditions (Federolf et al., 2013; Federolf et al., 2012).

The use of PCA in identifying subtle differences provides connections to the tendency of non-significant findings to be considered scientifically undesirable (Dufek, Bates, Davis, 1994; Scholes et al., 2012; Stergiou & Scott, 2005). Although in some cases Type II errors may be committed in research, it is also possible to obtain non-significant statistical results due to large between-subject variability as a result of participants within the sample having been drawn from different populations, or simply representing naturally occurring differences in the population (Dufek & Bates, 1992; Dufek et al., 1994). It is for this reason that examination of individual performers has been proposed in research, conducting within-subject analyses, where participants serve as their own controls (Dufek & Bates, 1992; Stergiou & Scott, 2005). Extensions upon PCA, including FDA have been cited as potential means of examining within-subject variation to more explicitly examine the role of variability in injury, while preserving the functional form of the data (Donoghue et al., 2008). Changes have been suggested to occur to baseline measures during individual performances therefore within-subject analyses may provide benefits over grouped analyses (Stergiou & Scott, 2005). This is not to suggest that between-subject comparisons are not essential in highlighting important phenomena that can be generalized to the population, but that conforming to typical research approaches at the cost of missing meaningful results should be avoided. Along these lines, approaches within PCA have been examined for gaining more complete access to individual trials in relation to extracted PCs. Biomechanical PCA approaches often interpret PCs by comparing raw waveforms from the 5th and 95th percentiles, approximately equivalent to ± 2 standard deviations (low and high PC scores), with single component reconstructions outlined as robust procedures for examining individual PC waveform patterns (Brandon et al., 2013). Constant improvements and modifications to existing methods emerge in the literature, though the foundations of PCA remain a driving force behind the exploration of PCA as a means of data reduction, allowing time series assessment of biomechanical variables.

Similar to single-subject analysis procedures, the use of PCA in small sample sizes is overcome with the use of a greater number of trials per participant (Bates, Dufek, & Davis, 1992; Dufek et al., 1994; Federolf et al., 2013). Trial size has been suggested to be a critical determinant of effective research design, which becomes of greater importance in low sample and single-subject designs. Specifically, it has been shown that increasing the number of trials per subject-condition can provide improvements in statistical power similar to increases in sample size (Bates et al., 1992; Dufek et al., 1994; Federolf et al., 2013; James, Herman, Dufek, & Bates, 2007). In addition to the greater number of observations (data points), the improved statistical power has been attributed to increased data reliability, an essential precondition for validity (Bates et al., 1992; Bates, Zhang, Dufek, & Chen, 1996; Dufek et al., 1994; Dufek et al., 1995). Within the context of PCA, the inclusion of a greater number of trials per participant has been linked to increases in statistical power, where variation among trials reveals correlations and more clearly defines PCs, allowing for differences to be detected in PC scores (Federolf et al., 2013). With respect to landing data, a minimum of four trials has been shown to be necessary for achieving performance stability (James et al., 2007). In the context of movement variability during landing, obtaining an appropriate number of trials is of particular importance in identifying performance variability once a stable movement pattern has been established. Excessive variability may be present during transitions between stable movement patterns, while the aim of research is to identify the changes in variability across task demands. An insufficient number of trials may limit the ability to draw appropriate conclusions about changes in variability across experimental conditions, which is of concern in designing research studies (Hamill et al., & Li, 1999; Li et al., 2005). The use of PCA in biomechanics provides new considerations for research design, including applications to small sample sizes, inclusion of an appropriate number of observations (trials and participants), as well as the inclusion of a greater number of variables.

The inclusion of multiple sources of data that can be reduced into a subset of relevant features is perceived as a considerable strength of PCA, serving as a method of data filtering (Cohen, 2014; Daffertshofer et al., 2004). The reduction and separation of relevant and irrelevant sources of variation may be useful in identifying systematic changes across task demands. PCA has shown flexibility in reducing very large datasets, including biomechanical data from contrasting sources (kinematics, kinetics, and EMG), among total body measurements, as well as more concentrated joint specific analyses

(Brandon et al., 2013; Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Federolf et al., 2012). For these reasons, PCA was used in exploring synergistic associations, sources of variation, and levels of control among the support limb joints during single-leg landings under contrasting mechanical task demands. The outcome of these analyses was interpreted in the context of potential injury mechanisms as outlined via the proposed relationship between movement variability and injury.

Summary

In reviewing the literature, human movement variability has been highlighted as means of evaluating functional control processes during repeated movements. The progression and development of variability examinations providing an indicator of movement function has led to the association between variability and injury. Despite examinations of movement variability having been conducted on a number of different tasks, the focus of this investigation was delimited to lower extremity function during single-leg landing. Landing has been extensively explored due to a high incidence of injury and the ability to modulate mechanical task demands in experimentally controlled settings. As well, running may be viewed as a series of single-leg landings providing potential generalizations to other movement patterns. The available measures of variability each demonstrate strengths and weaknesses in terms of the associated assumptions, limitations, and interpretability, with PCA demonstrating utility in reducing large datasets and providing interpretations in line with theories from motor control. Additionally, research design has illustrated the associated strengths and weaknesses of group analyses, though existing and emerging methods of analysis may provide new insight into movement control, both within and between individuals. As a result, it is acknowledged that biomechanical assessments using PCA involve a new set of limitations and considerations. The ultimate goal of research examining human movement variability is to better understand movement function and the fundamental processes controlling movement. Overall, this literature review served as the basis for the outlined examination of human movement variability. The aim was therefore to examine sources of functional movement variability under contrasting mechanical task demands, characterizing movement strategies as means of obtaining greater insight into movement control and potential mechanisms of injury in landing.

CHAPTER 3

Methods

Purpose

The purpose of this study was to evaluate changes in movement variability among kinematic, kinetic, and electromyographic (EMG) variables following mechanical task demand manipulations during single-leg drop landings. Biomechanical outcome variables included 3 kinematic (sagittal hip, knee, and ankle angles), 4 kinetic (sagittal hip, knee, ankle moments and vertical ground reaction force: GRF), and 5 EMG variables (gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior). Mechanical task demands were altered using load and landing height manipulations, computed as percentages of participant anthropometrics (bodyweight: BW, BW+12.5%, BW+25%, and height: H12.5% and H25%, respectively). Load accommodation strategies were characterized using impulse ratios relative to baseline (lowest task demands: BW•H12.5%) and 95% confidence intervals (CI) surrounding mechanically predicted impulse ratios. Collective movement variability among participants was assessed across the landing phase using principal component analysis (PCA). Follow-up analyses were carried out separately for each outcome variable, identifying changes in support limb movement patterns following mechanical task demand manipulations. PC loading vectors accessed sources of variation in the dataset, with inferential testing of PC scores identifying movement pattern changes. The number of extracted PCs and explained variance (EV) were used in interpreting movement control in each outcome variable and the associated lower extremity joints.

It was hypothesized that fewer emergent strategies would be observed with greater mechanical task demands, leading to decreased movement variability. Specifically, greater load and landing height were anticipated to result in more predictable response strategies (increasingly positive biomechanical) with decreased movement variability expressed by greater explained variance among fewer extracted PCs. Statistical differences were expected among conditions for PC scores explaining the greatest proportion of the variance in each landing condition. Additionally, greater PC scores were anticipated to occur with greater mechanical task demands. Follow up analyses for each outcome variable (kinematic, kinetic, and EMG), were expected to show greater EV among fewer PCs in proximal joints, relative to distal joints, with the number of PCs decreasing with greater load and landing height. PC loading vectors were predicted to show earlier increases in the landing phase at distal joints, with later increases at proximal joints. Joint specific strategies were expected from PC score differences among conditions. The outlined hypotheses were interpreted relative to functional variability surrounding movement repetitions, making associations between movement control strategies and potential injury mechanisms in single-leg drop landings.

Participants

Following approval from the Institutional Review Board at University of Nevada, Las Vegas (UNLV), participants were recruited through convenience sampling of the UNLV undergraduate and graduate student populations. Prior to participation, written informed consent was obtained from each participant. Twenty-two participants were recruited for participation with three participants removed from analysis due EMG signal losses during collection and/or reflective marker loss in kinematic and kinetic joint analyses; 19 participants were analyzed (15 male, 4 female, age: 24.3± 4.9 years, mass: 78.5 ± 14.7 kg, height: 1.73 ± 0.08m). Each participant completed 10 trials in each condition, 9 used during analysis, allowing problematic trials to be removed when necessary, otherwise the first collected trial in each condition was excluded from analysis, providing equal trial size among participants and conditions.

With respect to sample size, the 19 analyzed participants and 9 associated trials per condition provided 171 observations per condition. Correlation underlies PCA therefore 171 observations suggests 80% of the sample correlations will fall between ± 0.1 of the population *r* (*r* =0), providing adequate statistical power and allowing appropriate generalizations to be made from the data (Field, 2009; Hole, 2014). The collected sample size is considered adequate, though biomechanical PCA research has shown efficacy in smaller sample sizes, including fewer trials per participant (Federolf et al., 2013).

Inclusion criteria consisted of healthy adults aged 18-36 years of age, male or female, who performed voluntary exercise at least two times per week. Participants capable of performing single-leg drop landing trials from an elevated platform, while carrying a maximum external load of BW+25%, from a maximum landing height of H25% were recruited. Exclusion criteria included individuals with current lower extremity injuries, previous lower extremity injuries within the past 6 months, or lower extremity joint replacements. As well, females entering the second trimester of pregnancy were excluded from participation. Prior to participation, each participant was informed of the ability to withdraw from the study at any point, without consequence.

Instrumentation

Data were collected using three-dimensional motion capture, force platform analysis, and electromyography (EMG). Kinematic and kinetic data were simultaneously acquired using a 10-camera system (200Hz, MX T40-S, Vicon Motion Systems Ltd., Oxford, UK) and synchronized force platform (2000Hz, Kistler type 9281CA, Winterthur, Switzerland). Electrical muscle activity data were acquired using a 16-channel EMG system (2000Hz, Noraxon Myosystem 2000, Scottsdale, USA). Data collection was synchronized via Vicon Nexus (version 1.8.5), Noraxon MyoResearch XP Data & Acquisition software (version 1.08.35), and an external analog trigger.

Kinematic data collection included a 16-point lower body spatial model (Vicon Plug-in-Gait) and retro-reflective markers (14.0mm Pearl Markers, B & L Engineering, Santa Ana, USA) fixed using double-sided adhesive tape. Sites of marker attachment included the left and right anterior superior iliac spines (ASI), posterior superior iliac spines (PSI), lateral flexion-extension axis of the knees (KNE), lateral malleoli (ANK), second metatarsal heads (TOE), and calcaneus at the same height above the plantar surface of the foot as the toe marker (HEE). As well, left and right thigh (THI) markers were placed on the surface of the thigh in line with the hip and knee joint centers, with tibia markers placed on the surface of the leg in line with the knee and ankle joint centers (lower 1/3 surface on the left limb, upper 1/3 surface on the right limb). Lower extremity joint angles and kinetics were calculated using Vicon Nexus 1.8.5. Three-dimensional kinematic and kinetic data were collected, with analysis focusing on sagittal joint angles and kinetics, as well as vertical ground reaction forces.

Electrical muscle activity was collected using dual surface electrodes (Noraxon Dual Electrodes, Product # 272, disposable self-adhesive Ag/AgCL snap electrodes, Scottsdale, USA) placed on the muscle belly in line with the muscle fibers, at an inter-electrode spacing of 20mm at each muscle. Prior to electrode placement, the skin surface was shaved, cleaned with alcohol and abraded to minimize skin resistance. Electrical muscle activity was measured in the gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior muscles of the support limb. Sites of attachment were identified as outlined by Konrad (2005). Two single-surface electrodes (Noraxon Single Electrodes, Product # 270, disposable self-adhesive Ag/AgCL snap electrodes, Scottsdale, USA) were used for grounding, attached to the ipsilateral anterior superior iliac spine for the gluteus maximus muscle and the patella for channels measuring the remaining support limb muscles. Electrodes were secured to each participant with additional adhesive tape to secure leads in an attempt to minimize movement artifact.

Each participant was fitted with standardized footwear and clothing for testing. Small backpacks (CamelBak Magic, Petaluma, USA) were used for adding load with standard iron weight plates. Landing height manipulations were carried out using a platform with adjustable height.

Procedure

Upon arrival to the testing environment, informed consent was obtained, ensuring each participant met the inclusion criteria and was comfortable with the testing procedure. Participant anthropometric data characteristics were measured, including body mass and height, as well as leg length, knee width, and ankle width in each limb. Leg length was defined as the distance from the anterior superior iliac spine to the medial malleolus with the participant standing. Segment widths were measured using an anthropometer (Model 01291, Lafeyette Instrument Company, Indiana, USA), including knee width across the medio-lateral axis of the knee when standing, and ankle across the medial and lateral malleoli. The outlined anthropometric data were used in estimating joint centers during kinematic and kinetic analysis (Vicon Nexus 1.8.5), as well as in computing load and landing height manipulations.

Each participant completed a standardized warm-up, including approximately 5-minutes of treadmill running, as well as 1-2 single-leg landings prior to collected trials. Each participant identified a preferred support limb for completing single-leg landings from an elevated platform. All trials and conditions were performed using the preferred support limb. Following limb selection, electrode attachment sites were prepared (DeLuca, 1997; Konrad, 2005). Electrodes were attached and secured, followed by attachment of retro-reflective markers. Small backpacks were secured to the anterior and posterior aspects of the trunk for load adjustments.

Kinematic calibrations were carried out, identifying the location of each retro-reflective marker on the sites of attachment for each participant prior to data collection (Vicon Nexus 1.8.5). Zero offsets were obtained for GRF and EMG data prior to each trial (Vicon Nexus 1.8.5, Noraxon Myosystem 2000). Landing conditions included load and landing height manipulations, computed as percentages for each individual. For landing trials, participants were instructed to stand atop the adjustable platform on one leg, followed by leaning forward and dropping from the elevated platform, contacting the force platform and establishing balance. Participants were instructed to focus their attention on the force platform, aiming for center, while avoiding ground contact with the contralateral limb until beginning to stand upright. Movement of the arms was restricted, having participants fold their arms across their chest. No restrictions were be placed on the movement of the contralateral limb, though instructions will be provided to flex at the knee and hip to avoid contacting the ground with this leg during landing. Unsuccessful trials where the participant loses balance prior to standing upright, or where the contralateral limb contacts the ground, were repeated to a maximum of 90 trials during the testing session.

Load conditions included BW, BW+12.5%, and BW+25% (rounded to the nearest pound), evenly distributing standard iron weight plates to the anterior and posterior aspects of the trunk. Landing height conditions included H12.5% and H25% (participant height; rounded to the nearest centimeter). Each mass condition was carried out under each landing height condition, counterbalancing condition order for each participant in an attempt to minimize task familiarization. Condition counterbalancing was completed from a matrix of all possible condition combinations (6 condition combinations: BW•H12.5, BW12.5•H12.5, BW25•H12.5, BW•H25, BW12.5•H25, BW25•H25). The condition representing the greatest mechanical task demands BW25•H25 was excluded as the first testing condition out of consideration for participant safety; all other condition combinations were available for selection. Condition order was randomly assigned to each participant, without replacement. One to two practice trials were carried out prior to data collection in each condition, ensuring each participant was comfortable completing the task. Periods of rest from 30-seconds to 1-minute were provided between each trial, with 3-5 minutes of rest between conditions in an attempt to minimize participant fatigue. Ten blocked landing trials were carried out under each condition; 9 used during analysis. At the completion of testing, all instrumentation was removed from each participant, including adhesives.

Data reduction and analysis

Time series kinematic, kinetic, and EMG data were exported and processed using custom Matlab scripts (R2012a, MathWorks Inc., Natick, MA, USA). Data filtering and interpolation of kinematic and kinetic data were carried out via Vicon Nexus 1.8.5, with filtering cutoffs selected from residual analysis in Matlab R2012a. Data filtering was carried out via 4th order (zero lag) Butterworth filter (Vicon Nexus 1.8.5 and Matlab R2012a). Kinematic and kinetic data were filtered at 15 Hz and 50 Hz cutoffs, respectively. Joint moments were calculated with matched cutoffs (15 Hz) as a means of reducing the influence of impact artifact (Bisseling & Hof, 2006; Kristianslund, Krosshaug, van den Bogert, 2012).

EMG data were band-pass filtered (15 Hz & 300 Hz lower and upper cutoffs, respectively), full-wave rectified, and low-pass filtered (15 Hz cutoff), preserving the overall pattern of muscle activation and removing potential impact artifact and high-frequency noise in the signal (Kipp, 2014; Winter, 2009, p. 260-262; Winter & Patla, 1997, p. 21-35). Exemplar EMG processing figures are provided in Appendix I (data analysis, EMG processing).

The landing phase was defined from ground contact (GRFz > 20N) to the point vertical center of mass (COM) velocity crossed zero, following ground contact (Figure 3).

Vertical COM velocity was computed by subtracting the contribution of bodyweight and dividing by mass for each participant, followed by trapezoidal integration of COM acceleration versus time (Robertson et al., 2004). The landing phase was extracted from the time series of each outcome variable, providing distinct temporal ranges over which each variable was assessed.

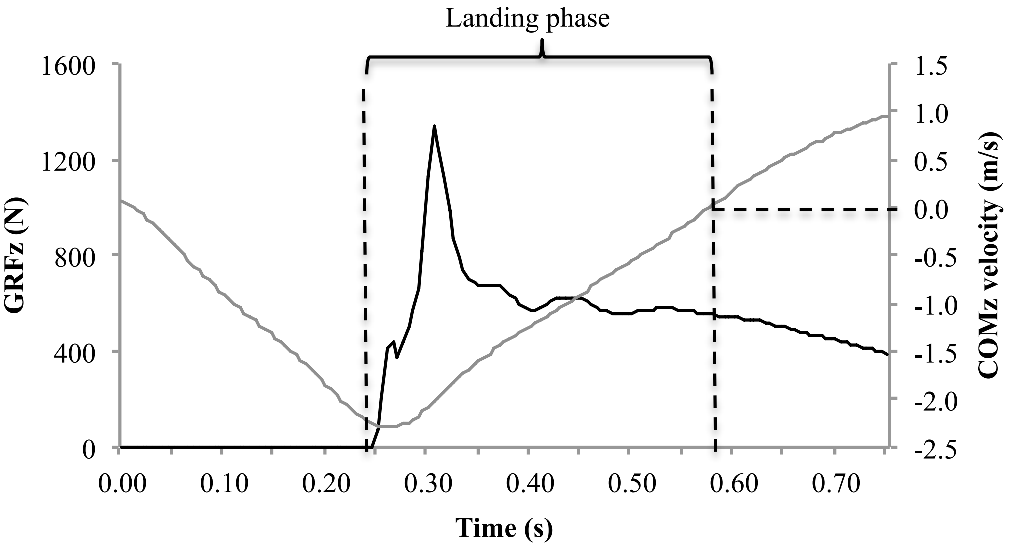


Figure 3: Landing phase identification.

(vertical ground reaction force [GRFz] and vertical center of mass velocity [vCOMz] vs. time)

Variables were each magnitude and temporally normalized. Magnitude normalization differed among variables. GRFz was normalized to participant bodyweight (BW) and joint moments were normalized to participant bodyweight and height (BW•H), removing sources of between-subject variability from each variable. EMG data were normalized to mean dynamic baseline activity for each participant (lowest mechanical task demands: BW•H12.5), which is considered preferential in the assessment of high-intensity, dynamic muscle actions (baseline multiple; BM; Ball & Scurr, 2013). Prior to temporal normalization, variable descriptive statistics were computed for each outcome variable in each condition (mean ± standard deviation), utilizing the integrated time series (area under the curve from trapezoidal integration). The integrated time series of GRFz represents landing impulse (BW•s), computed after removing the contribution due to the BW. Integrated joint moments represent angular impulse (BW•H•s) and integrated EMG (iEMG) represents the total muscle activity during landing (BM•s). Integrated joint angles represent the summation of joint angular position across the landing phase, multiplied by landing duration (deg•s). Following time series integration, each variable was temporally normalized to 101 data points via cubic spline interpolation. Overall, 12 outcome variables were analyzed including 3 kinematic variables (sagittal hip, knee, and ankle angles), 4 kinetic variables (sagittal hip, knee, ankle moments and GRFz), and 5 EMG variables (gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior).

Load accommodation strategies were defined following criteria outlined by James et al. (2003 & 2014) using GRFz landing impulse ratios (BW•s/ BW•s) relative to baseline (lowest mechanical task demands: BW•H12.5). The Newtonian strategy was defined based on conversion of gravitational potential to kinetic energy (Equations 1 and 2):

$$PE = mgh \tag{1}$$

where *PE* is gravitational potential energy, *m* is participant mass, *g* is acceleration due to gravity (9.81m/s^2), and h is initial height above the ground (Equation 1). Additionally,

$$KE = \frac{1}{2}mv^2 \tag{2}$$

where *KE* is kinetic energy, *m* is participant mass, and *v* is vertical center of mass velocity just prior to ground contact (Equation 2).

Rearranging Equations 1 and 2, vertical center of mass velocity was computed:

$$v = \sqrt{2gh} \tag{3}$$

The impulse-momentum relationship (Equation 3) then allowed landing impulse to be calculated:

$$\sum_i F_i \, \Delta t = m\Delta v \tag{4}$$

where landing momentum is the product of participant mass (*m*) and the change in vertical center of mass velocity from step off to ground contact (Δv; right side of Equation 3) and landing impulse is the cumulative product of the vertical ground reaction force at each data point (F_i) and the time between samples (Δt; left side of Equation 3). Defining the landing phase from initial ground contact to the point vertical center of mass velocity (vCOMz) reached zero then permitted prediction of landing impulse. As vertical vCOMz at step-off is assumed to be zero, Equation 3 can be substituted into Equation 4:

$$\sum_i F_i \, \Delta t = m\sqrt{2gh} \tag{5}$$

Equation 5 outlines the dependence of landing impulse on participant mass (*m*) and landing height ($\sqrt{h}$). Newtonian impulse ratios were therefore used in distinguishing observed responses from Newtonian predictions, computed from the product of percent increase in mass and the square root of percent increase in landing height (Equation 5). Newtonian impulse ratios were computed using the generalized form:

$$Impulse\ Ratio\ = \frac{condition\ impulse}{baseline\ impulse}$$

Condition impulse ratios therefore included:

$$1.125 = \frac{(\mathrm{BW12.5{\cdot}H25})}{baseline\ impulse}$$

$$1.250 = \frac{(\mathrm{BW25{\cdot}H12.5})}{baseline\ impulse}$$

$$1.414 = \frac{(\mathrm{BW{\cdot}H25})}{baseline\ impulse}$$

$$1.591 = \frac{(\mathrm{BW12.5{\cdot}H25})}{baseline\ impulse}$$

$$1.768 = \frac{(\mathrm{BW25{\cdot}H25})}{baseline\ impulse}$$

where impulse ratio is unit-less (BW•s/BW•s) and baseline is: (BW•H12.5).

Statistical analysis

Load accommodation strategy identification

Group and single-subject load accommodation strategies were identified using the 95% confidence interval (CI) surrounding the mean impulse ratio for the group and individual participants, respectively. Newtonian strategies in each condition (relative to baseline) were defined as previously outlined. During group and single-subject analyses, the 95% CI was computed for the 19-participant group mean impulse ratio. The group 95% CI was examined relative to the Newtonian impulse ratio as a means of evaluating group differences from Newtonian landing impulse predictions; a group 95% CI containing the Newtonian impulse ratio identified no significant difference from the Newtonian strategy (α = 0.05; Field, 2009; James et al., 2014). A similar approach was taken in single-subject analyses, using the 95% CI computed for the 9-trial mean impulse ratio of each participant. Each participant's 95% CI was examined relative to the Newtonian strategy; a 95% CI containing the Newtonian strategy indicated no significant difference (α = 0.05; Field, 2009; James et al., 2014).

Additional single-subject strategy classifications were identified using methods outlined by James et al. (2003, 2014). An individual 95% CI exceeding the Newtonian impulse ratio was classified as Super-Newtonian, a 95% CI lesser than Newtonian and greater than Fully Accommodating (impulse ratio 1.00) was classified as Positive Biomechanical, a 95% CI containing an impulse ratio of 1.00 was classified as Fully Accommodating, and a 95% CI lesser than1.00 was classified as Negative Biomechanical (α = 0.05; James et al., 2014).

Collective movement variability (PCA)

Prior to performing PCA, temporally and magnitude normalized time series variables were independently converted to z-scores for each subject-condition-variable, subtracting the subject's baseline mean (mean of the 9-trial ensemble time series) and dividing by the baseline standard deviation (mean standard deviation surrounding the 9-trial ensemble time series). Ensemble plots are provided in Appendix I (data analysis, time series z-score conversion), demonstrating time series z-score conversion for each variable and condition, maintaining relationships among conditions, while converting each variable to a standard scale.

The 12 outcome variables within each trial were then appended (linked end-to-end) creating a 1212-point vector length for each trial (Figure 4: 12 variables x 101 data points per variable; Federolf et al., 2013). Six independent analyses were carried out by condition, extracting PCs from 171 x 1212 dimension matrices (19 participants x 9 trials = 171). The first PCs explaining greater than 90% of the cumulative variance were interpreted relative to the collective movement variability among trials, variables, and participants (Brandon et al., 2013; Daffertshofer et al., 2004).

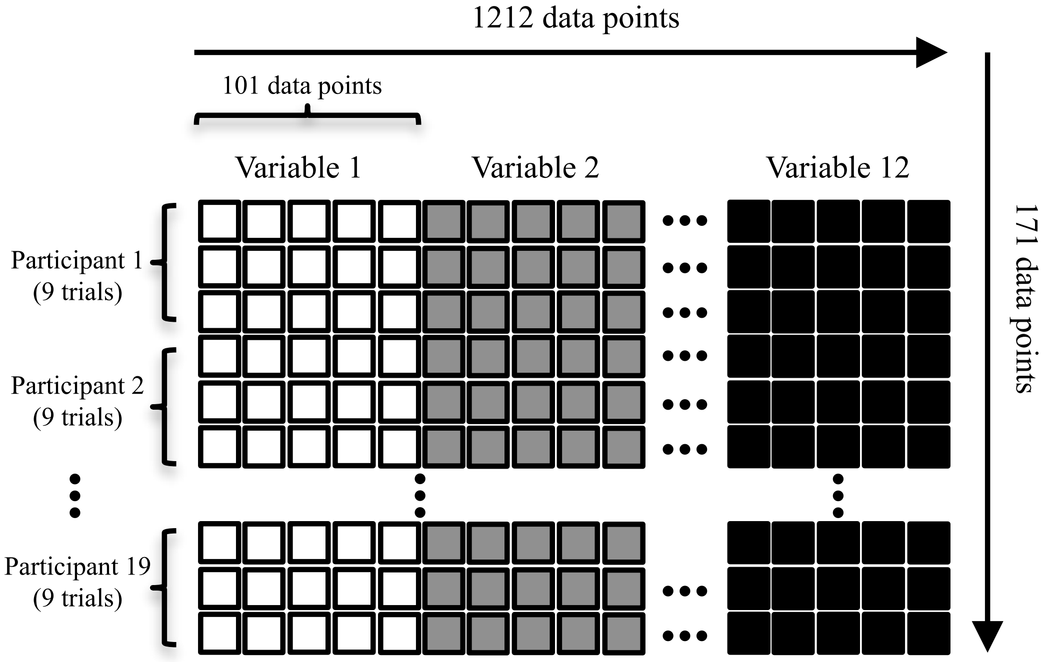


Figure 4: Collective movement variability input PCA matrix organization.

(independent analyses by condition)

Variable-specific adjustments (follow-up PCA)

Follow-up analyses were conducted for each variable (12 total: 3 kinematic, 4 kinetic, and 5 EMG), examining movement pattern differences among conditions. Prior to performing PCA, 1026 x 101 dimension matrices were assembled for each variable (Figure 5: 19 participants x 3 loads x 2 heights x 9 trials = 1026). PCs explaining greater than 90% of the cumulative variance were extracted for further analysis, with PC scores computed for each trial and PC (Matlab R2012a; Brandon et al., 2013; Daffertshofer et al., 2004). In each PC, means were computed for each participant from the 9 completed trials. PC score means were then used in subsequent inferential testing, evaluating differences among conditions (Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013).

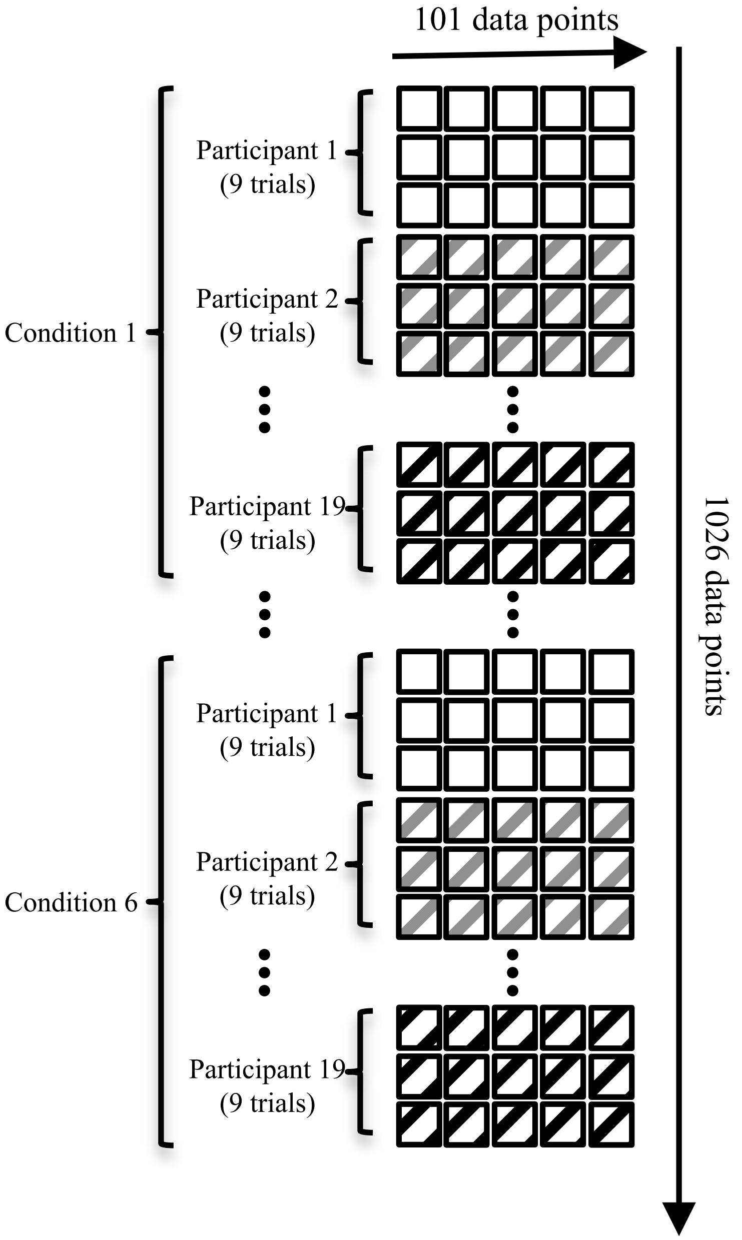


Figure 5: Follow-up PCA. (independent analyses by variable)

PC scores were assessed for normality, prior to statistical testing, though previous research suggests PC scores tend to meet the required parametric assumptions (Deluzio et al., in Robertson et al., 2014). Separate 3x2 (load x height) repeated measures factorial ANOVAs were conducted in each extracted PC, evaluating movement pattern differences among conditions. Follow-up one-way repeated measures ANOVAs and pairwise comparisons were performed as necessary. Degrees of freedom were adjusted as necessary using Huynh-Feldt corrections (Field, 2009). Bonferroni *post-hoc* adjustments were used during pairwise comparisons, controlling the family-wise error rate (Field, 2009).

The number of extracted PCs and associated explained variance were used in interpreting sources of variation and the level of movement control at each lower extremity joint. PC loading vectors were descriptively assessed and used in evaluating the manner in which movement pattern alterations occurred in each outcome variable. PC loading vectors were plotted alongside mean time series plots for each condition, identifying contributions of each source of variation in the corresponding outcome variable.

CHAPTER 4

Results

Purpose

The purpose of this study was to evaluate changes in movement variability among kinematic, kinetic, and electromyographic (EMG) variables following mechanical task demand manipulations during single-leg drop landings. Biomechanical outcome variables included 3 kinematic (sagittal hip, knee, and ankle angles), 4 kinetic (sagittal hip, knee, ankle moments and vertical ground reaction force: GRF), and 5 EMG variables (gluteus maximus, vastus medialis, biceps femoris, medial gastrocnemius, and tibialis anterior). Mechanical task demands were altered using load and landing height manipulations, computed as percentages of participant anthropometrics (bodyweight: BW, BW+12.5%, BW+25%, and height: H12.5% and H25%, respectively). Load accommodation strategies were characterized using impulse ratios relative to baseline (lowest task demands: BW•H12.5%) and 95% confidence intervals (CI) surrounding mechanically predicted impulse ratios. Collective movement variability among participants was assessed across the landing phase using principal component analysis (PCA). Follow-up analyses were carried out separately for each outcome variable, identifying changes in support limb movement patterns following mechanical task demand manipulations. PC loading vectors accessed sources of variation in the dataset, with inferential testing of PC scores identifying movement pattern changes. The number of extracted PCs and explained variance (EV) were used in interpreting movement control in each outcome variable and the associated lower extremity joints.

It was hypothesized that fewer emergent strategies would be observed with greater mechanical task demands, leading to decreased movement variability. Specifically, greater load and landing height were anticipated to result in more predictable response strategies (increasingly positive biomechanical) with decreased movement variability expressed by greater explained variance among fewer extracted PCs. Statistical differences were expected among conditions for PC scores explaining the greatest proportion of the variance in each landing condition. Additionally, greater PC scores were anticipated to occur with greater mechanical task demands. Follow up analyses for each outcome variable (kinematic, kinetic, and EMG), were expected to show greater EV among fewer PCs in proximal joints, relative to distal joints, with the number of PCs decreasing with greater load and landing height. PC loading vectors were predicted to show earlier increases in the landing phase at distal joints, with later increases at proximal joints. Joint specific strategies were expected from PC score differences among conditions. The outlined hypotheses were interpreted relative to functional variability surrounding movement repetitions, making associations between movement control strategies and potential injury mechanisms in single-leg drop landings.

Descriptive statistics

Contrasting integrated lower extremity movement variable trends were observed among conditions (mean and standard deviation; SD; Table 1). Landing duration (time from ground contact to zero vCOMz) was also summarized in each condition (mean and standard deviation; Table 1). Decreased landing duration trends were observed at greater loads in each height condition, with increased landing duration from greater landing height in each load condition. Landing impulse (integrated GRFz; BW•s) remained relatively consistent across load conditions, but showed increasing trends at greater landing height. Decreasing trends were observed for integrated hip, knee, and ankle angles (deg•s) at greater loads, while increasing trends were observed at greater landing height. In each condition, trends toward greater integrated knee angles were observed relative to the hip, which further exceeded the ankle. Decreasing angular impulse (BW•H•s) trends were observed at greater loads, with increasing trends at greater landing height. In each condition, trends toward greater ankle angular impulse were observed relative to the hip and knee, with similarities between hip and knee angular impulse across conditions. Contrasting trends were observed among muscles at greater loads for total muscle activity during the landing phase (integrated EMG; iEMG; baseline multiple; BM). When landing from H12.5%, decreasing trends were observed for biceps femoris and tibialis anterior iEMG, while inconsistent trends were observed for vastus medialis and medial gastrocnemius iEMG. Increasing iEMG trends were observed for the gluteus maximus muscle at greater loads from each landing height. When landing from H25%, decreasing trends were observed for vastus medialis, medial gastrocnemius, and tibialis anterior iEMG at greater loads, while inconsistent trends were observed for biceps femoris iEMG. Increasing iEMG trends were observed for each muscle with increasing landing height. With respect to iEMG trends among muscles, increasing trends were observed for gluteus maximus iEMG at greater loads and landing height, relative to the remaining lower extremity muscles.

Table 1: Integrated variable descriptive statistics by load (BW, BW+12.5%, BW+25%) and landing height (H12.5%, H25%) conditions

Landing Height	Variable	Load BW Mean	 SD	 BW+12.5% Mean	 SD	 BW+25% Mean	 SD
H12.5%	Landing duration (s)	0.22	0.05	0.18	0.03	0.15	0.02
	GRFz impulse (BW•s)	0.19	0.02	0.19	0.02	0.18	0.02
	Integrated hip angle (deg•s)	6.50	3.18	4.86	2.15	3.65	1.42
	Integrated knee angle (deg•s)	9.26	4.68	6.36	2.87	4.48	2.00
	Integrated ankle angle (deg•s)	3.53	1.54	2.23	1.07	1.36	0.80
	Hip angular impulse (BW•H•s)	0.013	0.008	0.009	0.005	0.007	0.004
	Knee angular impulse (BW•H•s)	0.013	0.006	0.010	0.005	0.007	0.004
	Ankle angular impulse (BW•H•s)	0.020	0.004	0.016	0.003	0.015	0.003
	Gluteus maximus iEMG (BM•s)	0.22	0.05	0.24	0.16	0.26	0.20
	Biceps femoris iEMG (BM•s)	0.22	0.05	0.18	0.07	0.16	0.05
	Vastus medialis iEMG (BM•s)	0.22	0.05	0.23	0.09	0.19	0.07
	Medial gastrocnemius iEMG (BM•s)	0.22	0.05	0.18	0.08	0.19	0.07
	Tibialis anterior iEMG (BM•s)	0.22	0.06	0.21	0.10	0.17	0.07
H25%	Landing duration (s)	0.26	0.05	0.20	0.03	0.16	0.02
	GRFz impulse (BW•s)	0.26	0.02	0.26	0.02	0.26	0.02
	Integrated hip angle (deg•s)	8.3	3.65	5.8	2.25	4.2	1.72
	Integrated knee angle (deg•s)	12.1	5.13	8.2	3.04	5.7	2.25
	Integrated ankle angle (deg•s)	4.7	1.64	3.0	1.16	1.9	0.90
	Hip angular impulse (BW•H•s)	0.018	0.009	0.012	0.006	0.008	0.005
	Knee angular impulse (BW•H•s)	0.016	0.007	0.014	0.006	0.012	0.005
	Ankle angular impulse (BW•H•s)	0.027	0.004	0.023	0.003	0.020	0.003
	Gluteus maximus iEMG (BM•s)	0.40	0.25	0.45	0.38	0.45	0.41
	Biceps femoris iEMG (BM•s)	0.28	0.11	0.22	0.08	0.24	0.18
	Vastus medialis iEMG (BM•s)	0.33	0.17	0.28	0.09	0.26	0.09
	Medial gastrocnemius iEMG (BM•s)	0.37	0.29	0.29	0.20	0.25	0.13
	Tibialis anterior iEMG (BM•s)	0.33	0.16	0.29	0.12	0.24	0.13

Note: SD is standard deviation, GRFz is vertical ground reaction force, BW is bodyweight, H is participant height, deg is degrees, s is seconds, BM is baseline multiple, baseline is BW•H12.5

Load accommodation strategy identification

Single-subject and group load accommodation strategy identification outlined a range of responses that diverged from Newtonian predictions ($p < 0.05$; Figure 6). Figure 6 summarizes the observed load accommodation strategies using the 95% confidence interval (CI) surrounding the impulse ratio from the 9 trials completed by each participant in each condition, and the 95% CI surrounding the mean impulse ratio from the 19 participants in each condition (single-subject and group analyses, respectively). In each condition comparison the Newtonian response, Fully Accommodating response, group mean (± 95% CI), and single-subject mean (± 95% CI) are shown along with load accommodation frequencies among participants (Figure 6).

None of the examined conditions demonstrated a group Newtonian load accommodation strategy, while five individual participants demonstrated Newtonian strategies in the BW•H25 condition. With respect to group load accommodation strategies, a group Fully Accommodating strategy was observed from BW12.5•H12.5, a group Negative biomechanical strategy was observed from BW25•H12.5, and group Positive Biomechanical strategies were observed from BW•H25, BW12.5•H25, BW25•H25 (Figure 6). From single-subject load accommodation strategy identification, strategies demonstrating the greatest frequency (mode) aligned with the group classification in each respective condition (Figure 6). Single-subject analysis, however, highlighted individual participants that diverged from the group response in all but one condition (BW25•H25), where exclusively Positive Biomechanical responses were observed among participants.

Additional assessments of the observed load accommodation strategies in each condition were performed in a supplementary statistical analysis, summarized in Appendix II (supplementary statistical analyses, load accommodation strategy identification). Fisher's exact tests were used in identifying associations among conditions and observed strategy frequencies. In each case, load accommodation strategies were associated with load and landing height (*p*<.001; Figure 6). The effect of load at H12.5% is summarized below the left column, the effect of load at H25% is summarized below the right column, the effect of height at BW+12.5% is summarized to the right of the middle row, the effect of height is summarized to the right of the bottom row (Figure 6; H12.5%: ↑ Load ↑ NB, *p*<.001; indicates Negative Biomechanical strategies were more frequently observed with greater load).

Overall, Negative Biomechanical strategies were observed with greater frequency at greater loads when landing from H12.5%. Positive Biomechanical strategies were observed with greater frequency at greater loads when landing from H25%, and with greater landing height when landing at BW+12.5% and BW+25% (Figure 6).

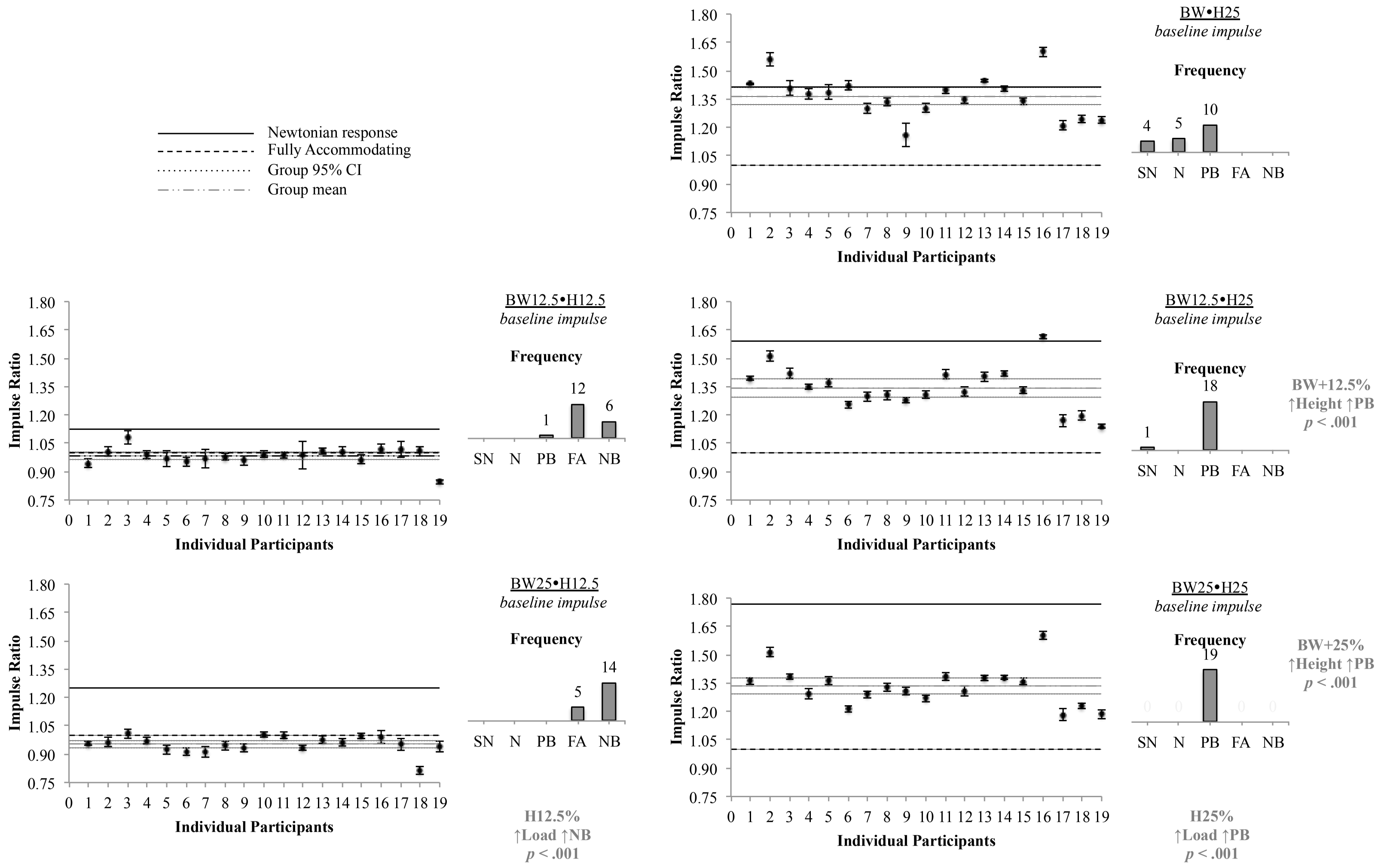


Figure 6: Load accommodation strategy identification and frequency of observed strategies by condition. (mean ± 95% CI; BW is bodyweight, H is participant height, SN is Super-Newtonian, N is Newtonian, PB is Positive Biomechanical, FA is Fully Accommodating, NB is Negative Biomechanical)

Collective movement variability (PCA)

Decreasing collective movement variability trends were observed with greater load and landing height, assessed from the number of PCs extracted in each condition. The number of extracted PCs explaining greater than 90% of the cumulative variance among trials, variables, and participants are summarized in each condition (Figure 7; top).

Supplementary analyses were performed using group and single-subject PCA approaches, examining synergies among variables in each condition (Appendix II, supplementary statistical analyses, collective movement variability [PCA]). From the group analysis, the number of extracted PCs explaining greater than 90% of the cumulative variance among trials, variables, and participants were summarized in each condition (Figure 7, middle). The results of the single-subject analyses summarize the number of extracted PCs explaining greater than 90% cumulative variance among trials and variables (Figure 7, bottom; mean among participants ± standard error). From single-subject PCA, decreased collective movement variability (# extracted PCs) was observed among participants with greater load and landing height during inferential testing using participant means (Figure 7, bottom; Friedman tests and Wilcoxon signed-rank tests, respectively; p<.001; Appendix II, supplementary statistical analyses, collective movement variability [PCA).

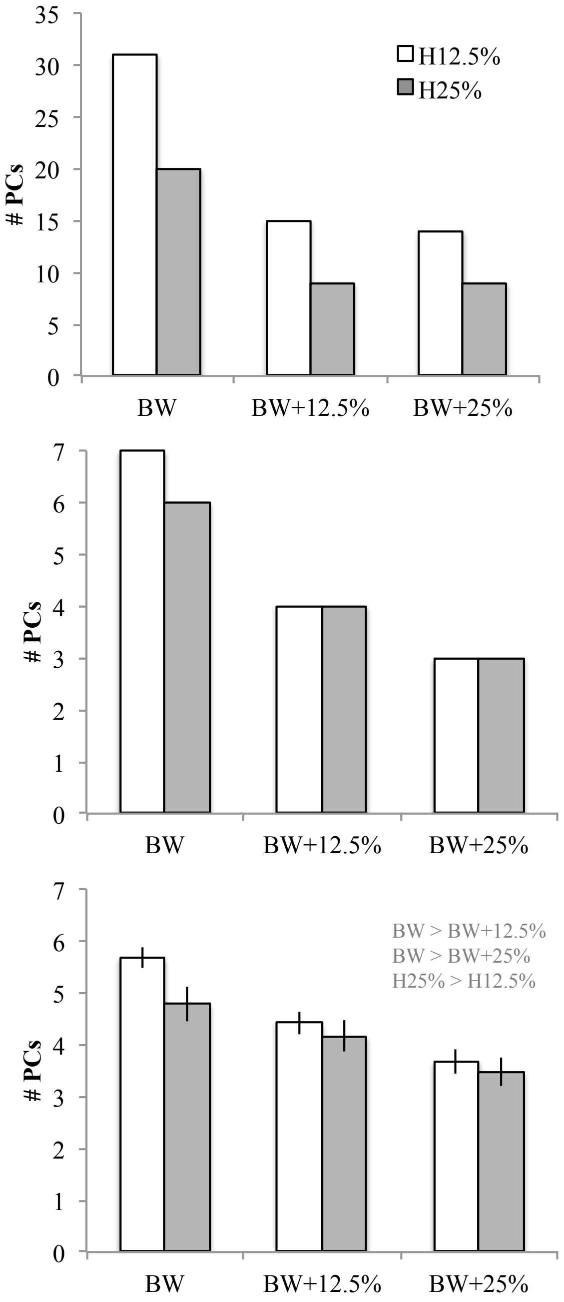


Figure 7: Collective movement variability summaries.

(number of principal components [# PCs] >90% cumulative explained variance; **Top**: group appended variable PCA; **Middle**: group synergy PCA; **Bottom**: mean (aggregated among participants; ± standard error) single-subject synergy PCA)

Collective movement variability assessed via PCA was examined alongside the coefficient of variation (CV; [standard deviation/baseline mean]*100) from the integrated time series for the 9 trials of each participant-condition-variable in a supplementary analysis (Appendix II, supplementary statistical analyses, within-subject variability). This approach provided a measure of within-subject variability, which demonstrated decreasing trends with greater mechanical task demands (load and landing height; Figure 8; mean within-subject CV ± standard error). Although CV provided a standardized scale for each variable (%), variability trends were explored in a 3x2 (load x height) repeated measures factorial MANOVA due to magnitude differences among variables. The results of this analysis are highlighted in Figure 8 (summarized by kinematic, kinetic, and EMG variables), indicating the location of statistical differences when present in each variable ($\alpha = 0.05$). Comprehensive results of this analysis are summarized in Appendices II and III (supplementary statistical analyses, within-subject variability; comprehensive statistical summary, within-subject variability, respectively).

Integrated hip, knee, and ankle angles, along with knee and ankle angular impulse, each demonstrated lesser within-subject variability at greater loads ($p \leq .016$) and landing height ($p \leq .044$; Figure 8). Lesser hip angular impulse variability was observed at BW+25% relative to BW ($p =.008$; Figure 8). Greater total muscle activity variability (iEMG) was observed in the gluteus maximus, vastus medialis, and tibialis anterior muscles at greater landing height ($p \leq .011$; Figure 8). Vertical ground reaction force (GRFz) impulse variability and biceps femoris iEMG variability failed to show differences among conditions ($p > .05$; Figure 8; Appendices II & III, supplementary statistical analyses & comprehensive statistical summary, within-subject variability).

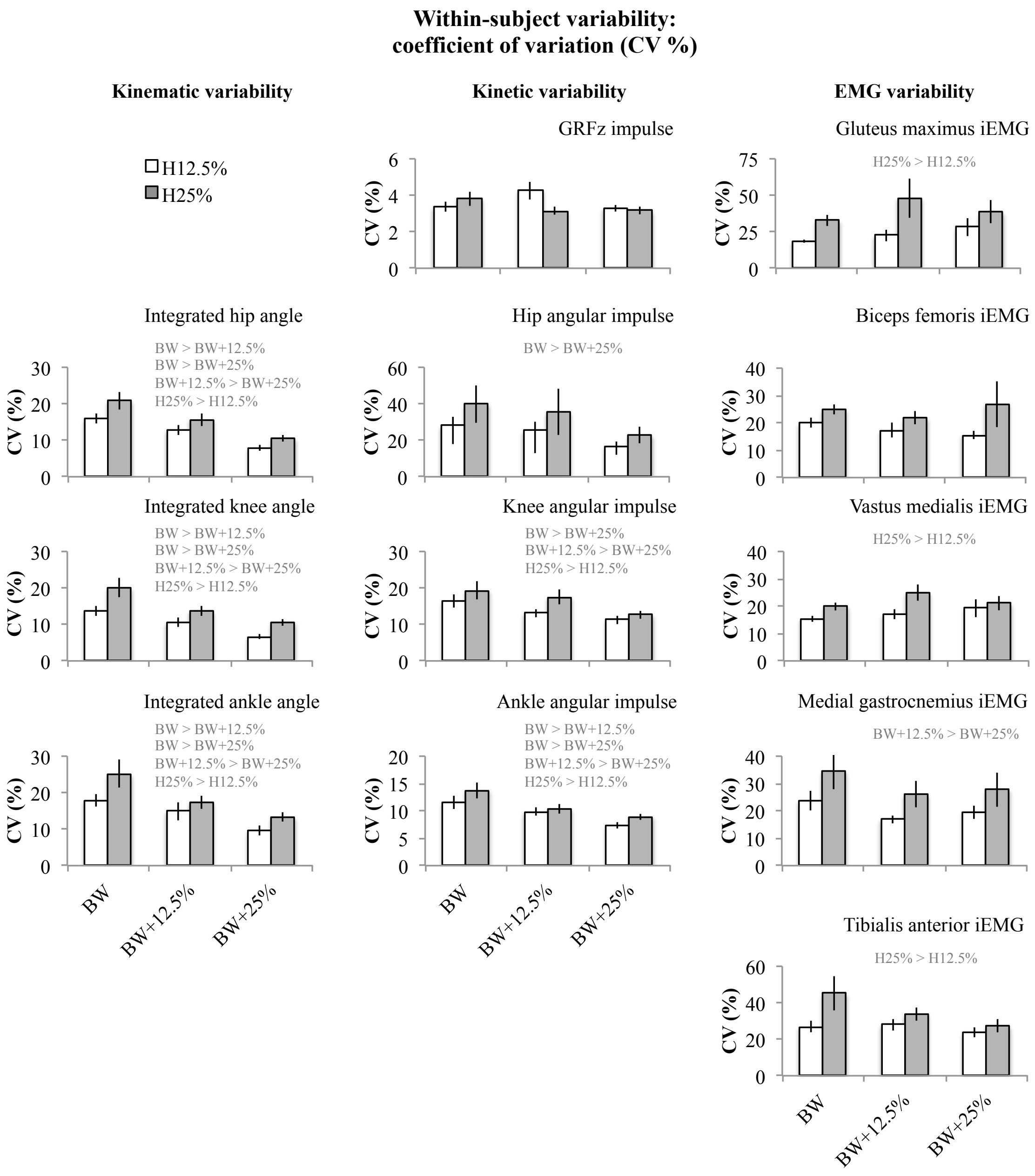


Figure 8: Integrated variable within-subject variability.

(coefficient of variation [CV%]; [condition standard deviation/baseline mean]*100; baseline is BW•H12.5%).

Variable-specific adjustments (follow-up PCA)

Separate follow-up assessments were performed for each variable via PCA, identifying contrasting trends among lower extremity movement variables following mechanical task demand manipulations. Magnitude differences among conditions were identified from PC score inferential testing (Figures 9-20). PC score differences are presented alongside time series PC loading vectors and ensemble time series means for each respective condition (Figures 9-20). PC loading vectors identify the source of variation captured by each PC, including accompanying temporal characteristics (i.e. % landing phase).

Figures 9 through 20 include ensemble time series mean plots (± standard deviation) for each condition (top left), PC loading vectors time series plots for each respective PC (presented in descending order of explained variance; left column), summary PC score means (± standard error) and the location of statistically significant differences among conditions (right column next to each respective PC loading vector plot). Significant interaction includes mean PC scores from each condition combination, while significant main effects for load and height include PC score means aggregated by load or landing height (Figures 9-20). Due to the number of variables and extracted PCs per variable, PC score inferential test results are summarized below, while comprehensive statistical results are provided in Appendix III (comprehensive statistical summary, variable specific adjustment [follow-up PCA]). Ensemble time series plots are provided in magnitude-normalized units, prior to z-score conversion (GRFz: BW, angles: degrees, moments: BW•H, EMG: BM [baseline multiple]). Ensemble time series z-score plots are presented in Appendix I (time series z-score conversion).

Supplementary PCA assessments were performed for each variable characterizing variability changes following mechanical task demand manipulations (Appendix II, variable-specific adjustments [follow-up PCA]). The numbers of extracted PCs in each condition are presented in Figures 9 through 20 (top right).

Vertical ground reaction force (GRFz)

Increasing trends were observed for GRFz magnitudes with greater load and landing height (Figure 9; top left). Later (% landing phase) GRFz magnitude increases were observed with greater load when landing from each height (Figure 9; top left). Contrasting trends were observed when examining GRFz landing patterns (Figure 9; top left) relative to landing impulse (Table 1). During PCA, four PCs provided 90% (Figure 9; left column). GRFz variability (number of extracted PCs) remained consistent among conditions (Figure 9; top right).

PC1: GRFz increases were observed from approximately 0-25% of the landing phase in the PC1 loading vector (45.5% EV; Figure 9; second row). PC1 scores were influenced by the interaction of load and landing height ($F[2, 36] = 8.3$, $p = .001$, $\eta^2 = .32$). When landing from both H12.5% and H25%, lesser PC scores were observed with greater load ($p < .001$; Figure 9; second row). When landing at BW and BW+12.5%, greater PC scores were observed at greater landing height ($p \leq .024$; Figure 9; second row). Greater GRFz was observed with lesser load and greater landing height indicating early GRFz increases from approximately 0-25% of the landing phase in PC1.

PC2: GRFz increases were observed from approximately 20-40% of the landing phase in the PC2 loading vector (24.9% EV; Figure 9; third row). PC2 scores were influenced by the interaction of load and landing height ($F[2, 36] = , p < .003, \eta^2 = .27$). At BW, BW+12.5%, and BW+25%, greater PC2 scores were observed from H25% relative to H12.5% ($p \leq .009$; Figure 9; third row). Greater GRFz was observed with greater landing height from approximately 20-40% of the landing phase in PC2.

PC3: GRFz increases were observed from approximately 0-25% and 40-100% of the landing phase in the PC3 loading vector (12.1% EV; Figure 9; fourth row). Significant load and landing height main effects were observed ($F[1.5, 26.2] = 9.8, p = .002, \eta^2 = .35$; $F[1, 18] = 15.3, p = .001, \eta^2 = .46$; respectively). Greater PC3 scores were observed at BW+25%, relative to BW and BW+12.5% ($p \leq .012$; Figure 9; fourth row). Greater PC3 scores were observed at H25% relative to H12.5% ($p = .001$; Figure 9; fourth row). Greater GRFz was observed with greater load and landing height from approximately 0-25% and 40-100% of the landing phase in PC3.

PC4: GRFz increases were observed from approximately 0-20% and 25-40% of the landing phase in the PC4 loading vector (7.9% EV; Figure 9; fifth row). PC4 scores were influenced by the interaction of load and landing height ($F[1.5, 27.3] = 3.9, p = .043, \eta^2 = .18$). From H12.5%, greater PC4 scores were observed at BW relative to BW+25% ($p = .037$; Figure 9; fifth row). As well, at H25% greater PC4 scores were observed at BW25% relative to BW12.5% ($p = .001$; Figure 9; fifth row).

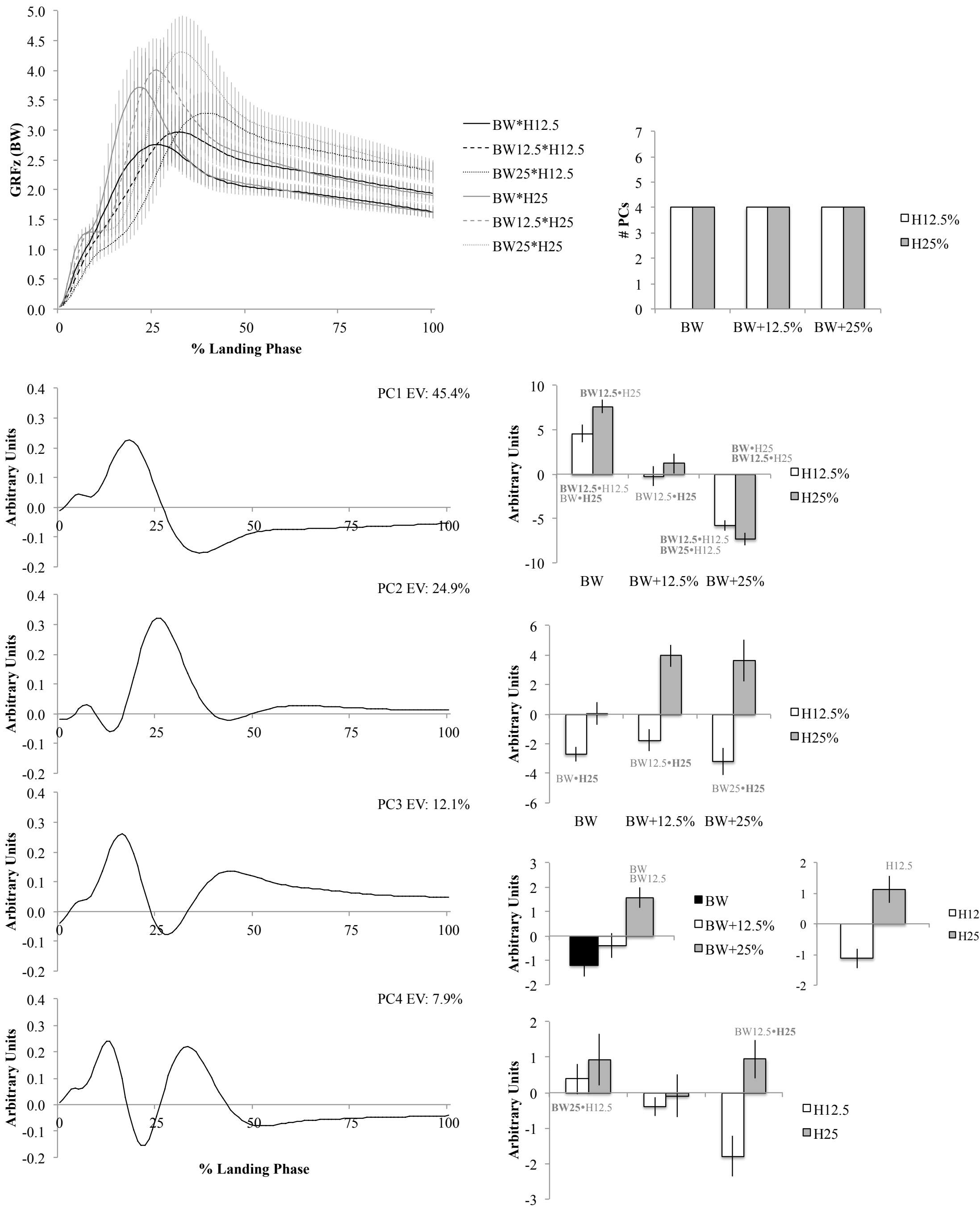


Figure 9: Vertical ground reaction force PCA.

(GRFz; EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Hip angle

Increasing trends were observed for hip flexion angles from greater landing height, with decreasing trends at greater load (Figure 10; top left). Later (% landing phase) increases were observed at greater load from each landing height (Figure 10; top left). Greater integrated hip angles were observed with lesser load and greater landing height (Table 1), demonstrating similarity to time series hip angle plots in each condition (Figure 10; top left). From PCA, two PCs provided 90% EV (Figure 10; left column). Decreasing hip angle variability trends were observed when landing from H12.5%, but remained consistent across load conditions from H25% (#PCs; Figure 10; top right).

PC1: Hip angle increases were observed across the landing phase in the PC1 loading vector (86.6% EV; Figure 10; second row). Significant load and landing height main effects were observed ($F[1.5, 26.3] = 18.3$, $p < .001$, $\eta^2 = .50$; $F[1, 18] = 12.0$, $p = .003$, $\eta^2 = .40$; respectively). PC1 scores decreased with load ($p \leq .007$; Figure 10; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 10; second row). Lesser hip angles were observed with greater load, while greater hip angles were observed at greater landing height in PC1.

PC2: Hip angle increases were observed from approximately 0-50% of the landing phase in the PC2 loading vector (5.9% EV; Figure 10; third row). Significant load and landing height main effects were observed ($F[2, 36] = 6.3$, $p = .005$, $\eta^2 = .26$; $F[1, 18] = 42.1$, $p < .001$, $\eta^2 = .70$; respectively). PC2 scores at BW exceeded BW+12.5% and BW+25% ($p \leq .028$; Figure 10; third row). Greater PC2 scores were observed from H12.5% relative to H25% ($p < .001$; Figure 10; third row). Greater hip angles were observed at BW and H12.5% from ~0-50% of the landing phase in PC2.

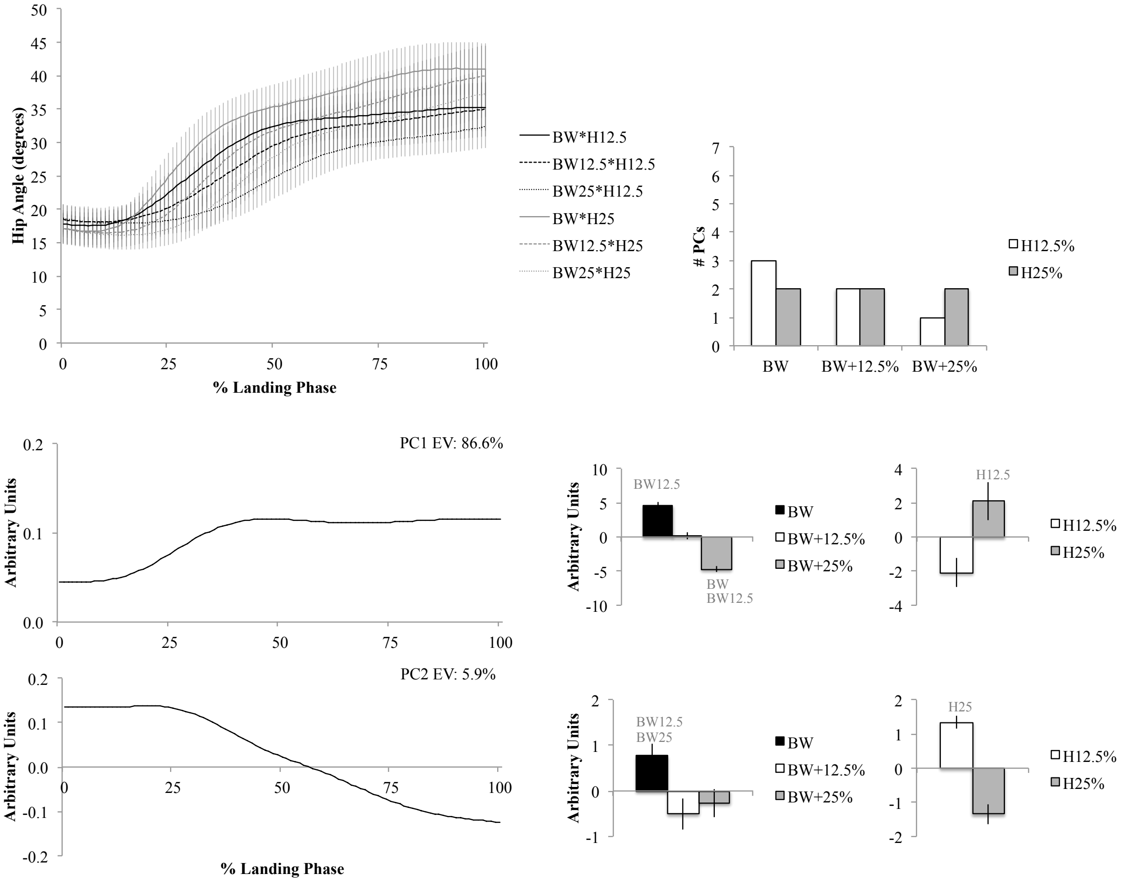


Figure 10: Hip angle PCA.

(EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Knee angle

Increasing trends were observed for knee flexion angles from greater landing height, with decreasing trends at greater load (Figure 11; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 11; top left). Similar trends were observed between time series knee angles plots (Figure 11; top left) and integrated knee angles (Table 1), decreasing with greater load and increasing from greater height. From PCA, two PCs provided 90% EV (Figure 11; left column). Contrasting knee angle variability trends were observed, decreasing at BW from H12.5% and increasing at BW+12.5% from H25% (# PCs; Figure 11; top right).

PC1: Increased knee angles were observed across the landing phase in the PC1 loading vector (86.7% EV; Figure 11; second row). Significant load and landing height main effects were observed ($F[2, 26] = 208.7$, $p < .001$, $\eta^2 = .92$; $F[1, 18] = 78.5$, $p < .001$, $\eta^2 = .81$; respectively). PC1 scores decreased with load ($p < .001$; Figure 11; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 11; second row). Lesser knee angles were observed with greater load, while greater knee angles were observed at greater landing height in PC1.

PC2: Knee angles increased from ~60-100% of the landing phase in the PC2 loading vector (6.6% EV; Figure 11; third row). PC2 scores were influenced by the interaction of load and landing height ($F[2, 36] = 13.7$, $p < .001$, $\eta^2 = .43$). At BW+12.5% and BW+25%, greater PC2 scores were observed from H25% relative to H12.5% (Figure 11; third row). From H12.5% and H25%, lesser PC2 scores were observed at BW relative to BW+12.5% and BW+25% (Figure 11; third row). Greater knee angles were observed at greater landing height and load from ~60-100% of the landing phase in PC2.

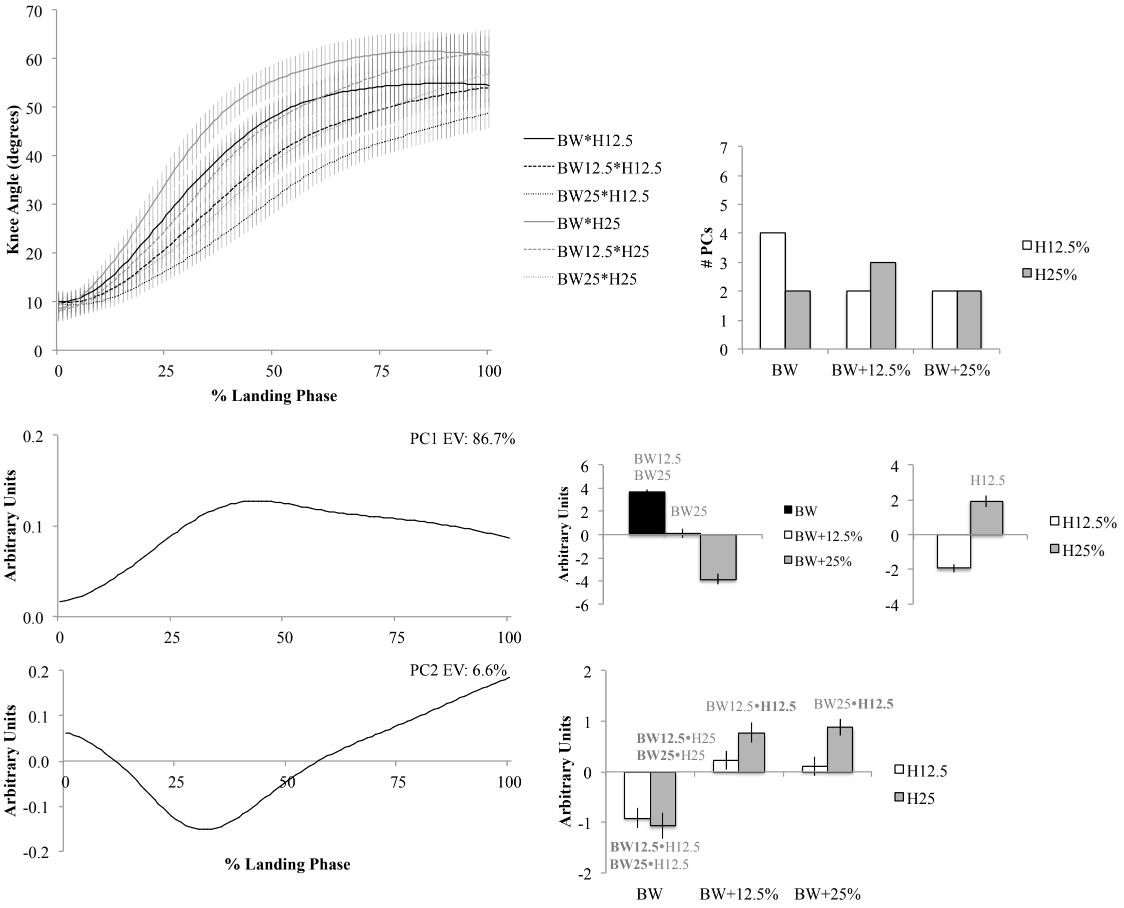


Figure 11: Knee angle PCA.

(EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Ankle angle

Increasing trends were observed for ankle flexion angles with greater load and landing height from approximately 25-100% of the landing phase, with lesser ankle dorsiflexion observed at ground contact (0% landing phase) when landing from H25% relative to H12.5% (Figure 12; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 12; top left). Common trends were observed between time series ankle angles plots (Figure 12) and integrated ankle angles, decreasing with greater load and increasing from greater height (Table 1). From PCA, three PCs provided 90% EV (Figure 12; left column). Decreasing trends were observed for ankle angle variability with increasing load from each landing height (number of extracted PCs; Figure 12; top right).

PC1: Increased ankle angles were observed across the landing phase in the PC1 loading vector (59.6% EV; Figure 12; second row). Significant load and landing height main effects were observed ($F[1.6, 28.9] = 79.2$, $p < .001$, $\eta^2 = .82$; $F[1, 18] = 38.8$, $p < .001$, $\eta^2 = .68$; respectively). PC1 scores decreased with load ($p < .001$; Figure 12; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 12; second row). Lesser ankle angles were observed with greater load, while greater ankle angles were observed with greater landing height in PC1.

PC2: Ankle angle increases were observed from approximately 0-25% of the landing phase in the PC2 loading vector (30.4% EV; Figure 12; third row). PC2 scores were influenced by the interaction of load and landing height ($F[1.5, 26.1] = 5.1$, $p = .021$, $\eta^2 = .22$). When landing from H25%, lesser PC2 scores observed at BW+12.5% relative to BW and BW+25%. In each load condition, PC2 scores were greater from H12.5%

relative to H25%. Lesser ankle angles were observed with greater landing height from approximately 0-25% of the landing phase in PC2.

PC3: Ankle angle increases were observed from approximately 0-10% of the landing phase in the PC3 loading vector (7.5% EV; Figure 12; fourth row). PC3 scores were influenced by the interaction of load and landing height ($F[1.4, 25.1] = 8.9$, $p = .004$, $\eta^2 = .33$). When landing from H25%, lesser PC2 scores observed at BW relative to BW+12.5% and BW+25% ($p < .001$; Figure 12; fourth row). At BW and BW+12.5%, PC2 scores were greater from H12.5% relative to H25% ($p < .001$; Figure 12; fourth row).

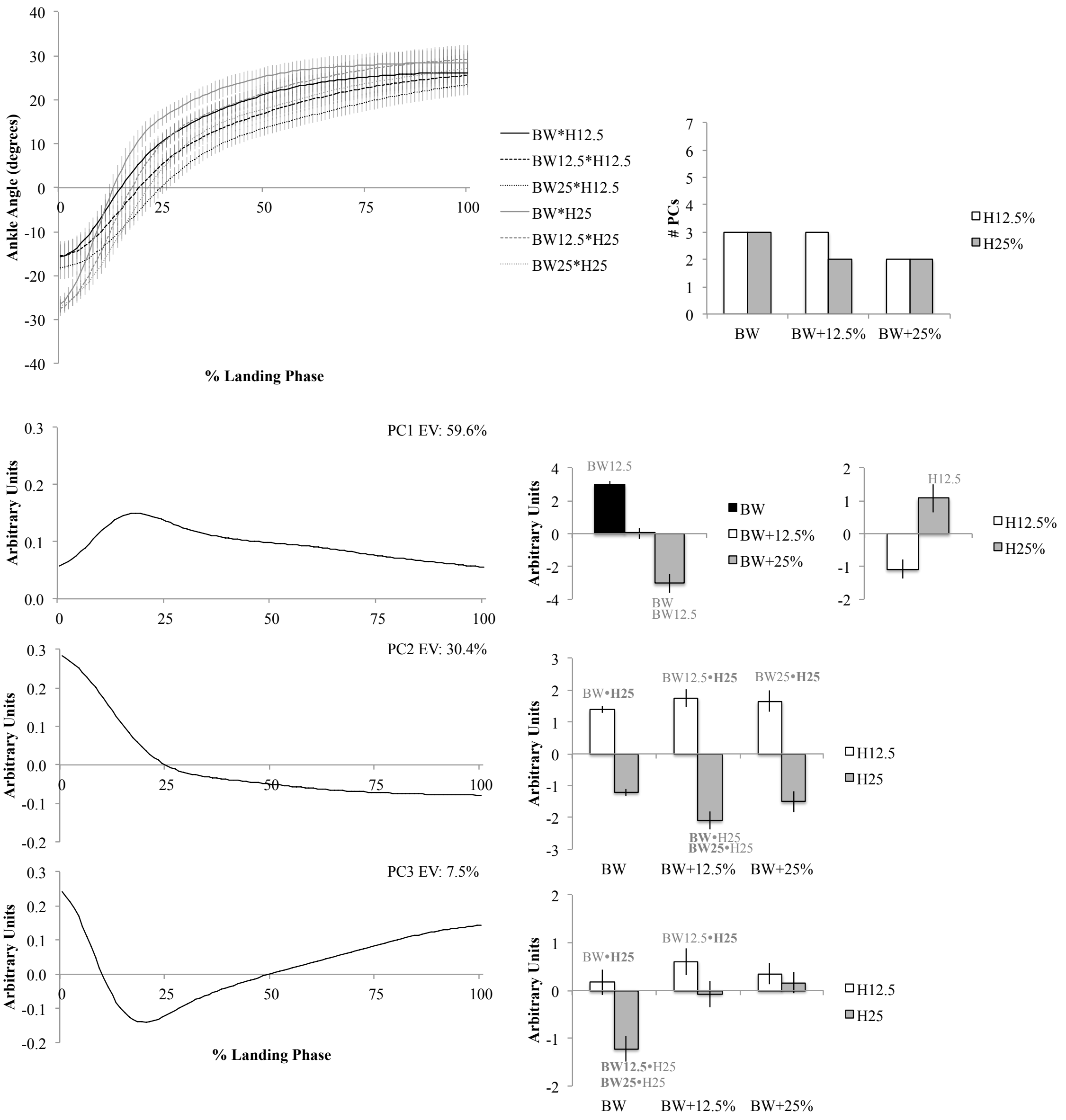


Figure 12: Ankle angle PCA.

(EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Hip moment

Increasing trends were observed for hip flexor moments (increasingly negative: ~0-25% landing phase) and hip extensor moments (increasingly positive: ~25-100% landing phase) with greater load and landing height (Figure 13a; top left). Later (% landing phase) hip moment increases (increasingly negative hip flexor moment and increasingly positive hip extensor moment) were observed with greater load from each landing height (Figure 13a; top left). Common trends were observed between hip moment time series plots (Figure 13a) and hip angular impulse values (Table 1), decreasing with greater load and increasing from greater height. From PCA, seven PCs provided 90% EV (Figure 13a and 13b; left columns). Decreasing trends were observed for hip moment variability from each landing height, relative to BW (number of extracted PCs; Figure 13a; top right).

PC1: Increased hip moments were observed across the landing phase in the PC1 loading vector (29.6% EV; Figure 13a; second row). Significant load and landing height main effects were observed ($F[2, 36] = 22.1$, $p < .001$, $\eta^2 = .55$; $F[1, 18] = 16.5$, $p = .001$, $\eta^2 = .48$; respectively). PC1 scores decreased with load ($p \leq .017$; Figure 13a; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p = .001$; Figure 13a; second row). Lesser hip moments were observed with greater load, while greater hip moments were observed with greater landing height in PC1.

PC2: Increased hip moments were observed from approximately 10-40% and 60-80% of the landing phase in the PC2 loading vector (17.5% EV; Figure 13a; third row). A significant landing height main effect was observed ($F[1, 18] = 15.6$, $p = .001$, $\eta^2 = .46$). Greater PC1 scores were observed from H12.5% relative to H25% ($p = .001$; Figure 13a;

third row). Greater hip moments were observed at lesser landing height from approximately 10-40% and 60-80% of the landing phase in PC2.

PC3: Increased hip moments were observed from approximately 0-30% and 40-70% of the landing phase in the PC3 loading vector (16.1% EV; Figure 13a; fourth row). A significant landing height main effect was observed ($F[1, 18] = 6.2$, $p = .023$, $\eta^2 = .26$). Greater PC3 scores were observed from H12.5% relative to H25% ($p = .023$; Figure 13a; fourth row). Greater hip moments were observed at lesser landing height from approximately 0-30% and 40-70% of the landing phase in PC3.

PC4: Increased hip moments were observed from approximately 10-30% of the landing phase in the PC4 loading vector (10.2% EV; Figure 13a; fifth row). Significant load and landing height main effects were observed ($F[2, 36] = 29.7$, $p < .001$, $\eta^2 = .62$; $F[1, 18] = 16.4$, $p = .001$, $\eta^2 = .48$). PC4 scores decreased with load ($p \leq .016$; Figure 13a; fifth row). Greater PC4 scores were observed from H12.5% relative to H25% ($p = .001$; Figure 13a; fifth row). Greater hip moments were observed at lesser load and landing height from approximately 10-30% of the landing phase in PC4.

PC5: Increased hip moments were observed from approximately 0-20%, 30-50%, and 65-90% of the landing phase in the PC5 loading vector (9.1% EV; Figure 13b; first row). PC5 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

PC6: Increased hip moments were observed from approximately 25-50% of the landing phase in the PC6 loading vector (5.8% EV; Figure 13a; second row). Significant load and landing height main effects were observed ($F[1.4, 25.2] = 4.8$, $p = .014$, $\eta^2 = .21$; $F[1, 18] = 20.2$, $p < .001$, $\eta^2 = .53$). At BW+12.5% PC6 scores exceeded BW+25%

(*p* = .004; Figure 13b; second row). Greater PC6 scores were observed from H12.5% relative to H25% (*p* < .001; Figure 13b; second row). Greater hip moments were observed at BW+12.5% relative to BW+25% and from lesser landing height across approximately 25-50% of the landing phase in PC4.

PC7: Increased hip moments were observed from approximately 30-75% of the landing phase in the PC7 loading vector (4.2% EV; Figure 13b; third row). A significant landing height main effect as observed ($F[1, 18] = 5.9$, $p = .026$, $\eta^2 = .25$). Greater PC7 scores were observed from H25% relative to H12.5% (*p* = .026; Figure 13b; second row). Greater hip moments were from greater landing height from approximately 30-75% of the landing phase in PC7.

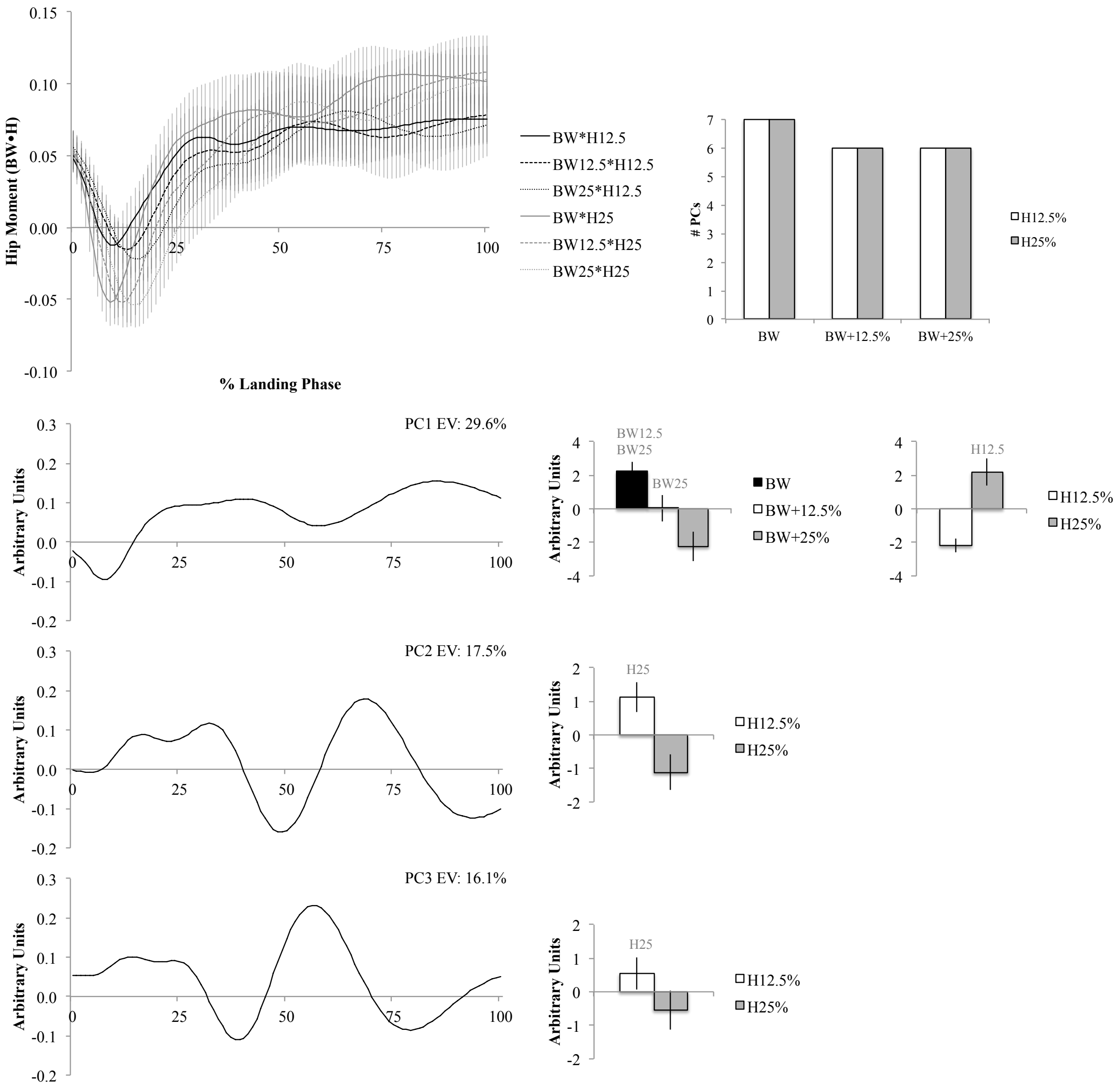


Figure 13a: Hip moment PCA a.

(PC1-4; EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

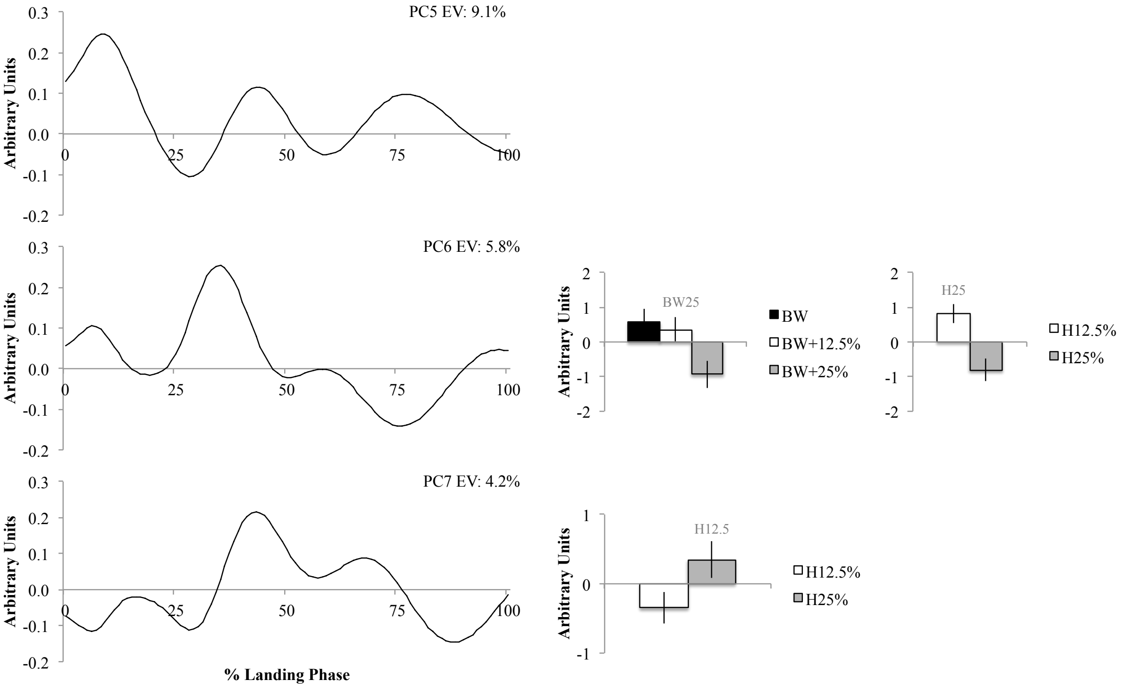


Figure 13b: Hip moment PCA b.

(PC5-7; EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Knee moment

Increasing trends were observed for knee extensor moments with greater load and landing height (Figure 14; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 14; top left). From PCA, four PCs provided 90% EV (Figure 14; left column). Knee moment variability remained consistent across conditions (number of extracted PCs; Figure 14; top right).

PC1: Increased knee moments were observed from approximately 0-50% of the landing phase in the PC1 loading vector (48.7% EV; Figure 14; second row). Significant load and landing height main effects were observed ($F[1.4, 25.9] = 121.6$, $p < .001$, $\eta^2 = .87$; $F[1, 18] = 118.7$, $p < .001$, $\eta^2 = .87$; respectively). PC1 scores decreased with load ($p < .001$; Figure 14; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 14; second row). Lesser knee moments were observed with greater load, while greater knee moments were observed with greater landing height from approximately 0-50% of the landing phase in PC1.

PC2: Knee moment increases were observed from approximately 30-100% of the landing phase in the PC2 loading vector (26.2% EV; Figure 14; third row). PC2 scores were influenced by the interaction of load and landing height ($F[2, 36] = 53.2$, $p < .001$, $\eta^2 = .75$). When landing from H12.5%, greater PC2 scores were observed at BW+12.5% relative to BW+25% ($p = .002$; Figure 14; third row). When landing from H25%, greater PC2 scores were observed at BW+12.5% and BW+25% relative to BW ($p < .001$; Figure 14; third row). At BW+12.5% and BW+25%, greater PC2 scores from H25% relative to H12.5% ($p = .009$; Figure 14; third row).

PC3: Knee moment increases were observed from approximately 15-30% and 60-100% of the landing phase in the PC3 loading vector (10.9% EV; Figure 14; fourth row). PC3 scores were influenced by the interaction of load and landing height (*F*[2, 36] = 3.5, $p = .040$, $\eta^2 = .16$). At BW and BW+25%, greater PC3 scores were observed from H25% relative to H12.5% ($p \leq .011$; Figure 14; fourth row).

PC4: Knee moment increases were observed from approximately 0-25% and 40-70% of the landing phase in the PC4 loading vector (5.7% EV; Figure 14; fifth row). A significant landing height main effect was observed (*F*[1, 18] = 39.1, $p < .001$, $\eta^2 = .69$). Greater PC4 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 14; fifth row). Greater knee moments were observed from greater landing height from approximately 0-25% and 40-70% of the landing phase in PC4.

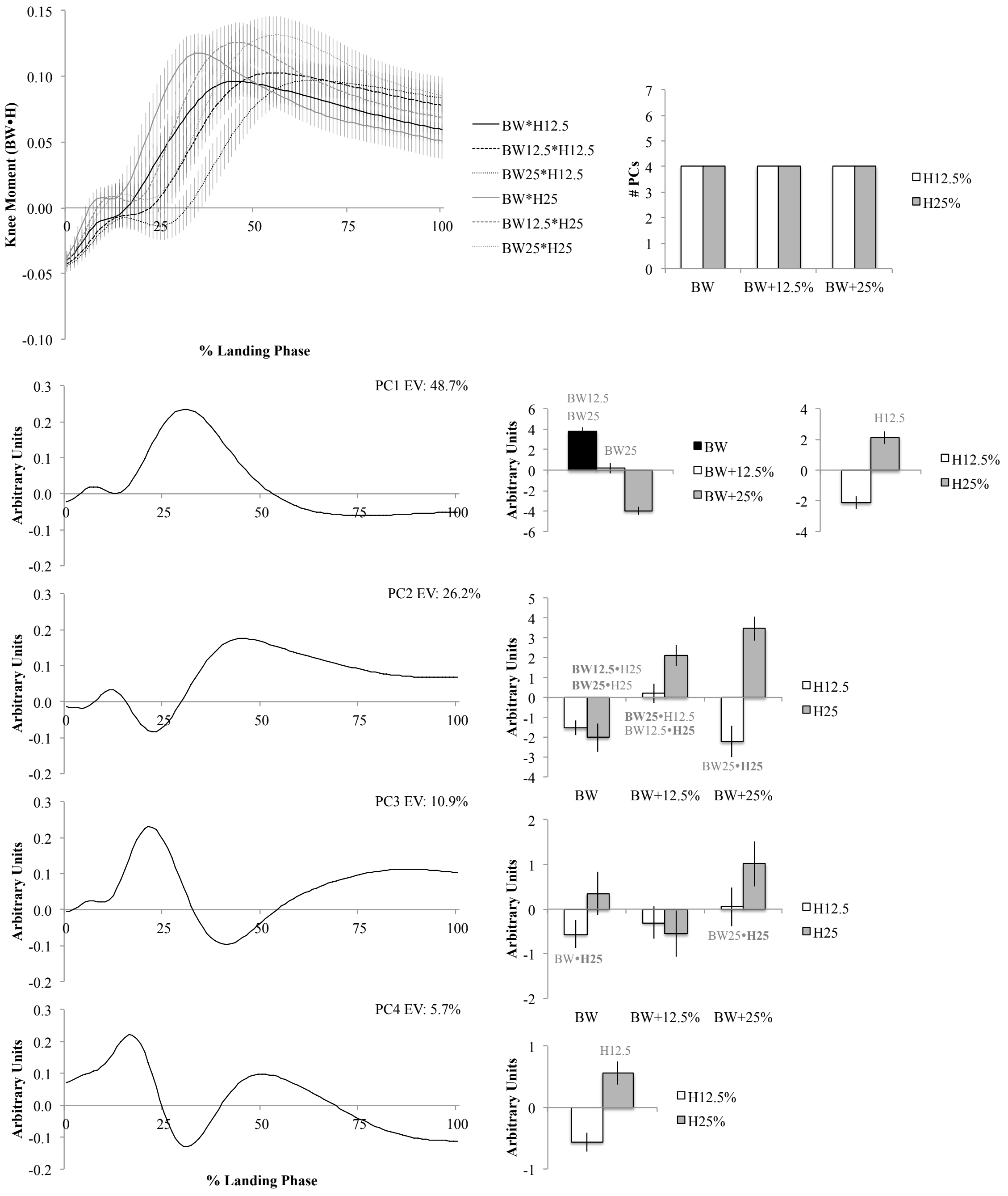


Figure 14: Knee moment PCA.

(EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Ankle moment

Increasing trends were observed for ankle plantarflexor moments with greater load and landing height (Figure 15; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 15; top left). From PCA, three PCs provided 90% EV (Figure 15; left column). Decreasing trends for ankle moment variability were observed at BW+25% from each landing height (number of extracted PCs; Figure 15; top right).

PC1: Increased ankle moments were observed across the landing phase in the PC1 loading vector (51.9% EV; Figure 15; second row). PC1 scores were influenced by the interaction of load and landing height ($F[2, 36] = 3.5$, $p = .042$, $\eta^2 = .16$). When landing from H12.5%, greater PC1 scores were observed at BW+25% relative to BW and BW+12.5% ($p \leq .003$; Figure 15; second row). When landing from H25%, PC2 scores increased with load ($p \leq .001$; Figure 15; second row). At BW, BW+12.5%, and BW+25% greater PC1 scores were observed from H25% relative to H12.5% ($p < .001$; Figure 15; second row). Greater ankle moments were observed with greater load and landing height across the landing phase in PC1.

PC2: Ankle moment increases were observed from approximately 0-40% of the landing phase in the PC2 loading vector (30.9% EV; Figure 15; third row). PC2 scores were influenced by the interaction of load and landing height ($F[1.6, 28.7] = 7.4$, $p = .004$, $\eta^2 = .29$). PC2 scores decreased with load at each landing height ($p < .001$; Figure 15; third row). In each load condition PC2 scores were greater from H25% relative to H12.5% ($p \leq .002$; Figure 15; third row). Lesser ankle moments were observed with greater load, while greater ankle moments were observed with greater landing height from approximately 0-40% of the landing phase in PC2.

PC3: Ankle moment increases were observed from approximately 0-25% and 60-100% of the landing phase in the PC3 loading vector (7.4% EV; Figure 15; fourth row). PC3 scores were influenced by the interaction of load and landing height ($F[1.5, 26.9] = 6.3$, $p = .010$, $\eta^2 = .26$). From H25%, PC3 scores were greater at BW relative to BW+12.5% and BW+25% ($p \leq .016$; Figure 15; fourth row). At BW, PC3 scores were greater from H25% relative to H12.5% ($p < .001$; Figure 15; fourth row).

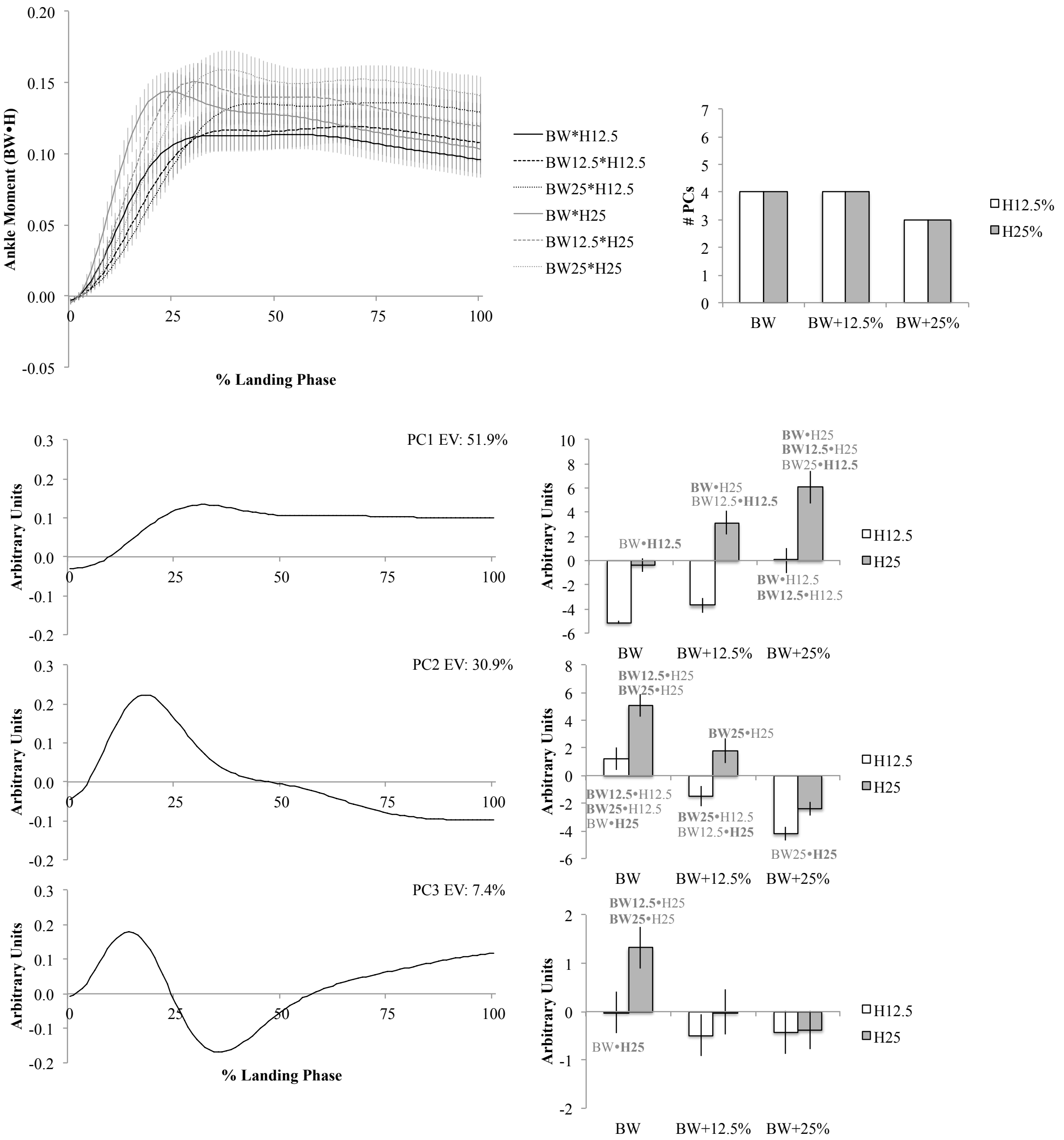


Figure 15: Ankle moment PCA.

(EV is explained variance, BW is bodyweight, H is participant height, # PCs is number of principal components)

Gluteus maximus

Increasing trends were observed for gluteus maximus muscle activity with greater load and landing height (Figure 16; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 16; top left). From PCA, three PCs provided 90% EV (Figure 16; left column). Decreasing trends for muscle activity variability were observed with greater mechanical task demands (# PCs; Figure 16; top right).

PC1: Increased gluteus maximus muscle activity was observed across the landing phase in the PC1 loading vector (82.7% EV; Figure 16; second row). A significant height main effect was observed ($F[1, 18] = 12.5$, $p = .002$, $\eta^2 = .41$). Greater PC1 scores were observed from H25% relative to H12.5% ($p = .002$; Figure 16; second row). Greater muscle activity was observed with greater landing height across the landing phase in PC1.

PC2: Increased gluteus maximus muscle activity was observed from approximately 50-100% of the landing phase in the PC2 loading vector (6.1% EV; Figure 16; third row). A significant load main effect was observed ($F[1.6, 29.0] = 15.5$, $p < .001$, $\eta^2 = .46$). Greater PC2 scores were observed across load conditions ($p \leq .017$; Figure 16; third row). Greater gluteus maximus muscle activity was observed with greater load approximately 50-100% of the landing phase in PC1.

PC3: Increased gluteus maximus muscle activity was observed from ~40-75% of the landing phase in the PC3 loading vector (4.8% EV; Figure 16; fourth row). A significant landing height main effect was observed ($F[1, 18] = 5.4$, $p = .032$, $\eta^2 = .23$). Greater PC3 scores were observed at lesser landing height ($p = .032$; Figure 16; fourth row).

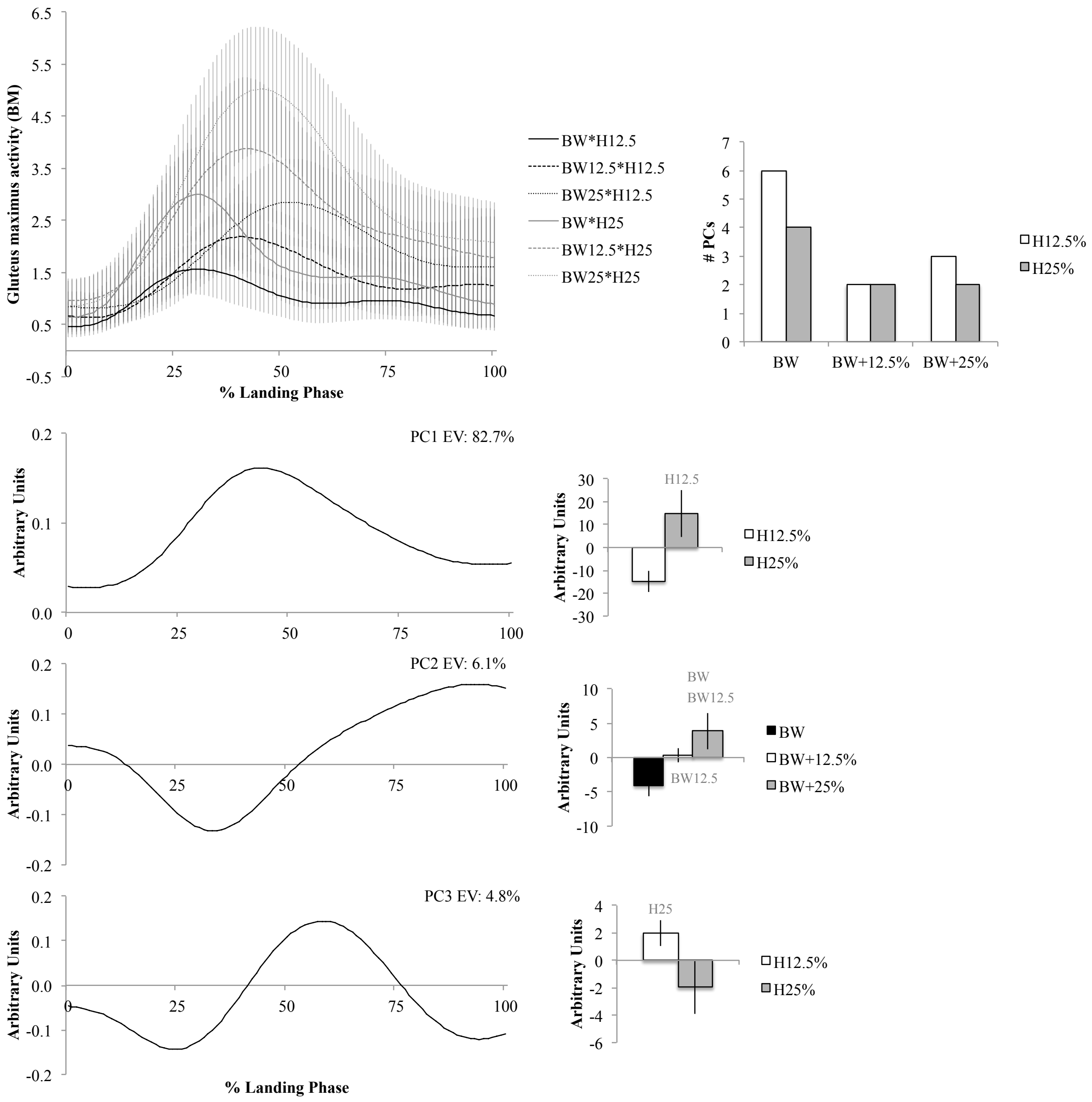


Figure 16: Gluteus maximus PCA.

(EV is explained variance, BW is bodyweight, H is participant height, BM is baseline multiple, # PCs is number of principal components)

Biceps femoris

Increasing trends were observed for mean peak biceps femoris muscle activity with greater load and landing height (Figure 17; top left). Later (% landing phase) mean peak magnitudes were observed with greater load from each landing height (Figure 17; top left). From PCA, five PCs provided 90% EV (Figure 17; left column). Decreasing trends for biceps femoris muscle activity variability were observed with increasing mechanical task demands (number of extracted PCs; Figure 17; top right).

PC1: Increased biceps femoris muscle activity was observed across the landing phase in the PC1 loading vector (52.1% EV; Figure 17; second row). Significant load and landing height main effects were observed ($F[1.2, 22.0] = 4.3$, $p = .043$, $\eta^2 = .19$; $F[1, 18] = 7.7$, $p = .013$, $\eta^2 = .30$; respectively). PC1 score differences among load conditions failed to be detected in pairwise comparisons ($p > .05$). Greater PC1 scores were observed from H25% relative to H12.5% ($p = .013$; Figure 17; second row). Greater biceps femoris muscle activity was observed with greater landing height across the landing phase in PC1.

PC2: Increased biceps femoris muscle activity was observed from approximately 20-60% of the landing phase in the PC2 loading vector (13.8% EV; Figure 17; third row). Significant load and landing height main effects were observed ($F[2, 35.6] = 13.9$, $p < .001$, $\eta^2 = .44$; $F[1, 18] = 10.0$, $p = .005$, $\eta^2 = .36$; respectively). Lesser PC2 scores were observed at BW+25% relative to BW and BW+12.5% ($p \leq .002$; Figure 17; third row). Greater PC2 scores were observed from H25% relative to H12.5% ($p = .005$; Figure 17; third row). Greater biceps femoris muscle activity was observed at BW+25% and with greater landing height from approximately 20-60% of the landing phase in PC2.

PC3: Increased biceps femoris muscle activity was observed from approximately 0-30% of the landing phase in the PC3 loading vector (11.1% EV; Figure 17; fourth row). PC3 scores were influenced by the interaction of load and landing height (*F*[2, 36] = 7.23, $p = .002$, $\eta^2 = .29$). From H25%, PC3 scores were lesser at BW+25% relative to BW and BW+12.5% ($p = .029$; Figure 17; fourth row). At BW, PC3 scores were greater from H25% relative to H12.5% ($p = .032$; Figure 15; fourth row), while at BW+25%, PC3 scores were lesser from H25% relative to H12.5% (*p*.024; Figure 17; fourth row).

PC4: Increased biceps femoris muscle activity was observed from approximately 40-75% of the landing phase in the PC4 loading vector (8.8% EV; Figure 17; fifth row). Significant load and landing height main effects were observed (*F*[1.6, 29.2] = 3.7, $p = .045$, $\eta^2 = .17$; *F*[1, 18] = 5.6, $p = .030$, $\eta^2 = .24$; respectively). Greater PC4 scores were observed at BW+12.5% relative to BW% ($p = .006$; Figure 17; fifth row). Greater PC4 scores were observed from H25% relative to H12.5% ($p = .030$; Figure 17; fifth row).

PC5: Increased biceps femoris muscle activity was observed from approximately 10-40%, and 60-80% of the landing phase in the PC6 loading vector (6.5% EV; Figure 17; sixth row). PC6 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

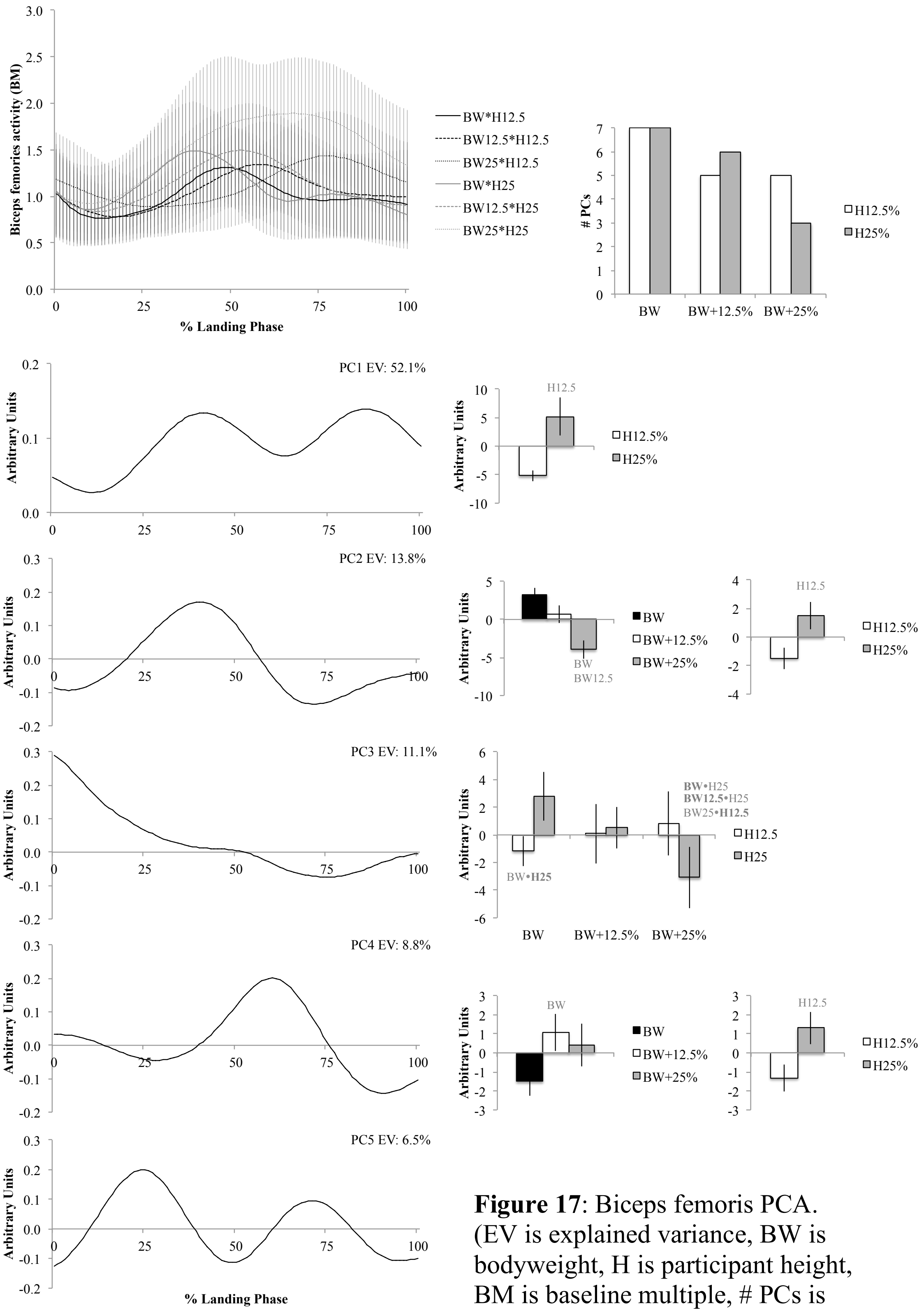


Figure 17: Biceps femoris PCA. (EV is explained variance, BW is bodyweight, H is participant height, BM is baseline multiple, # PCs is number of principal components)

Vastus medialis

Contrasting trends were observed for vastus medialis muscle activity among conditions (Figure 18; top left). Divergent trends were observed for mean peak magnitudes by load, while mean peak magnitudes showed increasing trends by landing height (Figure 18; top left). Later (% landing phase) increases were observed with greater load from each landing height (Figure 18; top left). From PCA, five PCs provided 90% EV (Figure 18; left column). Decreasing trends for vastus medialis muscle activity variability were observed with increasing mechanical task demands (number of extracted PCs; Figure 18; top right).

PC1: Increased vastus medialis muscle activity was observed across the landing phase in the PC1 loading vector (56.4% EV; Figure 18; second row). Significant load and landing height main effects were observed ($F[1.6, 29.3] = 7.1$, $p = .005$, $\eta^2 = .28$; $F[1, 18] = 10.3$, $p = .005$, $\eta^2 = .36$; respectively). Lesser PC1 scores were observed at BW relative to BW+12.5% and BW+25% ($p \leq .034$; Figure 18; second row). Greater PC1 scores were observed from H25% relative to H12.5% ($p = .005$; Figure 18; second row). Greater vastus medialis muscle activity was observed at BW+12.5% and BW and with greater landing height across the landing phase in PC1.

PC2: Increased vastus medialis muscle activity was observed from approximately 65-100% of the landing phase in the PC2 loading vector (12.1% EV; Figure 18; third row). A significant load main effect was observed ($F[2, 36] = 5.4$, $p = .009$, $\eta^2 = .23$). Greater PC2 scores were observed at BW+25% relative to BW ($p = .027$; Figure 18; second row). Greater vastus medialis muscle activity was observed at BW+25% relative to BW from approximately 65-100% of the landing phase in PC2.

PC3: Increased vastus medialis muscle activity was observed from approximately 0-35% of the landing phase in the PC3 loading vector (10.7% EV; Figure 18; fourth row).). Significant load and landing height main effects were observed ($F[2, 36] = 6.9$, $p = .003$, $\eta^2 = .28$; $F[1, 18] = 13.2$, $p = .002$, $\eta^2 = .42$; respectively). Greater PC3 scores were observed at BW relative to BW+25% (p =.019; Figure 18; fourth row). Greater PC3 scores were observed from H25% relative to H12.5% ($p = .002$; Figure 18; fourth row). Greater vastus medialis muscle activity was observed at BW and with greater landing height from approximately 0-35% of the landing phase in PC3.

PC4: Increased vastus medialis muscle activity was observed from approximately 0-25% and 50-80% of the landing phase in the PC4 loading vector (7.9% EV; Figure 18; fifth row). A significant load main effect was observed ($F[2, 36] = 7.9$, $p = .001$, $\eta^2 = .30$). Greater PC4 scores were observed at BW% relative to BW+25% ($p = .004$; Figure 18; fifth row). Greater vastus medialis muscle activity was observed at BW from 0-25% and 50-80% of the landing phase in PC4.

PC5: Increased vastus medialis muscle activity was observed from approximately 25-45%, and 65-85% of the landing phase in the PC5 loading vector (5.5% EV; Figure 17; fifth row). PC5 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

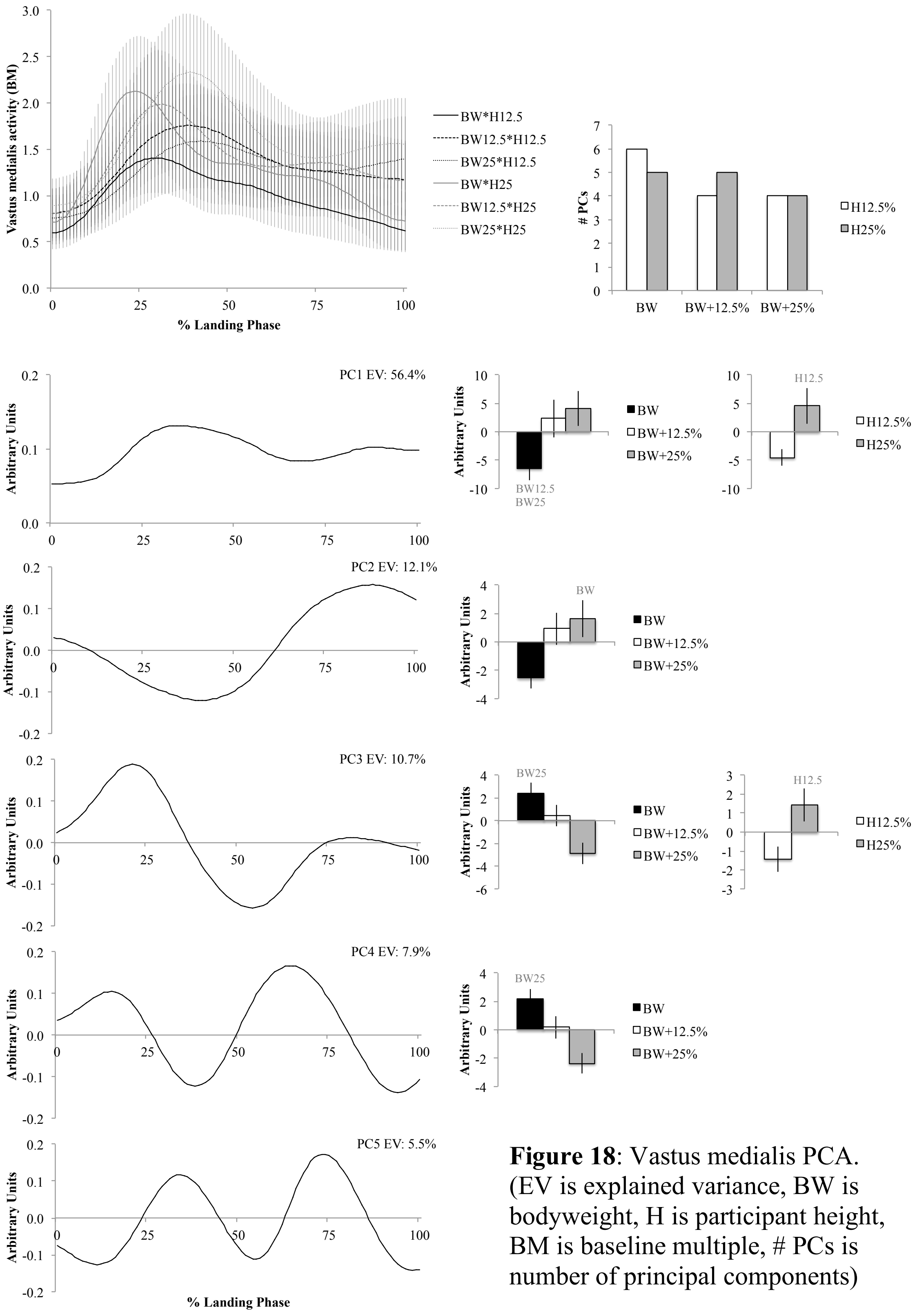


Figure 18: Vastus medialis PCA. (EV is explained variance, BW is bodyweight, H is participant height, BM is baseline multiple, # PCs is number of principal components)

Medial gastrocnemius

Contrasting trends were observed for medial gastrocnemius muscle activity among conditions (Figure 19; top left). Mean magnitudes generally decreased across the landing phase, with a local peak occurring from approximately 20-70% of the landing phase. Local mean peak magnitudes (~20-70% landing phase) showed increasing trends by landing height (Figure 19; top left), while decreasing trends were observed with increasing load at H25%, in contrast to H12.5% (Figure 19; top left). Later (% landing phase) local mean peak values were observed with increasing load from each landing height (Figure 19; top left). From PCA, five PCs provided 90% EV (Figure 19; left column). Decreasing trends were observed for medial gastrocnemius muscle activity variability when landing from H12.5%, while medial gastrocnemius muscle activity variability from H25% remained consistent across load conditions (number of extracted PCs; Figure 19; top right).

PC1: Increased medial gastrocnemius muscle activity was observed across the landing phase in the PC1 loading vector (45.8% EV; Figure 19; second row). A significant landing height main effect was observed ($F[1, 18] = 6.1$, $p = .024$, $\eta^2 = .25$; respectively). Greater PC1 scores were observed from H25% relative to H12.5% ($p = .024$; Figure 19; second row). Greater medial gastrocnemius muscle activity was observed with greater landing height across the landing phase in PC1.

PC2: Increased medial gastrocnemius muscle activity was observed from approximately 0-25% of the landing phase in the PC2 loading vector (21% EV; Figure 19; third row). PC2 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

PC3: Increased medial gastrocnemius muscle activity was observed from approximately 0-20% and 40-100% of the landing phase in the PC3 loading vector (16.3%% EV; Figure 19; fourth row). PC3 scores were influenced by the interaction of load and landing height (*F*[2, 36] = 4.7, $p = .016$, $\eta^2 = .21$). From H12.5% and H25%, PC3 scores were greater at BW+25% relative to BW+12.5% ($p \leq .0.010$; Figure 19; fourth row). From H25%, PC3 scores were greater at BW+25% relative to BW ($p = 006$; Figure 19; fourth row). Greater medial gastrocnemius muscle activity was observed at BW+25% from approximately 0-20% and 40-100% of the landing phase in PC3.

PC4: Increased medial gastrocnemius muscle activity was observed from approximately 30-70% of the landing phase in the PC4 loading vector (6.6% EV; Figure 19; fifth row). PC4 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

PC5: Increased medial gastrocnemius muscle activity was observed from approximately 10-25% and 50-80% of the landing phase in the PC5 loading vector (4.5% EV; Figure 19; sixth row). PC5 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

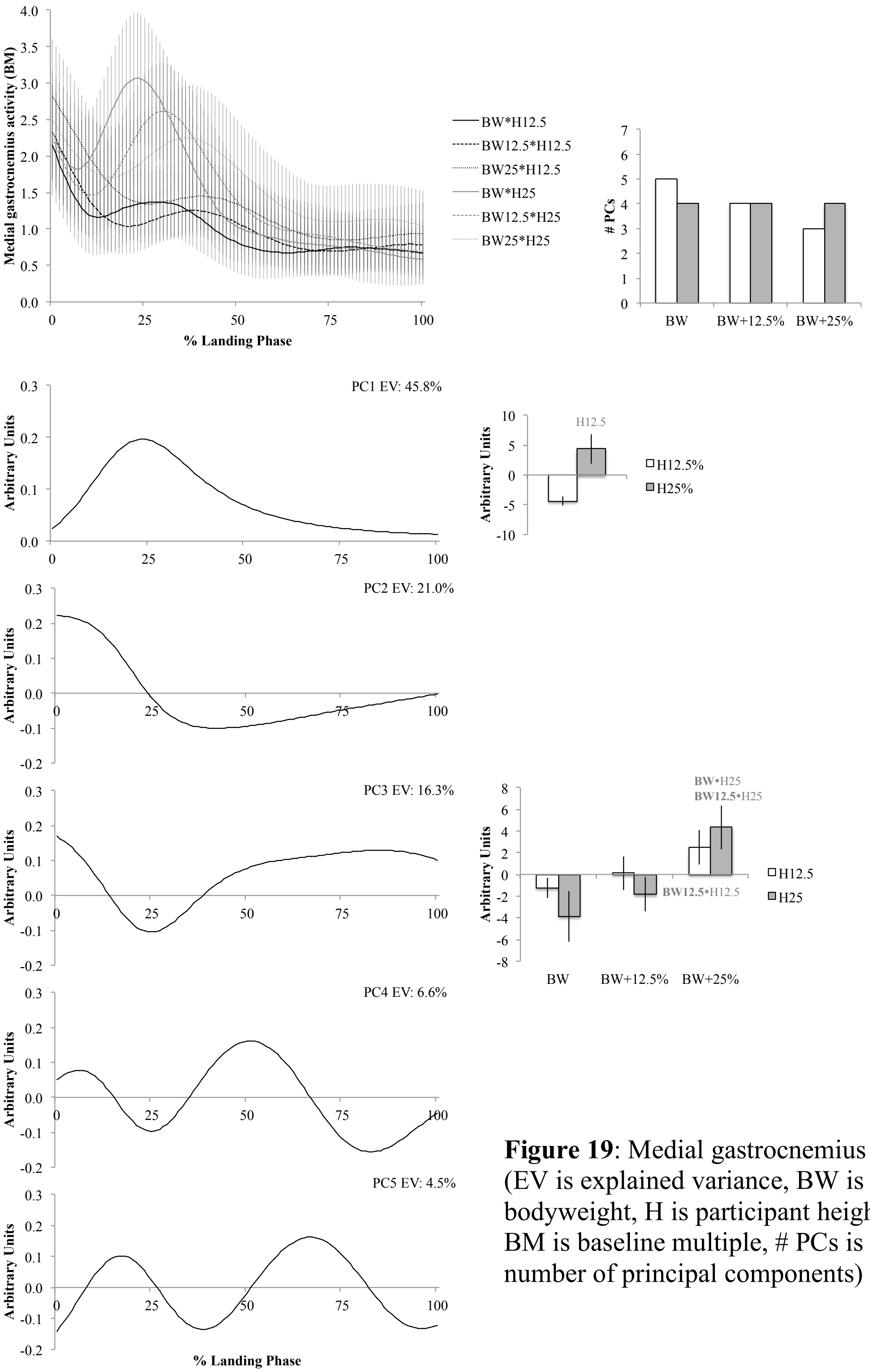


Figure 19: Medial gastrocnemius PCA. (EV is explained variance, BW is bodyweight, H is participant height, BM is baseline multiple, # PCs is number of principal components)

Tibialis anterior

Tibialis anterior muscle activity showed increasing trends by load and landing height (Figure 20; top left). Later (% landing phase) increases were observed with increasing load from each landing height (Figure 20; top left). From PCA, three PCs provided 90% EV (Figure 20; left column). Decreasing trends were observed for tibialis anterior muscle activity variability with increasing mechanical task demands (number of extracted PCs; Figure 18; top right).

PC1: Increased tibialis anterior muscle activity was observed across the landing phase in the PC5 loading vector (66.6% EV; Figure 20; second row). PC1 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

PC2: Increased tibialis anterior muscle activity was observed from approximately 50-100% of the landing phase in the PC2 loading vector (18.0% EV; Figure 20; second row). A significant load main effect was observed ($F[2, 36] = 8.0, p = .001, \eta^2 = .31$; respectively). Lesser PC2 scores were observed at BW relative to BW+12.5% and BW+25% ($p \leq .031$; Figure 19; second row). Greater tibalis anterior muscle activity was observed with greater load height from approximately 50-100% of the landing phase in PC2.

PC3: Increased tibialis anterior muscle activity was observed from approximately 0-35% and 70-100% of the landing phase in the PC3 loading vector (7.1% EV; Figure 20; fourth row). PC3 was not influenced by the interaction of load and landing height; neither load nor landing height main effects were significant ($p > .05$).

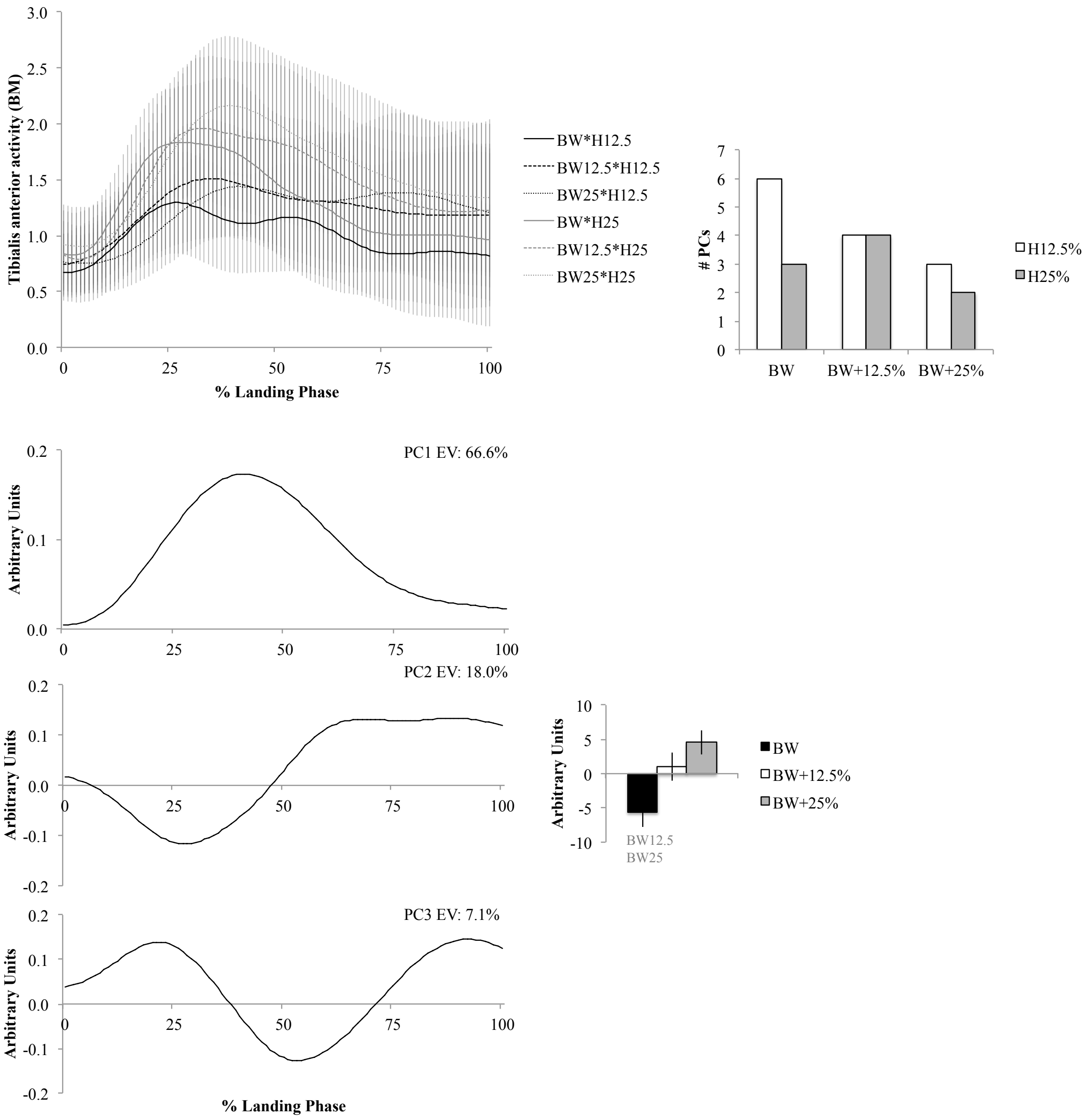


Figure 20: Tibialis anterior PCA.

(EV is explained variance, BW is bodyweight, H is participant height, BM is baseline multiple, # PCs is number of principal components)

CHAPTER 5

Discussion

The comprehensive results from this study have presented a scenario in single-leg landing where participants demonstrated fewer emergent load accommodation strategies by way of lower extremity mechanical and neuromuscular adjustments, potentially highlighting characteristics of movement control with implications for mechanisms of injury. The purpose of this study was to evaluate changes in movement variability among kinematic, kinetic, and electromyographic (EMG) variables following mechanical task demand manipulations during single-leg drop landings. As a result, collective movement variability changes were assessed from each of these sources, alongside variable-specific adjustments following mechanical task demand manipulations. Assessments among integrated time series variables (Table 1), subject-specific load accommodation strategies (Figure 6), collective movement variability (Figure 7), and variable-specific modifications (Figures 9-20) each provided insight into the nature and sources of movement variability changes in single-leg landing. Descriptive and statistical analyses were conducted, with the use of PCA providing multiple avenues of data exploration. In research, PCA has been used as a powerful data reduction tool that can be applied in a number of ways, dependent on matrix assembly procedures (Brandon et al., 2013; Cohen, 2014; Daffertshofer et al., 2004; Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Kipp et al., 2014; Li, 2006; Lohse et al., 2013; Molenaar et al., 2013; Robbins et al., 2013; Wang, Molenaar, & Newell, 2013). Throughout this investigation, five separate PCA procedures were used, including supplementary analyses, which differed in matrix assembly and interpretable outcomes (Appendix II, supplementary statistical analyses, collective movement variability, and variable-specific adjustments [follow-up PCA]).

1.) Collective movement variability, appended time series variables (# PCs; Figure 7, top).
    a. Collective movement variability, group synergies (# PCs; Figure 7, middle; Appendix II, supplementary statistical analyses, collective movement variability [PCA]).
    b. Collective movement variability, single-subject synergies (# PCs; Figure 7, bottom; Appendix II, supplementary statistical analyses, collective movement variability [PCA]).
2.) Variable-specific adjustments (follow-up PCA), PC scores (Figures 9-20, right column).
    a. Variable-specific adjustments (follow-up PCA), variability (# PCs; Figures 9-20, top right; Appendix II, supplementary statistical analyses, variable-specific adjustments [follow-up PCA]).

Collective movement variability assessments were carried out from the number of extracted PCs in each load and landing height condition, while follow-up PCA assessed changes in variable-specific movement patterns through inferential testing of PC scores in relation to the extracted PC loading vectors (Deluzio et al., in Robertson et al., 2014). Follow-up PCA therefore highlighted the ability of PCA to identify subtle biomechanical changes in the time series of each variable across mechanical task demand manipulations (Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Federolf et al., 2012). Overall, the following discussion aims to synthesize the results from each analysis, reducing the interpretable information into functional outcomes. As well, the shared results among movement pattern adjustments will be combined in the general discussion.

Descriptive statistics

Integrated time series movement variables provided insight into the effects of mechanical task demand manipulations on single-leg landing biomechanics (Table 1). In contrast to conventional analysis of peak magnitudes extracted from the time series of each respective outcome variable, variable integration simultaneously assessed magnitude and temporal characteristics within each variable. A limitation of integrated variable assessment, however, is the inability to distinguish between magnitude and temporal contributions (James et al., 2003). Further, specific insight into movement variable patterns was not gained, rather the product of magnitude and temporal characteristics were evaluated together. Despite these limitations, variable integration allowed mechanical prediction of landing impulse (integrated GRFz), which was used in load accommodation strategy identification (James et al., 2003; James et al., 2014). As a result, each outcome variable was consistently expressed along with landing impulse as a means of descriptively examining biomechanical changes following mechanical task demand manipulations (Table 1).

Relatively consistent landing impulse (GRFz; BW•s) with greater loads at each landing height provided initial insight into load accommodation, while greater landing height identified predictable trends toward greater landing impulse (Table 1; James et al., 2000; James et al. 2003; Seegmiller & McCaw, 2003). Decreasing trends for integrated hip, knee, and ankle angles with greater load (deg•s), alongside decreasing trends for hip, knee, and ankle angular impulse (BW•H•s), as well as landing duration (s), highlight tradeoffs occurring during movement pattern adjustments (James et al., 2000; James et al. 2003). Increasing trends were, however, observed when landing from greater height among integrated hip, knee, and ankle angles, angular impulse variables, and landing duration, which demonstrated similarity to previous research (James et al., 2000; James et al., 2003). Increasing trends were also observed among integrated muscle activities (iEMG) when landing from greater height, though contrasting trends were observed among muscles with greater load. Notably, gluteus maximus iEMG increased with both load and landing height in relation to the remaining lower extremity muscle, while decreasing trends were observed at greater loads for biceps femoris and tibialis anterior muscles from H12.5% and vastus medialis, medial gastrocnemius, and tibialis anterior muscles from H25% (Table 1). Decreasing iEMG trends among muscles controlling the knee and ankle joints fall in line with predictions of decreased muscle activity during eccentric activity (Bishop et al., 2000; Guilhem et al., 2010; James et al., 2006; Linnamo et al., 2000; Mianfang & Li, 2010; Westing et al., 1991), which may have implications on the observed load accommodation strategies and movement variability (James et al., 2000; James et al., 2003). Although descriptive assessment of the integrated biomechanical variables does not provide comprehensive insight into load accommodation strategies, variable specific alterations during landing were assessed in detail during follow-up PCA.

Load accommodation strategy identification

In agreement with the outlined hypotheses, fewer emergent strategies were observed with greater mechanical task demands, assessed from single-subject load accommodation strategy classifications in each condition. Greater occurrence of Fully Accommodating and Negative Biomechanical responses when landing from H12.5%, and greater occurrence of Negative Biomechanical responses with increased load from H12.5%, may highlight protective neuromuscular response strategies, accommodating loads as a result of perceived dangers to the system (Figure 6; Caster & Bates, 1995; James et al., 2003). Positive Biomechanical strategies were observed with greater occurrence among participants with greater landing height and load (Figure 6). As hypothesized, constraints imposed on the biomechanical system through neuromuscular activity appeared to limit emergent strategies, exceeding participant capacities for full accommodation, while protectively reducing landing impulse relative to Newtonian and Super-Newtonian responses (Figure 6; Caster & Bates, 1995; James et al., 2003; James et al., 2014). Greater occurrence of Positive Biomechanical responses imply partial neuromuscular accommodation with greater load and landing height, with the rate of landing impulse increase lesser than predicted (Caster & Bates, 1995; James et al., 2003).

Although variable-specific adjustments were explored during the landing phase (ground contact to zero vCOMz), post-landing adjustments (after ground contact) were not responsible for the observed landing strategies predicted from impulse ratios. Mechanical predictions of landing impulse from the conversion of gravitational potential to kinetic energy, and subsequent landing momentum, explicitly outline the dependence of landing impulse on ground contact velocity (ground contact vCOMz; v in Equations 3, 4, and 5; James, 2003; Niu, Feng, Jiang, & Zhang, 2014). Adjustments made after ground contact alter the GRFz-time pattern, while landing impulse is determined by ground contact vCOMz (mass remains constant in each trial-condition; Equations 4 and 5). As a result, supplementary analysis was performed on pre-landing strategy (Appendix II, supplementary statistical analyses, pre-landing strategy), as a means of identifying mechanisms by which participants altered landing impulse relative to mechanical predictions. Observed vCOMz was evaluated from kinematic analysis, while expected vCOMz was mechanically predicted (Equation 3; James, 2003; Niu et al., 2014). Mean percent difference between observed and expected vCOMz at ground contact indicated that participants contacted the ground with lesser velocity than predicted, which decreased with greater load and landing height (Table1, Appendix II, supplementary statistical analyses, pre-landing strategy). The manner in which this was accomplished can be attributed to body segment configuration adjustments at step-off and ground contact, components of the overall landing strategy, which decrease effective landing height, flight time, and ground contact velocity (Dufek & Bates, 1990; Dufek & Zhang, 1996). Body position at step-off was not evaluated during kinematic analysis, while segment configurations will be discussed when considering variable-specific adjustments in follow-up PCA.

Additional load accommodation strategy verification was carried out using observed ground contact vCOMz from kinematic analysis as a means of computing predicted landing impulse (Appendix II, supplementary statistical analyses, pre-landing strategy; James, 2003; Niu et al., 2014). Observed ground contact vCOMz was used in computing expected landing impulse (Equation 4), while observed landing impulse was computed from force platform analysis (GRFz impulse, summarized in Table 1; Appendix II, supplementary statistical analyses, pre-landing strategy). Mean percent difference was computed between observed and expected landing impulse in each trial, demonstrating agreement among means from trials and participants in each condition (-3.1% to 0.2% range in mean difference among conditions; Table 3; Appendix II, supplementary statistical analyses, pre-landing strategy). A strong relationship was demonstrated between observed and expected landing impulse ($r = 0.995$; Appendix II, supplementary statistical analyses, pre-landing strategy) after accounting for body segment configuration adjustments prior to ground contact.

Limitations are acknowledged in the interpretations of observed landing strategies from the assessment of landing impulse ratios relative to the mechanically predicted values. Figure 6 identifies wider impulse ratio ranges defining Positive Biomechanical load accommodation strategies with greater load and landing height (95% CI < Newtonian & > Fully Accommodating). Notably, Fully Accommodating impulse ratios (1.00) are equivalent in each condition comparison, while Newtonian impulse ratios increase with load and landing height (Figure 6). Lesser sensitivity in detecting non-Positive Biomechanical responses in each condition comparison may have implications on interpretations surrounding the occurrence of each load accommodation strategy (Figure 6, frequency in each condition). An alternative approach may have limited comparisons to adjacent load conditions in each respective height condition, but would have discarded information provided by the study design.

Collective movement variability (PCA)

Decreasing trends for collective movement variability, evaluated from the number of PCs extracted in each experimental condition, fall in line with the outlined hypotheses and concepts drawn from the motor control literature. Lesser collective movement variability with increasing mechanical task demands were therefore considered in the context of movement control, relating functional movement outcomes to factors potentially linked to injury (Barrett et al., 2008; Bartlett et al., 2007; Brown et al., 2012; James et al., 2000; James, in Stergiou, 2004, p. 29-62). Decreased movement variability with greater load and landing height (Figure 7; top) theoretically provide insight into lesser system flexibility, limiting adaptability to unexpected changes in the environment under greater task demands (Barrett et al., 2008; James et al., 2000; Li et al., 2005; Stergiou & Decker, 2011; Stergiou et al., 2006). Emergent movement solutions arise in response to constraints shaping movement patterns, where lesser collective movement variability observed at greater task demands identified dimension reduction, protectively freezing functional DOF via neural and mechanical synergies, in an attempt to accommodate greater external forces (Bernstein, 1967, p. 161-164; Chvatal & Ting, 2012; Diedrichsen et al., 2009; Todorov, 2006; Turvey, 1990). The numbers of extracted PCs in each condition were therefore considered indirectly representative of a number of functional DOF available in completing the task (Diedrichsen et al., 2009; Latash, 2010; Li, 2006; Lohse et al., 2013; Scott, 2012; Todorov, 2006; Turvey, 1990). Additionally, the proposed anisotropic nature of movement variability was highlighted from PCA, where a greater number of orthogonal PCs with lesser mechanical task demands fall in line with concepts and theories from motor control (Deluzio et al., in Robertson et al., 2014; Diedrichsen et al., 2009; Federolf et al., 2013; Li, 2006; Lohse et al., 2013; Todorov, 2006). Accumulated variation on task irrelevant dimensions appeared to be identified at lesser mechanical task demands during single-leg landing, expressed through a greater number of PCs.

Although cognitive processing was not assessed in this investigation, the experimental manipulations of load and landing height may have understandably required greater conscious control and pre-planning prior to movement execution at greater mechanical task demands, particularly with the eccentric requirements of the movement (Fang et al., 2004; Guilhem et al., 2010). Further evidence may be provided from lesser ground contact vCOMz with greater load and landing height (Appendix II, supplementary statistical analyses, pre-landing strategy). Decreased functional variability may therefore be partially attributed to conscious control of movements at greater mechanical task demands, potentially highlighting processes indicative of lesser movement automaticity as outlined in the Constrained Action Hypothesis (CAH; Lohse et al., 2013; Wulf et al., 2001). Previous assessments of movement efficiency, including levels of muscle activation and force production, have been linked to factors related to functional variability, attributed to contrasting levels of brain activation, which control muscle force during eccentric muscular contractions (Olsson et al., 2012). Inhibitory mechanisms at muscle and spinal level levels, as well as selective recruitment of larger motor units (Guilhem et al., 2010; Komi et al., 2000; Linnamo et al., 2003), provide further explanations for lesser load accommodation capacity, and decreased movement variability at greater mechanical task demands.

Decreased collective movement variability assessed from appended neural and mechanical outcome variables successfully highlighted dimension reduction that provided interpretations for fewer functional DOF when performing single-leg landings under greater mechanical task demands (Latash, 2010; Lohse et al., 2013; Turvey, 1990). Neuromechanical synergies were additionally assessed during supplementary analysis, identifying common sources of variation that underscore collective movement coordination (Appendix II, supplementary statistical analyses, collective movement variability). Fewer control units (PCs) under greater mechanical task demands were once again demonstrated (Figure 7, Middle), highlighting dimension reduction when examining co-variation among time series variables across the landing phase (Diedrichsen et al., 2009; Latash, 2010; Li, 2006; Turvey, 1990).

Further follow-up analysis examining neuromechanical synergies among time series variables at the level of individual participants was carried out as a means of identifying subject-specific collective movement variability changes following mechanical task demand manipulations (Appendix II, supplementary statistical analyses, collective movement variability). Single-subject PCA has been identified as a potential research avenue following previous recommendations (Trudeau, von Tscharner, Vienneau, Hoerzer, & Nigg, 2015), which identify single-subject analysis as a valuable research tool capable of identifying individual response strategies that may otherwise be overlooked during group analysis (Dufek et al., 1995; Scholes et al., 2012; Stergiou & Scott, 2005). In this instance, single-subject PCA identified a variety of variability responses under contrasting mechanical task demands, with aggregated single-subject PCA results promisingly demonstrating agreement with group assessments of neuromechanical synergies and collective movement variability among the appended time series variables (Figure 7, bottom). During each assessment, lesser collective variability trends were observed under greater mechanical task demands (Figure 7), meeting the research hypotheses and identifying movement control mechanisms that meet predictions from motor control theories, including the concepts, Optimal Control Theory (OCT), Optimal Feedback Control Theory, the Uncontrolled Manifold Hypothesis (UCM), and indirectly, the Constrained Action Hypothesis (CAH; Diedrichsen et al., 2009; Federolf et al., 2013; Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999; Scott, 2012; Todorov, 2006; Wulf et al., 2001).

Alongside collective movement variability assessments, within-subject variability was evaluated using coefficient of variation (CV). Decreased within-subject variability with greater load among integrated hip, knee, and ankles, as well as angular impulse at each joint, demonstrated agreement with the results from collective movement variability via PCA (Figure 8 and Appendix II, supplementary statistical analyses, within-subject variability). Contrasting trends were, however, observed for iEMG variability at greater landing height for the gluteus maximus, vastus medialis, and tibialis anterior muscles. Limitations must, however, be acknowledged in the use of CV in variables with baseline means close to zero, which is of particular concern for GRFz, joint angular impulse, and iEMG variables (Table 1; Brown et al., 2012). As a result, consideration for within-subject variability, assessed using CV, was limited as a comparator for the novel PCA approach taken in this investigation. Importantly, a limitation of collective movement variability PCA assessments included the inability to separate between and within-subject variability (Figure 7 and Appendix II, supplementary statistical analyses, collective movement variability [PCA]; Brandon et al., 2013; Deluzio et al., in Robertson et al., 2014; Bates, 1996; Bates et al., in Stergiou et al., 2004, p. 5-28; James & Bates, 1997). The use of single-subject-PCA provided a remedy to this limitation, while sacrificing the number of observations and subsequent sources of variation in each independent analysis. Nevertheless, the aggregated results of single-subject PCA demonstrated agreement with trends from group neuromechanical synergies and appended time series analyses (Figure 7). In combination, results from single-subject PCA and within-subject variability (CV) suggest the normalization procedures used during time-series z-score conversion (Appendix I, time series z-score conversion) effectively limited sources of between-subject variability, providing insight into within-subject variability changes among participants, relative to baseline.

Overall, lesser collective movement variability with greater mechanical task demands, assessed using PCA, appeared to demonstrate freezing among functional degrees of freedom, potentially limiting system flexibility and adaptation to the applied stressors (Latash, 2010; Lohse et al., 2013; Turvey, 1990). These results may have implications for injury mechanisms under greater task demands (James et al., 2000; James in Stergiou, 2004, p. 29-62). Variable and joint-specific alterations therefore provided additional explanations for the observed load accommodation strategies (Figure 6) and changes in collective movement variability (Figure 7).

Variable-specific adjustments (follow-up PCA)

Although comprehensive statistical results for inferential tests among PC scores were provided (Figures 9-20, and Appendix III, comprehensive statistical summary, variable-specific adjustments [follow-up PCA]), the following sections address the observed biomechanical changes within and among outcome variables. Comparisons were therefore made with previous literature, while examining the variable-specific adjustments in the context of the observed collective movement variability changes.

Within each variable, PCA was capable of identifying multiple independent sources of variation among participants and conditions, allowing subtle-movement pattern differences to be identified (Deluzio et al., in Robertson et al., 2014; Federolf et al., 2013; Federolf et al., 2012). Limitations surrounding interpretations of latter PCs, however, should be addressed. Depending on the research aims, it may be more economical to select a lower threshold for explained variance (i.e. suppress PCs with low explained variance), rather than examining variables identified with an upper threshold (i.e. retain PCs exceeding a cumulative explained variance value; Kipp et al., 2012). Instances where a large proportion of the variance was explained in a small number of PCs leave the remaining PCs to fulfill the cumulative upper threshold while providing lesser interpretable information. Likewise, variation may be spread among a larger number of PCs, which sum to the required cumulative upper threshold, while only a small subset of these PCs exceed a lower threshold. Suppressing PCs falling below a lower explained variance threshold may therefore be worthwhile in certain instances (Kipp et al., 2012). Nevertheless, the sensitivity of PCA in detecting movement pattern adjustments was highlighted from the extracted sources of variation, providing interpretations for changes in movement variability during single-leg landings.

Vertical ground reaction force (GRFz)

Contrasting GRFz patterns were observed with greater load and landing height in several distinct landing phases. Early GRFz increases were observed with greater landing height (Figure 9, PC1, 2, 3, 4), in agreement with previous research (Ali et al., 2014), while GRFz trends diverged in specific landing phases with greater load (Figure 9, PC1, 3, 4). Early GRFz increases (Figure 9, PC3) with greater load and greater landing height provided initial insight into pre and post-landing strategies, carried out via lower extremity adjustments before and after ground contact (Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996).

Opposing GRFz patterns (Figure 9) and landing impulse trends (Table 1), highlight the interactive influence of GRFz magnitudes and timing as well as the effects of pre and post-landing lower extremity adjustments (Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996). Landing duration decreased with greater load, and increased with greater landing height, which was masked by temporal normalization in Figures 9-20. Consistent GRFz variability among conditions (Figure 9; top right), provided similarity to the within-subject variability results (Figure 8).

Hip angle

Greater hip flexion with lesser load and greater landing height highlighted postural adjustments during landing accommodation (Figure 10; PC1; Ali et al., 2014;

Devita & Skelly, 1992). Lesser hip flexion with greater load may identify biomechanical freezing of the hip joint with greater load (Figure 10; PC1), with potential implications on landing stiffness, which was considered further when examining joint moments (ratio of joint moment to angular displacement; $k_{\text{joint}} = \Delta M_{\text{joint}}/\Delta\theta_{\text{joint}}$, where k_{joint} is joint stiffness, is ΔM_{joint} is change in joint moment, and $\Delta\theta_{\text{joint}}$ is change in joint angular displacement; Devita & Skelly, 1992; Lyle, Valero-Cuevas, Gregor, & Powers, 2014; Wang & Peng, 2014). Although joint stiffness was not explicitly computed, trends among joint angles and moments across the landing phase provided insight into joint stiffness modifications. Lesser hip flexion during early landing (Figure 10; PC2) with greater load and height provided insight into pre-landing strategies, identifying a mechanism for lesser ground contact velocities observed during supplementary analysis (Appendix II, supplementary statistical analyses, pre-landing strategy; Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996). Although relative sagittal hip angle does not provide insight into absolute trunk and thigh positions, it can be understood that lesser hip flexion at ground contact likely corresponds with a more erect landing posture (Devita & Skelly, 1992).

Similar trends were observed when examining integrated hip angles (Table 1) and time series hip angle landing patterns, decreasing with load and increasing with height (Figure 10; PC1). Contrasting trends were observed for hip angle variability at each height (Figure 10, top right), providing partial agreement with decreased within-subject variability at greater load and landing height (Figure 8). Fewer extracted PCs highlight dimension reduction associated with fewer functional degrees of freedom at greater mechanical task demands (Figure 8; Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990).

Knee angle

Similar to the hip joint, greater knee flexion with lesser load and greater landing height identified landing modifications carried out by the knee joint (Figure 11; PC1; Ali et al., 2014; Devita & Skelly, 1992; Lyle et al., 2014; Wang & Peng, 2014). The knee joint has been identified as largely responsible for landing stiffness, which was considered further when examining knee joint moments (Devita & Skelly, 1992; Lyle et al., 2014; Wang & Peng, 2014). Contrasting knee angle adjustments were observed during late landing (Figure 11; PC2), where greater knee flexion was observed with both load and landing height, indicating differential rates of knee angle change across the landing phase (Figure 11; top left).

Similar trends were observed when examining integrated knee angles (Table 1) and knee angle landing patterns, decreasing with load and increasing with height (Figure 11; PC1). Contrasting knee angle variability trends at each height (Figure 11, top right) showed partial agreement with decreased within-subject variability at greater load and landing height (Figure 8), interpreted in the context of fewer functional degrees of freedom with greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990). Similarity between the numbers of extracted PCs at the hip and knee joint may provide additional interpretations regarding the functional degrees of freedom at each lower extremity joint (Li, 2006), but should be considered alongside each respective joint moment.

Ankle angle

Greater ankle dorsiflexion across the landing phase with lesser load and greater landing height demonstrated agreement with hip and knee angles, identifying similar lower extremity kinematic adjustments among conditions (Figure 12; PC1). Greater ankle plantarflexion (lesser ankle angle) was identified with greater landing height during early landing (Figure 12; PC2 and PC3, respectively), in agreement with previous research (Ali et al., 2014), providing further insight into pre-landing strategies and explanations for lesser ground contact velocity observed with greater load and landing height (Appendix II, supplementary statistical analyses, pre-landing strategy). Combined interpretations from the hip and ankle joints, suggest participants made lower extremity adjustments, extending at the hip and ankle in attempt to reduce the effective landing height, subsequent flight time, and ground contact velocity (Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996).

Collectively, greater lower extremity joint flexion with lesser load and greater landing height provided insight into the GRFz differences among conditions. Greater GRFz with greater load and landing height follow Newtonian predictions, however, GRFz patterns are dictated by the acceleration of the system COM, controlled by the support limb. As a result, greater GRFz with greater load may also be attributed to lesser overall lower extremity joint flexion (associated with lower extremity stiffness), which also reduced the landing phase duration (Table 1; Devita & Skelly, 1992; Lyle et al., 2014; Wang & Peng, 2014).

Similar trends were observed between integrated ankle angles (Table 1) and ankle angle landing patterns, each decreasing with load and increasing with landing height

(Figure 12; PC1). Decreased ankle angle variability trends from each landing height (Figure 12, top right) demonstrated agreement with decreased within-subject variability at greater load and landing height (Figure 8), associated with fewer functional degrees of freedom at greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990). Similarity among the numbers of extracted PCs from each lower extremity joint kinematics may provide insight into the functional degrees of freedom at each lower extremity joint (Li, 2006), with additional information gained from kinetic and EMG variables.

Hip moment

Considerable variation was observed among participants and conditions, expressed through the number of extracted PCs, potentially providing interpretations for the functional degrees of freedom at the hip joint (Li, 2006). Similar hip moment patterns across the landing phase were observed in relation to previous research, with an initial hip flexor moment followed by a peak hip extensor moment (Sinsurin, Vachalathiti, Jalayondela, Limroongreungrat, 2013). Contrasting hip moment increases were observed during several distinct landing phases, identifying greater hip moments with lesser load and greater landing height (Figure 13a; PCs 1, 7), despite greater observed hip moments from lesser height during several distinct landing phases, as captured by latter PCs with lesser explained variance (Figures 13a and 13b; PCs 2, 3, 6).

Similar trends were observed when examining hip angular impulse (Table 1) and hip moment landing patterns, decreasing with greater load and increasing with greater height (Figure 13a; PC1). Greater hip moments at lesser load and greater landing height

(Figures 13a, b; PC1, 7), in combination with greater hip flexion at lesser load and greater landing height (Figure 10, PC1), suggest hip joint stiffness was relatively consistent among conditions (similar trends in joint moments and angles; Devita & Skelly, 1992; Lyle et al., 2014). Consistent hip joint stiffness among conditions suggests greater stiffness was required at distal joints, when considered alongside increased GRFz at greater load and landing height (Schmitz et al., 2007).

Despite the greater number of extracted PCs among hip joint moments (relative to each other outcome variable), decreased hip moment variability trends were observed at each landing height (Figure 13a, top right). Lesser variability at greater mechanical task demands demonstrated agreement with decreased within-subject variability at greater load and landing height (Figure 8), identifying dimension reduction associated with fewer functional degrees of freedom at greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990).

Knee moment

Contrasting knee moment increases were identified in specific portions of the landing phase, with similar landing phase patterns in relation to previous research (Brown, McClean, & Palmieri-Smith, 2014; Sinsurin et al., 2013). Early knee moment increases were observed at lesser load (Figure 14; PC1), while knee moment increases were observed during specific landing phases at greater load (Figure 14; PCs 2 and 3) and landing height (Figure 14; PCs 2, 3, and 4). When assessed in combination with lesser knee flexion at greater load and lesser landing height, knee stiffness remained relatively consistent among conditions (Figures 11 and 14; Devita & Skelly, 1992; Lyle et al., 2014; Wang & Peng, 2014).

Similar trends were observed when examining knee angular impulse (Table 1) and knee moment landing patterns, decreasing with greater load and increasing with greater height (Figure 14; PC1). Consistent knee moment variability trends were observed among conditions (Figure 14, top right), in contrast to decreasing within-subject variability trends at greater load and landing height (Figure 8). Despite this finding, fewer extracted knee joint moment PCs, relative to hip moments (Figure 13a), may identify distal kinetic dimension reduction with potential implications for factors relating to injury (James et al., 2000; James in Stergiou, 2004, p. 29-62; Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990).

Ankle moment

Similar to the hip and knee joints, contrasting ankle moment increases were observed in specific portions of the landing phase, with landing phase patterns in agreement with previous research (Sinsurin et al., 2013). Greater ankle moments were observed at greater load and landing height (Figure 15, PC1), while lesser ankle moments were observed early in the landing phase with greater load and lesser landing height (Figure 15, PC 2, 3). In combination with greater ankle plantarflexion at ground contact and lesser ankle dorsiflexion across the landing phase, when landing with greater load (Figure 12, PC 3) and landing height (Figure 12; PC 2, 3), greater ankle stiffness was inferred (Devita & Skelly, 1992; Lyle et al., 2014; Wang & Peng, 2014).

Contrasting trends were observed between ankle angular impulse and ankle moment landing patterns, highlighting interactions between magnitude and temporal characteristics (Devita & Skelly, 1992; Schmitz et al., 2007). Ankle angular impulse decreased with greater load and increased with greater height (Table 1), while ankle moments generally increased with both load and landing height (Figure 15, PC1). Decreased ankle moment variability was observed with greater load at each landing height (Figure 15, top right), associated with fewer functional degrees of freedom at greater mechanical task demands (Figure 8; Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990). Lesser ankle moment variability assessed from the number of extracted PCs in each condition demonstrated similarity to within-subject variability (Figure 8). As well, fewer extracted PCs were observed relative to hip and knee moments (Figure 13a, 14), further identifying distal kinetic dimension reduction that may have implications for factors relating to injury (James et al., 2000; James in Stergiou, 2004, p. 29-62; Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990).

Load accommodation strategies therefore appeared to involve altered lower extremity segmental configuration adjustments with both load and landing height increases during pre-landing. Participants demonstrated greater hip and ankle extension (ankle plantarflexion), decreasing the effective landing height, as well as the observed landing impulse (Table 1; Figure 6; Figure 10, PC2; Figure 12, PC 2, 3; Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996) Additionally, greater ankle joint stiffness with greater load and landing height, may provide explanations for GRFz increases at greater load and landing height (Figure 9, PC3; Devita & Skelly, 1992; Decker et al., 2003; Dufek & Bates, 1990; Dufek & Zhang, 1996; Lyle et al., 2014;

Schmitz et al., 2007; Wang & Peng, 2014). The neuromuscular control mechanisms underlying the observed kinematic and kinetic movement pattern adjustments were addressed among follow-up EMG variable analyses. Pre-landing strategies were inferred from muscle activation at ground contact, with post-landing strategies identified across the landing phase.

Gluteus maximus

Gluteus maximus muscle activity was considered in relation to the observed changes in the hip angle and moments due to its major role in controlling hip extension (Healy & Harrison, 2014; Zazulak et al., 2005). Greater gluteus maximus activation was observed earlier in the landing phase with greater landing height, in agreement with previous literature (Figure 16, PC1; Zazulak et al., 2005) and later in the landing phase with greater load (Figure 16, PC2). Greater activation at lesser height was observed later in the landing phase (Figure 16; PC3). Viewed alongside hip joint alterations, increased hip angles (Figure 10, PC1) and moments (Figure 13a, PC1) at greater landing height align with greater gluteus maximus activation (Figure 16; PC1). Decreased hip angles (Figure 10, PC1) and moments (Figure 13a, PC1) with greater load, indicating consistent hip stiffness (Devita & Skelly, 1992; Wang & Peng, 2014), are also explained by greater gluteus maximus activation with greater load (Figure 16, PC2; Iida et al., 2011; Zazulak et al., 2005).

Similar trends were observed between gluteus maximus iEMG (Table 1) and the time series activation patterns during landing, increasing with greater load and landing height (Figure 16; PC1, 2). Decreased gluteus maximus activation variability was observed with greater load and landing height (Figure 16, top right), in opposition to within-subject variability (Figure 8). Limitations regarding the use of coefficient of variation for variables with a mean near zero may therefore limit within-subject variability interpretations for iEMG (Brown et al., 2012), or may highlight contrasts between the results from each analysis. From PCA, fewer functional degrees of freedom were interpreted at greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990). Decreased gluteus maximus variability at greater load and landing height also provided insight into potential neuromuscular mechanisms for lesser observed variability among hip angle and moment variability (Figure 8, 10, 13a).

Biceps femoris

Biceps femoris muscle activity was interpreted in relation to the hip and knee joints (angles and moments) due to the biarticular nature of this muscle, serving as both a hip extensor and a knee flexor (Brown et al., 2014). Greater biceps femoris muscle activation was observed from greater landing height (Figure 17, PC1, 2, 4) and with lesser load (Figure 17, PC2). The level of pre-activation (at ground contact) was also identified, demonstrating contrasting trends among load and landing height conditions, decreasing with greater load when landing from greater height (Figure 17, PC3). Greater biceps femoris activity with greater landing height was interpreted in the context greater hip extension requirements (Brown et al., 2014), while lesser biceps femoris activation with greater load may be attributed to previously identified ankle joint adjustments with greater loads (Lyle et al., 2014; Yeow, Lee, & Goh, 2011; Wang & Peng, 2014).

Similar trends were observed between biceps femoris iEMG (Table 1) and activation patterns during landing, decreasing with greater load and increasing with greater landing height (Figure 17; PC1, 2). Lesser biceps femoris activation variability was observed with greater load and landing height (Figure 16, top right), in opposition to within-subject variability (Figure 8), further identifying coefficient of variation limitations (Brown et al., 2012). From PCA, fewer functional degrees of freedom were interpreted at greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990), aligning with kinematic and kinetic trends at the hip and knee joints (Figure 8, 10, 11, 13a), while kinetic knee joint variability remained consistent across load and landing height conditions (Figure 14).

Vastus medialis

Vastus medialis activity was viewed in the context of the knee joint, serving as a knee extensor (Yeadon et al., 2010). Greater vastus medialis activation was observed from greater landing height, in agreement with previous investigations (Figure 18, PC1, 3; de Britto, Carpes, Koutras, & Papas, 2014), and with greater load (Figure 18, PC1, 2). Greater activation was present, however, early in the landing phase at lesser loads (Figure 18, PC3, 4). Viewed alongside kinematic and kinetic adjustments of the knee joint at greater loads, greater knee moments during late landing (Figure 14, PC2) with lesser knee flexion (Figure 11, PC1) are explained by increased vastus medialis activity (Figure 18, PC1, 2; de Britto et al., 2014; Yeadon et al., 2010), maintaining knee position under greater loads. Similarly, greater knee moments from greater height (Figure 14, PC1, 2, 3, 4), although with greater knee flexion (Figure 11, PC1, 2), also required greater vastus medialis activity (Figure 18, PC1, 3; de Britto et al., 2014; Yeadon et al., 2010).

Divergent trends were observed for vastus medials iEMG (Table 1) and activation patterns across the landing phase (Figure 18, PC1, 2), highlighting physiological considerations for eccentric muscular contractions. Muscle activation under greater eccentric loads (greater mechanical task demands) resulted in predictable decreases in iEMG, attributed to neural control mechanisms, including inhibitory feedback from joint receptors, free nerve endings in muscle, pain receptors, and Golgi tendon organs (Ball & Scurr, 2013; Bishop et al., 2000; Westing et al., 1991). As well, preferential recruitment of larger, high-threshold motor units has been attributed to neural strategies that better distribute mechanical stresses (Ball & Scurr, 2013; Guilhem et al., 2010; Linnamo et al., 2003). As a result, lesser iEMG activity alongside greater local activation peaks is likely explained by motoneuron facilitation via muscle spindles, while the muscle actively lengthens and develops tension (Bishop et al., 2000).

Physiological underpinnings may therefore provide partial explanations for lesser observed vastus medials activation variability (Figure 18, top right), where decreased total muscle activity (iEMG) limits neuromuscular accommodation at greater mechanical task demands. Fewer functional degrees of freedom were therefore interpreted at greater mechanical task demands (Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990), aligning with kinematic trends at the knee joint (Figure 8, 10, 11, 13a), while kinetic variability remained consistent among conditions (Figure 14; top right). Greater within-subject variability (Figure 8) at greater landing height highlights coefficient of variation limitations for values with a mean near zero (Brown et al., 2012).

Medial gastrocnemius

Medial gastrocnemius muscle activation was viewed alongside ankle joint kinematics and kinetics, due to its role as an ankle plantarflexor (Mokhtarzadeh et al., 2013; Santello & McDonagh, 1998). Greater medial gastrocnemius activation was observed from greater landing height, in agreement with previous research (Figure 19, PC1, Santello & McDonagh, 1998), and with greater load (Figure 19, PC3). Particular attention can be directed to levels of medial gastrocnemius pre-activation (at ground contact; 0% landing phase; Figure 19; top left). Greater pre-activation with greater load (Figure 19, PC3) provides additional explanation for greater ankle stiffness during early landing (Figures 12 and 15), in addition to the greater observed plantarflexion at ground contact when landing from greater height (Figure 11, PC2, 3; Devita & Skelly, 1992; Lyle et al., 2014; Santello & McDonagh, 1998; Schmitz et al., 2007; Wang & Peng, 2014). Previous research has demonstrated distal to proximal transfer of mechanical energy, with greater reliance on the ankle joint during braking at greater landing stiffness (Iida et al., 2011; Lyle et al., 2014; Schmitz et al., 2007; Yeow et al., 2011b; Zhang, 2000).

Divergent trends for medial gastrocnemius iEMG (Table 1) and landing phase activation patterns (Figure 19, PC) further identify physiological considerations for iEMG and peak muscle activity. Greater peak EMG (Figure 19, PC1, 3) was observed at greater landing height and load, while iEMG (Table1) increased at greater landing height, but decreased with load. Integrated EMG decreases were therefore attributed to neural inhibition, while recruitment of larger motor units, and muscle spindle facilitation of motoneurons, provide explanations for greater peaks (Ball & Scurr, 2013; Bishop et al., 2000; Guilhem et al., 2010; Linnamo et al., 2003; Westing et al., 1991). As well, these physiological mechanisms may underlie the decreasing trends observed among kinematic and kinetic ankle variability (Figure 8, Figure 12, Figure 15). Decreasing medial gastrocnemius variability trends were observed among conditions (Figure 19, top right), in agreement with decreased within-subject variability at greater load, while greater within-subject variability was observed at greater landing height (Figure 8).

Tibialis anterior

Tibialis anterior muscle activation was interpreted alongside ankle joint kinematics and kinetics due to its role as an ankle dorsiflexor (Iida et al., 2011; Santello & McDonagh, 1998). Greater tibialis anterior activation was observed with greater load (Figure 20, PC2). Viewed alongside ankle joint angles and moments, as well as medial gastrocnemius activation patterns, increased tibialis anterior activity with greater load can likely be attributed to co-contraction with ankle plantarflexors (i.e. medial gastrocnemius) during landing (Iida et al., 2011; Santello & McDonagh, 1998). Although the level of muscle activation peaked during mid landing (~45% landing phase; Figure 20, PC1), the observed changes in ankle angle at ground contact and inferred changes in ankle stiffness (Figure 12, PC2, 3; Figures 12 and 15, PC1), suggest tibialis anterior activity provided contributions via co-contraction (Iida et al., 2011; Santello & McDonagh, 1998).

Divergent trends between tibialis anterior iEMG (Table 1) and peak activation patterns (Figure 20, PC2) draw attention to neural inhibition of total muscle activity, with simultaneous increases in peak activity as a result of larger motor unit recruitment and motor neuron facilitation via muscle spindles (Ball & Scurr, 2013; Bishop et al., 2000;

Guilhem et al., 2010; Linnamo et al., 2003; Westing et al., 1991). Greater EMG magnitudes were observed with greater load (Figure 20, PC2), while iEMG decreased (Table1). Further, decreasing trends were observed among kinematic and kinetic ankle variability (Figure 8, 12, 15). Contrasting tibialis anterior variability trends were observed at each height, while variability decreased at greater load from lesser height (Figure 20, top right). Greater within-subject variability was observed at greater landing height, identifying coefficient of variation limitations (Figure 8).

General discussion and conclusions

Throughout this investigation, an interdisciplinary approach was taken in providing interpretations surrounding the observed biomechanical changes following mechanical task demand manipulations, and the subsequent changes in movement variability. Interpretations were therefore drawn from domains including neuroscience, motor control, physiology, anatomy, and biomechanics, providing proposed mechanisms and implications for changes among movement patterns and the associated variability surrounding movement repetitions. Although links between movement variability and injury remain largely theoretical (James et al. 2000; James, in Stergiou, 2004, p. 29-62), it is hoped that systematic evaluations of motor responses under experimentally controlled laboratory settings may provide a body of evidence that can be applied to more ecologically relevant investigations, including but not limited to potential injury mechanisms in landing activities.

In agreement with the outlined research hypotheses, fewer emergent strategies were identified under greater mechanical task demands, defined using the load accommodation strategies model (James et al., 2003; James et al., 2014). Increasingly positive biomechanical responses were observed with greater load and landing height (Figure 6), alongside decreased collective movement variability assessed using PCA among neural and mechanical outcome variables. As well, joint-specific biomechanical adjustments were identified among conditions, through assessment of PC score differences, highlighting mechanisms for the observed load accommodation strategies and changes in movement variability. PC scores tended to increase with greater mechanical task demands, in accordance with increased biomechanical outcome variable magnitudes, but were often variable-specific, highlighting tradeoffs among variables by condition. In opposition to the outlined hypotheses, greater explained variance among fewer PCs was generally detected among variables associated with distal joints. This finding, however, seems to support the notion of additional functional degrees of freedom among proximal joints, controlled by larger biarticular muscles (James et al., 2000; Mills et al., 2009; Zhang et al., 2000). Finally, earlier (% landing phase) PC loading vector increases were generally observed from greater landing height, in agreement with the outlined hypotheses, though with lesser load. Later increases among time series variables at greater load can be attributed to the nature of the observed biomechanical alterations.

Specific biomechanical alterations following mechanical task demand manipulations included decreased lower extremity joint angles at greater load, increased lower extremity joint angles with greater landing height, and contrasting decreases among hip and knee joint moments despite increased ankle joint moments. The observed mechanical adjustments suggested participants used an increasingly upright landing posture, extending at the hip and plantarflexing at the ankle, decreasing effective landing height and reducing landing impulse. Postural adjustments prior to landing therefore provided insight into the mechanisms by which participants carried out load accommodation strategies (Devita & Skelly, 1992; Dufek & Bates, 1990; Dufek & Zhang, 1996). Particular attention can be paid to the kinetic contributions of each lower extremity joint, interpreted from peak joint moment and angular impulse, identifying similar trends in relation to previous research (Devita & Skelly, 1992; James et al., 2000; James et al., 2003). The increasingly upright landing posture under greater mechanical task demands highlights energy absorbing contributions from the ankle joint, exceeding proximal joints, likely as a result of decreased moment arms about the hip and knee joints, leading to lesser joint moments (Devita & Skelly, 1992; Zhang et al., 2000).

With respect to electrical muscle activity, contrasting gluteus maximus activation patterns across the landing phase provided insight into phase dependent adjustments at the hip joint at greater load and height, with implications for biceps femoris activation, decreasing with greater load and increasing with greater landing height. Importantly, participants appeared to adopt a stiffer ankle joint through medial gastrocnemius pre-activation at greater load and landing height, along with greater tibialis anterior co-contraction with greater load (Iida et al., 2011; Lyle et al., 2014; Santello & McDonagh, 1998). Greater peak vastus medialis activity at greater load and landing height, alongside decreased iEMG activity in medial gastrocnemius, tibialis anterior, and vastus medialis muscles also highlighted physiological considerations for the observed load accommodation strategies and movement variability changes following mechanical task demand manipulations. Greater peak activation and lesser iEMG under greater eccentric loads were attributed to large motor unit recruitment and inhibitory neural feedback mechanisms, respectively (Ball & Scurr, 2013; Bishop et al., 2000; Guilhem et al., 2010; Linnamo et al., 2003; Westing et al., 1991). The observed physiological responses among muscle controlling distal joints likely lead to lesser neuromuscular accommodation capacities and more frequently observed positive biomechanical load accommodation strategies. Additionally, the outlined neural considerations provided a physiological basis for the observed decreases in movement variability, amalgamating notions from motor control theories (Feldman, in Latash, 2010). The additive influence of implicit or explicit pre-movement planning at greater mechanical task demands, may have additively contributed to the observed variability decreases, as outlined in the Constrained Action Hypothesis (Lohse et al., 2013; Wulf et al., 2001).

The observed alterations in movement variability were interpreted in the context of the available functional degrees of freedom when landing from contrasting mechanical task demands. Emergent movement strategies were therefore attributed to the constraints imposed by individual morphology, biomechanics, and the surrounding environment (Caster & Bates, 1995; James et al., 2003; James et al., 2014), with lesser movement variability accomplished via synergistic associations among patterns of muscle activation, applied loads, and segmental configurations (Kipp et al., 2012; Kipp et al., 2014; Latash, 2010; Li, 2006; Lohse et al., 2013; Turvey, 1990). The experimental procedures therefore appeared to elicit the anticipated decreases in functional movement variability with greater mechanical task demands aligning with predictions from the OCT and UCM (Diedrichsen et al., 2009; Latash, 2010; Lohse et al., 2013; Scholz & Schoner, 1999). Additionally, the manner in which participants accomplished the observed load accommodation strategies was indicative of a movement solution that minimized potential energy and muscle activation, as outlined in the equilibrium-point hypothesis (Feldman, in Latash, 2010). Specifically, participants reduced effective landing height through segmental configuration adjustments in pre-landing, as well as landing duration, through reduced joint flexion and pre-activation of distal muscles, used in absorbing energy prior to proximal muscles (Iida et al., 2011; Lyle et al., 2014; Schmitz et al., 2007; Yeow et al., 2011; Zhang, 2000). Collectively, these results demonstrate agreement with the outlined research hypotheses, aligning with physiological predictions and theories drawn from motor control.

Overall, a more holistic approach was taken in this investigation, evaluating changes in collective movement variability among neural and mechanical sources by way of statistical dimension reduction of times series data. Bates (1996) importantly identified that "all measurement schemes, no matter how thorough and well-conceived, lose more information than they gather", the goal of the researcher should therefore be to perform human movement analyses in a manner that furthers the understanding of the underlying control mechanisms while minimizing indiscriminate information loss. A systematic, multi-step approach, from multiple perspectives, was therefore used in this investigation, aimed at gaining more comprehensive understanding of movement control mechanisms in single-leg landing by way of greater variable inclusion and time series analysis. From this perspective, the aim of evaluating changes in movement variability among kinematic, kinetic, and electromyographic sources was successful in gaining a more complete understanding of mechanisms contributing to changes in movement variability and factors that may underlie landing injuries. Although the variability and overuse injury hypothesis was not explicitly addressed in this investigation, connections between participant load accommodation strategies and movement variability were highlighted, demonstrating lesser load accommodation capacities and lesser movement variability under greater mechanical task demands. In this context, the application of repetitive loads to identical structures as a result of fewer available movement options may understandably lead to the deterioration and eventual failure of tissues. Future work establishing direct connections between movement variability and overuse injuries is therefore considered important, applying the experimentally controlled variability changes in more ecologically relevant situations.

APPENDICES

Appendix I

Data analysis

EMG processing

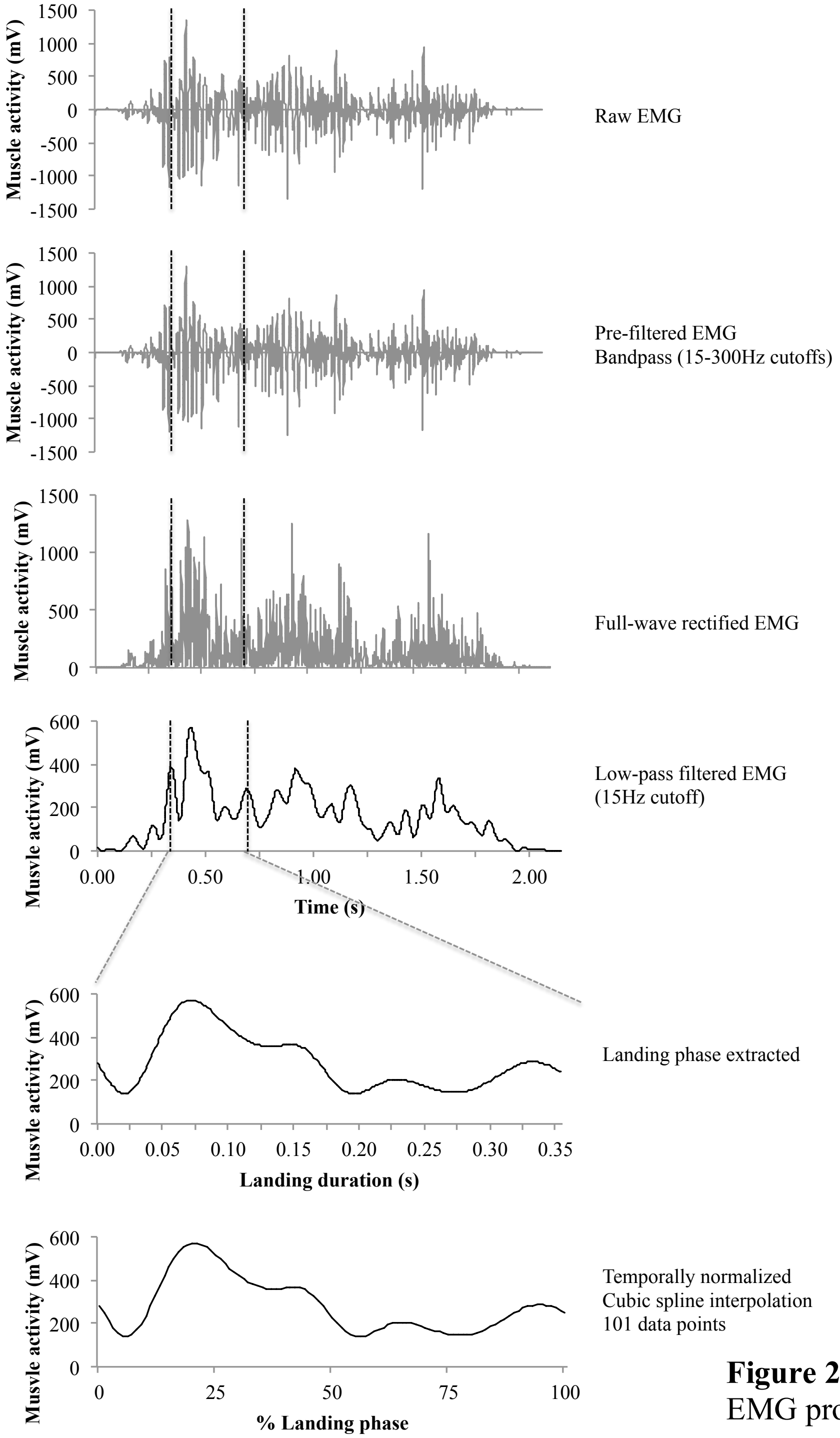


Figure 21: Exemplar EMG processing.

Time series z-score conversion

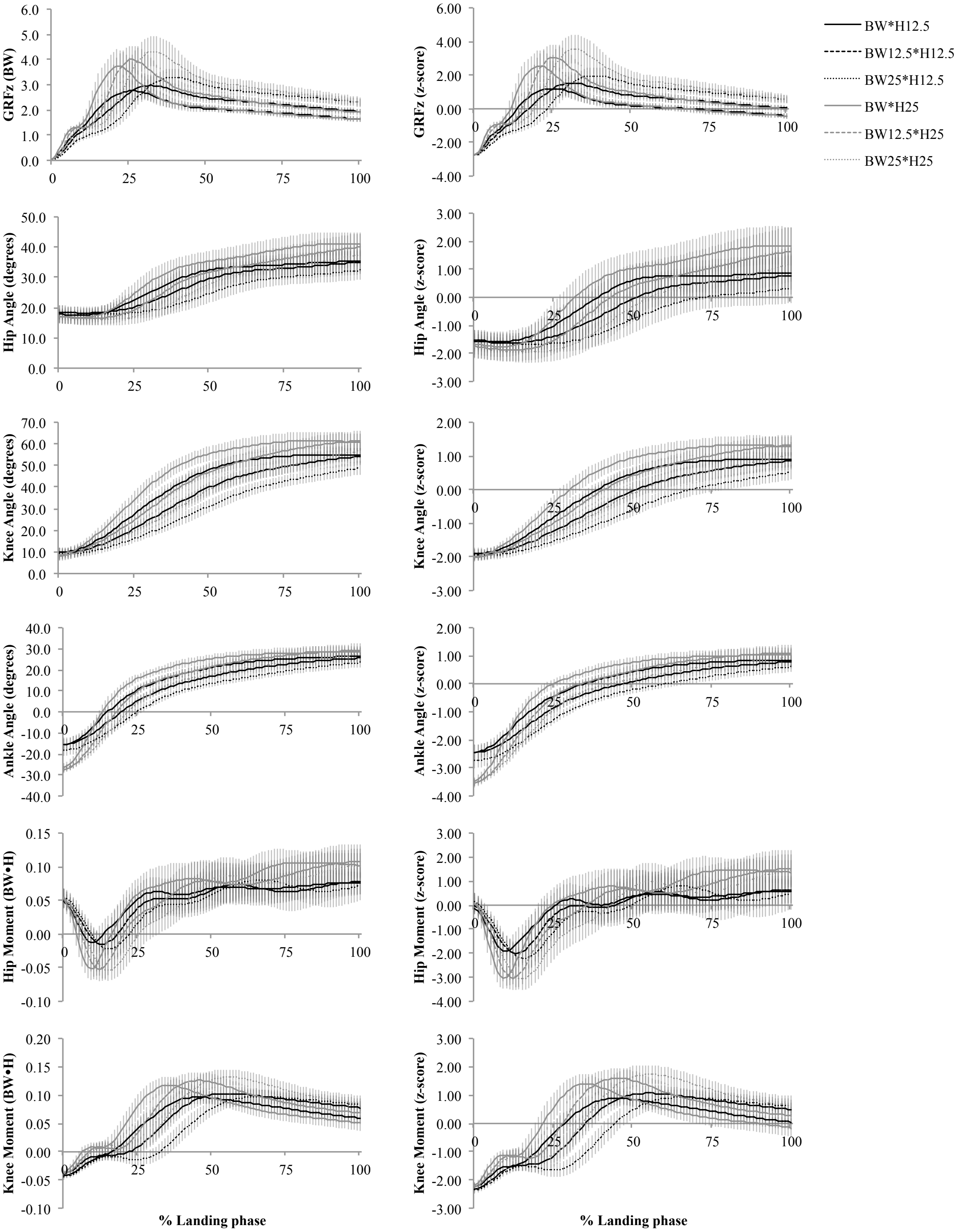


Figure 22a: z-score conversion by variable and condition a.
(condition mean ± standard deviation; GRFz, hip angle, knee angle, ankle angle, hip moment, knee moment)

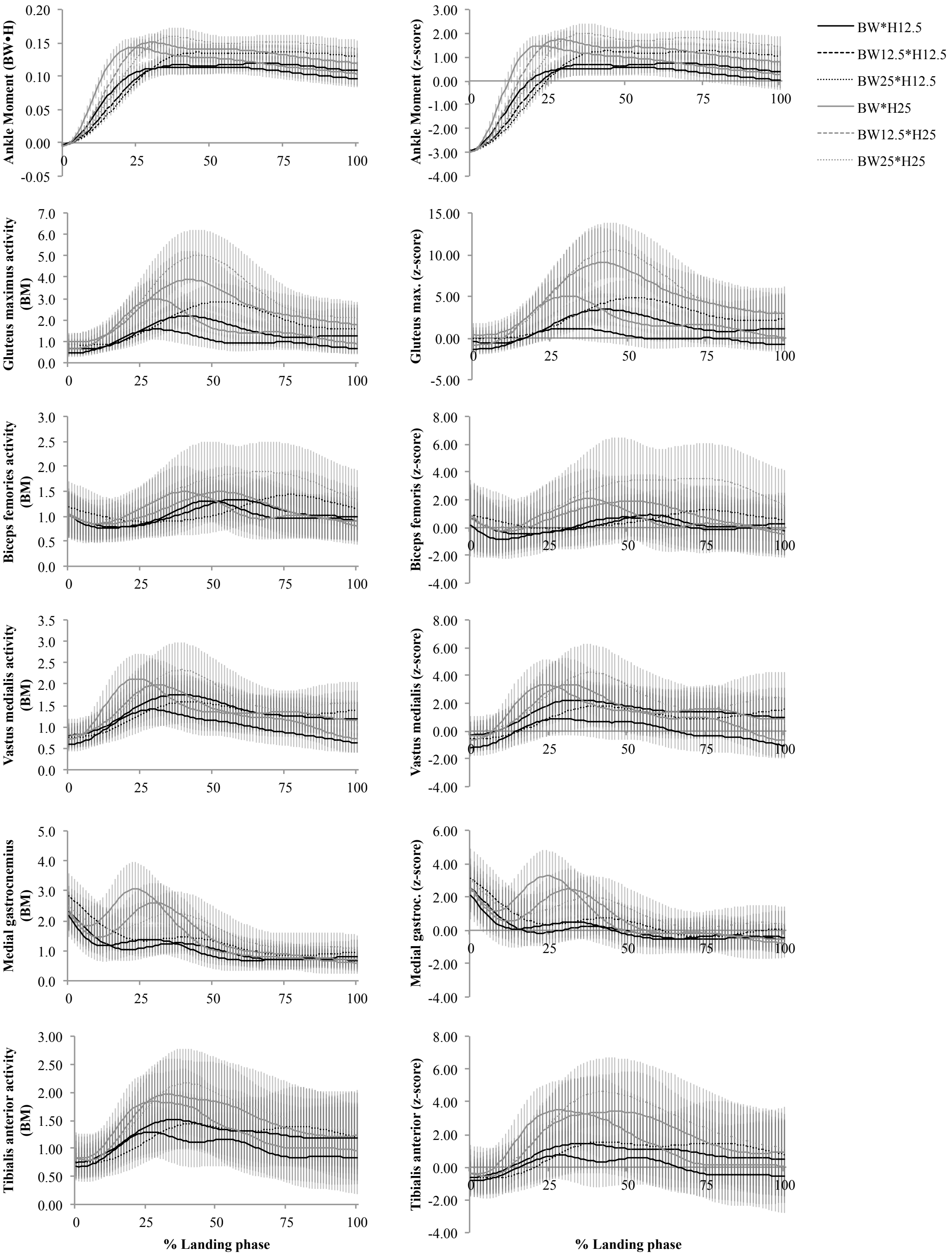


Figure 22b: z- score conversion by variable and condition b.
(mean ± standard deviation; ankle moment, gluteus maximus, biceps femoris, vastus medialis, medial gastrocnemius, tibialis anterior muscles)

Appendix II

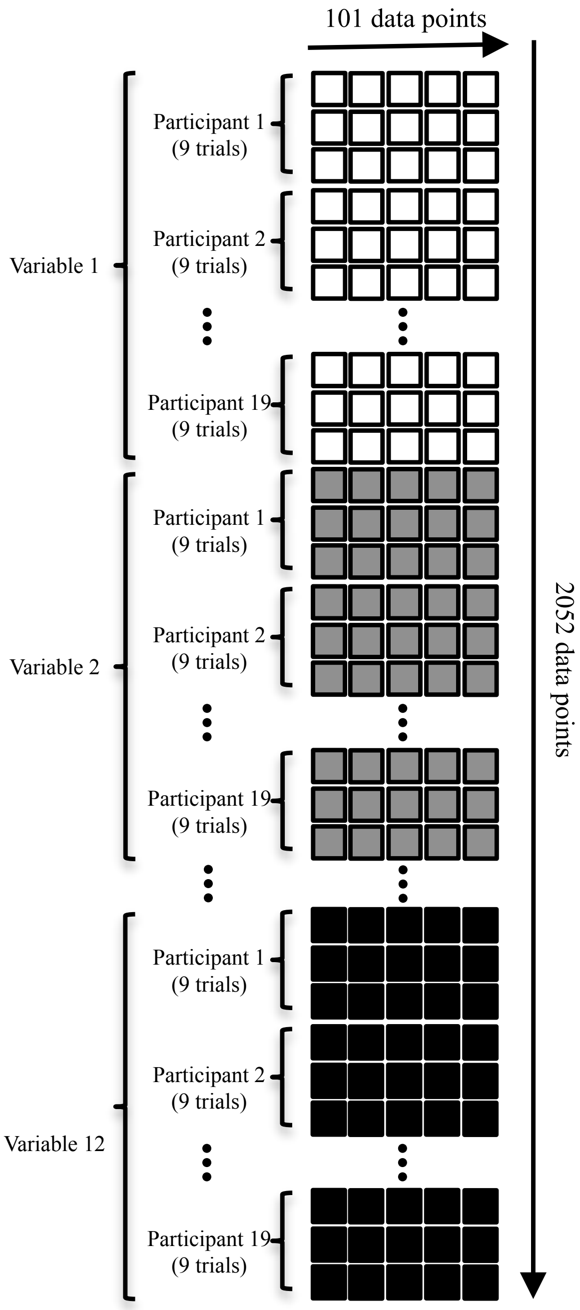
Supplementary statistical analyses

Load accommodation strategy identification

Following loading accommodation strategy identification, using 95% confidence intervals surrounding the mean impulse ratio (condition impulse/ baseline impulse) from the 9 trials completed by each participant in each condition, the frequency of observed strategies was summarized among participants (Figure 6). Associations among conditions and observed strategy frequencies were assessed using Fisher's exact tests ($\alpha = 0.05$). Separate tests were carried out at each load and landing height. Four total tests were performed: 1) effect of load at H12.5% (2x5 [load x strategy] contingency table), 2) effect of load at 25% (3x5 [load x strategy] contingency table), 3) effect of height at BW+12.5% (2x5 [height x strategy] contingency table), 4) effect of height at BW+25% (3x5 [height x strategy] contingency table). The effect of load at H12.5% was conducted on a 2x5 (load x strategy) contingency table as each impulse ratio was computed relative to baseline (BW•H12.5) prohibiting strategy identification at baseline. The results of Fisher's exact tests are summarized in Figure 6.

Collective movement variability (PCA)

Assessment of collective movement variability using the number of PCs extracted following PCA on the matrix of appended time series variables in each condition was followed up with analyses in line with assessments of synergies among time series variables (Kipp et al., 2014). This approach was taken from both group and single-subject approaches as a means of further identifying changes in collective movement variability across changes in mechanical task demands (load and landing height). During group analysis, matrices consisting of the time series from all 12 variables were assembled, placing the time series from each trial, variable, and participant in a separate row, creating 2052 x 101 dimension matrices for each condition (19 participants x 9 trials x 12 variables = 2052; normalized time series length = 101; Figure 23a). Six independent analyses were carried out by condition, extracting PCs that explained greater than 90% of the variance among trials, variables, and participants using the same procedure as was performed on the matrix of appended variables. During single-subject analyses, matrices were assembled by the same procedure, placing the time series from each trial and variable on a separate row, creating 108 x 101 dimension matrices for each participant in each condition (9 trials x 12 variables = 108; normalized times series length = 101; Figure 23b). Independent analyses were carried out by condition, with the same explained variance criterion (>90% EV). In both group and single-subject approaches the number of extracted PCs were interpreted relative to the collective movement variability among trials and variables (and participants in group analysis; Brandon et al., 2013; Daffertshofer et al., 2004; Kipp et al., 2014).



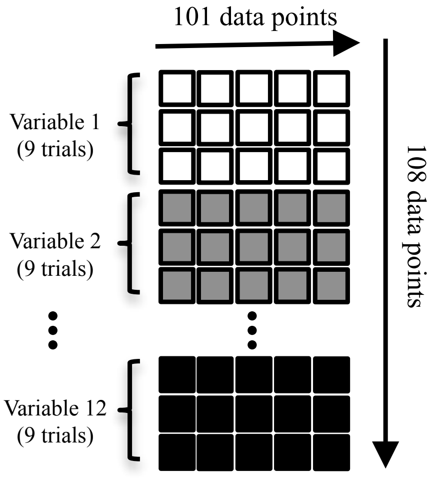


Figure 23a (left): Group synergy PCA. (independent analyses by condition)

Figure 23b (right): Single-subject synergy PCA. (independent analyses by condition)

Following single-subject PCA, additional statistical analysis was performed on the number of extracted PCs in each condition. In accordance with the study design, the interaction of load and landing height was explored in a 3x2 (load x height) repeated measures ANOVA (α=0.05). Due to the ordinal nature of the number of extracted PCs, however, main effects were examined using Friedman tests and follow-up Wilcoxon signed-rank tests (α=0.05). Then number of extracted PCs in each condition was not influenced by the interaction of load and landing height ($F(2,36) = 3.0$, $p = .063$, $\eta^2 = .14$). Collective single-subject movement variability decreased at greater loads ($\chi^2[2]=36.1$, $p<.001$; BW > BW+12.5%, $p<.001$, BW > BW+25%, $p<.001$) and landing height (H25% > H12.5%; $Z=-5.1$, $p<.001$), demonstrating fewer extracted PCs (Figure 7).

Within-subject variability

A traditional measure of within-subject variability was assessed using coefficient variation (CV; [standard deviation/mean]*100) of the integrated time series from the 9 trials of each participant-condition-variable. The standard deviation of each participant was specifically normalized to each participant's baseline mean (BW•H12.5%), providing similarity to the normalization procedure used during time series z-score conversion prior to PCA. Within-subject variability (CV) differences among conditions were evaluated using a 3x2 (load x height) repeated measures factorial MANOVA, with follow-up factorial (3x2: load x height) and one-way repeated measures ANOVAs, as well as pairwise comparisons for each variable following statistically significant differences among conditions ($\alpha = 0.05$). Degrees of freedom were adjusted via Huynh-Feldt corrections as necessary, with Bonferroni adjusted pairwise comparisons.

The comprehensive results from the 3x2 repeated measure factorial MANOVA are summarized in Appendix III (comprehensive statistical summary, within-subject variability), with significant pairwise comparisons highlighted in Figure 8, presenting within-subject CV means (± standard error) for each of the 12 integrated variables in each condition. Decreased within-subject variability was observed at greater loads (Figure 8; $F(12,26) = 8.1, p<.001, \eta^2 = .79$), while the main effect of landing height marginally missed reaching significance ($F[12,7] = 3.2, p=.064, \eta^2 = .85$). For this reason, the main effect of landing height was explored in follow-up 3x2 repeated measure factorial ANOVAs in each variable (summarized below and in Figure 8). Within-subject variability was not influenced by the interaction of load and landing height ($F[12,26] = 1.7, p=.125, \eta^2 = .44$).

Variable-specific adjustments (follow-up PCA)

Variable-specific movement variability was assessed separately in each condition using the number of PCs extracted during PCA. Matrices consisting of the time series from all 19 participants and their respective 9 trials were placed in separate rows, creating 171 x 101 dimension matrices for each condition (19 participants x 9 trials = 171; normalized time series length = 101; Figure 24). Six independent analyses were carried out by condition in each variable, extracting PCs that explained greater than 90% of the variance among trials and participants.

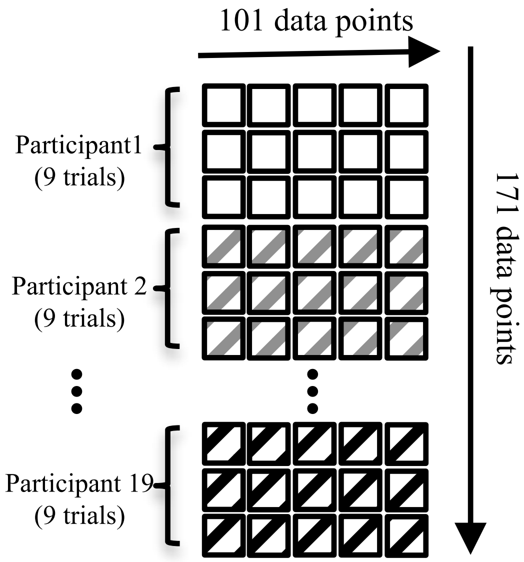


Figure 24: Variable-specific PCA. (independent analyses by condition)

Pre-landing strategy

Supplementary analysis was performed using maximum downward velocity of the pelvis segment origin at ground contact from kinematic analysis, as a surrogate for vertical center of mass velocity (vCOMz). The use of maximum downward pelvis origin velocity at contact allowed comparison of observed versus expected velocity at ground contact, as a means of accounting for the observed load accommodation landing strategies. Expected vCOMz at contact was computed using Equation 3 ($v = \sqrt{2gh}$), with landing height (h) calculated as a percentage of participant height, corresponding to the appropriate landing height condition (H12.5% and H25%). Mean percent difference was calculated among trials and participants in each (Table 2; 100*[observed-expected]/expected).

Table 2: Mean (± standard deviation) percent difference between observed and expected ground contact velocity

Landing Height	Load					
	BW		BW+12.5%		BW+25%	
	Mean (%)	SD (%)	Mean (%)	SD (%)	Mean (%)	SD (%)
H12.5%	-1.6	7.2	-4.7	7.1	-8.4	5.9
H25%	-9.5	5.4	-12.5	5.6	-13.8	6.2

Note: SD is standard deviation, BW is bodyweight, % is percent difference

Further analysis was performed for landing impulse, comparing observed (GRFz impulse from force platform analysis) versus expected landing impulse (predicted from kinematic analysis), as an additional means of accounting for the observed load accommodation strategies. During this analysis, expected landing impulse was computed using Equation 3 ($\sum_i F_i \, \Delta t = m\Delta v$), where landing velocity (Δv) from the kinematic analysis (maximum downward velocity of the pelvis origin; surrogate for vCOMz) was used along with participant mass (*m*) associated with each corresponding load condition (BW, BW+12.5%, BW+25%; *m*, *m*+12.5%, *m*+25%, respectively). Mean percent difference was calculated among trials and participants in each condition (Table 2; 100*[observed-expected]/expected). The relationship between observed and expected landing impulse was also investigated using the bivariate Pearson correlation from all trials, conditions, and participants ($r = 0.995$).

Table 3: Mean (± standard deviation) percent difference between observed and expected landing impulse

Landing Height	Load BW		BW+12.5%		BW+25%	
	Mean (%)	SD (%)	Mean (%)	SD (%)	Mean (%)	SD (%)
H12.5%	-3.1	3.0	-2.7	3.1	-2.4	3.4
H25%	-1.2	1.4	-0.3	1.4	0.2	1.9

Note: SD is standard deviation, BW is bodyweight, % is percent difference

Appendix III

Comprehensive statistical summary

Within-subject variability

Statistically significant main effects and pairwise comparisons are highlighted in bold.

3x2 repeated measures factorial MANOVA
Load: **F(12,26) = 8.1, p<.001, η^2 = .79**
Height: F(12,7) = 3.2, **p=.064**, η^2 = .85
Load x Height: F(12,26) = 1.7, p=.125, η^2 = .44

3x2 repeated measures factorial ANOVAs
GRFz impulse variability
Load: F(2,36) = 1.1, p=.321, η^2 = .06
Height: F(1,18) = 2.0, p=.178, η^2 = .10
Load x Height: F(2,36) = 3.1, p=.057, η^2 = .15

Integrated hip angle variability
Load: **F(2,36) = 27.5, p<.001, η^2 = .60**
BW > BW+12.5% p = .007
BW > BW+25% p <.001
BW+12.5% > BW+25% p = .001
Height: **F(1,18) = 6.5, p=.020, η^2 = .27**
H25% > H12.5% p = .020
Load x Height: F(2,36) = 0.5, p=.627, η^2 = .03

Integrated knee angle variability
Load: **F(1.4,25.0) = 24.2, p<.001, η^2 = .57**
BW > BW+12.5% p = .009
BW > BW+25% p < .001
BW+12.5% > BW+25% p < .001
Height: **F(1,18) = 13.5, p=.002, η^2 = .43**
H25% > H12.5% p = .002
Load x Height: F(1.6,28.8) = 0.7, p=.478, η^2 = .04

Integrated ankle angle variability
Load: **F(2,36) = 15.9, p<.001, η^2 = .47**
BW > BW+12.5% p = .047
BW > BW+25% p < .001
BW+12.5% > BW+25% p = .003
Height: **F(1,18) = 7.2, p=.015, η^2 = .29**
H25% > H12.5% p = .015
Load x Height: F(1.5,27.9) = 0.8, p=.451, η^2 = .04

Hip angular impulse variability
Load: **$F(1.3,22.8) = 6.1, p=.016, \eta^2 = .25$**
BW > BW+25% $p = .008$
Height: $F(1,18) = 2.6, p=.125, \eta^2 = .13$
Load x Height: $F(2,36) = 0.5, p=.557, \eta^2 = .03$

Knee angular impulse variability
Load: **$F(2,36) = 11.7, p<.001, \eta^2 = .39$**
BW > BW+25% $p = .001$
BW+12.5% > BW+25% $p = .003$
Height: **$F(1,18) = 4.7, p=.044, \eta^2 = .21$**
H25% > H12.5% $p = .044$
Load x Height: $F(2,36) = 0.5, p=.640, \eta^2 = .02$

Ankle angular impulse variability
Load: **$F(1.6,29.6) = 16.7, p<.001, \eta^2 = .48$**
BW > BW+12.5% $p = .009$
BW > BW+25% $p = .009$
BW+12.5% > BW+25% $p = .005$
Height: **$F(1,18) = 6.2, p=.023, \eta^2 = .26$**
H25% > H12.5% $p = .023$
Load x Height: $F(2,36) = 0.8, p=.464, \eta^2 = .04$

Gluteus maximus iEMG variability
Load: $F(1.5,32.2) = 1.1, p=.353, \eta^2 = .06$
Height: **$F(1,18) = 11.8, p=.003, \eta^2 = .40$**
H25% > H12.5% $p = .003$
Load x Height: $F(2,36) = 1.4, p=.272, \eta^2 = .07$

Biceps femoris iEMG variability
Load: $F(2,36) = 0.4, p=.684, \eta^2 = .02$
Height: $F(1,18) = 3.3, p=.085, \eta^2 = .16$
Load x Height: $F(1.5,27.0) = 0.6, p=.436, \eta^2 = .03$

Vastus medialis iEMG variability
Load: $F(2,36) = 1.3, p=.298, \eta^2 = .07$
Height: **$F(1,18) = 8.6, p=.009, \eta^2 = .32$**
H25% > H12.5% $p = .009$
Load x Height: $F(2,36) = 1.4, p=.249, \eta^2 = .07$

Medial gastrocnemius iEMG variability
Load: **$F(2,36) = 3.3, p=.048, \eta^2 = .16$**
BW > BW+12.5% $p = .062$
Height: $F(1,18) = 2.9, p=.105, \eta^2 = .14$
Load x Height: $F(1.6,28.1) = 0.1, p=.897, \eta^2 = .004$

Tibialis anterior iEMG variability

Load: $F(1.6,29.2) = 2.2$, $p=.138$, $\eta^2 = .11$

Height: $\boldsymbol{F(1,18) = 8.1}$, $\boldsymbol{p=.011}$, $\boldsymbol{\eta^2 = .31}$

H25% > H12.5% $p = .011$

Load x Height: $F(1.5,27.8) = 3.4$, $p=.061$, $\eta^2 = .16$

Variable-specific adjustments (follow-up PCA)

Results of the separate 3x2 (load x height) repeated measures factorial ANOVAs are summarized below by variable and principal component (PC#). Statistically significant interaction, main effects, and pairwise comparisons are highlighted in bold. Statistically significant pairwise comparisons are also identified in Figures 9 through 20.

GRFz PC1

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.6, 29.5) = 172.5, p < .001, \eta^2 = .91$
  - Height $F(1, 18) = 1.9, p = .190, \eta^2 = .09$
  - **Load x Height:** **$F(2, 36) = 8.3, p = .001, \eta^2 = .32$**
- Main effect
  - Load: 100 > 112.5 $p < .001$
    - 100 > 125 $p < .001$
    - 112.5 > 125 $p < .001$
- Simple main effects
  - **Load @ H12.5%:** **$F(2, 36) = 62.0, p < .001, \eta^2 = .78$**
    - **Load: 100 > 112.5** **$p < .001$**
      - **100 > 125** **$p < .001$**
      - **112.5 > 125** **$p < .001$**
  - **Load @ H25%:** **$F(2, 36) = 157.5, p < .001, \eta^2 = .90$**
    - **Load: 100 > 112.5** **$p < .001$**
      - **100 > 125** **$p < .001$**
      - **112.5 > 125** **$p < .001$**
  - **Height @ L100%:** **$F(1, 18) = 13.7, p = .002, \eta^2 = .43$**
    - **Height: 12.5 < 25** **$p = .002$**
  - Height @ L112.5%: $F(1, 18) = 1.2, p = .295, \eta^2 = .06$
  - **Height @ 125%:** **$F(1, 18) = 6.1, p = .024, \eta^2 = .25$**
    - **Height: 25 < 12.5** **$p = .024$**

GRFz PC2

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.2, 20.9) = , p = .104, \eta^2 = .12$
  - Height $F(1,18) = , p < .001, \eta^2 = .87$
  - **Load x Height:** **$F(2, 36) = , p < .003, \eta^2 = .27$**
- Main effect
  - Height: $12.5 < 25$ $p < .001$
- Simple main effects
  - Load @ H12.5%: $F(1.3, 23.9) = 1.3, p = .280, \eta^2 = .07$
  - Load @ H25%: $F(1.2, 21.9) = 4.0, p = .050, \eta^2 = .18$
    - Load: $100 < 112.5$ $p = .019$
  - **Height @ L100%:** **$F(1, 18) = 8.5, p = .009, \eta^2 = .32$**
    - **Height:** **$12.5 < 25$ $p = .009$**
  - **Height @ L112.5%:** **$F(1, 18) = , p < .001, \eta^2 = .28$**
    - **Height:** **$12.5 < 25$ $p < .001$**
  - **Height @ 125%:** **$F(1, 18) = , p < .001, \eta^2 = .81$**
    - **Height:** **$12.5 < 25$ $p < .001$**

GRFz PC3

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.5, 26.2) = 9.8, p = .002, \eta^2 = .35$**
  - **Height** **$F(1, 18) = 15.3, p = .001, \eta^2 = .46$**
  - Load x Height: $F(2, 36) = 1.5, p = .247, \eta^2 = .08$
- Main effect
  - **Load:** **$125 > 100$ $p = .010$**
  - **$125 > 112.5$ $p = .012$**
  - **Height:** **$25 > 12.5$ $p = .001$**

GRFz PC4

- 3x2 repeated measures factorial ANOVA
  - Load: $F(2, 36) = 1.9, p = .162, \eta^2 = .10$
  - Height $F(1, 18) = 11.4, p = .003, \eta^2 = ..39$
  - **Load x Height:** **$F(1.5, 27.3) = 3.9, p = .043, \eta^2 = .18$**
- Main effect
  - Height: $25 > 12.5$ $p = .003$
- Simple main effects
  - **Load @ H12.5%:** **$F(1.5, 27.7) = 6.3, p = .009, \eta^2 = .26$**
    - **Load:** **$100 > 125$ $p = .037$**
  - Load @ H25%: $F(2, 36) = 0.9, p = .417, \eta^2 = .05$
  - Height @ L100%: $F(1, 18) = 0.6, p = .458, \eta^2 = .03$
  - Height @ L112.5%: $F(1, 18) = 0.3, p = .612, \eta^2 = .02$
  - **Height @ 125%:** **$F(1, 18) = 17.2, p = .001, \eta^2 = .49$**
    - **Height:** **$25 > 12.5$ $p = .001$**

Hip angle PC1
3x2 repeated measures factorial ANOVA
Load: **$F(1.5, 26.3) = 18.3, p < .001, \eta^2 = .50$**
Height **$F(1, 18) = 12.0, p = .003, \eta^2 = .40$**
Load x Height: $F(1.6, 28.6) = 0.6, p = .50, \eta^2 = .04$
Main effect
Load: **100 > 112.5** **$p = .003$**
112.5 > 125 **$p = .007$**
100 > 125 **$p = .001$**
Height: **25 > 12.5** **$p = .003$**

Hip angle PC2
3x2 repeated measures factorial ANOVA
Load: **$F(2, 36) = 6.3, p = .005, \eta^2 = .26$**
Height **$F(1, 18) = 42.1, p < .001, \eta^2 = .70$**
Load x Height: $F(2, 36) = 1.4, p = .272, \eta^2 = .07$
Main effect
Load: **100 > 112.5** **$p = .027$**
100 > 125 **$p = .028$**
Height: **12.5 > 25** **$p < .001$**

Knee angle PC1

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(2, 26) = 208.7, p < .001, \eta^2 = .92$**
  - **Height** **$F(1, 18) = 78.5, p < .001, \eta^2 = .81$**
  - Load x Height: $F(2, 26) = 0.2, p = .82, \eta^2 = .01$
- Main effect
  - **Load:** **100 > 112.5 $p < .001$**
    - **112.5 > 125 $p < .001$**
    - **100 > 125 $p < .001$**
  - **Height:** **25 > 12.5 $p < .001$**

Knee angle PC2

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.3, 23.4) = 59.7, p < .001, \eta^2 = .77$
  - Height $F(1, 18) = 9.3, p = .007, \eta^2 = .34$
  - **Load x Height:** **$F(2, 36) = 13.7, p < .001, \eta^2 = .43$**
- Main effect
  - Load: 100 < 112.5 $p < .001$
    - 100 < 125 $p < .001$
  - Height: 12.5 < 25 $p = .007$
- Simple main effects
  - **Load @ H12.5%:** **$F(1.5, 27.0) = 20.7, p < .001, \eta^2 = .54$**
    - **Load:** **100 < 112.5 $p < .001$**
    - **100 < 125 $p = .002$**
  - **Load @ H25%:** **$F(1.6, 28.7) = 84.9, p < .001, \eta^2 = .83$**
    - **Load:** **100 < 112.5 $p < .001$**
    - **100 < 125 $p < .001$**
  - Height @ L100%: $F(1, 18) = 0.7, p = .414, \eta^2 = .04$
  - **Height @ L112.5%:** **$F(1, 18) = 11.0, p = .004, \eta^2 = .38$**
    - **Height:** **25 > 12.5 $p = .004$**
  - **Height @ 125%:** **$F(1, 18) = 25.0, p < .001, \eta^2 = .58$**
    - **Height:** **25 > 12.5 $p <.001$**

Ankle angle PC1

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.6, 28.9) = 79.2$, $p < .001$, $\eta^2 = .82$**
  - **Height** **$F(1, 18) = 38.8$, $p < .001$, $\eta^2 = .68$**
  - Load x Height: $F(1.5, 27.3) = 0.04$, $p = .919$, $\eta^2 = .002$
- Main effect
  - **Load:** **100 > 112.5** **$p < .001$**
    - **112.5 > 125** **$p < .001$**
    - **100 > 125** **$p < .001$**
  - **Height:** **25 > 12.5** **$p < .001$**

Ankle angle PC2

- 3x2 repeated measures factorial ANOVA
  - Load: $F(2, 36) = 0.8$, $p = .474$, $\eta^2 = .04$
  - Height $F(1, 18) = 22.4$, $p < .001$, $\eta^2 = .55$
  - **Load x Height:** **$F(1.5, 26.1) = 5.1$, $p = .021$, $\eta^2 = .22$**
- Main effect
  - Height: 12.5 > 25 $p < .001$
- Simple main effects
  - Load @ H12.5%: $F(1.5, 26.3) = 0.5$, $p = .565$, $\eta^2 = .03$
  - **Load @ H25%:** **$F(2.0, 35.2) = 8.0$, $p = .001$, $\eta^2 = .31$**
    - **Load:** **100 > 112.5** **$p = .005$**
      - **125 > 112.5** **$p = .013$**
  - **Height @ L100%:** **$F(1, 18) = 13.7$, $p = .002$, $\eta^2 = .43$**
    - **Height:** **12.5 > 25** **$p = .002$**
  - **Height @ L112.5%:** **$F(1, 18) = 18.6$, $p < .001$, $\eta^2 = .51$**
    - **Height:** **12.5 > 25** **$p < .001$**
  - **Height @ 125%:** **$F(1, 18) = 42.8$ $p < .001$, $\eta^2 = .70$**
    - **Height:** **12.5 > 25** **$p < .001$**

Ankle angle PC3

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.6, 29.2) = 12.0$, $p < .001$, $\eta^2 = .40$
  - Height $F(1, 18) = 23.7$, $p < .001$, $\eta^2 = .57$
  - **Load x Height: $F(1.4, 25.1) = 8.9$, $p = .004$, $\eta^2 = .33$**
- Main effect
  - Load: 100 < 112.5 $p < .001$
    100 < 125 $p = .009$
  - Height: 12.5 > 25 $p < .001$
- Simple main effects
  - Load @ H12.5%: $F(1.2, 22.2) = 1.3$, $p =.277$, $\eta^2 = .07$
  - **Load @ H25%: $F(1.7, 29.9) = 28.7$, $p < .001$, $\eta^2 = .61$**
    - **Load: 100 < 112.5 $p < .001$**
      100 < 125 $p < . 001$
  - **Height @ L100%: $F(1, 18) = 21.0$, $p < .001$, $\eta^2 = .54$**
    - **Height: 25 < 12.5 $p < .001$**
  - **Height @ L112.5%: $F(1, 18) = 17.5$, $p = .001$, $\eta^2 = .49$**
    - **Height: 25 < 12.5 $p =.001$**
  - Height @ 125%: $F(1, 18) = 1.0$, $p = .337$, $\eta^2 = .05$

Hip moment PC1

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(2, 36) = 22.1, p < .001, \eta^2 = .55$**
  - **Height** **$F(1, 18) = 16.5, p = .001, \eta^2 = .48$**
  - Load x Height: $F(2, 36) = 2.6, p = .092, \eta^2 = .126$
- Main effect
  - **Load:** **100 > 112.5** **$p = .003$**
  - **112.5 > 125** **$p = .017$**
  - **100 > 125** **$p < .001$**
  - **Height:** **12.5 < 25** **$p = .001$**

Hip moment PC2

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.6, 29.5) = 3.1, p = .072, \eta^2 = .15$
  - **Height** **$F(1, 18) = 15.6, p = .001, \eta^2 = .46$**
  - Load x Height: $F(2, 36) = 2.8, p = .079, \eta^2 = .14$
- Main effect
  - **Height:** **12.5 > 25** **$p = .001$**

Hip moment PC3

- 3x2 repeated measures factorial ANOVA
  - Load: $F(2, 36) = 0.2, p = .783, \eta^2 = .01$
  - **Height** **$F(1, 18) = 6.2, p = .023, \eta^2 = .26$**
  - Load x Height: $F(1.3, 22.6) = 0.3, p = .615, \eta^2 = .02$
- Main effect
  - **Height:** **12.5 > 25** **$p = .023$**

Hip moment PC4

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(2, 36) = 29.7, p < .001, \eta^2 = .62$**
  - **Height** **$F(1, 18) = 16.4, p = .001, \eta^2 = .48$**
  - Load x Height: $F(2, 36) = 0.2, p = .791, \eta^2 = .01$
- Main effect
  - **Load:** **100 > 112.5** **$p = .016$**
  - **100 > 125** **$p < .001$**
  - **112.5 > 125** **$p < .001$**
  - **Height:** **25 > 12.5** **$p = .001$**

Hip moment PC5

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.6, 28.7) = 1.0, p = .353, \eta^2 = .05$
  - Height $F(1, 18) = 1.2, p = .294, \eta^2 = .06$
  - Load x Height: $F(1.8, 32.9) = 0.1, p = .875, \eta^2 = .01$

Hip moment PC6

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.4, 25.2) = 4.8$, $p = .014$, $\eta^2 = .21$**
  - **Height** **$F(1, 18) = 20.2$, $p < .001$, $\eta^2 = .53$**
  - Load x Height: $F(2, 36) = 0.1$, $p = .913$, $\eta^2 = .01$
- Main effect
  - **Load:** **112.5 > 125** **$p = .004$**
  - **Height:** **12.5 > 25** **$p < .001$**

Hip moment PC7

- 3x2 repeated measures factorial ANOVA
  - Load: $F(2, 36) = 0.4$, $p = .652$, $\eta^2 = .02$
  - **Height** **$F(1, 18) = 5.9$, $p = .026$, $\eta^2 = .25$**
  - Load x Height: $F(2, 36) = 0.8$, $p = .468$, $\eta^2 = .04$
- Main effect
  - **Height:** **25 > 12.5** **$p = .026$**

Knee moment PC1

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.4, 25.9) = 121.6, p < .001, \eta^2 = .87$**
  - **Height** **$F(1, 18) = 118.7, p < .001, \eta^2 = .87$**
  - Load x Height: $F(2, 36) = 0.7, p = .734, \eta^2 = .02$
- Main effect
  - **Load:** **100 > 112.5 $p < .001$**
  - **112.5 > 125 $p < .001$**
  - **100 > 125 $p < .001$**
  - **Height:** **25 > 12.5 $p < .001$**

Knee moment PC2

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.4, 25.0) = 8.5, p = .004, \eta^2 = .32$
  - Height $F(1, 18) = 22.7, p < .001, \eta^2 = .56$
  - **Load x Height:** **$F(2, 36) = 53.2, p < .001, \eta^2 = .75$**
- Main effect
  - Load: 100 < 112.5 $p = .001$
  - Height: 25 > 12.5 $p < .001$
- Simple main effects
  - **Load @ H12.5%:** **$F(1.6, 28.1) = 4.6, p = .025, \eta^2 = .21$**
    - **Load:** **112.5 > 125 $p = .002$**
  - **Load @ H25%:** **$F(1.5, 27.3) = 25.1, p < .001, \eta^2 = .58$**
    - **Load:** **100 < 112.5 $p < .001$**
    - **100 < 125 $p < .001$**
  - Height @ L100%: $F(1, 18) = 0.7, p = .427, \eta^2 = .04$
  - **Height @ L112.5%:** **$F(1, 18) = 8.6, p = .009, \eta^2 = .32$**
    - **Height:** **25 > 12.5 $p = .009$**
  - **Height @ 125%:** **$F(1, 18) = 101.8, p < .001, \eta^2 = .85$**
    - **Height:** **25 > 12.5 $p < .001$**

Knee moment PC3

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.5, 26.8) = 1.8, p = .186, \eta^2 = .09$
  - Height $F(1, 18) = 5.0, p = .037, \eta^2 = .22$
  - **Load x Height:** **$F(2, 36) = 3.5, p = .040, \eta^2 = .16$**
- Main effect
  - Height: 25 > 12.5 $p = .037$
- Simple main effects
  - Load @ H12.5%: $F(1.5, 27.2) = 0.7, p = .509, \eta^2 = .04$
  - Load @ H25%: $F(1.6, 27.1) = 3.3, p = .063, \eta^2 = .15$
  - **Height @ L100%:** **$F(1, 18) = 5.1, p = .036, \eta^2 = .22$**
    - **Height:** **25 > 12.5 $p = .036$**
  - Height @ L112.5%: $F(1, 18) = 0.4, p = .547, \eta^2 = .02$
  - **Height @ 125%:** **$F(1, 18) = 8.0, p = .011, \eta^2 = .31$**
    - **Height:** **25 > 12.5 $p = .011$**

Knee moment PC4

3x2 repeated measures factorial ANOVA

Load:	$F(2, 36) = 0.7$, $p = .528$, $\eta^2 = .04$
Height	**$F(1, 18) = 39.1$, $p < .001$, $\eta^2 = .69$**
Load x Height:	$F(2, 36) = 2.4$, $p = .103$, $\eta^2 = .12$

Main effect

Height:	**25 > 12.5**	**$p < .001$**

Ankle moment PC1

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.5, 26.7) = 26.5, p < .001, \eta^2 = .60$
  - Height $F(1, 18) = 98.6, p < .001, \eta^2 = .85$
  - **Load x Height:** **$F(2, 36) = 3.5, p = .042, \eta^2 = .16$**
- Main effect
  - Load: $100 < 112.5$ $p = .004$
    - $112.5 < 125$ $p < .001$
    - $100 < 125$ $p < .001$
  - Height: $12.5 < 25$ $p < .001$
- Simple main effects
  - **Load @ H12.5%:** **$F(1.6, 28.6) = 16.6, p < .001, \eta^2 = .48$**
    - **Load:** **$100 < 125$ $p = .001$**
    - **$112.5 < 125$ $p = .003$**
  - **Load @ H25%:** **$F(1.6, 28.9) = 27.9, p < .001, \eta^2 = .61$**
    - **Load:** **$100 < 112.5$ $p = .001$**
    - **$100 < 125$ $p < .001$**
    - **$112.5 < 125$ $p = .001$**
  - **Height @ L100%:** **$F(1, 18) = 54.9, p < .001, \eta^2 = .75$**
    - **Height:** **$12.5 < 25$ $p < .001$**
  - **Height @ L112.5%:** **$F(1, 18) = 66.4, p < .001, \eta^2 = .79$**
    - **Height:** **$12.5 < 25$ $p < .001$**
  - **Height @ 125%:** **$F(1, 18) = 63.9, p < .001, \eta^2 = .78$**
    - **Height:** **$25 > 12.5$ $p < .001$**

Ankle moment PC2
- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.5, 26.5) = 93.2, p < .001, \eta^2 = .84$
  - Height $F(1, 18) = 31.2, p < .001, \eta^2 = .63$
  - **Load x Height:** **$F(1.6, 28.7) = 7.4, p = .004, \eta^2 = .29$**
- Main effect
  - Load: 100 > 112.5 $p < .001$
    - 112.5 > 125 $p < .001$
    - 125 > 112.5 $p < .001$
  - Height: 12.5 < 25 $p < .001$
- Simple main effects
  - **Load @ H12.5%:** **$F(2, 36) = 53.5, p < .001, \eta^2 = .75$**
    - **Load:** **100 > 112.5** **$p < .001$**
      - **112.5 > 125** **$p < .001$**
      - **100 > 125** **$p < .001$**
  - **Load @ H25%:** **$F(2, 36) = 86.9, p < .001, \eta^2 = .83$**
    - **Load:** **100 > 112.5** **$p < .001$**
      - **112.5 > 125** **$p < .001$**
      - **100 > 125** **$p < .001$**
  - **Height @ L100%:** **$F(1, 18) = 76.2, p < .001, \eta^2 = .81$**
    - **Height:** **25 > 12.5** **$p < .001$**

  - **Height @ L112.5%:** **$F(1, 18) = 14.9, p = .001, \eta^2 = .45$**
    - **Height:** **25 > 12.5** **$p = .001$**
  - **Height @ 125%:** **$F(1, 18) = 13.3, p = .002, \eta^2 = .43$**
    - **Height:** **12.5 < 25** **$p = .002$**

Ankle moment PC3
- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.3, 23.4) = 4.2, p = .043, \eta^2 = .19$
  - Height $F(1, 18) = 4.8, p = .041, \eta^2 = .21$
  - **Load x Height:** **$F(1.5, 26.9) = 6.3, p = .010, \eta^2 = .26$**
- Main effect
  - Load: 100 > 112.5 $p = .018$
  - Height: 25 > 12.5 $p = .041$
- Simple main effects
  - Load @ H12.5%: $F(1.3, 23.2) = 0.7, p = .464, \eta^2 = .04$
  - **Load @ H25%:** **$F(1.5, 26.5) = 9.1, p = .002, \eta^2 = .34$**
    - **Load:** **100 > 112.5** **$p = .001$**
      - **100 > 125** **$p = .016$**
  - **Height @ L100%:** **$F(1, 18) = 19.2, p < .001, \eta^2 = .52$**
    - **Height:** **25 > 12.5** **$p < .001$**
  - Height @ L112.5%: $F(1, 18) = 1.3, p = .272, \eta^2 = .07$
  - Height @ 125%: $F(1, 18) = 0.02, p = .883, \eta^2 = .001$

Gluteus maximus PC1

- 3x2 repeated measures factorial ANOVA
  - Load: $F(2, 36) = 2.6, p = .086, \eta^2 = .13$
  - **Height** **$F(1, 18) = 12.5, p = .002, \eta^2 = .41$**
  - Load x Height: $F(1.2, 21.2) = 0.5, p = .535, \eta^2 = .03$
- Main effect
  - **Height:** **12.5 < 25** **$p = .002$**

Gluteus maximus PC2

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.6, 29.0) = 15.5, p < .001, \eta^2 = .46$**
  - Height $F(1, 18) = 1.0, p = .334, \eta^2 = .05$
  - Load x Height: $F(2, 36) = 0.9, p = .372, \eta^2 = .05$
- Main effect
  - **Load:** **100 < 112.5** **$p = .017$**
    - **100 < 125** **$p = .001$**
    - **112.5 < 125** **$p = .009$**

Gluteus maximus PC3

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.2, 21.9) = 3.5, p = .067, \eta^2 = .16$
  - **Height** **$F(1, 18) = 5.4, p = .032, \eta^2 = .23$**
  - Load x Height: $F(2, 36) = 0.8, p = .437, \eta^2 = .05$
- Main effect
  - **Height:** **12.5 > 25** **$p = .032$**

Biceps femoris PC1

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.2, 22.0) = 4.3, p = .043, \eta^2 = .19$**
  - **Height** **$F(1, 18) = 7.7, p = .013, \eta^2 = .30$**
  - Load x Height: $F(1.2, 20.7) = 1.1, p = .318, \eta^2 = .06$
- Main effect
  - **Height:** **25 > 12.5** **$p = .013$**

Biceps femoris PC2

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(2, 35.6) = 13.9, p < .001, \eta^2 = .44$**
  - **Height** **$F(1, 18) = 10.0, p = .005, \eta^2 = .36$**
  - Load x Height: $F(2, 36) = 0.2, p = .806, \eta^2 = .01$
- Main effect
  - **Load:** **100 > 125** **$p < .001$**
    112.5 > 125 **$p = .002$**
  - **Height:** **12.5 < 25** **$p = .005$**

Biceps femoris PC3

- 3x2 repeated measures factorial ANOVA
  - Load: $F(1.5, 27.5) = 1.4, p = .265, \eta^2 = .07$
  - Height $F(1, 18) = 0.03, p = .864, \eta^2 = .002$
  - **Load x Height:** **$F(2, 36) = 7.23, p = .002, \eta^2 = .29$**
- Simple main effects
  - Load @ H12.5%: $F(2, 36) = 0.7, p = .480, \eta^2 = .04$
  - **Load @ H25%:** **$F(1.4, 24.9) = 7.3, p = .007, \eta^2 = .29$**
    - **Load:** **100 > 125** **$p = .029$**
      112.5 > 125 **$p = .029$**
  - **Height @ L100%:** **$F(1, 18) = 5.4, p = .032, \eta^2 = .23$**
    - **Height:** **12.5 < 25** **$p = .032$**
  - Height @ L112.5%: $F(1, 18) = 0.1, p = .719, \eta^2 = .01$
  - **Height @ 125%:** **$F(1, 18) = 6.1, p = .024, \eta^2 = .25$**
    - **Height:** **25 < 12.5** **$p = .024$**

Biceps femoris PC4

- 3x2 repeated measures factorial ANOVA
  - **Load:** **$F(1.6, 29.2) = 3.7, p = .045, \eta^2 = .17$**
  - **Height** **$F(1, 18) = 5.6, p = .030, \eta^2 = .24$**
  - Load x Height: $F(2, 36) = 2.3, p = .113, \eta^2 = .11$
- Main effect
  - **Load:** **100 < 112.5** **$p = .006$**
  - **Height:** **12.5 < 25** **$p = .030$**

Biceps femoris PC5

3x2 repeated measures factorial ANOVA

Load: $F(2, 36) = 0.6$, $p = .560$, $\eta^2 = .03$
Height $F(1, 18) = 3.9$, $p = .064$, $\eta^2 = .18$
Load x Height: $F(2, 36) = 1.6$, $p = .224$, $\eta^2 = .08$

Vastus medialis PC1
3x2 repeated measures factorial ANOVA
Load: **$F(1.6, 29.3) = 7.1, p = .005, \eta^2 = .28$**
Height **$F(1, 18) = 10.3, p = .005, \eta^2 = .36$**
Load x Height: $F(1.5, 26.5) = 1.3, p = .271, \eta^2 = .07$
Main effect
Load: **$100 < 112.5$ $p = .011$**
$100 < 125$ $p = .034$
Height: **$12.5 < 25$ $p = .005$**

Vastus medialis PC2
3x2 repeated measures factorial ANOVA
Load: **$F(2, 36) = 5.4, p = .009, \eta^2 = .23$**
Height $F(1, 18) = 3.5, p = .079, \eta^2 = .16$
Load x Height: $F(2, 36) = 0.7, p = .522, \eta^2 = .04$
Main effect
Load: **$100 < 125$ $p = .027$**

Vastus medialis PC3
3x2 repeated measures factorial ANOVA
Load: **$F(2, 36) = 6.9, p = .003, \eta^2 = .28$**
Height **$F(1, 18) = 13.2, p = .002, \eta^2 = .42$**
Load x Height: $F(2, 36) = 2.7, p = .079, \eta^2 = .13$
Main effect
Load: **$100 > 125$ $p = .019$**
Height: **$25 > 12.5$ $p = .002$**

Vastus medialis PC4
3x2 repeated measures factorial ANOVA
Load: **$F(2, 36) = 7.9, p = .001, \eta^2 = .30$**
Height $F(1, 18) = 0.3, p = .580, \eta^2 = .02$
Load x Height: $F(2, 36) = 3.0, p = .060, \eta^2 = .14$
Main effect
Load: **$100 > 125$ $p = .004$**

Vastus medialis PC5
3x2 repeated measures factorial ANOVA
Load: $F(2, 36) = 1.1, p = .357, \eta^2 = .06$
Height $F(1, 18) = 3.4, p = .080, \eta^2 = .16$
Load x Height: $F(2, 36) = 0.6, p = .547, \eta^2 = .03$

Medial gastrocnemius PC1
3x2 repeated measures factorial ANOVA
Load: $F(2, 36) = 0.8, p = .476, \eta^2 = .04$
Height $F(1, 18) = 6.1, p = .024, \eta^2 = .25$
Load x Height: $F(1.6, 29.4) = 1.2, p = .312, \eta^2 = .06$
Main
Height: 12.5 < 25 $p = .024$

Medial gastrocnemius PC2
3x2 repeated measures factorial ANOVA
Load: $F(2, 36) = 0.2, p = .838, \eta^2 = .01$
Height $F(1, 18) = 4.1, p = .057, \eta^2 = .19$
Load x Height: $F(2, 36) = 2.6, p = .091, \eta^2 = .13$

Medial gastrocnemius PC3
3x2 repeated measures factorial ANOVA
Load: $F(1.3, 23.1) = 10.1, p = .002, \eta^2 = .36$
Height $F(1, 18) = 0.6, p = .453, \eta^2 = .03$
Load x Height: $F(2, 36) = 4.7, p = .016, \eta^2 = .21$
Main effect
Load: 100 < 125 $p = .010$
112.5 < 125 $p < .001$
Simple main effects
Load @ H12.5%: $F(1.3, 23.9) = 4.2, p = .041, \eta^2 = .19$
Load: 125 > 112.5 $p = 0.010$
Load @ H25%: $F(1.5, 27.4) = 11.1, p = .001, \eta^2 = .38$
Load: 100 < 125 $p = .006$
112.5 < 125 $p < .001$
Height @ L100%: $F(1, 18) = 2.1, p = .161, \eta^2 = .11$
Height @ L112.5%: $F(1, 18) = 3.5, p = .077, \eta^2 = .16$
Height @ 125%: $F(1, 18) = 1.4, p = .252, \eta^2 = .07$

Medial gastrocnemius PC4
3x2 repeated measures factorial ANOVA
Load: $F(1.4, 24.8) = 5.2, p = .023, \eta^2 = .22$
Height $F(1, 18) = 2.1, p = .161, \eta^2 = .11$
Load x Height: $F(2, 36) = 0.2, p = .832\ \eta^2 = .01$

Medial gastrocnemius PC5
3x2 repeated measures factorial ANOVA
Load: $F(1.6, 29.0) = 3.0, p = .075, \eta^2 = .14$
Height $F(1, 18) = 0.2, p = .642, \eta^2 = .01$
Load x Height: $F(1.5, 26.6) = 2.8, p = .092, \eta^2 = .14$

Tibialis anterior PC1

3x2 repeated measures factorial ANOVA

Load: $F(1.4, 25.5) = 1.5, p = .232, \eta^2 = .08$

Height $F(1, 18) = 3.2, p = .088, \eta^2 = .15$

Load x Height: $F(1.2, 21.5) = 0.3, p = .609, \eta^2 = .02$

Tibialis anterior PC2

3x2 repeated measures factorial ANOVA

Load: **$F(2, 36) = 8.0, p = .001, \eta^2 = .31$**

Height $F(1, 18) = 0.02, p = .882, \eta^2 = .001$

Load x Height: $F(1.5, 26.8) = 1.1, p = .341, \eta^2 = .06$

Main effect

Load: **100 < 112.5** **$p = .031$**

100 < 125 **$p = .010$**

Tibialis anterior PC3

3x2 repeated measures factorial ANOVA

Load: $F(1.4, 24.9) = 0.03, p = .923, \eta^2 = .002$

Height $F(1, 18) = 1.5, p = .242, \eta^2 = .08$

Load x Height: $F(1.4, 24.6) = 2.5, p = .119, \eta^2 = .12$

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Winter, D. A. (2009). *Biomechanics and motor control of human movement* (4th Eds.). Hoboken, NJ: John Wiley & Sons, Inc.

Winter, D. A., Patla, A. E. (1997). *Signal processing and linear systems for the movement sciences*. Waterloo, ON: Graphic Services, University of Waterloo.

Wulf, G., Höβ, M., & Prinz, W. (1998). Instructions for motor learning: differential effects of internal versus external focus of attention. Journal of Motor Behavior

Wulf, G., McNevin, N., & Shea, C. H. (2001). The automaticity of complex motor skill learning as a function of attentional focus. *The Quarterly Journal of Experimental Psychology*, *54*A(4), 1143-1144.

Yeadon, M. R., King, M. A., Forrester, S. E., Caldwell, G. E., & Pain, M. T. G. (2010). The need for muscle co-contraction prior to a landing. *Journal of Biomechanics*, *43*, 364-369.

Yeow, C. H., Lee, P. V. S., & Goh, J. C. H. (2009). Effect of landing height on frontal plane kinematics, kinetics and energy dissipation at lower extremity joints. *Journal of Biomechanics*, *42*, 1967-1973.

Yeow, C. H., Ng, K. S., Cheong, C. H., Lee, P. V. S., & Goh, J. C. H. (2009). Repeated application of incremental landing impact loads to intact knee joints induces anterior cruciate ligament failure and tibiofemoral cartilage deformation and damage: A preliminary cadaveric investigation. *Journal of Biomechanics*, *42*, 972-981.

Yeow, C. H., Lee, P. V. S., Goh, J. C. H. (2011). An investigation of lower extremity energy dissipation strategies during single-leg and double-leg landing based on sagittal and frontal plane biomechanics. *Human Movement Science*, *30*, 624-635.

Zazulak, B. T., Ponce, P. L., Straub, S. J., Medvecky, M. J., Avedisian, L., & Hewett, T. E. (2005). Gender comparison of hip muscle activity during single-leg landing. *Journal of Orthopaedic & Sports Physical Therapy*, *35*(5), 292-299.

Zhang, S., Bates, B. T., & Dufek, J. S. (2000). Contributions of lower extremity joints to energy dissipation during landings. *Medicine & Science in Sports & Exercise*, *32*(4), 812-819.

ACADEMIC VITA
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University of Nevada, Las Vegas
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EDUCATION

Ph.D. Candidate	**Doctor of Philosophy, Kinesiology**, Biomechanics, 2014 University of Nevada, Las Vegas, Las Vegas, NV, USA Dissertation title: Investigating single-leg landing strategies and movement control across changes in task demands Advisor: Janet S. Dufek, Ph.D.
M.Sc.	**Master of Science, Kinesiology**, 2011 Lakehead University, Thunder Bay, ON, Canada Thesis title: Comparing vertical jump height measurement methods Advisor: Ian Newhouse, Ph.D.
H.B.K. (First class standing)	**Honours Bachelor of Kinesiology**, 2009 Lakehead University, Thunder Bay, ON, Canada Thesis title: Exploring rhythmic skill acquisition using music Advisor: Jane Taylor, Ph.D.
B.Sc.	**Bachelor of Science, Physics**, 2009 Lakehead University, Thunder Bay, ON, Canada

PROFESSIONAL EXPERIENCE

2014-2015	University of Nevada, Las Vegas Foundation President's Graduate Research Fellow, University of Nevada, Las Vegas, Las Vegas, Nevada, USA
2014	Lecturer, Department of Kinesiology and Nutrition Sciences, University of Nevada, Las Vegas, Las Vegas, Nevada, USA
2012-2014	Graduate Assistant, Department of Kinesiology and Nutrition Sciences, University of Nevada, Las Vegas, Las Vegas, Nevada, USA
2009-2011	Graduate Assistant, School of Kinesiology, Lakehead University, Thunder Bay, Ontario, Canada

GRANTS & AWARDS

Nordin, A.D. Kinetic variability during walking in casual footwear as a function of speed. *Graduate & Professional Student Association (GPSA) Sponsorship Funding*, University of Nevada, Las Vegas, $300, Summer 2012.

Nordin, A.D. Continuous variability assessment of vertical ground reaction forces during landing. *Graduate & Professional Student Association (GPSA) Sponsorship Funding*, University of Nevada, Las Vegas, $210, Spring 2013.

Nordin, A.D. Nebraska Biomechanics Core Facility Nonlinear Analysis Summer Workshop 2013. *Graduate & Professional Student Association (GPSA), School of Allied Health Sciences, and Department of Kinesiology and Nutrition Sciences Sponsorship Funding*, University of Nevada, Las Vegas, $1120, Summer 2013.

Nordin, A.D. Implications of increased lower extremity movement variability on fall susceptibility at increased stride lengths during locomotion. *Graduate & Professional Student Association (GPSA) Sponsorship Funding*, University of Nevada, Las Vegas, $400, Fall 2013.

Nordin, A.D. University of Nevada, Las Vegas (UNLV) Summer Session Scholarship, *UNLV Graduate College*, University of Nevada, Las Vegas, $2000, Summer, 2013.

Nordin, A.D. Changes in impact kinetics with altered foot strike patterns in running. *Southwest Chapter of the American College of Sports Medicine (SWACSM)*, Norman James Research Award, $500, Fall, 2013.

Nordin, A.D. Characterizing impact kinetics with changes in footwear and foot-strike using principal component analysis. *New Balance Footwear Research Award*. Production Footwear Award Category, Spring, 2014.

Nordin, A.D. Associating kinematic and kinetic synergies with movement control strategies in drop landing. *Graduate & Professional Student Association (GPSA) Sponsorship Funding*, University of Nevada, Las Vegas, $800, Summer, 2014.

Nordin, A.D. University of Nevada, Las Vegas (UNLV) Foundation President's Graduate Research Fellowship, *UNLV Graduate College*, University of Nevada, Las Vegas, $23,000, Fall 2014- Spring 2015.

Nordin, A.D. University of Nevada, Las Vegas (UNLV) Summer Session Scholarship, *UNLV Graduate College*, University of Nevada, Las Vegas, $2000, Summer, 2014.

Nordin, A.D. 2015 Regents' Graduate Scholar Award. *Nevada System of Higher Education (NSHE)*, University of Nevada, Las Vegas, $5000, Spring, 2015.

PUBLICATIONS

Refereed journal articles:

Nordin, A.D., Dufek, J.S. Theoretical perspectives surrounding gross motor movement strategies and functional variability. In preparation for *Journal of Motor Behavior*.

Nordin, A.D., Dufek, J.S. Subject-specific landing strategies identify changes in movement control. In preparation for *Human Movement Science*.

Nordin, A.D., Dufek, J.S. Loading patterns vary by direction, footwear, and foot-strike in running. In preparation for *Footwear Science*.

Nordin, A.D., James, C.R., Bates, B.T., Dufek, J.S. Identifying performer strategies in drop landing activities. In preparation for *Journal of Sports Sciences*.

Nordin, A.D., Dufek, J.S. Mercer, J.A. Three-dimensional impact kinetics with foot-strike manipulations during running. *Journal of Sport and Health Science*. JSHS.2015-0107. In review.

Nordin, A.D., Dufek, J.S. Lower extremity kinetic variability changes with drop landing height manipulations. *The Knee*. THEKNE-S-00281. In review

McKay, B., Wulf, G., Lewthwaite, R., **Nordin, A.D.**, The Self: Your Own Worst Enemy? A Test of the Self-Invoking Trigger Hypothesis. *Quarterly Journal of Experimental Psychology*.
doi: 10.1080/17470218.2014.997765.

Refereed technical paper:

Nordin A.D., Bailey, J., & Dufek, J.S. Implications of increased lower extremity movement variability on fall susceptibility at increased stride lengths during locomotion. *2013 American Society of Mechanical Engineers (ASME) International Mechanical Engineering Congress and Exposition Conference Proceedings (IMECE)*, November 2013. IMECE2013-63804.

PRESENTATIONS

Session Presentations:

National

Freedman Silvernail, J., **Nordin, A.D.**, & Dufek, J.S. Alterations in movement coordination due to increasing landing height. *American College of Sports Medicine (ACSM) 62nd Annual Meeting, 6th World Congress on Exercise is Medicine and World Congress on the Basic of Exercise Fatigue*, San Diego, USA, May 2015. Accepted.

Dufek, J.S., **Nordin, A.D.**, James, C.R., & Bates, B.T. Does landing strategy change with increased mechanical task demands? *American College of Sports Medicine (ACSM) 62nd Annual Meeting, 6th World Congress on Exercise is Medicine and World Congress on the Basic of Exercise Fatigue*, San Diego, USA, May 2015. Accepted.

Regional

Nordin, A.D., Castro, S.B., Dufek, J.S., & Mercer, J.A. Changes in impact kinetics with altered foot strike patterns in running. *Southwest Chapter of the American College of Sports Medicine (SWACSM) 32nd Annual Meeting*, Student Research Award Presentation, Norman James Research Award Winner 2013, Newport Beach, USA, October 2013.

Local

Nordin, A.D., Dufek, J.S., & Mercer, J.A. Foot-strike kinetics in running: making an impact via 3D analysis. University of Nevada, Las Vegas, *Graduate & Professional Student Research Forum, Dental Medicine, Nursing and Science Platform Session*, Las Vegas, USA, March 2014.

Nordin, A.D., & Dufek, J.S. Minimalist vs. cushioned running shoes: impact loads vary with foot-strike pattern. University of Nevada, Las Vegas, *Graduate & Professional Student Research Forum, Dental Medicine, Nursing and Science Platform Session*, Las Vegas, USA, March 2015.

Refereed poster presentations:

International

Nordin, A.D., & Dufek, J.S. Continuous variability assessment of vertical ground reaction forces during landing. *Computer Methods in Biomechanics and Biomedical Engineering (CMBBE) 2013, 11th International Symposium*, Salt Lake City, USA, April 2013.

Nordin, A.D., & Dufek, J.S. Examining lower extremity range of motion and movement variability changes due to focus of attention during landing. *International Society of Motor Control (ISMC), Progress in Motor Control IX*, Montreal, Canada, July 2013.

Nordin, A.D., & Dufek, J.S. Investigating lower extremity functioning via frontal plane movement variability asymmetries during landing. *43rd annual meeting of the Society for Neuroscience (SfN)*, San Diego, USA, November 2013.

Nordin A.D., Bailey, J., & Dufek, J.S. Implications of increased lower extremity movement variability on fall susceptibility at increased stride lengths during locomotion. *2013 American Society of Mechanical Engineers (ASME) International Mechanical Engineering Congress and Exposition (IMECE)*, San Diego, USA, November 2013.

Bailey, J.P., **Nordin, A.D.**, & Dufek, J.S. Effects of stride length perturbations on anterior-posterior components during the stance phase of walking. *American College of Sports Medicine (ACSM) 61st Annual Meeting, 5th World Congress on Exercise is Medicine and World Congress on the Role of Inflammation in Exercise, Health and Disease*, Orlando, USA, May 2014.

Nordin, A.D., Kivi, D.K., Zerpa, C.Z., & Newhouse, I.J. Comparison of methods for assessing vertical jump height performance. *32nd Conference of the International Society of Biomechanics in Sports (ISBS)*, Johnson City, USA, July 2014.

Nordin, A.D., & Dufek, J.S. Associating kinematic and kinetic synergies with movement control strategies in drop landing. *7th World Congress of Biomechanics*, Boston, USA, July 2014.

Nordin, A.D., Bailey, J.P., Dufek, J.S., & Mercer, J.A. A novel approach to characterizing 3D impact kinetics during foot-strike manipulations in running. *International Calgary Running Symposium*, Calgary, Canada, August 2014.

Nordin, A.D., & Dufek, J.S. Joint-specific kinetic adjustments following landing height manipulations. *American College of Sports Medicine (ACSM) 62nd Annual Meeting, 6th World Congress on Exercise is Medicine and World Congress on the Basic of Exercise Fatigue*, San Diego, USA, May 2015. Accepted.

National

Nordin, A.D., Kivi, D.M.R., Newhouse, I.J., Zerpa, C. Comparing maximum and takeoff centre of mass velocities during jumping in female university volleyball players. *2011 Canadian Society for Exercise Physiology (CSEP) Annual General Meeting*, Quebec City, Canada, October 2011.

Nordin, A.D., & Dufek, J. S. Kinetic variability during walking in casual footwear as a function of speed. *17th Biannual Canadian Society of Biomechanics/Societe Canadienne de Biomechanique (CSB/SBC)*, Burnaby, Canada, June 2012.

Nordin A.D., & Dufek, J.S. Gender difference in frontal plane lower extremity kinetic variability during landing. *2013 Meeting of the American Society of Biomechanics (ASB)*, Omaha, USA, September 2013.

Nordin A.D., Bailey, J., & Dufek, J.S. Variations in leg stiffness and lower extremity range of motion variability from stride length perturbations during gait. *2013 Meeting of the American Society of Biomechanics (ASB)*, Omaha, USA, September 2013.

Bailey, J.P., **Nordin, A.D.**, & Dufek, J.S. Step length perturbations alter variations in center of mass horizontal velocity. *2013 Meeting of the American Society of Biomechanics (ASB)*, Omaha, USA, September 2013.

Nordin, A.D., & Dufek, J.S. Loading patterns vary direction, footwear, and foot-strike. *39th Annual Meeting of the American Society of Biomechanics*, Columbus, USA, August 2014. Submitted.

Nordin, A.D., & Dufek, J.S. Subject-specific landing strategies identify changes in movement control. *39th Annual Meeting of the American Society of Biomechanics*, Columbus, USA, August 2014. Submitted.

Regional

Nordin, A.D., & Dufek, J.S. Examining lower extremity joint range of motion variability during landing. *Southwest Chapter of the American College of Sports Medicine (SWACSM) 31st Annual Meeting*, Newport Beach, USA, October 2012.

Bailey, J., **Nordin, A.D.**, Lee, D., & Dufek, J.S. Kinematic effects of stride length perturbations on system COM horizontal velocity during locomotion. *Southwest Chapter of the American College of Sports Medicine (SWACSM) 31st Annual Meeting*, Newport Beach, USA, October 2012.

Nordin, A.D., & Dufek, J.S. Knee joint moment variability during landing for females. *Noraxon Fall 2012 Research Symposium*, Las Vegas, USA, October 2012.

Coupe, A., Montes, A., **Nordin, A.D.**, Dufek, J.S. Effect of simulated obesity on the double support phase of gait. *Southwest Chapter of the American College of Sports Medicine (SWACSM) 32nd Annual Meeting*, Newport Beach, USA, October 2013.

Holt, J.L., **Nordin, A.D.**, & Dufek, J.S. Effects of added mass and horizontal displacement on landing strategies. *Southwest Chapter of the American College of Sports Medicine (SWACSM) 32nd Annual Meeting*, Newport Beach, USA, October 2013.

Wilson, J.N., **Nordin, A.D.**, & Dufek, J.S. Do landing strategies vary with increased system energy? *Southwest Chapter of the American College of Sports Medicine (SWACSM) 32nd Annual Meeting*, Newport Beach, USA, October 2013.

Local

Nordin, A.D., Dufek, J.S., & Mercer, J.A. Inter-axis changes in impact kinetics with foot-strike manipulations in running. *7th Annual Interdisciplinary Research & Scholarship Day (IRSD),* 3rd place Student Poster Award, University of Nevada, Las Vegas, USA, April 2014.

Nordin, A.D., & Dufek, J.S. Exploring movement control strategies via principle component analysis in landing. *7th Annual Interdisciplinary Research & Scholarship Day (IRSD)*, University of Nevada, Las Vegas, USA, April 2014.

Non-refereed poster presentations:

Local

Bailey, J., **Nordin A.D.**, and Dufek, J.S. The effects of altering stride length on system COM vertical and horizontal velocity. *2012 Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program*, University of Nevada, Las Vegas, USA, August 2012.

Comanescu, T., **Nordin, A.D.**, Bailey, J., Dufek, J.S, and Lee, D. Influence of step length manipulations on the frequency components of ground reaction force- Do dynamic oscillations change with step length? *2012 Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program*, University of Nevada, Las Vegas, USA, August 2012.

Wilson, J.N., **Nordin, A.D.**, & Dufek, J.S. Landing strategy patterns at selected bodyweight percentages. *2013 Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program*, University of Nevada, Las Vegas, USA, August 2013.

Holt, J.L., **Nordin, A.D.**, & Dufek, J.S. Effects of added mass and horizontal displacement on landing strategies. *2013 Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program*, University of Nevada, Las Vegas, USA, August 2013.

Castro, S.B., **Nordin, A.D.**, & Mercer, J.A. Kinematics and kinetics of foot strike patterns during running. *2013 Institutional Development Award Network of Biomedical Research Excellence (INBRE) Undergraduate Research Opportunity Program*, University of Nevada, Las Vegas, USA, August 2013.

UNIVERSITY TEACHING EXPERIENCE

Graduate

KIN 743	Research Techniques in Biomechanics	S-13, S-14

Undergraduate

KIN 346	Biomechanics	SI-12, 14, S-12-14, F-12-13
KINE 1010	Introduction to Kinesiology	W-10, W-11
KINE 1035	Physical Growth and Motor Development	W-10
KINE 1711	Movement Observation and Assessment	W-10
KINE 2015	Introduction to Biomechanics	F-10
KINE 2711	Qualitative Analysis	W-11
KINE 3230	Research Processes	F-10

KINE 4310	Ergonomics	F-09, F-10
KINE 4714	Clinical Exercise Therapy	F-10

*KINE: Kinesiology, School of Kinesiology, Lakehead University, Thunder Bay, Ontario, Canada. (F: Fall semester, W: Winter semester, -Year)

*KIN: Kinesiology, Department of Kinesiology and Nutrition Sciences, School of Allied Health Science, University of Nevada, Las Vegas, Nevada, USA. (F: Fall semester, S: Spring semester, SI: Summer semester I, - Year)

PROFESSIONAL ORGANIZATIONS

International Society of Biomechanics (ISB); 2012-present
International Society of Biomechanics in Sports (ISBS); 2013-present
International Society of Motor Control (ISMC); 2013-present
Society for Neuroscience (SfN); 2013-present
American Society of Biomechanics (ASB); 2012-present
Canadian Society for Biomechanics (CSB); 2012-present
American College of Sports Medicine (ACSM); 2012-present
Southwest Chapter of the American College of Sport Medicine; 2012-present
American Society of Mechanical Engineers (ASME); 2013-present

PROFESSIONAL SERVICE

University, Department, College

Nebraska Biomechanics Core Facility (NBCF) Nonlinear Analysis Summer Workshop, University of Nebraska, Omaha, USA, June 2013.

University of Nevada, Las Vegas, Graduate & Professional Student Association (GPSA), Department of Kinesiology & Nutrition Sciences Representative, Elections Committee; 2013.

University of Nevada, Las Vegas, Rebel Science, Technology, Engineering, & Math (STEM) Academy, School of Allied Health Sciences: Biomechanics Lab Representative; 2012-2013.

12th Motor Control Summer School (MCSS-XII), Pennsylvania, USA, May 2015. Invited.

Non-University

Manuscript Reviewer

Medicine and Science in Sports and Exercise; 2013-present.

HONOURS, AWARDS, AND DISTINCTIONS

2003	International Baccalaureate Programme Diploma Graduate
2003	Lakehead University Academic Entrance Scholarship
2006	Lakehead University Male Indoor Track & Field Athlete of the Year
2006	Lakehead University Track & Field Team Captain
2007- 2008	Academic All-Canadian, Lakehead University
2008	Canadian National Coaching Certification Program Level II Sprints and Hurdles
2009	Honours Bachelor of Kinesiology- First Class Standing, Lakehead University
2013	Southwest Chapter of the American College of Sports Medicine (SWACSM) Norman James Research Award
2014	New Balance Footwear Research Award
2014	University of Nevada, Las Vegas (UNLV) Foundation President's Graduate Research Fellowship
2014	University of Nevada, Las Vegas, Division of Health Sciences, Interdisciplinary Research & Scholarship Day, 3rd place Student Poster Award
2015	Nevada System of Higher Education (NSHE) Regents' Graduate Scholar Award, University of Nevada, Las Vegas