Biomechanical Analyses of Bodyweight Unilateral Lower Limb Exercise Tasks – Comparison of Common Squatting and Lunting Movements

by

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A thesis proposal submitted in conformity with the requirements for the degree of Master of Science
Department of Exercise Sciences
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Abstract

Bodyweight unilateral lower limb exercise tasks (BULLETs) are used in movement assessment and (re)training contexts. Since kinematic characteristics of individual BULLETs are different, it was hypothesized that lower extremity biomechanical loading patterns would differ between BULLETs. Thirty-two participants performed bodyweight forward lunges, backward lunges, split squats, single-leg squats, and double-leg squats. Body segment kinematics and ground reaction forces were used to quantify body-size normalized, lower limb net joint moments. Peak and average hip, knee, ankle, and total support moments differed across BULLETs ($p < 0.05$). Sex-based differences were observed for peak and average support moment, and peak knee moment ($p < 0.05$). Characterizing BULLETs based on joint moment magnitudes can be used to improve the sensitivity and specificity of movement assessments and to design movement (re)training programs. Further work is needed to address cause(s) of between-sex differences in joint moment magnitudes given that body size differences were not explanatory.
Acknowledgments

This manuscript is dedicated to my grandparents:

John (Jack) and Shirley MacFarlane

I would not be the person I am today without their love, support, influence and inspiration.
# Table of Contents

Acknowledgments ........................................................................................................ iii

Table of Contents ........................................................................................................ iv

List of Figures .............................................................................................................. vii

List of Tables ............................................................................................................... vii

List of Appendices ..................................................................................................... x

1 Introduction .............................................................................................................. 1

  1.1 Questions ........................................................................................................... 4

      1.1.1 Question 1 – Joint Kinetics ................................................................. 4

      1.1.2 Question 2 – Sex Differences ............................................................. 4

1.2 Hypotheses ......................................................................................................... 4

      1.2.1 Null Hypothesis 1 ............................................................................. 4

      1.2.2 Null Hypothesis 2 ............................................................................. 4

2 Review of Literature .............................................................................................. 5

  2.1 Single-Leg Squat (SLS) ................................................................................... 5

  2.2 Forward Lunge / Backward Lunge / Split Squat (FL / BL / SS) .................... 8

  2.3 Double-Leg squat (DLS) ................................................................................. 10

  2.4 Current Application of BULLETs ................................................................. 12

      2.4.1 Qualitative Movement Assessment ............................................... 13

      2.4.2 Exercise Prescription ................................................................. 14

  2.5 Justification for Methodology ...................................................................... 15

      2.5.1 Experimental Conditions ............................................................. 15

      2.5.2 Kinematic Variables ........................................................................ 16

      2.5.3 Kinetic Variables ............................................................................ 18

      2.5.4 Sex Differences ............................................................................. 20

3 Methods ............................................................................................................... 23
B. Physical Activity Readiness Questionnaire (PAR-Q) .................................................. 75
C. Images of Exercise Tasks (BULLETs)............................................................................ 76
D. Verbal Instructions for Exercise Tasks........................................................................ 78
   Task: Single-Leg Squat (SLS) .................................................................................... 78
   Task: Double-Leg Squat (DLS) .................................................................................. 79
   Task: Split Squat (SS) ............................................................................................... 80
   Task: Forward Lunge (FL) ....................................................................................... 81
   Task: Backward Lunge (BL) ..................................................................................... 82
E. BULLET Kinematics................................................................................................... 83
F. Alternate BULLET Normalized Joint Moments ......................................................... 84
List of Figures

Figure 1: Visual 3D™ screen captures of linked segment models of the four BULLETs and double-leg squat .......................................................... 35

Figure 2: (Top) Retroreflective marker placement for data collection. Colour of marker denotes purpose: red = segment calibration; yellow = segment tracking; green = dual purpose segment calibration and tracking. (Bottom) Image of participant performing a BULLET task with markers attached .................................. 37

Figure 3a: Visual 3D™ screen captures denoting event detections ON and OFF for Double-Leg Squat (DLS), Single-Leg Squat (SLS), and Split Squat (SS) using processed kinematic signals .............................................. 41

Figure 3b: Visual 3D™ screen captures denoting event detections ON and OFF for Backward Lunge (FL) using processed kinematic signals .............................................. 41

Figure 3c: Visual 3D™ screen captures denoting event detections ON and OFF for forward lunge (FL) using processed kinetic signals. Force onset as shown on left by first captured frame detecting a force vector (blue arrow) .............................................. 41

Figure 4: (Left) Peak and (Right) average body-size normalized support moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters above bars (p > 0.05), while different letters indicate BULLET tasks were statistically different (p < 0.05). Men were significantly greater than women across all TASK conditions for peak (p = 0.009) and average (p = 0.013) normalized support moment magnitudes ...................................................................... 46

Figure 5: (Left) Peak and (Right) average body-size normalized hip net joint moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters above bars (p > 0.05), while different letters indicate BULLET tasks were statistically different (p < 0.05). No significant difference were observed between men and women across all conditions for peak and average values (p > 0.05) ............................................. 47

Figure 6: (Left) Peak and (Right) average body-size normalized knee net joint moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters above bars (p > 0.05), while different letters indicate BULLET tasks were statistically different (p < 0.05). A significant difference for SEX existed in the peak knee moments, (p < 0.018), but between-sex differences in average knee moment magnitudes were not statistically significant (p > 0.069) .................................................. 49
Figure 7: (Left) Peak and (Right) average body-size normalized ankle net joint moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) participants are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters below bars ($p > 0.05$), while different letters indicate BULLET tasks were statistically different ($p < 0.05$). No significant difference existed between men and women across all conditions for peak and average values ($p > 0.05$)
List of Tables

Table 1: Participant characteristics presented as mean (standard deviation). N = numbers of subjects.

Table 2: Peak joint flexion angles and ranges of motion for the hip, knee, ankle and trunk during the four BULLET conditions (SLS = single leg squat; SS = split squat; BL = backward lunge; FL = forward lunge) and a baseline double leg squat task (DLS). Data included are the mean (standard deviation) values calculated across all participants (N = 32) as no SEX*TASK interaction and no main effect for SEX was detected from two-way ANOVAs computed for each joint individually (p > 0.05). Superscript lowercase letter indicates post-hoc results as a main effect for TASK was found for each joint individually (p < 0.05).

Table 3: Peak and average net joint extension/plantarflexion moments for the hip, knee, ankle and total support during the four BULLET conditions (SLS = single-leg squat; SS = split squat; BL = backward lunge; FL = forward lunge) and a baseline double-leg squat task (DLS). Data are normalized to body-mass (N·m/kg) and included are the mean (standard deviation) values calculated for men (N = 16), women (N = 16), and all participants combined (N = 32).
List of Appendices

A: Informed Consent .................................................................................................................. 82
B: Physical Activity Readiness Questionnaire ........................................................................... 85
C: Images of Exercise Tasks (BULLETs) .................................................................................. 86
D: Verbal Instructions for Exercise Tasks .................................................................................. 88
E: BULLET Kinematics .............................................................................................................. 93
F: Alternate BULLET Normalized Joint Moments ................................................................... 94
1 Introduction

Resistance training is a fundamental intervention used in the health care and fitness industries, and is recommended to the general population as part of a healthy, active lifestyle (CSEP, 2014; Garber et al., 2011). Bodyweight unilateral lower limb exercise tasks, referred to herein as BULLETs (e.g., lunge, split squat, single-leg squat), are commonly used in exercise prescription to elicit specific adaptations such as increasing lower limb strength and muscle mass (Baechle & Earle, 2008; Bizzini, Junge, & Dvorak, 2007; Tsatsouline, 2003). Furthermore, these tasks are included in clinical assessments in an attempt to determine an individual’s risk of sustaining an injury based on the patterns of movement coordination and control exhibited during task execution (M. A. Clark & Lucett, 2010; Comerford & Mottram, 2012; Cook, Burton, Kiesel, Rose, & Bryant, 2010; Kritz, 2012; Mischiati et al., 2015).

If the human body is modeled as a complex mechanical linkage, the kinematics and kinetics of its motion are intrinsically linked (Zajac, 1993). Since the kinematic characteristics of individual BULLETs are inherently different, it is therefore likely that the biomechanical loading patterns amongst lower limb joints differ between tasks. For example, between-task differences in the configuration of body segments suggests that whole-body center of mass (COM) trajectories would differ across BULLETs and result in different ground reaction force profiles to maintain postural stability; this would affect the external joint moment of force and associated (i.e., equilibrating) muscle force patterns. Moreover, muscle lengths and orientations would vary between tasks due to different body segment and joint angles, causing differences in the muscle moment arms, contraction velocities, and thus their force- and moment-producing capabilities.

Although acknowledged that the addition of external resistance or variations in movement speeds could affect the abovementioned relationships in complex ways (Frost, Beach, Callaghan, & McGill, 2013), it is nevertheless beneficial to investigate biomechanical differences between BULLETs because performing exercises without added resistance and at low speeds often constitutes a baseline condition for novice or uninjured trainees. Further, bodyweight tasks are commonly incorporated as part of a battery of tasks in qualitative movement assessments (M. A. Clark & Lucett, 2010; Comerford & Mottram, 2012; Cook et al., 2010; Kritz, 2012). In this specific case, understanding how similar or different these tasks are
in lower limb loading may enhance the rationale for their inclusion or exclusion, which in turn could enhance the validity, specificity or sensitivity of assessments that incorporate one or more of these movement tasks.

Lower limb internal net joint moments have been used to great extent in biomechanical research as a simple but crude means to quantify the total outcome effect of all internal force-producing and -bearing structures surrounding a joint (Andriacchi, Andersson, Fermier, Stern, & Galante, 1980; S. P. Flanagan & Salem, 2008; Gullett, Tillman, Gutierrez, & Chow, 2009; Mathiyakom, McNitt-Gray, Requejo, & Costa, 2005; Riemann, Congleton, Ward, & Davies, 2013; Salem, Salinas, & Harding, 2003; Singh, Yack, Francis, & Janz, 2015; Winter, 1980). Knowledge of the lower limb net joint moments associated with performing BULLETs, in particular, could aid practitioners in the design and administration of these exercises (Herbert, Moore, Moseley, Schurr, & Wales, 1993). As opposed to primarily manipulating external resistance and/or movement speed of a given exercise, different BULLETs could also be selected to progress/regress the crude biomechanical demands imposed on the basis of the associated net joint moment magnitudes. This would be especially useful for individuals with limited experience (e.g., novice trainees) and/or those who are not physically capable to safely execute exercises at high movement speeds and/or against additional external resistance (e.g., sedentary or previously injured trainees). Additionally, using different BULLETs as a means to alter biomechanical demands could provide a method of progressive overload when equipment and coaching are unavailable. Having greater control over progressive overload and a further understanding of the imposed lower limb demands during such exercise tasks could also prove useful in the prevention of injuries, as the application and rate of application of excessive load without adequate adaptation or recovery is suggested to be a primary cause of training related injuries (Blanch, 2014; Kumar, 2001; McGill, 1997).

Quantifying and comparing the lower limb net joint moments across BULLETs can bring practitioners a step closer to integrating more biomechanical logic into planning exercise programs (i.e., periodization). Current practice focuses more on manipulating external loading parameters (e.g., added mass, system mass, etc.), with less regard for the associated joint loading demands (e.g., net joint moments). In cases where specific musculoskeletal exercise adaptations are desired, using net joint moment magnitudes to inform exercise prescription and progression is arguably more ecologically valid. Given that the majority of joint motion during
BULLETs is confined to the sagittal plane, net joint extension (hip and knee) and plantarflexion (ankle) moment magnitudes could be considered for each of the primary lower limb joints individually as a means to crudely quantify the task demand. Knowledge of these specific joint moment magnitudes would support a practitioner/clinician, for example, during a rehabilitation or task-transfer scenario where the attentional focus and desired outcomes are targeted locally at a single joint (e.g., improved hip extensor strength). In cases when the objective is to elicit a more global adaptation (e.g., lower limb concentric/eccentric strength improvement), the practitioner could use knowledge of a combination of moments in the exercise planning process. For example, the “support moment” can be calculated by adding the net joint extension (hip and knee) and plantarflexion (ankle) moments (Winter, 1980), and used a valid measure of total lower limb mechanical demand (S. Flanagan & Salem, 2005).

One important consideration when using bodyweight exercises for the abovementioned purposes is that segment masses and lengths differ between individuals – especially between men and women (Sizer & James, 2014) – which would affect net joint moment magnitudes. It is common practice to normalize net joint moment magnitudes by dividing them by body mass and/or size; this is done to account for differences in anthropometry so that inter-individual comparisons can be made (Moisio, Sumner, Shott, & Hurwitz, 2003). However, likely related to sexual dimorphism in the human musculoskeletal system (Sizer & James, 2014), even when biomechanical data are normalized with respect to body size and shape characteristics, between-sex differences still exist in single-leg squat and lunge mechanics (Bolga, Cook, Hogarth, Scott, & West, 2014; Dwyer, Boudreau, Mattacola, & Uhl, 2010; Graci, Van Dillen, & Salsich, 2012; Nakagawa, Moriya, Maciel, & Serrao, 2012; Zeller, McCrory, Kibler, & Uhl, 2003). Thus, it is important to consider if between-sex differences exist in the lower limb net joint moments of various BULLETs, as interpretation and application of results may vary between men and women.

To date, no research has directly compared the lower limb net joint moments and support moment between the various BULLETs performed for use in the aforementioned scenarios. Thus, the primary objective of this research was to quantify lower limb net joint moments during commonly prescribed BULLETs, and to compare their magnitudes, and a summation of these magnitudes between BULLETs. A secondary objective was to make between-sex
comparisons in these quantities to determine if net joint moment magnitudes differ between men and women when normalized for body size.

1.1 Questions

1.1.1 Question 1 – Joint Kinetics

Do mean and peak lower limb net joint extension (hip and knee) and plantarflexion (ankle) moments, and their algebraic summation (i.e., total support moment), differ between four common BULLETs (i.e. single-leg squat, split squat, forward lunge, backward lunge) and a baseline non-BULLET task (i.e., double-leg squat)?

1.1.2 Question 2 – Sex Differences

Are any between-BULLET differences in the abovementioned kinetic quantities dependent on the sex of the performer when body-size and mass are accounted for by dividing net joint moment magnitudes by body mass × leg length?

1.2 Hypotheses

1.2.1 Null Hypothesis 1

There will be no difference in the mean or peak net joint extension (hip and knee), plantarflexion (ankle), or total support moment between the BULLETs.

1.2.2 Null Hypothesis 2

Any between-BULLET differences in the abovementioned kinetic quantities will be consistent between men and women.
2 Review of Literature

The objective of this section is three-fold. First, a summary of the current state of knowledge and gaps that exist with respect to selected BULLETs is provided. Second, an overview of how practitioners currently use BULLETs in exercise paradigms is offered as way to demonstrate the relevance of the work undertaken. Third, justification of the methodology employed is provided.

BULLET is an acronym used in this thesis for a collection of “bodyweight unilateral lower limb exercise tasks”, which include kinematically asymmetrical, ground-based, free-standing movements whose motion primarily involves flexion/extension at the hip and knee and dorsiflexion/plantarflexion at the ankle. ‘Bodyweight’ implies that no additional external resistance is added to the body, only an individual’s weight (i.e., body mass × acceleration due to gravity) offers the resistance sustained. The exercise tasks investigated herein are the forward lunge (FL), backward lunge (BL), split squat (SS), and single-leg squat (SLS; Appendix C). The double-leg squat [DLS] is used as a baseline task for comparison. Rationale for the selection of these BULLETs will be provided in justification of methodology.

All BULLETs consist of the following positions and motions in sequential order: an initial starting position; an eccentric (downward) phase of movement; a transition phase from eccentric to concentric also known as the bottom position; a concentric (upward) phase of movement; and ending with the final position which is identical to the initial position. For the purposes of this study, hands were placed on the pelvis (i.e., hands on “hips”) in order to minimize variations in upper limb positioning and its influence on kinematics and COM motion. The dominant leg was considered the “stance-leg” moving forward. The specific kinematic variation and a summary of the available literature to date for each task will be explained subsequently in its respective section.

2.1 Single-Leg Squat (SLS)

The SLS begins with an initial position of upright standing with feet flat on the ground, hip-to-shoulder-width apart. The hip and knee joints are initially extended while the ankle joint remains in a neutral position. The non-stance leg is then elevated slightly so that it does not come in contact with the ground at any point during the movement task. Movement is initiated
by lowering the COM towards the ground via (dorsi)flexion about the hip, knee and ankle (i.e., triple flexion) while attempting to maintain a neutrally-oriented thoracolumbar spine and permitting minimal mediolateral motion of the knee joint. The transition point or bottom of movement occurs once a maximum controllable squat depth is reached. All three lower limb joints then subsequently extend to produce upward COM motion. The movement ends when the final position is reached, which is identical to the initial position. For the purposes of this study, the participant was instructed to maintain postural stability without allowing the elevated foot to touch the ground for the duration of the movement task.

Variations in the position of the elevated foot in the SLS do exist (i.e., foot in front of stance leg, foot behind the stance leg). One unpublished pilot study by the author has shown that while stance leg kinematics and inter-joint load sharing remain largely unaffected by elevated foot position, body mass-normalized sagittal net joint moments differ at the hip and knee (Chapman et al., 2014). Although this pilot study was statistically underpowered, the results were confirmed in an appropriately powered study (A. Khuu, Foch, & Lewis, 2016), and thus provides rationale to control position of the elevated foot during SLS performance.

During quantitative kinetic and kinematic analysis, all joint angles, moments and ground reaction forces can be reliably measured between sessions within a given day and within a given week for the SLS (Alenezi, Herrington, Jones, & Jones, 2014). These measurement methods have been widely employed to justify its use as a qualitative movement screen, more so than the other BULLETs studied herein. Visual observation of the SLS has shown to be reliable between raters for both inferring joint angle/segment positions (Ageberg et al., 2010; Edmondston et al., 2013; Poulsen & James, 2011) and qualitative scoring scales (Almangoush, Herrington, & Jones, 2014; Crossley, Zhang, Schache, Bryant, & Cowan, 2011; Junge, Balsnes, Runge, Juul-Kristensen, & Wedderkopp, 2012). This reliability increases with education and experience (Weeks, Carty, & Horan, 2012). Using visual observation of the SLS in movement screens is supported as it sensitive to assessing knee valgus or medial knee deviation (Ugalde, Brockman, Bailowitz, & Pollard, 2014), which has been shown to be a risk factor for lower limb non-contact knee injuries (K.R. Ford, Myer, & Hewett, 2003; Hewett et al., 2005).

SLS performance is also related to hip muscle function (Crossley et al., 2011; Hollman, Galardi, Lin, Voth, & Whitmarsh, 2014; Stickler, Finley, & Gulgin, 2015). People who exhibit
less medial knee displacement during the SLS have faster onset of hip abductor activation and stronger hip muscles. Hip muscle function is greatly compromised in people with anterior knee pain (Nakagawa et al., 2012; Piva, Goodnite, & Childs, 2005; Prins & van der Wurff, 2009). Individuals who have patellofemoral pain syndrome (PFPS) exhibit approximately 20% less strength in hip abduction and external rotation compared to healthy controls, and show greater knee valgus (Levinger, Gilleard, & Coleman, 2007; Nakagawa et al., 2012), in addition to displaying limited ankle dorsiflexion (Mauntel et al., 2013). This would suggest that all primary joints of the lower limb need to be considered when practitioners interpret SLS performance. In addition to exhibiting greater knee valgus and less ankle dorsiflexion than their unaffected counterparts, individuals with PFPS also show greater ipsilateral trunk lean, contralateral pelvic drop, and hip adduction. A more upright trunk position during SLS has also been shown to increase the force and strain of the anterior cruciate ligament (Kulas, Hortobagyi, & DeVita, 2012).

Further evidence from an intervention study supports the link between hip musculature and performance during the SLS (Willy & Davis, 2011). Specifically, healthy women were assessed with excessive hip adduction using instrumented motion capture during running and placed into an intervention group. This group completed a hip movement and strengthening program focused on the abductors and external rotators that consisted of progressions from side lying to standing, from supported to non-supported, and from unloaded to added resistance. This program was performed 3 times per week for 6 weeks and a control group comprised of participants who did not display excessive hip adduction during the SLS were included. While the intervention group showed no changes in hip adduction during running, they did significantly decrease hip adduction, hip internal rotation and contralateral pelvic drop during SLS task. The training intervention was designed to improve hip muscle capacity, and resulted in improvements in kinematics of SLS performance, but did not influence the kinematics of running performance.

There is more SLS literature that is predominantly frontal plane focused, but is not relevant to the study at hand and therefore not reported. Surprisingly, with the vast amount of research on the kinematics and muscle activation of SLS performance, only one study has quantified the net joint moments of the hip, knee and ankle during this exercise task (A. Khuu et al., 2016). The results of this study showed that changing the elevated foot/leg position during a SLS changed
the net joint moments observed, in addition to other kinetic and kinematic variables. While this provides published values to compare the results herein against, the authors only extracted the moment values at two discrete kinematic time points, 60 degrees knee flexion and peak knee flexion. For the purpose herein, the peak and average net joint moments were examined to understanding the magnitude of maximum loading during the SLS and the average total loading for the duration of the task. While there is some evidence to suggest the former may occur at peak flexion angles during squatting tasks (Cotter, Chaudhari, Jamison, & Devor, 2013; Scaglioni-Solano, Ferris, Dunn, & Salem, 2005), the latter does not relate to these discrete time points. Therefore, the current study will add to the body of literature of net joint moment responses observed during SLS exercise tasks. Furthermore, to date no studies have compared the net joint moments of SLS to other BULLETs nor DLS in order to determine and differences and potentially categorize the observed magnitudes relative to one another.

2.2 Forward Lunge / Backward Lunge / Split Squat (FL / BL / SS)

The FL and BL tasks involve the same initial position of upright standing with feet hip- to shoulder-width apart. Movement is initiated with the stance-leg stepping and COM moving down and forward (FL), or the non-stance leg stepping and COM moving down and backward (BL). Similar to SLS, an attempt is made to control frontal plane knee motion and maintain a neutral spine throughout the duration of the exercise task. The transition phase, or bottom of the movement, occurs either when the back knee (nearly) contacts ground, and/or when the front knee reaches a specific joint angle, usually at or around 90 degrees (Graham, 2002; Keogh, 1999).

The SS slightly differs in initial position as it begins with feet displaced in the anterior-posterior direction at lunge step-length, with the stance leg being the anterior leg. This results in the front-leg hip being flexed and the rear-leg hip is extended, while an attempt is made to maintain the trunk in an upright position. However, both knees are fully extended and this is used as the cue for initial and final position being achieved. Feet remain in place while COM motion is initiated downward to the same transition phase as the lunge, followed by upward COM motion returning to the starting position (Schütz et al., 2014).
When step length for lunging patterns is standardized, there isn’t agreement among the chosen distance. Some studies use a set length of 1.0-1.2 x tibia length (Singh et al., 2015) or 65% ± 5% of leg length (Bouillon et al., 2012), while others use a set knee angle anywhere from 45 to 105 degrees (Schütz et al., 2014; Thijs, Van Tiggelen, Willems, De Clercq, & Witvrouw, 2007). However, when participants self-select step length, it results in a 10% distance decrease compared to the standardized lengths mentioned above (Riemann et al., 2013). While this could be due to many things such as hip extension ROM and trunk orientation, the shorter distance achieved during self-selection is interesting to note as a shorter stance is related to increased patellofemoral (PF) force and strain (Escamilla et al., 2008). This also suggests that the standardized step lengths typically used are perceived as too constraining since the shorter step is favourable when the performer self-selects. The effect of step length on knee angles remains unclear (Riemann et al., 2013; Schütz et al., 2014), and could be due to inter-individual variation in lower limb bone segment lengths or joint geometry. One study has shown an interaction between step length and knee angle, further complicating any standardization of step length (Schütz et al., 2014).

The FL and SS have previously been compared in a patellofemoral (PF) injury context, and no difference was found in PF force and stress between the two tasks (Escamilla et al., 2008). However, the researchers did suggest that a longer distance between foot position in the step lunge and split squat should be performed to decrease both the force and stress on the PF joint at greater knee angles. Contrary to the common belief that the knee should not move in front of the toes, there is no supporting evidence for this anecdotal guideline. Furthermore, the knee joint moments are greater in the rear limb during all variations of this task performed, and therefore the rear knee should be the one of consideration when high forces are contraindicated (Schütz et al., 2014).

Forward lunges are considered to be a hip extensor dominant exercise with the largest moments consistently occurring in the front hip (Riemann et al., 2013; Singh et al., 2015). The greatest range of motion occurs about the knee during the FL, but knee moment impulses (i.e., moment-time integrals) were the lowest among the three primary joints (Riemann et al., 2013). The addition of external load does not change the kinematics of the task (Riemann et al., 2013), however normalized joint and support moment are higher in obese individuals compared to normal body-mass index controls (Singh et al., 2015). This is most likely due to the increased
body mass per segment altering their mass-inertial properties, and relatedly due to the between-group differences in ground reaction force magnitudes, directions, and points of application.

Trunk position also modulates the biomechanics of lunging tasks. A more forward trunk angle increases hip extensor and plantar flexor moment-impulse and increases gluteus maximus and biceps femoris normalized muscle activation (i.e., percentage of maximum voluntary isometric contraction, %MVIC) when compared to an upright or extended posture (Farrokhi et al., 2008). This is similar to the SLS in which a more upright trunk position increases the forces and strain of the ACL and in that case more forward trunk lean is recommended (Kulas et al., 2012).

Very few studies have compared FL, BL, and/or SS to other BULLETs. Only one study has compared joint kinetics of the bodyweight DLS with a bodyweight FL (Singh et al., 2015). They found the squat had slightly higher knee moments while lunge had higher hip, ankle and support moments within individuals. An issue arises however as the authors do not indicate if they are averaging both legs or looking within a single limb during task performance, leaving the reader to speculate which case is true. To date, no study has compared net joint moments of lunge pattern variations such as the FL, BL, and SS.

### 2.3 Double-leg squat (DLS)

The DLS may be assumed to be identical in positions and motions to the SLS, aside from having both feet planted on the ground for the duration of the movement. It begins with an initial position of upright standing with feet flat on the ground and hip- to shoulder-width apart. The hip and knee joints are extended while the ankle joint remains in a neutral position. Movement is initiated by lowering the COM towards the ground, moving through flexion of the hip, knee, and dorsiflexion of the ankle. The transition point or bottom of movement occurs once a maximum controllable squat depth is reached. All three joints then subsequently extended/plantarflexed to produce upward COM motion. An attempt is made to maintain the thoracolumbar spine in a neutral position during the duration of the movement. The movement ends when the final position is reached which is identical to the initial position (Czaprowski, Biernat, & Kedra, 2012; Schoenfeld, 2010).
Due to the prevalence of acute and chronic knee injuries in physically active populations, research has focused on the biomechanics of the knee joint during DLS. Escamilla and colleagues have done much of this work (Escamilla, 2001; Escamilla, Fleisig, Lowry, Barrentine, & Andrews, 2001; Escamilla, Fleisig, Zheng, et al., 2001) and recommend that knee rehabilitation patients squat in the range between 0 and 50° knee flexion. This is due to knee forces being minimal within this range of motion. Tibiofemoral (TF) and PF compressive and shear forces progressively increase as the knee is flexed, reaching peak values near maximum knee flexion, and decreases as the knee is extended. Similar to the other BULLETs, knee extension moments increase in magnitude as squat depth increases (Cotter et al., 2013; Singh et al., 2015). While one study examined knee flexion moments during the first half of descent, they instructed subjects to descend as fast as they could (Dionisio, Almeida, Duarte, & Hirata, 2008). Their electromyography (EMG) data showed the quadriceps weren’t active during this time and therefore not resisting the external moments created by gravity. The hamstrings were active indicating a potential acceleration in addition to gravity (i.e., pulling COM down opposed to slowly resisting downward motion). The velocity of the movement in this case changed the characteristics of motion as previously demonstrated (Frost et al., 2013).

Other research looking specifically at muscle activation has shown EMG of the primary movers increasing concurrently with an increase in DLS depth (Escamilla, 2001) and external load (D. R. Clark, Lambert, & Hunter, 2012). Further, muscle activation has been shown not to change with common DLS variations such as modified stance width, hip rotation, and front barbell rack position does not significantly affect muscle activation (D. R. Clark et al., 2012). Muscle activation of DLS has been compared to SLS in women (McCurdy et al., 2010). SLS showed higher mean and peak EMG activity overall, while DLS had higher quadriceps EMG activity and higher quadriceps to hamstring ratios. However, no joint moments were compared between the two exercise tasks.

Kinetic and kinematic differences have been described looking at legs individually during DLS among populations suffering from various injuries. In ACL injured people, peak sagittal knee moments are 25% less in surgically repaired limbs, while the hip exhibits non-significant trend towards greater peak moment and the ankle moments remained unchanged (Salem et al., 2003). The movement strategy becomes more hip-dominant (i.e., larger hip moments and lower knee moments) to reduce the load placed upon the previously injured knee
The effect of external resistance on lower limb net joint moments during DLS has been given some attention in the literature. Greater body-mass normalized support moments were found in obese individuals versus non-obese healthy controls, suggesting the distribution and density of body mass can affect net joint moments (Singh et al., 2015). When relative amounts of external mass are added to a squat, the hip, knee, and ankle net joint moments show a linear response (S. P. Flanagan & Salem, 2008). The hip and ankle become more dominant as net joint moments increase, while the knee moment decreases as the external resistance is added. Moving the external mass forward into a front squat position resulted in a decrease in knee extension moments (Gullett et al., 2009). By utilizing a front squat position which involves holding the mass more forward, knee loads can be modulated while muscle recruitment remains unchanged.

Although the quantity is greater, the research on DLS is rather similar to that done on BULLETs. It primarily focuses on the knee joint and mitigating forces to decrease injury risk or support rehabilitation in addition to assessing activation levels of lower limb muscles. The DLS has the most literature quantifying net joint moments in various conditions, but has not been compared and contrasted to many of the BULLETs examined herein.

Looking across all BULLETs, most of the literature considering net joint moment magnitudes is comparing various external loading conditions. To date, no research has been done on comparing the net joint moments of four commonly performed BULLETs, therefore this study appears to be the first. The results garnered from this study will contribute new knowledge related to the characterization of joint kinetics of BULLET performance. This new knowledge will serve as valuable information used by practitioners who use these tasks in exercise prescription.

2.4 Current Application of BULLETs

Two primary applications of BULLETs exist among current exercise prescription by practitioners. The first is using them as a means of assessing potentially aberrant movement and injury risk in a clinical setting. The second is within an exercise prescription paradigm. Although these purposes are often interrelated (e.g., using movement assessments to prescribe exercises), they may also be used independently to maximize health, fitness and performance benefits of resistance training, or to identify injury risk factors that may be modifiable via
exercise (e.g., frontal plane knee motion). These applications will be explained in greater detail in the subsequent sections.

**2.4.1 Qualitative Movement Assessment**

Over the last decade, qualitative movement assessments have become increasingly popular among paramedical and exercise professional communities. Physiotherapists, chiropractors, strength coaches and personal trainers have adopted this type of tool and associated paradigm as they believe there is value in doing so for their targeted populations. The primary goal of a qualitative movement assessment is to infer whether or not movement strategies used during a movement task are “good” or “bad” based on visual observation of key features. The information is then typically used to drive an intervention strategy. A more simplified approach to this assessment is qualitative movement screening. Screening is essentially a green/red light system, triaging the individual allowing them to either proceed with training as is, or referring them forward for diagnosis or advanced assessment by a qualified practitioner. This is what makes it a beneficial tool, as it is rather quick and easy to implement on a large number of people in a confined space over a short period of time. It also requires minimal equipment and financial costs, making it widely accessible and implementable.

BULLETs have become commonplace in the majority of movement assessments’ battery of tasks. The use of BULLETs is not limited to the assessments subsequently listed, but these are the most utilized, published, researched and/or commercially available. While other and lesser known movement screens and assessments utilizing BULLETs could fit into this category, for the purposes of this document they will be left out and can be considered redundant in scope and practice.

The Movement Competency Screen (MCS), appears to be the most peer-reviewed and scientifically-driven movement assessment to date (Kritz, 2012). It is also the only movement assessment that makes use of three of the BULLET tasks: SLS; DLS; and FL. The Functional Movement Screen™ (FMS) and the National Academy of Sport Medicine (NASM) Corrective Exercise Training manual incorporates variants of the SS and DLS (M. A. Clark & Lucett, 2010; Cook et al., 2010), whereas The Foundation Matrix (TFM) uses variations of the SLS and SS to assess movement function (Mischiati et al., 2015).
Single task movement assessments such as the Y-Balance Test (YBT), and its precursor the Star Excursion Balance Test (SEBT), both use multiple variations of SLS to quantify distance reached by the elevated foot while qualitatively assessing the participants ability to balance or dynamically control their whole body during performance (Earl & Hertel, 2001; Ganesh, Chhabra, & Mrityunjay, 2015; Ness, Taylor, Haberl, Reuteman, & Borgert, 2015; Plisky et al., 2015). It is clearly demonstrated that the use of BULLETs is pervasive in contemporary qualitative movement assessment. As such, findings from this study could be valuable to the practitioners creating, performing and interpreting these assessments.

### 2.4.2 Exercise Prescription

BULLETs are a universally accessible means for able-bodied individuals to perform lower limb resistance training. They are taught and prescribed in many exercise programs to achieve health, fitness and performance related goals (Baechle & Earle, 2008; Bizzini et al., 2007; Tsatsouline, 2003). They are often prescribed as an entry point to exercise due to the unloaded nature of these tasks (i.e., no external resistance), and progressed by adding volume, external resistance, or movement speed. This is also why they are used in many rehabilitation programs as they can provide low level stimulus to individuals whose capacity/ability is diminished from injury.

Corrective exercise paradigms have also adopted the use of BULLETs (Bizzini et al., 2007; Cook et al., 2010; Soligard et al., 2008). One thought process of these programs is that aberrant movement strategies exhibited during the performance of BULLETs may be a cause or consequence of musculoskeletal injuries. The aforementioned movement assessments are used to elicit and detect these potentially dysfunctional movement strategies. Then using this exercise paradigm, they are improved or ‘corrected’ by getting the individual to practice a movement or exercise task while learning to control certain key kinematic features (Frost, Beach, Callaghan, & McGill, 2015) or use specific musculature through instruction and feedback (Willy & Davis, 2011). It is believed by the proponents of this model that this will teach the individual to coordinate and control their movements in a safer and more effective way, and that these adaptations will transfer to athletic, occupational, recreational, and other physical activities of daily living. While the efficacy of this approach is not established and is beyond the scope of this manuscript, BULLETs are being used daily by practitioners as suggested in this manner,
and knowledge of the joint loading magnitude could aid practitioners in decision making when prescribing these exercises for desired outcomes.

2.5 Justification for Methodology

2.5.1 Experimental conditions

The four BULLETs chosen as experimental conditions include the forward lunge (FL), backward lunge (BL), split squat (SS), and single-leg squat (SLS). While the individual tasks have been examined both kinetically and kinematically as described above, to date no direct biomechanical comparisons of net joint and support moments have been made between bodyweight versions of these tasks. The three lunge pattern tasks – FL, BL, and SS – were chosen because they are qualitatively similar exercise tasks with exception of the initial/final position and whole-body direction of movement (Figure 1). This often leads to these exercises being used interchangeably with one another in professional practice. It is of interest to describe the joint kinetics associated with the performance of different BULLETs, as it may inform practitioners and clinicians for use in many scenarios as previously mentioned.

SLS is the only single-limb support task in this experiment, as the others have the non-stance-leg in contact with the ground for support for at least a portion of the movement. Comparing how single- vs double-limb support changes joint loads among similar bodyweight exercise tasks is of interest to practitioners prescribing these exercises. Logic would suggest that using only one limb to support the same amount of mass would results in greater loading of that limb given similar movement speed and depth, but to what magnitude is unknown. Conversely, if joint moments of SLS are similar to other BULLETs, it might suggest that any perceived difference is due to the balance and control challenge of the task from having a smaller base of support. Additionally, if the tasks are similar, an individual who has trouble with balance could use one of the lunging tasks with bilateral support to achieve the comparable joint loading outcomes with less challenge. As with the lunge patterns, knowledge of the net joint moment magnitudes could be useful for the progression and regression of lower limb joint loads, as single-limb support is typically considered progression from bilateral support in daily practice.

The double-leg squat (DLS) was chosen as a baseline/comparison task. DLS is not considered a BULLET based on the definition provided above (i.e., it is a kinematically symmetrical task). As mentioned in the review of literature, given the lack of data concerning
the net joint moments of BULLETs, using DLS also provides previous data to validate results by comparing the measured values to determine if they fall within expected and/or realistic ranges.

Among the more applied reasons for using DLS in this study, it is one of the most commonly used exercise tasks in the abovementioned movement assessments. Understanding the magnitudes observed in a single limb of a symmetrically performed double limb movement will give more context to the joint loading demand of BULLETs, and how they might be used together or interchangeably to achieve outcomes of a movement assessment. Knowledge of how this task compares to other bodyweight exercises may give practitioners more options for increasing or decreasing loading demand. If the unilateral tasks are similar, yet different from the bilateral task, it would indicate that single- versus double-leg support characteristic is important in grouping exercise tasks based on kinetic characteristics of those tasks.

It is a common scenario that individuals injure one leg and not both. DLS exercises are also used in rehabilitation as a means of increasing support or dividing load amongst the uninjured limb. This more easily controlled and coordinated exercise task also allows the individual to progress external resistance with minimal change to the functional base of support. Conversely, since the relationship between ‘external’ (e.g., added mass) and ‘internal’ (e.g., net joint) loading is highly non-linear and difficult to estimate (Cholewicki, McGill, & Norman, 1991), this may be unnecessary as another BULLET could potentially provide the desirable increase in loading without the need for adding external load in certain cases. Finally, at the time of writing, the joint kinetics of a symmetrical support bodyweight exercise task have only been compared against a FL (Singh et al., 2015), but have yet to be compared to BL, SS and SLS.

### 2.5.2 Kinematic Variables

The kinematic performance of the exercise tasks were both instructed and demonstrated to the participant by the primary investigator to ensure consistency. Both methods of explaining the BULLETs were used to mimic standard daily practice in the exercise professional and paramedical clinical settings. Since this is commonplace in the real world, it was considered the most ecologically valid delivery model. While the author is aware that variation in both instruction and demonstration can affect movement outcomes (S. Khuu, Musalem, & Beach, 2015; Newell, Carlton, & Antoniou, 1990), these were both standardized and delivered by the
same expertly trained individual in order to control as best as possible (Appendix D). The goal of this study was not to evaluate the efficacy or manipulation of the exercise task description delivery, but to evaluate the biomechanics of these tasks given a consistent delivery model. Exercise technique instructions were chosen based on what could be found in the literature (Czaprowski et al., 2012; Graham, 2002; Keogh, 1999; Schoenfeld, 2010; Schütz et al., 2014). To fill in any gaps or discrepancies in the instructions, information from a pilot study (Chapman et al., 2014) and consultation with professional Certified Strength and Conditioning Coaches from the Canadian Sport Institute Ontario was used.

The kinematics of each task were measured in order to quantify and describe the resulting joint angles of the task being performed. This allowed for potential explanations for any sex-based differences in the kinematic performance of BULLETs, which could potentially drive any significant results in kinetic discrepancies. Furthermore, this would help with the interpretation of joint moment data, since it has been clearly demonstrated the depth and knee angle of the exercise performed can affect joint loading (Cotter et al., 2013; Scaglioni-Solano et al., 2005; Singh et al., 2015).

While encouraged to move through as large of a range of motion as possible during BULLET tasks (i.e., “squat as low as you can go”), the resulting vertical center of mass (COM) displacement was self-selected. This technique has been used elsewhere and instructs squat depth primarily for the reason of not forcing participants into a range past their comfort level and/or physical abilities (Dwyer et al., 2010). While standardizing a movement depth might seem like a control at first glance, individual anthropometrics (e.g., bony structural limitations) and muscle mechanical properties (e.g., muscle origins/insertions) are unlikely similar between individuals at a given depth of ROM, and therefore the sustained loading may still differ. Second, standardizing around a single joint angle for exercise tasks with varying kinematics could have added a joint constraint which could have affected movement ability in the other joints and potentially forced participants to move in an unnatural way. Furthermore, adding a depth control or target gave the participants an additional external focus of attention and feedback for each repetition, which could have influenced the performance of each subsequent repetition (Peh, Chow, & Davids, 2011). For these reasons, setting exercise depth in this case was considered a confounder opposed to a control. Further research is required to determine the
interrelatedness of these variables, find a balance between internal and ecological validity, and reconcile this problem for future research.

Considerations were given for lunging pattern step length for FL, BL and SS. First, there is no agreement on a standard step length as no studies use the same methodology (Bouillon et al., 2012; Riemann et al., 2013; Schütz et al., 2014; Singh et al., 2015; Thijs et al., 2007). This combined with the points previously mentioned, result in athletes self-selecting step length. When athletes self-select step range it has been shown to result in 10% less distance than suggested standardized lengths (Riemann et al., 2013). Similar to depth, a standard length may not be achievable for certain individuals due to restrictions such as rear hip extension ROM. Further, since evidence suggests shorter step length decrease patellofemoral force and strain (Escamilla et al., 2008), individuals may be optimizing for a decrease in these magnitudes. To further support this decision, one study that computed the same lower leg normalized joint moments during FL showed that changing step length does not change these magnitudes in the majority of cases (Singh et al., 2015).

COM motion was computed using a linked-segment model through measurement of individual segment motion of all tracked segments. This was the primary rationale behind a large, full-body marker set and kinematic measurement of both legs. The vertical and horizontal COM motion was used to determine the onset and completion of task repetitions during analysis, as discussed in detail the methods section below.

2.5.3 Kinetic Variables

Time histories of moments of force at a joint give valuable insight into the net effect of muscle activity acting upon that joint, and signify an integration of all the neural control acting at each joint (Winter, 1980). The net joint moment is a representation of the total outcome effect all the force-producing and -bearing structures surrounding a joint. Net joint moments are computed using inverse dynamical procedures combining anthropometric, kinematic, and ground reaction force data (D.G.E. Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2014).

Peak and average net joint moments were examined in the sagittal plane for the hip, knee and ankle, in addition to their summation to compute a total support moment. From an exercise perspective, this is where the largest range of motion occurs during these tasks. Given this,
BULLETS are used in current exercise paradigms to impose a loading stimulus on the sagittal plane flexion/extension musculature about these joints. Whether for specific physiological adaptations to that musculature (e.g., hypertrophy of leg extensors, rehabilitation of knee musculature after ACL surgery), or increasing performance of a related task performance (e.g., lunge in a tennis shot, climbing stairs), there are many reasons why these exercise tasks are chosen for sagittal plane driven adaptations.

Peak net joint moments were considered to represent a crude measure of the net maximum instantaneous biomechanical demand imposed on these joints during the exercise tasks. Repeated exposures to submaximal load or “cumulative load” can also cause an anatomical structures to positively or negatively adapt (McGill, 1997). In addition to knowing the single maximal load imposed, average net joint moments were used to examine the difference in loading over the duration of a single repetition. This represented an estimate of the total net biomechanical load imposed on a single lower limb joint during the performance of BULLET tasks. It must be noted that the flight phase of the forward lunge was not included, and time normalization was used from 0 to 100% of time using force onset and offset.

In addition to the kinetic variables above, a support moment was computed through algebraic summation of the hip, knee and ankle net joint moments, and the peak and average values were analyzed. A support moment can be considered a gross representation of the total lower limb demand during a ground-based, free-standing movement task (S. Flanagan & Salem, 2005). Since practitioners may want to consider or target the whole leg opposed to a single joint during assessment or prescription, this variable would be useful in further characterizing each BULLET for this purpose. Additionally, a support moment could be of greater value to practitioners as it gives a quantifiable representation of “triple extension”, which is commonly used both as a feedback tool (i.e., verbal cue, attentional focus) and a transfer principle in exercise intervention.

There are limitations when using a net joint moment for the abovementioned purposes (Bryanton & Chiu, 2014). First, the net joint moment cannot discern the gross quantity of agonist and antagonistic activity. It is the net difference between the two groups of actuators and dictates which direction the limb will have tendency to accelerate around the joint. It therefore represents the minimum muscular effort required to accelerate a joint assuming no antagonistic
contraction, or it will underestimate the actual muscular effort if antagonist contraction is present. Further, it also does not adequately represent biarticular muscles, such as the quadriceps and hamstrings which can produce moments of force around both the hip and knee joint. Additional muscle modeling (e.g., single equivalent muscle model, EMG-assisted, optimization-assisted) is required to partition the net joint moment amongst all force-generating and -bearing structures (Dowling, 1997; Reeves & Cholewicki, 2003). Therefore, practitioners are not able to make muscle force inferences via visual observation because many more assumptions are required beyond those necessary to make inferences about net joint moments (Herbert et al., 1993). This makes the net joint moment magnitudes valuable both literally and conceptually for the practitioner to understand and consider when prescribing exercise. Knowledge of what one can and can’t infer with respect to exercise kinetics brings greater validity to prescription, potentially enhancing the desired outcomes.

Kinetic data of this nature comes with another caveat, as various normalization methods are commonly applied when biomechanical variables are examined (Moisio et al., 2003). This is typically done to account for variation in body size and mass between subjects as mass-inertial properties play a role in driving the computations of joint moments. In contrast, it has been shown that normalization to body mass and/or size can change the relationship between variables, and may not be recommended for widespread use as it currently is (Davidson, Myers, Shelburne, Van Lieshout, & Curran-Everett, 2014). Further, some metrics better discriminate populations when left absolute, such as external adduction moments differentiating severity of knee osteoarthritis (Robbins, Birmingham, Maly, Chesworth, & Giffin, 2011). However, as per the secondary objective to determine if the net joint moment response is different between men and women, body mass and size need to be accounted for in order to remove it as a confounding factor. Further, normalization is much more commonly used in the related literature, and rationale provide support for the use of normalization in this case. Body mass × leg length was used to normalize all peak and average joint moment data, as all kinetic analyses were constrained to the lower limb.

2.5.4 Sex Differences

Evidence shows that men and women perform lower limb movement tasks differently when considering both kinetic and kinematic variables. Thus, it could not be assumed that the men and women in this study would have shown the same results with respect to net joint moment.
magnitudes. For this reason, “sex” was included as a factor in the statistical model, even after net joint moment magnitudes were normalized as described above.

In healthy women compared to men, less trunk flexion, greater trunk rotation towards the stance limb, greater hip adduction and greater knee adduction occur when performing a SLS (Graci et al., 2012). When attempting to move through the maximum range of motion while performing SLS and lunges, women exhibit larger knee flexion, knee valgus, ankle dorsiflexion and ankle pronation angles with smaller hip flexion angles (Dwyer et al., 2010; Zeller et al., 2003). They also show significantly different muscle activation patterns, specifically greater normalized muscle activation (%MVIC) in gluteus maximus, rectus femoris, and trunk extensors compared to men for the same tasks (Bolga et al., 2014; Zeller et al., 2003). In contrast, one study did show no difference between %MVIC in eight muscles during FL and SLS (Bouillon et al., 2012), bringing contention into whether or not true sex based differences exist.

Comparable trends of biomechanical differences between sexes have been shown in the performance of similar yet higher demand tasks such as cutting (Beaulieu, Lamontagne, & Xu, 2009; Kevin R. Ford, Myer, Toms, & Hewett, 2005) and both double-leg (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003; Sigward, Pollard, & Powers, 2012) and single-leg (K. R. Ford et al., 2006) drop landings. While the aforementioned studies show these kinematic and muscle activation trends are similar across all lower limb tasks regardless of the level of demand, the kinetic findings of the high demand tasks similarly occurring during low demand BULLETs have yet to be examined.

The collective evidence would suggest that men and women use different movement strategies when performing BULLETs. While none of the studies mentioned give a determinate mechanism for these differences, anatomical proportions and geometries do differ between men and women (Sizer & James, 2014). Men are on average bigger, and women carry a larger proportion of their mass in their lower body. It is possible the differences observed in biomechanical variables are largely due to the fact that men are larger than women. This warrants the use of body-size normalized values as previously mentioned, since it removes it as a confounding factor to determine if a true sex-based difference exists.
Women also have a wider pelvis which in some cases increases the angle that the femur sits in the frontal plane relative to the pelvis and the tibia (i.e., Q-angle). Therefore, joint angles could be inherently different due to sex-based anatomy, potentially altering kinematics of performance between men and women. Conversely, it is possible that these results are due to a lack of standardized instruction or demonstration among the literature published to date. By giving all subjects standardized instructions, and by quantifying the kinematics of BULLET performance, it can be determined if these are possibly contributing factors to any sex-based differences in joint kinetics observed. By analyzing sex-based differences using body-size normalized net joint moments, this could inform the practitioner if sex-based considerations should be taken when assessing or prescribing BULLETs and to what magnitudes those differences might be. Finally, sex-based differences in normalized net joint moments have yet to be examined for any BULLET tasks.
3 Methods

3.1 Study Design

Mean and peak lower limb net joint extension (hip and knee) and plantarflexion (ankle) moments, and their algebraic summation (i.e., total support moment) were compared within participants between four common BULLETs and a baseline non-BULLET task. This design allowed for anthropometrics and motor abilities of the participants to be controlled between experimental conditions. Body segment motion and ground reaction force data were collected while participants performed the 5 exercise tasks: single-leg squat (SLS); split squat (SS); forward lunge (FL); backward lunge (BL); and double-leg squat (DLS) (Figure 1). An inverse dynamical linked-segment model was used to generate dependent variables of interest, and analyses of variance (ANOVA) with one within-participant (TASK) and one between-participant (SEX) factor was used to make the comparisons needed to address the study objectives. Funding for this study was provided by the Canadian Sport Institute Ontario, and it was reviewed and approved by the University of Toronto Office of Research Ethics prior to commencement.

3.2 Participants

A convenience sample of 32 recreationally active students (16 men and 16 women) from the University of Toronto population were recruited (Table 1). The recruitment process involved placing posters around the University of Toronto campus and delivering presentations during undergraduate lectures (Appendix A). For inclusion, participants indicated they were at the time – and for the three months preceding the study – injury- and pain-free (i.e., injury diagnosed by a medical profession or suspected/believed by the individual), as well as absent of any existing medical conditions that may be exacerbated by study participation. Additionally, for the same three months preceding the study, they were required to recreationally exercise at a minimum of three times per week for inclusion to ensure that exercise was not a novel stimulus and the chances of injury were minimized.
Participants were required to fill out an informed consent document (Appendix A) and a Physical Activity Readiness Questionnaire (PAR-Q; Appendix B). If they did not consent, or answered “yes” to any of the questions on the PAR-Q, they were excluded from the study. They were also asked to report their age, in addition to being asked if they agreed to permit video and/or pictures to be taken during their session for use in dissemination and education. Following consent, participants were assigned an alphanumeric code in order to depersonalize data, in addition to the option of allowing their depersonalized data available for use in future studies.

Individuals were excluded if they experienced any pain during the familiarization or data collection and they were able to withdraw from participation at any point during the study. No participants were excluded nor did any withdraw from the study. If the participant completed 60 minutes of the data collection session, they received $20 remuneration for their time. If the participant withdrew after this point they retained remuneration.

### 3.3 Procedures

Participants attended one 60-minute testing session in the Musculoskeletal Biomechanics and Injury Prevention Laboratory on the Downtown (St. George) Campus of the University of Toronto. Upon participants’ arrival at the laboratory, the primary investigator described all study objectives and testing procedures, presented informed consent (Appendix A), PAR-Q documentation (Appendix B), and answered any questions posed by participants. After giving consent, the participants changed into the requested attire and were assessed for any reflective material that may have delivered false marker trajectories. Participants were asked to bring running shoes and spandex shorts to data collection, and women were also asked to bring a sports brassiere while men were shirtless. This attire was necessary to attach reflective markers directly to the skin in order to minimize motion artifact.

<table>
<thead>
<tr>
<th></th>
<th>N</th>
<th>Age (yrs)</th>
<th>Mass (kg)</th>
<th>Height (m)</th>
<th>Leg length (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Men</td>
<td>16</td>
<td>23.7 (3.5)</td>
<td>73.8 (11.1)*</td>
<td>1.79 (0.08)*</td>
<td>0.92 (0.05)*</td>
</tr>
<tr>
<td>Women</td>
<td>16</td>
<td>22.8 (5.6)</td>
<td>64.6 (9.3)*</td>
<td>1.70 (0.07)*</td>
<td>0.87 (0.05)*</td>
</tr>
</tbody>
</table>

* denotes that values were statistically different ($p < 0.05$) between men and women assessed via t-tests.
Leg dominance was assessed in order to determine the “stance leg” for BULLETs, with the participant self-reporting followed by 5 in-place unilateral hops on each leg to determine which leg the participant felt more control, comfort, and dominant jumping off of. In addition to being used as a control, this was primarily due to the fact that only one in-ground force platform was available for this study, so the stance leg for each BULLET was selected for kinetic analysis and placed upon the force platform during data collection trials. Once leg dominance was determined, participants were led through a standardized warm-up consisting of 5 repetitions each of the four BULLETs and the baseline task (Figure 1; Appendix C).

The warm up also served as exercise familiarization, so the participants were able to get comfortable with all the movement tasks prior to data collection. The primary investigator demonstrated the movements tasks required and gave a standardized verbal description of foot placement, initial/final and transition positions, and direction of movement as described in the aforementioned literature review and Appendix D. The instructions and demonstration were given to ensure participants understood the exercise tasks and were able perform the movement sufficiently, safely and in control. Exercise technique instructions were chosen based on what could be found in the literature (Czaprowski et al., 2012; Graham, 2002; Keogh, 1999; Schoenfeld, 2010; Schütz et al., 2014), in addition to consultation with professional Certified Strength and Conditioning Coaches (NSCA) at the Canadian Sport Institute Ontario. Furthermore, since instruction and demonstration is standard practice in the exercise coaching and paramedical clinical settings, it was deemed the most authentic and ecologically valid delivery model. No additional information or augmented feedback concerning BULLET technique was given after the initial demonstration and instruction if performed correctly in order to avoid the effect of coaching on the participants’ performance of the tasks. If, during familiarization, the participant was not performing the tasks as directed, the primary investigator reiterated the parts of the instruction that were not complied with, and demonstration was repeated if necessary. Care was taken to ensure the participant did not make stepping movements (i.e., forward lunge and backward lunge) with a two-phase step then lower. Repetitions needed to be performed in a smooth (i.e., controlled) and continuous manner to be acceptable. With respect to ROM, participants were instructed on all tasks to “move as low as you can go while maintaining full control of your body”. Hands were placed on the hips for all trials, fixated to the pelvis in order to minimize upper limb influence on kinematics and COM motion.
During the familiarization and subsequent data collection, each of the five tasks were performed for five repetitions, as a minimum of three are necessary in order to increase the reliability of biomechanical measurements such as joint moment data (Alkjaer, Henriksen, Dyhre-Poulsen, & Simonsen, 2009), while accounting for some of the inter-repetition variability in performance. Exercise tempo was set with a digital metronome using a two-second eccentric, two-second concentric tempo in order to best control for movement speed through standard exercise prescription methodology. These durations were selected using data from an unpublished pilot study wherein participants elected to perform the SLS at this tempo when permitted to self-select movement speed (Chapman et al., 2014). This is equivalent to, or slightly faster that what is shown elsewhere for a similar population (Graci et al., 2012; Riemann et al., 2013). If the participant missed a repetition or started the task off-tempo, another repetition was added to the end of the set to ensure five completed repetitions were recorded at the desired tempo. A two second pause was taken in between the completion of a repetition and the initiation of the subsequent one to control inter-repetition rest time. The

![Figure 1: Visual 3D™ screen captures of linked segment models of the four BULLETs and double-leg squat.](image)
order of the tasks was randomized between participants to prevent an order effect, but all repetitions of each task were blocked together within participants to streamline the data collection. A two minute break was given between each task block in order to avoid the accumulation of acute fatigue. This duration was selected using ATP-CP recovery timelines (Dawson et al., 1997), and was supported in pilot testing wherein participants’ ratings of perceived exertion always returned to a 1 out of 10 on the BORG CR-10 scale (Buckley & Borg, 2011), regardless of the task performed.

3.4 Data Acquisition

Body segment kinematics were tracked using 14 mm retroreflective markers secured to participants by 2.54 cm non-allergenic double-sided tape (3M Canada Corporate, London, ON, CA). Markers were attached following familiarization and prior to data collection. The marker set consisted of 56 calibration, tracking and dual-purpose (i.e., calibration and tracking) markers attached directly to skin overlying anatomical landmarks (Figure 2). A full body marker set was used, excluding arms given they were fixed, in order to determine COM motion for event detections as subsequently described.

The primary investigator, who has previous training as an athletic therapist, performed all landmarking via manual palpation for consistency across participants. Three-dimensional marker trajectories were sampled at a rate of 200 Hz using eight Qualisys Oqus 1 cameras (Qualisys AB, Gothenburg, Sweden). The motion capture system was calibrated using a 300.1 mm wand and L-frame (Qualisys AB, Gothenburg, Sweden) using the factory standard protocol with the maximum allowable standard deviation of marker displacement set at 0.80 mm. The calibration L-Frame was placed in the lower left hand corner of the force plate in order to create the global coordinate system (GCS). International Society of Biomechanics (ISB) convention with x-axis forward and y-axis up using a right-handed coordinate system was adopted for the GCS (Wu & Cavanagh, 1995). Following calibration, spatial synchronization of the force plate within the GCS was achieved using an MTD-2 CalTester Rod (Motion Lab Systems, Inc., Baton Rouge, LA) via CalTesterPlus software (C-Motion, Inc., Germantown, MD) and protocols (Holden, Selbie, & Stanhope, 2003). Ground reaction forces were sampled at rate of 2000 Hz using one in-ground force plate (BP600900, Advanced Mechanical Technology Inc., Watertown, Massachusetts). Force plate data was temporally synchronized with marker position data via a ±10V, 16-bit analog-to-digital conversion system (Analog Acquisition Interface Unit,
Qualisys AB, Gothenburg, Sweden). The kinetic and kinematic data was temporally synchronized, digitized, and stored for post-processing using Qualisys Track Manager software (version 2.9, Qualisys AB, Gothenburg, Sweden).

Figure 2: (Top) Retroreflective marker placement for data collection. Colour of marker denotes purpose: red = segment calibration; yellow = segment tracking; green = dual purpose segment calibration and tracking. (Bottom) Image of participant performing a BULLET task with markers attached.
For each participant, a static calibration trial was conducted with the participant standing motionless in an anatomically neutral position, facing the positive x-axis in the GCS. Feet were aligned parallel to the lateral edge of the force plate, and aligned facing anterior. Following this trial, the 10 calibration-only markers were removed. Subsequently, a series of standardized hip and knee motion trials were recorded that were used to compute joint center locations and axes of rotation using functional methods (Begon, Monnet, & Lacouture, 2007; Schache, Baker, & Lamoreux, 2006). The functional joint trails consisted of ten repetitions of knee flexion/extension and hip abduction/adduction/circumduction on the left and right legs individually, while keeping the rest of the body as motionless as possible.

### 3.5 Data Processing and Analysis

All signal conditioning, model-building procedures, and calculations of dependent variables were performed using Visual3D™ software (Version 5, C-Motion, Inc., Germantown, MD). Three-dimensional kinematic and kinetic data were used to construct a “bottom-up”, 8-segment, 6 degree-of-freedom, inverse dynamical linked-segment model (LSM) of the human body. The midpoint between the medial and lateral static calibration markers at the proximal and distal ends of each segment were used to define segment endpoints. Segment-fixed coordinate systems (SCS) were initially defined using the proximal segment endpoints. Segment-fixed coordinate systems (SCS) were initially defined using the proximal segment endpoints. Subsequently, hip and knee joint centers along with knee flexion/extension axes were computed using the functional joint trials with the default algorithms in Visual3D™ (Schwartz & Rozumalski, 2005). Virtual markers were then created using the functionally derived joint and axes modified from the original calibration markers in order to adjust the SCS based on individual motion characteristics. Ankle joint centers were defined as the midpoint between medial and lateral malleoli markers. Two virtual landmarks were created as the midpoint between the left and right anterior superior iliac spines (ASISmid) and the left and right posterior superior iliac spines (PSISmid) respectively. These virtual landmarks were then used to calculate the depth of the pelvis segment. Using a similar procedure, the xiphoid (XYPH) and 10th thoracic spinous process (T10) markers were used to calculate the depth of the trunk segment. Participant height and leg length were computed using the distance from force plate to the HEAD marker and to the proximal endpoint of the stance leg thigh segment respectively. Participant mass was calculated from the measured weight during the static calibration trial. Body segment parameters (i.e., segment mass-inertial properties) used during inverse dynamics procedures
were defined using the default regression and geometric models within Visual 3D™ inputting participant height, mass, and depth of the pelvis and trunk (Dempster, 1955; Hanavan, 1964).

A zero-lag, fourth order low-pass Butterworth filter at a cut-off frequency of 10 Hz was used to smooth the marker position data to minimize high-frequency noise from motion artifact when position data is double-differentiated (Pezzack, Norman, & Winter, 1977). Kinetic signals were also processed through this filter at a matched cut-off in order to minimize errors propagating in the joint moment calculations (Bisseling & Hof, 2006; Kristianslund, Krosshaug, & van den Bogert, 2012, 2013). This was further justified by the slow movement velocity and lack of impact during exercise tasks (e.g., drop landing).

The LSM was used to generate time histories of lower limb net joint moments, angular displacements, and trunk segment angular displacements using methods described elsewhere (Beach, Frost, Clark, Maly, & Callaghan, 2014). Joint moments were calculated using the Newton/Euler equations of motion and resolved in the distal segment-fixed coordinate system (D.G.E. Robertson et al., 2014; Winter, 1991). The calculations started at the foot to quantify the ankle joint reaction kinetics, and then proceeded proximal through the shank and the thigh to produce the knee and hip joint reaction kinetics respectively.

In addition, the net extension (hip and knee) and plantarflexion (ankle) moments were summed at each instant to yield support moment time histories. The support moment was included herein as a crude measure of the overall lower limb muscular demand required to simultaneously prevent “collapse” and to accelerate the COM during task performance (Winter, 1980). The potential utility of this is that practitioners may consider the lower limb in totality when designing and administering assessments or exercise interventions, especially in cases where between-BULLET differences in hip, knee, and ankle joint moments vary in a complex way.

Ensemble averages of the data derived from five repetitions of each task were created for each subject in order to increase stability of the variables (Alkjaer et al., 2009). The beginning and end of each repetition were first defined using automated event detection algorithms in Visual3D™, then manually assessed and adjusted if necessary. These endpoints (ON, OFF) were defined based on COM motions. DLS, SLS and SS solely used vertical COM motion (Figure 3a), while BL used vertical and horizontal COM motion to account for the backwards
stepping motion (Figure 3b). Since the stance leg during FL initiates using a flight phase, this task used the onset and conclusion of a ground reaction force to determine when the stance leg was in contact with the ground (Figure 3c). While this created a temporal discrepancy between this task and the others, it was accounted for by rubber-banding (time-normalizing) the data, which ultimately enabled dependent variables to be compared during the contact phase of each BULLET. Rubber-banding also accounted for any minor repetition-to-repetition variations in tempo. From the ensemble averages, the peak and average net extension (hip and knee), plantarflexion (ankle), and support moment were calculated in absolute units (N·m), extracted, then body-size normalized using participant’s body mass × leg length (N·m/kg·m) before being included in statistical analyses. Although such normalization procedures can distort data distributions (Davidson et al., 2014; Robbins et al., 2011), it has also shown to account for the variance in biomechanical quantities due to height and mass (Moisio et al., 2003). Further, this method was used to account for size differences before investigating if sex-based differences existed in the joint moments.

Figure 3a: Visual 3D™ screen captures denoting event detections ON and OFF for Double-Leg Squat (DLS), Single-Leg Squat (SLS), and Split Squat (SS) using processed kinematic signals.
Figure 3b: Visual 3D™ screen captures denoting event detections ON and OFF for Backward Lunge (FL) using processed kinematic signals.

Figure 3c: Visual 3D™ screen captures denoting event detections ON and OFF for forward lunge (FL) using processed kinetic signals. Force onset as shown on left by first captured frame detecting a force vector (blue arrow).

Although not the primary focus of this study, time histories of trunk segment, hip, knee, and ankle joint angles were also quantified in order to help interpret any differences in the dependent variables of interest. Since anthropometrics were controlled using within-participant study design, and between sexes using body-size normalization, differences or similarities in kinematic performance between BULLETs and between sexes could provide rationale for the
results observed. Joint angles were computed as the orientation of the distal SCS with respect to that of the adjacent proximal segment (Woltring, 1991) via a Cardan/Euler rotation sequence (Davis et. al., 1991; Engstrom et. al., 1989). Using an equivalent decomposition method, trunk segment angle was defined as orientation of the thorax relative to the GCS. Peak and ROM (peak-to-peak) angles about forward/backward inclination (trunk), flexion/extension (hip and knee) and dorsiflexion/plantarflexion (ankle) axes were extracted for analyses.

### 3.6 Statistics

Statistical analyses were performed using SPSS statistics software (Version 21, IBM, Markham, Canada). Dependent variables were compared via analyses of variance (ANOVA) containing one within-participant factor (TASK) and one between-participant factor (SEX). TASK contained five levels representing each exercise task condition (DLS, SLS, SS, BL, FL) and SEX contained two levels representing the participant being men or women (MEN, WOMEN).

For an ANOVA containing a single within-between interaction, a sample size of 22 was deemed sufficient to detect a small effect size ($f^2 > 0.25$), with an alpha ($\alpha$) level of 0.05, and power (1-$\beta$) of 0.80 (G*Power V3.1.9.2, Dusseldorf, Germany). Data were gathered from 32 participants in order to ensure we were above our desired power number while accounting for error in a priori calculations and any data losses not detected during collection sessions.

Screening for statistical outliers was conducted via visual inspection of boxplots (SD > ±3.0) and normality was assessed using visual inspection of histograms and Q-Q plots. Homogeneity of variances and covariances were assess using Levene’s test ($p < 0.05$) and Box’s M test ($p < 0.001$), respectively. Assumptions of sphericity were assessed using Mauchly’s test ($p < 0.05$), and if the assumption was violated, a Greenhouse-Geisser correction was used to adjust the degrees of freedom for the two-way ANOVA. If an interaction was found between SEX and TASK, data were separated by SEX and an ANOVA containing one within-participant factor (TASK) was conducted. Tukey’s HSD post-hoc tests were run to interpret any statistically significant main effects. Significance was set at $\alpha < 0.05$ for interaction and main effects, and $p < 0.05$ was considered to be statistically significant.
4 Results

4.1 Global Findings

Prior to running the two-way mixed ANOVAs, outliers in both SS and BL conditions were detected in one participant’s peak support moment values. However, these values were within an expected range across conditions and removal would have violated key assumptions of the ANOVA. Further, removal of the outlier did not change statistical significance for any interaction or main effects when processed without the data, so data points were retained for final analysis. Data for all conditions were approximately normally distributed as assessed by visual inspection of histograms and normal Q-Q plots.

Upon examination of results across all dependent variables, some consistent findings arose for the peak and average normalized hip, knee, ankle and support moments (Figures 4-7). The results of two-way mixed ANOVAs showed that no interactions occurred between TASK and SEX ($p > 0.05$), while a main effect for TASK was present for all cases ($p < 0.05$). A main effect for SEX was only found in normalized peak and average support moment, and normalized peak knee moment ($p < 0.05$). In these cases, data from men were always of significantly greater magnitudes than those from women, while no sex-based differences were observed in all other cases ($p > 0.05$).

Analyses of the main effects for TASK using post-hoc testing revealed similar trends across dependent variables (Figures 4-7). Performing the DLS always resulted in the smallest normalized net joint moment magnitude when compared to all BULLETs ($p < 0.05$). For both peak and average normalized knee, ankle, and total support moments, a general trend was found for task specific magnitudes when rank ordered from smallest to largest: DLS < SS ≤ BL ≤ FL < SLS. The primary difference between dependent variables was the BL task, which in every case except average ankle moment was grouped with either the SS task or the FL task ($p > 0.05$). Despite the similarities observed in the lunge pattern tasks, SS and FL were significantly different in all cases ($p < 0.05$). The hip was the only joint examined which did not follow the aforementioned joint moment magnitude trend, ordering from smallest to largest: DLS < SLS ≤ SS ≤ BL ≤ FL.
Ancillary kinematic results for the hip, knee, ankle and trunk shown in Appendix E. When looking across all joints examined for both peak flexion angles and total range of motion exhibited, no two BULLETs were deemed identical. Furthermore, given standardized instructions, both men and women performed the BULLET movements using similar kinematics, as no SEX * TASK interaction, nor a main effect for SEX was observed for all joint angles examined ($p > 0.05$).

Below, results of the comparisons in joint moment data are presented in detail. To facilitate more direct comparisons between the data collected in this thesis and those reported elsewhere, additional data are provided in Appendix F (i.e., joint moment magnitudes normalized with respect to body mass). While rationale is provided in the Review of Literature and Methods sections to support the method of normalization used to answer the research questions, many other studies use the method of joint moment normalization reported in Appendix F.

### 4.2 Support Moment

All descriptive results in subsequent sections are presented as mean ± standard deviation for body size-normalized values for moment magnitudes (units = N·m/kg·m). Beginning with the overall measure of lower limb loading, normalized support moment results of BULLETs were similar for both peak and average values (Figure 4). The normalized peak support moment for all subjects in the DLS was $2.00 ±0.28$. The SS ($2.83 ±0.42$) and BL ($2.84 ±0.46$) were both 43% greater than in DLS, while FL ($3.21 ±0.42$) and SLS ($3.56 ±0.60$) were 61% and 79% greater, respectively. The mean normalized peak support moment across all BULLETs was $2.74 ±0.60$ for women and $3.03 ±0.73$ for men.

For peak support moment, no interaction was observed for TASK * SEX, $F(2.068, 62.035) = 0.981$, $p = 0.383$, partial $\eta^2 = 0.032$, but there was a main effect for TASK, $F(2.068, 62.035) = 90.882$, $p < 0.001$, partial $\eta^2 = 0.752$. A main effect for SEX, $F(1, 30) = 7.907$, $p = 0.009$, partial $\eta^2 = 0.209$, was also observed. The normalized peak support moment had both homogeneity of variances ($p > 0.05$) and covariances ($p = 0.721$). However, Mauchly’s test indicated the assumption of sphericity was violated, $\chi^2(9) = 46.265$, $p = <0.001$; therefore, a Greenhouse-Geisser adjustment was used $\varepsilon = 0.517$. 
The normalized average support moment for all subjects in the DLS task was 1.14 ±0.15. The SS (1.60 ±0.26) and BL (1.63 ±0.29) were 40% and 43% greater than DLS, while FL (2.24 ±0.32) and SLS (2.19 ±0.28) were 96% and 93% greater, respectively. The normalized average support moment for all BULLETs was 1.68 ±0.47 for women and 1.84 ±0.49 for men.

For average support moment, no interaction was observed for TASK * SEX, $F(3.077, 92.299) = 0.684, p = 0.568$, partial $\eta^2 = 0.022$, but there was a main effect for TASK, $F(3.077, 92.299) = 172.588, p < 0.001$, partial $\eta^2 = 0.852$, and for SEX, $F(1, 30) = 6.947, p = 0.013$, partial $\eta^2 = 0.188$. Normalized average support moment had homogeneity of variances ($p > 0.05$) and covariances ($p = .281$). Mauchly’s test indicated a violation of sphericity $\chi^2(9) = 17.756, p = 0.038$, and therefore the Greenhouse-Geisser adjustment was used, $\varepsilon = 0.769$.

Using a post-hoc test to determine where significant differences for TASK occurred, a slightly different trend existed for peak support moment across TASK conditions (Figure 4). SLS was significantly different than FL ($p = .004$) in peak magnitudes, whereas no significant difference occurred for these two conditions in average support moment magnitudes ($p = 0.501$). The peak support moment had a magnitude rank-order trend of DLS < SS ≤ BL < FL < SLS ($p < 0.05$), while the average support moment trend was DLS < SS ≤ BL < FL ≤ SLS ($p < 0.05$).

**Figure 4:** (Left) Peak and (Right) average body-size normalized support moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters above bars ($p > 0.05$), while different letters indicate BULLET tasks were statistically different ($p < 0.05$). Men were significantly greater than women across all TASK conditions for peak ($p = 0.009$) and average ($p = 0.013$) normalized support moment magnitudes.
4.3 Hip Moment

Considering BULLET performance by individual lower limb joints beginning proximally, the results for normalized peak and average hip net joint moment magnitudes are presented in Figure 5. The normalized peak hip moment for the DLS task was -0.70 ±0.14. The SLS (-1.00 ±0.44) and SS (-1.14 ±0.25) were 42% and 61% greater than DLS, while BL (-1.35 ±0.30) and FL (-1.35 ±0.27) were both 92% greater than this baseline value. The mean normalized peak hip moment for women was -1.06 ±0.40 and for men was -1.15 ±0.36.

For peak hip moment, no interaction was observed for TASK * SEX, $F (2.229, 66.872) = 1.930, p = 0.148$, partial $\eta^2 = 0.060$. A main effect was observed for TASK, $F (2.229, 66.872) = 54.302, p < 0.001$, partial $\eta^2 = 0.644$, while no significant differences existed for SEX, $F (1, 30) = 1.321, p = 0.259$, partial $\eta^2 = 0.042$. Normalized peak hip moment had homogeneity of variances ($p > 0.05$) and covariances ($p = 0.023$). Mauchly’s test indicated the assumption of sphericity was violated, $\chi^2 (9) = 42.671, p < 0.001$; therefore, a Greenhouse-Geisser adjustment was used, $\varepsilon = 0.557$.

The normalized average hip moment for the DLS task was -0.35 ±0.09. The SLS (-0.55 ±0.23), SS (-0.69 ±0.20), BL (-0.74 ±0.17), and FL (-0.93 ±0.20) were 54%, 95%, 108%, and 162% greater than DLS, respectively. Women’s mean values were -0.63 ±0.29 while men were -0.67 ±0.24 for normalized average hip net joint moments.

No interaction was observed for TASK * SEX, $F (3.079, 92.370) = 1.532, p = 0.211$, partial $\eta^2 = 0.049$, while a main effect was found for TASK, $F (3.079, 92.370) = 80.134, p < 0.001$, partial $\eta^2 = 0.738$. Similar to peak values, no significant differences occurred for SEX, $F (1, 30) = 0.639, p = 0.430$, partial $\eta^2 = 0.021$. Normalized average hip moments also had homogeneity of variances ($p > 0.05$), covariances ($p = 0.440$), and a violation of sphericity $\chi^2 (9) = 17.279, p = 0.045$; therefore, the Greenhouse-Geisser adjustment was used, $\varepsilon = 0.0770$.

While TASK showed similar trends in magnitudes for normalized hip net joint moments across BULLETs for men and women (Figure 5), post-hoc testing resulted in different groupings of BULLETs between the peak and average values. For peak magnitudes, SLS and SS were statistically similar ($p = 0.071$), as were BL and FL tasks ($p = 0.921$). However, for average hip net joint moments, SS and BL were the only tasks deemed similar ($p = 0.104$).
peak hip moment had a different magnitude rank-order trend than the rest of the joints: DLS < SLS ≤ SS < BL < FL ≤ SLS (p < 0.05). The BULLETs grouped slightly differently in the average hip moment with a trend of: DLS < SLS < SS ≤ BL < FL (p < 0.05).

Figure 5: (Left) Peak and (Right) average body-size normalized hip net joint moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) are plotted; error bars represent the standard deviation. *Post-hoc* results for TASK show no statistical difference between BULLETs with the same alphabetical letters above bars (p > 0.05), while different letters indicate BULLET tasks were statistically different (p < 0.05). No significant difference were observed between men and women across all conditions for peak and average values (p > 0.05).

4.4 Knee Moment

Moving distally on the lower limb, the results for normalized peak and average knee net joint moment magnitudes are presented in Figure 6. The normalized peak knee moment for the DLS task was 1.10 ±0.20. Lunge tasks FL (1.30 ±0.25) and BL (1.29 ±0.26) were both 18% greater than DLS, while SS (1.18 ±0.19) and SLS (1.70 ±0.30) were 8% and 55% greater, respectively. Women’s and men’s mean values for peak moments across BULLETs were 1.23 ±0.29 and 1.39 ±0.32, respectively.

No interaction was observed for TASK * SEX, \( F (2.430, 672.889) = 1.327, p = 0.273 \), partial \( \eta^2 = 0.042 \). A main effect was observed for TASK, \( F (2.430, 672.889) = 54.315, p < 0.001 \), partial \( \eta^2 = 0.644 \), as was a main effect for SEX, \( F (1, 30) = 6.258, p = 0.018 \), partial \( \eta^2 = \)
Normalized peak knee moment had homogeneity of variances \((p > 0.05)\) and covariances \((p = 0.324)\). Mauchly’s test indicated the assumption of sphericity was violated, \(\chi^2(9) = 32.592, p < 0.001\); therefore, a Greenhouse-Geisser adjustment was used \(\varepsilon = 0.607\).

The normalized average knee moment for the DLS task was \(0.69 \pm 0.11\). TASK conditions SS \((0.77 \pm 0.11)\), BL \((0.78 \pm 0.14)\), FL \((0.93 \pm 0.14)\) and SLS \((1.04 \pm 0.19)\) were 12\%, 14\% 35\%, and 52\% greater than DLS, respectively. Women’s mean values were \(0.81 \pm 0.18\) and those for men were \(0.87 \pm 0.20\) for normalized average knee joint moments.

No interaction was observed for \(\text{TASK} \times \text{SEX}\), \(F(2.409, 72.263) = 0.888, p = 0.432\), partial \(\eta^2 = 0.029\). There was a main effect for \(\text{TASK}\), \(F(2.409, 72.263) = 47.040, p < 0.001\), partial \(\eta^2 = 0.611\), but in contrast to peak values, a main effect was not found for \(\text{SEX}\), \(F(1, 30) = 3.548, p = 0.069\), partial \(\eta^2 = 0.106\).

Normalized average knee moment did not have homogeneity of variances for DLS \((p = 0.20)\) and SLS \((p = 0.046)\), while the other tasks met the condition \((p > 0.05)\). However, since the assumptions of normality are met for DLS and SLS, the between-participant condition sample numbers are equal, and the departure from homogeneity is rather small, the data were left as-is since ANOVA is considered robust to the violation in this case. The homogeneity of covariances did exist \((p = 0.518)\), and a violation of sphericity occurred, \(\chi^2(9) = 32.700, p < 0.001\), so the Greenhouse-Geisser adjustment was used, \(\varepsilon = 0.602\).

The results from \textit{post-hoc} testing for \(\text{TASK}\) indicate that for peak normalized knee net joint moments (Figure 6), BL and FL were the only conditions statistically similar \((p = 0.828)\). For the average normalized knee joint moments, BL and SS were the only similar tasks \((p = 0.457)\). The peak knee moment had a magnitude rank-order trend of DLS < SS < BL ≤ FL < SLS \((p < 0.05)\), while the average knee moment trend was DLS < SS ≤ BL < FL < SLS \((p < 0.05)\).
Finally, moving to the most distal lower limb joint analyzed during BULLET performance, the results for normalized peak and average ankle net joint moments are presented in Figure 7. The normalized peak ankle moment for the DLS task was \(-0.31 \pm 0.08\), while TASK conditions SS \((-0.56 \pm 0.20\), BL \((-0.64 \pm 0.14\), FL \((-0.68 \pm 0.21\) and SLS \((-0.94 \pm 0.19\) were 82\%, 105\% 119\%, and 204\% greater than the DLS, respectively. Mean peak net joint moments across BULLETs for women were \(-0.61 \pm 0.26\) and for men were \(-0.64 \pm 0.27\).

No interaction was observed for TASK * SEX, \(F(4, 120) = 1.938, p = 0.108\) partial \(\eta^2 = 0.061\). A main effect was observed for TASK, \(F(4, 120) = 83.500, p < 0.001\), partial \(\eta^2 = 0.736\). There was no main effect observed for SEX, \(F(1, 30) = 0.318, p = 0.577\), partial \(\eta^2 = 0.010\). Normalized peak ankle moment had homogeneity of variances \((p > 0.05)\) and covariances \((p = 0.764\). Mauchly’s test indicated the assumption of sphericity was met, \(\chi^2(9) = 9.736, p = <0.373\), and the only case where a correction factor was not necessary.

The normalized average ankle moment for the DLS task was \(-0.18 \pm 0.06\). Compared to the DLS, TASK conditions SS \((-0.32 \pm 0.12\), BL \((-0.36 \pm 0.11\), FL \((-0.45 \pm 0.15\) and SLS \((-0.75\).
±0.14) were 74%, 98% 147%, and 309% greater in magnitudes, respectively. Mean values for women and men were -0.41 ±0.23 and -0.42 ±0.22, respectively, for average ankle moments across all tasks.

No interaction was observed for TASK * SEX, $F (3.049, 91.485) = 0.723, p = 0.578$, partial $\eta^2 = 0.024$, but a main effect for TASK was found, $F (3.049, 91.485) = 160.636, p < 0.001$, partial $\eta^2 = 0.843$. No significant differences existed for SEX, $F (1, 30) = 0.109, p = 0.743$, partial $\eta^2 = 0.004$. Normalized average ankle moments also had homogeneity of variances ($p > 0.05$) and covariances ($p = 0.788$). Mauchly’s indicated a violation of sphericity, $\chi^2 (9) = 20.393, p = 0.016$; therefore, a Greenhouse-Geisser adjustment was used, $\varepsilon = 0.762$.

The results from post-hoc testing for TASK (Figure 7) show that all BULLETs were significantly different for normalized average ankle moments ($p < 0.05$), while BL and FL were the only statistically similar conditions for peak moment magnitudes ($p = 0.162$). The peak ankle moment had a magnitude rank-order trend of: DLS < SS < BL ≤ FL < SLS ($p < 0.05$). The average knee moment trend was: DLS < SS < BL < FL < SLS ($p < 0.05$).

Figure 7: (Left) Peak and (Right) average body-size normalized ankle net joint moments across tasks: Double Leg Squat (DLS); Single Leg Squat (SLS); Split Squat (SS); Backward Lunge (BL) and Forward Lunge (FL). Mean values of all 16 men (white bars) and 16 women (grey bars) participants are plotted; error bars represent the standard deviation. Post-hoc results for TASK show no statistical difference between BULLETs with the same alphabetical letters below bars ($p > 0.05$), while different letters indicate BULLET tasks were statistically different ($p < 0.05$). No significant difference existed between men and women across all conditions for peak and average values ($p > 0.05$).
5 Discussion

To the author’s knowledge, this was the first study to compare the mean and peak net joint extension (hip and knee) and plantarflexion (ankle) moments together with the mean and peak of their summed value (support moment) between four common BULLETs and a DLS task. The null hypotheses were rejected, as statistically significant differences existed in moment magnitudes between BULLETs within individuals and between sexes. Comparison of the results to previous related work is first discussed below as a means to verify the calculations made and address the study hypotheses. Subsequently, the practical applications and clinical relevance of the study results are discussed for exercise and paramedical professionals. Finally, study limitations are acknowledged and recommendations for future research are offered.

5.1 Comparison to Previous Research

5.1.1 Joint Kinetics and Kinematics

Since this was the first known study to directly make between-BULLET comparisons in the biomechanical quantities analyzed, results from the individual tasks can be compared to those reported in previous research for verification and interpretation. However, the majority of previous studies measuring joint moments reported their magnitudes normalized magnitudes with respect to body mass (i.e., N·m/kg) and not body size (i.e., N·m/kg·m). The use of body-size normalization methodology was justified in the Review of Literature (Section 2.5.3) and Methods (Section 3.5) of this document; however, this is not the most common method used. For this reason, body mass-normalized data are provided in Appendix F to facilitate comparisons between data reported herein and previously. Additionally, if new literature was found as a result of recent publication or inadvertent omission a priori, consideration was taken in the discussion of results and reflection of the study methodology.

Four studies containing a DLS tasks did report body size-normalized joint moments, however only two reported sagittal plane kinetics (Almosnino, Kingston, & Graham, 2013; Butler, Plisky, Southers, Scoma, & Kiesel, 2010), while two did not (Donohue et al., 2015; Gooyers, Beach, Frost, & Callaghan, 2012). Almosnino et al. (2013) examined the effect of foot position and stance-width on three-dimensional knee joint kinetics during performance of a DLS. In all stance-width and foot position conditions, average peak knee moments ranged from approximately 0.80 to 0.90 N·m/kg·m (These values were interpreted from time-series figures as
discrete values were not reported). The average peak knee moments of the DLS task observed herein were $1.10 \pm 0.20 \text{ N}\cdot\text{m/kg}\cdot\text{m}$. The slight discrepancies could be explained by the procedures, as it was reported Almosnino et al. (2013) controlled movement tempo similar to the current study, however a specific tempo time for each repetition cannot be determined from the information provided. It is not uncommon to prescribe slower tempos to focus on task control and minimize any effects of inertia and velocity, which would lower the resulting moments. The 2-second up, 2-second down tempo in the current study was chosen based on pilot data where subjects self-selected tempo (Chapman et al., 2014). This is the same or slightly faster than what has been observed elsewhere (Graci et al., 2012; Riemann et al., 2013). Therefore, if self-selected tempos were used in Almosnino et al (2013), they could be slightly slower potentially lowering the knee joint moment as seen in the comparison. Tempo or movement speed should be reported in methods when biomechanical assessments of exercise tasks are completed. It provides information which can directly affect the joint moment magnitudes and necessary for interpretation of the results. Another potential cause for the lower values is that participants in Almosnino et al. (2013) were slightly different, with twice as many males than females and both sexes being slightly heavier, but shorter when compared to the current study. Regardless of these discrepancies, the peak normalized knee moment magnitudes in the current study are comparable to those reported by Almosnino et al. (2013).

Butler et al. (2010) quantified peak joint moments during a deep bilateral squat task between three groups who performed with visually observable kinematic differences. Group 3 was defined as non-compensatory, while group 2 and 1 were defined as using compensatory movement patterns based on qualitative scoring criteria, with the latter being more limited in performance. The peak joint moment magnitudes observed (group 3; group 2; group 1; units) for the hip (-0.54; -0.56; -0.36; N·m/kg·m), the knee (0.63; 0.56; 0.45; N·m/kg·m), and the ankle (-0.21; -0.25; -0.27; N·m/kg·m). Although similar, they were all slightly lower than those observed herein for the hip (-0.70±0.14 N·m/kg·m), knee (1.10±0.20 N·m/kg·m), and ankle (-0.31±0.08 N·m/kg·m). The pattern of moment magnitude loading was the same in both studies (i.e., knee > hip > ankle), and the group with no ‘compensations’ was the closest to our values in the hip and knee joint moments. Multiple reasons stand out for the slight discrepancy between the current study and Butler et al. (2010). First, in their study the arm position was raised overhead and conjoined by a dowel rod, while arms were placed on the hips for the duration of
the movement task in the current study. Second, different instructions were used, and no tempo or movement velocity information is given in the publication. Similar to the Almosnino et al. (2013) study mentioned previously, it is possible the participants self-selected a slower speed in their study and thus could exhibit lower moment magnitudes. While the exact cause of the discrepancy cannot be pinpointed, the moment magnitudes observed in the current study still fall within a reasonable range compared to Butler et al. (2010).

As mentioned above, body mass-normalized joint moment data were added in Appendix F to assist with further interpretation of the results observed herein. Body mass-normalized lower limb net joint moments have been analyzed at peak knee flexion (79.2±8.7º) during SLSs using very similar procedures to the current study (A. Khuu et al., 2016). It was reported that joint moments were 1.11±0.54 N·m/kg for hip extension, 1.51±0.30 N·m/kg for knee extension, and 0.71±0.26 N·m/kg for ankle plantarflexion. A support moment was not computed in Khuu et al. (2016), but it is likely that the values would be comparable to those reported herein given the close similarity in hip, knee, and ankle joint moment magnitudes between studies. Kinetic comparison between the two studies assumes that our peak moments occurred at peak knee flexion angles, which is the discrete time point of measurement by Khuu et al. (2016). While this would be a logical assumption to make as supported in the literature (Cotter et al., 2013; Scaglioni-Solano et al., 2005; Singh et al., 2015), the discrete joint angle of peak moment was not examined in the current study (i.e., a “global” peak was extracted herein).

Kinematic data in the Khuu et al. (2016) study were similar to those reported in Appendix E. In both studies, participants’ self-selected SLS squat depth and almost identical peak knee flexion angles were observed (79.2±8.7º vs. 80.1±12.1º). Further supporting this, the other lower limb kinematics at peak knee flexion in Khuu et al. (2016) are very similar to the peak angles measured in this study: 59.8±16.1º vs. 53.8±12.4º at the hip joint; 29.3±5.8º vs. 29.8±5.9º at the ankle joint; and 24.3±14.4º vs. 28.2±11.0º for the trunk angle (former versus latter, respectively). Exercise task performance instructions were much more detailed herein, as Khuu et al. (2016) did not report using a script, nor did they describe much detail regarding their instructions aside from asking participants to squat as low as they can. This specific instruction was also used in our study, and perhaps a driver of the kinematic similarities described as it relates to the primary direction of COM displacement observed during this exercise task.
One study has previously compared a BULLET task (i.e., FL) to a DLS task and is the closest resemblance to the current study (Singh et al., 2015). However, instead of multiple BULLETs, they looked at three different depths for the squat (i.e., knee angle of 60º, 70º, 80º) and three different lengths for the lunge step (i.e., 1.0, 1.1, 1.2 × tibia length). Furthermore, Singh et al. (2015) made comparisons between obese and non-obese individuals as determined by Body Mass Index (BMI). In participants without obesity, Singh et al. (2015) reported peak extensor/plantar flexor moments that ranged from 0.96-1.14 N·m/kg for the hip, 0.52-0.64 N·m/kg for the knee, 0.40-0.45 N·m/kg for the ankle, and 2.05-2.07 N·m/kg for the total support moment. The values in Singh et al. (2015) are approximately 0.05-0.20 N·m/kg lower for the hip and ankle, 0.50-0.60 N·m/kg for the knee and 0.80 N·m/kg for the support moment, slightly lower in magnitude compared to those herein (Appendix F). These discrepancies are likely due to the event selection methods of choice and kinematics observed. The moments were only measured at the bottom or transition phase of the task and averaged during a 3-second pause in this position. Deceleration into and acceleration out of this position were not accounted for in the cited study, essentially making the task an isometric positional hold. Knowing this, it is perhaps better to compare to the SS from the current study to their FL, as the SS task herein consisted almost exclusively of vertical COM motion since the feet were statically placed, and it was always the lowest in magnitude out of the lunge pattern tasks examined. When compared to the SS magnitudes, Singh et al. (2015) data fall in the exact values for hip joint moments, are 0.35-0.50 N·m/kg lower for knee and support moment, and 0.05-0.10 N·m/kg lower for the ankle. This brings the values even closer to each other. Another likely contributor to the discrepancies is the peak flexion angles, as SS and FL averages across all subjects were very similar at 104.4º and 107.4 º, but approximately 25º-45º degrees greater than those observed in Singh et al. (2015). Greater knee flexion angles have shown to increase the joint loading magnitudes (Cotter et al., 2013; Scaglioni-Solano et al., 2005), and Singh et al. (2015) also confirmed this, which support the rationale for the observed discrepancies.

Similar trends were observed when comparing the DLS joint magnitudes from Singh et al. (2015) to those herein. Performers of the DLS in the current study exhibited an average peak knee flexion angle of 99.4º, which was also 20º greater than the largest angle observed in Singh et al (2015). Given this, it would suggest that the peak moments herein would be higher in magnitude, which was the case as the values for the lowest depth squat condition (i.e., 80º) in
Singh et al. (2015) were 0.24-0.37 N·m/kg for the hip, 0.75-0.80 N·m/kg for the knee, 0.20-0.34 N·m/kg for the ankle, and 1.18-1.53 N·m/kg for the support moment. Aside from the ankle falling in the exact range of the current study, these values are 0.20-0.50 N·m/kg lower for the hip, knee, and support moment.

It is interesting to note that the standard deviations (SD) in Singh et al. (2015) are consistently larger than those in the current study. In fact, this same trend holds true for most of kinetic and even kinematic data reported above, as the SD’s in the current study seem to be the smallest among the metrics contrasted between the cited studies. One possible reason is the current study incorporated the most detailed and robust instructions, in the addition to demonstration for the performance of the exercise tasks presented. There is evidence in motor learning to suggest that giving participants’ pre-practice demonstration results in more consistent outcomes (Newell et al., 1990). Additionally, the role of specific cueing and wording area is starting to receive attention in other exercise tasks (S. Khuu et al., 2015), however the effect of specific instruction and cueing on BULLET task performance has yet to be examined. Given the direct implications for coaching and exercise prescription, this would be a logical direction for future research.

When compared to the current body of literature on individual BULLETs, the results of this study fall within expected ranges for all joint moment and support moment magnitudes observed. Where small discrepancies existed, methodological inconsistencies and kinematic differences between studies likely account for the majority of the discrepancies. Knowing that the quantities derived in the current study are largely consistent with what has been reported previously makes it possible to offer practical recommendations related to the prescription and progression of uni- and bi-lateral lower limb bodyweight exercises. This will be done in more detail in Section 5.2, but a few general remarks are warranted below.

Aside from the hip joint, the results of the research done to date indicate that single-leg support exercises result in greater joint loading than double-leg support tasks, which is expected based on physical laws (i.e., Newtonian mechanics). When considering only double-leg support tasks, asymmetrical limb position results in greater joint loading that than symmetrical limb positioning. Even further, when considering only the asymmetrical tasks (i.e., lunge pattern), those containing horizontal COM motion (i.e., BL, FL) produced greater joint loading than tasks
with minimal to no horizontal COM motion (i.e., SS). Of the lunge tasks, the FL had the
greatest lower limb joint moment magnitudes, which could be due to a greater mechanical
impulse. Given that the same tempo was used during all of the tasks and that the FL was the
only task containing a flight phase, the force observed for the duration of contact time would be
greater than the other lunge patterns. While this was dealt with through time normalization (i.e.,
rubber-banding) and represents the force sustained during the task, it does still represent a
discrepancy between FL and the other lunge tasks. While the grouping of tasks may have
slightly differed between the variables, the peak and average joint moments displayed the same
overall trend in joint loading response. Further discussion of this observed similar rank-ordering
and the practical applications stemming from it will be discussed in Section 5.2.

5.1.2 Sex Differences

Despite body size-normalizing net joint moment magnitudes (i.e., dividing moments by
body mass × leg length factor), between-sex differences emerged in some of the net joint and
support moments examined. Specifically, significant differences between men and women were
detected in three of the eight dependent variables, namely the peak and average support moment
and the peak knee moment. However, between-sex comparisons in the average knee moment
also approached significance ($p = 0.069$) with a medium effect size (partial $\eta^2 = 0.106$), which
closely resembled that of the peak knee moment (partial $\eta^2 = 0.173$). In cases where significant
differences were not detected between men and women, $p$-values exceeded 0.20 with small
effect sizes (partial $\eta^2 < 0.04$). Since the observed between-sex differences in individual joint
moments only occurred at the knee, it is plausible that the knee moments were also the main
driver of discrepancies observed in support moments as well.

No statistically significant TASK*SEX interactions were observed, as men always
exhibited greater magnitudes than women whenever differences existed. On first glance this is
not surprising, as men on average are larger and are comprised of a greater proportion of lean
mass than women (Sizer & James, 2014). However, given the methods used herein to remove
these potential confounders, observed differences in the support and knee moments are likely
attributed to other factors. The goal of normalization was to account for the size differences
inherent between men and women. Previous work has shown that height and mass can represent
up to 82% of the variance in joint moments (Moisio et al., 2003). When accounted for using
body-mass normalization, this number was reduced to <6% in their study. Further, using body
size (i.e., either height or leg length in addition to body mass) for the normalization of joint moments removed an even greater amount of the variability, and this factor is usually significantly different between men and women (Moisio et al., 2003). However, results of the current study suggest that body mass and leg length are not the key driver of the differences in knee joint and support moments observed between sexes during BULLET performances.

Other components of sex-based anthropometric dimorphism could be responsible for the observed differences. It is known that skeletal geometries differ between men and women, for example women have a wider pelvis which contributes to a larger Q-angle in the frontal plane (Sizer & James, 2014). If muscle attachments are approximately the same, yet shapes of bony segments are different, the factors affecting moment of force generation such as muscle lines-of-action, lengths and effective moment arms could all differ. Furthermore, the distribution of mass also differs between men and women (Miller, MacDougall, Tarnopolsky, & Sale, 1993). Given the contribution of segmental mass-inertial properties to joint kinetics, this could also drive the discrepancies. Advanced diagnostic methods such as dual x-ray absorptiometry (DXA) and magnetic resonance imaging (MRI) would be required in order to determine the participant- and sex-specific parameters necessary to accurately conclude the contribution of these factors. These methods are extremely expensive, require a technician to operate, and time-consuming in procedure; therefore not realistically accessible or usable in most laboratory settings, such as the case herein. Laboratories that have access and support for this type of individual anthropometric modelling should be encouraged to do so in order to provide more valid results. Further support of this exists as the default anthropometric models used in Visual3D (Dempster, 1955; Hanavan, 1964), while supported in the literature and widely used, are rather dated. Individualized models would bring less estimation and more accuracy to the computation of joint moments, and are starting to be explored and utilized in related biomechanics research (Laschowski, 2016; Laschowski & McPhee, 2016; Riemer & Hsiao-Wecksler, 2009).

Joint kinematics were measured in order to assist with the interpretation of kinetics. If men and women exhibit different motion patterns when performing the exercise tasks, it could explain the net joint moment differences observed. However, sagittal plane kinematics of the ankle, knee, hip, and trunk did not differ between men and women during BULLETs (Appendix E). While the evidence of divergent sex-based performance is equivocal across other athletic and locomotive tasks, our results contrast some of the current research regarding BULLET
performance. Women have been shown to exhibit less trunk flexion (Graci et al., 2012), greater knee flexion, greater ankle dorsiflexion, and less hip flexion than men when performing both SLS and lunge tasks (Dwyer et al., 2010; Zeller et al., 2003). One possibility is that our study involved consistent, robust instructions and demonstrations of task performances delivered to both men and women and a lack thereof in the comparable literature. Most of the studies cited herein including those mentioned above provided minimal detail, if any at all, about how they communicated task positions, motions and general performance to the participants. Most do not disclose if feedback or coaching was given, which can also change movement outcomes (Frost et al., 2015). Since the task instructions herein were rather detailed (Appendix D), evidence based-based (Czaprowski et al., 2012; Graham, 2002; Keogh, 1999; Schoenfeld, 2010; Schütz et al., 2014), with no additional feedback beyond that provided to ensure that tasks were performed as instructed, it is conceivable this was the reason for the similar kinematic responses observed between sexes.

While only the body-fixed sagittal plane data was examined in this study, the literature does offer evidence of sex-based biomechanical differences in other planes of motion during BULLET and DLS performances (Dwyer et al., 2010; Graci et al., 2012; Zeller et al., 2003). It is plausible these differences also existed in this study and warrant further exploration. Data from injury biomechanics do support frontal and transverse biomechanics potentially affecting the sagittal plane kinetics (Hewett et al., 2005; Quatman, Quatman-Yates, & Hewett, 2010; Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006). While not examined within the current study, the effect of non-sagittal plane kinematics on BULLET kinetic performance is a logical step for the progression of this research.

Other muscle-based considerations with evidence to support a potential sex-effect on kinetic measures are an individual’s muscle activation and strength levels. When performing the same BULLET tasks, significantly different muscle activation patterns of lower limb musculature have been observed between men and women (Bolga et al., 2014; Zeller et al., 2003). Conversely, one study showed no sex-based differences between the EMG of eight muscles during FL and SLS tasks (Bouillon et al., 2012). Further work is needed to determine if this alone is a factor, or if the activation levels are more closely related to, or a direct response from another factor such as the volitional effort of the task, differing anthropometric geometries or mechanics of task performance as mentioned above.
Strength is the maximum amount of force a muscle or group of muscles can produce. While men have greater absolute strength, there is evidence to show they have greater relative strength as well (Miller et al., 1993). This discrepancy is primarily attributed to the differences in lean body mass between sexes. This would suggest once again that body mass is a potential culprit, albeit in a slightly different context. Instead of all body mass, variance in the distribution of lean body mass specifically could be considered to influence joint moment magnitudes. As mentioned above, MRI or DXA machines would be required, and accessibility to equipment and trained staff create barriers to implementation. This was beyond the scope of this project, but is suggested for future exploration to determine the mechanisms contributing to sex-based discrepancies observed in kinetic measures during exercise task performance.

Another method of accounting for anthropometric variability exists called allometric scaling. This method uses various coefficients to adjust individual data for their body size and shape. It is based on the principle that organisms of the same species are isometric, that is geometrically similar in form (Alexander & Jayes, 1983). However, aside from these methods not being commonly used in biomechanics research (Moisio et al., 2003), its complexity due to interactions based on the varying individual components make it difficult to practically apply, as scaling differs between populations based on geographical origin, sex, and body part(s) of interest (Sylvester, Kramer, & Jungers, 2008). This over-complicates the application of allometric scaling and the derived results, and make it difficult to justify its methodological use at this time.

The results from this study showed that sex-based differences existed in knee and support moments, while hip and ankle moments remained unchanged during BULLET performance. As summarized above, there are many plausible reasons for this occurrence, but the data collected in this study are not sufficient to explain these findings. It could be suggested that body size and mass play an insignificant part in driving the observed differences, as they were accounted for using validated normalization techniques. However, consensus on suggested methodology to account for all of the anthropometric dimorphism between men and women does not exist. It is clear further work is required to account for all of the confounding variables when studying sex-based differences in human biomechanics. Until then, exercise practitioners and clinicians should still give consideration to the varying knee joint and support moment response of men and women when performing BULLETs targeted for kinetic outcomes.
5.2 Practical Applications

The results garnered from this study can be used to aid and inform exercise professionals, injury rehabilitation clinicians, and any other individuals who are prescribing BULLETs in their daily testing and training environment. Consideration is given primarily to the common paradigms that BULLETs are used in, and not every possible use case will be covered.

5.2.1 Biomechanical Progression

Progressive overload is a concept used to safely and effectively deliver an increasing stimulus over time in order to create continuous improvements in training adaptations (Kraemer et al., 2002; Ratamess et al., 2009). This is done in order to combat the ‘plateau’, or absence of improvement as no requirement for increased adaptation from the movement system would occur if progression is not employed. If all training variables remain constant overtime, the maintenance of abilities would remain unchanged. Mechanical loading is one variable that is commonly manipulated and monitored by the exercise professional in this fashion. Part of the challenge is when quantifying the stimulus for the progression of load, a practitioner typically considers only the added external load, and may not give consideration to the individual joint loading directly. This is understandable, as the transformation between ‘external’ and ‘internal’ loading is highly non-linear, individualized, and thus difficult to estimate in a practical setting (Cholewicki et al., 1991). It requires biomechanical modeling procedures such as those utilized herein for general or crude load measures (e.g., net joint moments), and more complex modelling musculoskeletal modelling for specific tissue load measures (Dowling, 1997; Reeves & Cholewicki, 2003). However, with data garnered from this study, practitioners can use exercise variations to manipulate mechanical loading in a more targeted and progressive manner.

To date, only one study has attempted to evaluate the progression of various single-leg exercise tasks: SLS; FL; and a step-up-and-over task (Boudreau et al., 2009). The goal of this study was to categorize the tasks based on hip muscle activation using EMG for the purpose of rehabilitation and strengthening of specific hip muscles. Individual activation levels of muscle do yield valuable information about the timing and magnitude of muscle action. However, the relationship between EMG signals and muscle forces is not well-defined in dynamic movements, as it is dependent on properties such as activation state, contraction velocity,
muscle length, and pennation angle among a host of other variable muscle properties (Dowling, 1997). As such, without corresponding kinematic and kinetic data combined with advanced musculoskeletal modelling procedures, it is difficult to interpret the results of EMG studies for the purposes of characterizing exercises based on load response.

The DLS task has been rank-ordered based on vertical ground reaction force magnitudes among a battery of locomotive (i.e., walk, jog, and sprint) and jumping tasks (Scarfe, Li, Reddin, & Bridge, 2011). DLS ranked the lowest among all tasks for both absolute and relative peak vertical force and rate of force development. While conceptually this progression is similar to that observed within the current experiment, the kinetic dependent variables were whole-body point-mass measures opposed to the specific lower limb joint moments herein. Regardless, the kinetic results of both studies align as the DLS also had the lowest net joint moments across all tasks performed within this study. The other tasks compared in the Scarfe et al. (2001) study do not relate to the other BULLETs examined, and as such the current study was the first to rank-order the four BULLET tasks and a DLS based on net joint moment magnitudes.

Using these net joint moment magnitudes, the tasks were rank-ordered in the current study from smallest to largest. For the knee, ankle, and support moments, combining peak and average results, a general trend of DLS < SS ≤ BL ≤ FL ≤ SLS was observed. The primary difference between dependent variables was the BL task, which in every case was grouped with either the SS task or the FL task (p > 0.05). Other kinematic characterizations of this trend include double-leg support being lower in magnitude to single-leg support, and within the double-leg support the symmetrical task was always lower in magnitude to asymmetrical support.

The hip was the only joint examined which did not follow the aforementioned net joint moment magnitude trend. When combining peak and average results, the following ordering from smallest to largest resulted: DLS < SLS ≤ SS ≤ BL ≤ FL. The characteristics of this trend are a little different, as squat pattern tasks are lower in magnitude than lunge pattern tasks. An overall trend was observed where vertical dominant COM displacement tasks (i.e., DLS, SLS, SS) were lower in magnitude compared to those that with horizontal COM displacement (i.e., BL, FL).

Despite the similarities observed in the lunge pattern tasks, SS and FL were significantly different in all cases (p < 0.05). One practical consideration, since BL is equal in almost every
case to SS or FL, it could be removed from the progression to simplify it for practitioners. This would remove redundancy of BL and still maintain a similar loading progression. In doing this, the progression for knee, ankle and support moments becomes DLS < SS ≤ FL ≤ SLS, while the progression for the hip becomes DLS < SLS ≤ SS ≤ FL. In all of the above cases, SS was the lowest magnitude lunge task, suggesting that the added horizontal movement of BL and FL increases the loading demand of the lunge task.

Each exercise variation examined in the current study has the potential to improve different biomechanical demands throughout the musculoskeletal linkage. The recommendations above are based solely on the lower limb joint moment magnitudes analyzed. Knowledge of these observed progression gives practitioners the ability to progress and regress the crude loading of individual lower limb joints. While the potential utility of this will be explored in the subsequent sections, it must be noted that practitioners may have other rationale for using specific BULLETs or a DLS task (e.g., coordination and control), in which case these observed progressions may not apply for the desired adaptations.

5.2.2 Qualitative Movement Assessment

The most direct place for application of the observed results is with the movement assessments that use BULLETs and DLS directly in their battery of tasks, since these movements are already being implemented. Almost every mainstream formalized movement assessment to date contains one or more of these tasks. The Movement Competency Screen, which is the most evidence-based movement assessment published to date, is the only screen that makes use of all the task groupings examine herein: DLS; SLS; and the lunge pattern (Kritz, 2012). The grouping of exercise tasks resulting from this study would support the use of all three from a loading progression standpoint, as each one places a different loading demand on the lower limb joints. Further, given the MCS considers increasing demand by adding external load and speed, an initial internal loading progression could be built in by ordering the tasks based on the increasing joint loading magnitudes. Therefore, by completing a DLS, followed by a lunge task (i.e., SS, BL, FL), then a SLS, a crude internal loading progression is inherently built into the MCS.

The Functional Movement Screen (FMS) and the NASM Corrective Exercise Training Manual (NASM) utilize forms the SS and DLS in their assessment (M. A. Clark & Lucett, 2010;
Based on differential loading magnitudes, our results would suggest that the addition of an SLS could provide an increased loading demand on the lower limb, in turn potentially exposing a desirable/undesirable response. Additionally, neither contain a single limb support flexion and extension task like the SLS, so it would provide a unique movement challenge not yet considered in their battery of tasks. Given the goals of these types of assessments, and the fact that human locomotion comprises of many single limb support flexion/extension tasks (e.g., stair climbing, running, bounding), it would seem logical to add in a SLS-type task to assess a low demand version of this movement pattern.

In a similar fashion, The Foundation Matrix (TFM) uses a SS and a SLS pattern task for assessment (Mischiati et al., 2015). The results herein would suggest the DLS as a lower demand loading options to initiate assessment, or to regress to a DLS if a test participant were having difficulty with the task. Moreover, the Y-Balance (YBT) and Star Excursion Balance Test (SEBT) both use SLS exclusively (Chimera, Smith, & Warren, 2015; Ness et al., 2015). Given the nature of the test in creating a SLS balance challenge and measuring elevated limb reach, a DLS may not fit the imposed testing stimulus. If a test participant could not successfully complete the assessment of YBT or SEBT, either regressing to or training with a DLS and/or lunge pattern could be done to initiate a starting point or build capacity respectively, progressing towards the SLS-based assessment and intervention tasks if they were deemed too challenging.

As previously mentioned, a general guideline that could be applied to any movement assessment is the grouping or removal of a lunge task(s). Backward lunges could be removed from any screen as the loading pattern is redundant to SS and FL as the results here have shown. However, in consideration of all abovementioned suggestions, the assessor or exercise professional may have other reasons for using one or more of the exercise tasks studied. Perhaps there is a kinematic quality, irrespective of loading, they are interested in imposing upon the participant to potentially expose a specific movement behavior or dysfunction. Conversely, the addition of more BULLET tasks could also be supported from a loading standpoint. As previously discussed load and speed are modifiers of movement behavior (Frost et al., 2013) and increase joint loading (Schoenfeld, 2010). When compared to external loading conditions, changing the BULLET performed can also increase joint loads, but to a lesser amount with smaller increments. Including an array of BULLETs in movement assessments would expose the participant to a variety of joint loading conditions and provide a less vigorous and more
stepwise progression prior to adding external load and movement speed. From a lower limb joint loading standpoint, the evidence would support the use of a DLS, SLS, and a lunge pattern among a battery of tasks in qualitative movement assessments.

**5.2.3 Exercise Prescription and Load Monitoring**

Knowledge of the magnitudes and rank-ordering of these exercise tasks could prove useful to practitioners who prescribe exercise. BULLETs are a mode of lower limb resistance training that is universally accessible to able-bodied individuals. The abovementioned biomechanical loading trend based on joint moment magnitudes would allow such an individual to progress the sustained demand by simply changing the exercise task. When utilizing BULLETs in a battery of tasks, the observed progression results in smaller increments and lower joint loading than that observed when adding external load (S. P. Flanagan & Salem, 2008; Gullett et al., 2009; Riemann et al., 2013). Greater control can be given by the practitioner when applying load, and consideration for a more ecologically valid loading paradigm is given as the ‘internal’ loading of the joint(s) are being considered. Progressive overload could then be initiated without added external load and/or increased movement speed. The modulating variable then becomes a desired volume of repetitions or the time-under-tension (TUT), which could be planned and progressed to achieve a target goal prior to moving on to the next BULLET task to progress the magnitude of loading. This is arguably much simpler for the practitioner and participant when prescribing an exercise program for novice, elderly or rehabilitating populations, as well as those who are remote or do not have access to equipment.

In cases where a ‘weak link’ (i.e., joint or limb) in the kinetic chain needs to be addressed, incremental loading could be provided and progressed based on the ability to control and/or master the specific exercise task. It is not uncommon for a practitioner to consider an individual limb or joint when prescribing exercise. They may look to increase the strength or capacity of one joint or limb to tolerate loading for a specific imposed demand or transfer-to-task. Additionally, musculoskeletal injury and rehabilitation from such an injury may be unilateral in nature, therefore the interventions may also be primarily unilateral. For example, bilateral asymmetries in maximal knee flexor torque and other kinetic measures have been observed in skiers who have sustained an anterior cruciate ligament (ACL) injury and subsequent reconstruction (Jordan, Aagaard, & Herzog, 2015a, 2015b, 2017). The intervention required to bring the asymmetry into an acceptable functional threshold would logically have to focus on
the previously injured limb, as training both equally may continue to promote any asymmetry and subsequent risk.

Using BULLETs based on magnitudes for progressive overload in this manner can provide utility for practitioners looking to build robustness or prevent against musculoskeletal (re-) injury. Mechanical injury can be defined as damage occurring when an applied load exceeds the tolerance of human musculoskeletal tissue. A single supramaximal load, or cumulative (repetitive and/or sustained) submaximal load over time are two of the primary mechanisms that drive mechanical tissue damage or injury (Blanch, 2014; Kumar, 2001; McGill, 1997). Tissue can also fatigue lowering the failure tolerance for damage to occur. If ample recovery time is not given in between successive bouts of loading, the tissue does not have time to adapt by repairing and remodeling in order to handle increased stimulus. By starting to understanding the magnitude of loading when prescribing specific BULLETs, and how it compares to both external loading conditions and other exercise variations, it gives practitioners greater confidence and control in planning and applying loads on individuals to avoid the potential for injury causation and build resilience.

A good example of the abovementioned logic in current practice is with knee osteoarthritic (OA) populations. Knee adduction moment magnitudes increase with knee OA severity (Foroughi, Smith, & Vanwanseele, 2009). Part of the goal when prescribing exercise to knee OA populations is to limit the knee adduction moments, minimizing potential symptoms and progression of the injury. Yoga poses have been previously examined and categorized based on the moment magnitudes in order to find the poses that provided small moments with large muscle activations (Longpre, Brenneman, Johnson, & Maly, 2015). Squatting and lunging poses such as ‘chair’ and ‘warrior’ were found to be more ideal using these criteria than single-leg poses such as ‘tree’, having a direct and immediate impact on clinical populations. This knowledge was applied into longer-term programming, and a 12-week modified yoga program resulted in decreased pain and increased strength (Brenneman, Kuntz, Wiebenga, & Maly, 2015). Aside from the immediate outcomes and application of such research, it demonstrates that knowledge of joint moments associated with performing exercise tasks can conceptually bridge a gap between exercise modalities, as it provides a common language for the various practitioners to unify around, utilizing the underlying first principles that dictate all human movement.
Pedagogically, the concept of understanding and applying moments of force when loading individuals in all aspects of exercise prescription is surprisingly not given much attention. While defined and explained in mechanical terms in most academic biomechanics textbooks, it is absent from more practitioner related strength and conditioning texts and certification manuals (Baechle & Earle, 2008; CANFITPRO, 2014; Crossfit, 2016; CSEP, 2013). Further, to the best knowledge of the author, the specific application and craft around daily utilization of joint moments has only been explained in detail in one practitioner education program and supporting text (Purvis, 2011). To note, this statement is not exhaustive by any means, as language, geography, and access to non-peer reviewed literature, text and educational material may result in the unknown omission of material claiming otherwise. However, the author is confident even if missed, additional material would still represent a large minority of that which is available to the industry professionals and public-at-large.

Given the clear limitations for practitioners inferring muscle forces and actions (Herbert et al., 1993), the reliance on force (i.e., resistance and gravity) as a tool providing stimulus for adaptation, and therefore the governing principles of Newtonian mechanics in all aspects of exercise, it would suggest that the understanding and subsequent application of ‘moments of force’ into the daily exercise environment would seem logical and only beneficial to all parties involved. Net joint moments provide a more simplistic concept of how the sum of linear muscular forces are expressed around an angular joint. For the exercise coach, understanding what inferences about the kinetics of exercise tasks can and cannot be made through human observation and interaction is important to avoid misinterpretation of causality and outcomes. While it can be argued that inferring muscle function is futile due to the unobservable nature of the numerous variables required to do so, using minimal assumptions and knowledge of the exercisers intentions while garnering feedback, directionality of planar moments of force around a joint can be deduced (Herbert et al., 1993). While inferring the associated magnitudes is more questionable, results obtained from research such as that conducted herein can help to fill in the gap to provide characterizations about exercise tasks and their joint moment responses to specific kinematic conditions.

To support this point further, the concept of a support moment is uncommon and not only absent from most exercise pedagogy, but rather uncommon contemporary biomechanics. However, an analogous concept of ‘triple extension’ is used regularly both as a rationale for task
transfer and in-practice as a desired motion or outcome of exercise. A support moment, or the algebraic summation of the hip and knee extension moments, and the ankle plantarflexion moment into a single variable, provides a metric which can be considered similar to triple extension. More thought and possible adoption of this metric could drive more research to provide further characterization of exercise tasks and better understanding about triple extension, which is so widely adopted among strength and conditioning practitioners. In general, knowledge of the associated joint moment magnitudes of exercise tasks such as BULLETs give the practitioner more information and help to characterize these tasks in order to make more appropriate and better-informed exercise prescriptions in daily use.

5.3 Limitations

There were a number of limitations to the current study. Consideration was taken a priori and rationale has been provided to delimit wherever possible. From commencement, only a single force plate was available for use during this study and this limited the collection of bilateral kinetic data. Having a second force plate would have allowed the non-stance leg kinetics to be measured during BULLETs, and could have provided more detail regarding how the whole lower body is bearing load between the two limbs. Weight shifts or uneven load sharing between legs may have provided further insight for interpretation of the joint moment magnitudes examined. However, the ground reaction forces measured and net joint moments computed were still those observed on the stance leg during the BULLETs task performance. This methodology has been used previously (Gullett et al., 2009; Singh et al., 2015), and beyond the limitation of only having a single force plate, scientists and practitioners often only consider a single leg for specific outcomes or adaptations from exercise prescription.

The dominant leg was analyzed as the stance leg, but the mirror image condition (i.e., stance leg = non-dominant leg) for BULLETs was not. There is evidence to support that any differences between dominant and non-dominant legs are not significant in BULLET-type tasks (Khuu 2016). This would suggest that the mirror image performance of BULLETs may be similar in moment magnitudes. In contract, there are data to show significant dominance asymmetries in other human movement tasks (Demura, Yamaji, Goshi, & Nagasawa, 2001; Sadeghi, Allard, Prince, & Labelle, 2000). Since the evidence at this point is inconclusive, further research is required regarding the relationship between dominant and non-dominant stance leg during BULLET performance. Furthermore, with the use of bilateral force plates,
examination of the kinetic relationship of the non-stance leg and stance leg within a single version of the task deserves attention. Specifically, the effect of instruction and/or demonstration from the researcher, coach or practitioner, in addition to how other modifiers might change the load sharing between the two limbs during both versions of the exercise task is suggest for future research.

It is standard practice in exercise prescription to use both instruction and demonstration when communicating the positions and motions desired for performance of an exercise task. Both methods are supported in the literature for use in this manner (Al-Abood, Davids, & Bennett, 2001; Hodges & Franks, 2002), and were used in this study to maintain ecological validity for participants and the subsequent application of the results. However, it has been shown that variation in both instruction and demonstration can change the way people move and affect the resulting biomechanics of task performance (Al-Abood et al., 2001; S. Khuu et al., 2015). While some might disagree with the instructions used in this study (i.e., suggest the addition, change, or subtraction of individual components), they provide the most detail out of any related study cited herein. The exercise task instructions were driven by literature as much as possible (Czaprowski et al., 2012; Graham, 2002; Keogh, 1999; Schoenfeld, 2010; Schütz et al., 2014), then supplemented with expert opinion through consultation with the professional Strength and Conditioning Coaches at the Canadian Sport Institute Ontario. Further, to maintain consistency (i.e., control), the primary investigator delivered the standardized instructions and provided a corresponding demonstration of task performance. One method that would have ensured exact consistency in demonstration is consistent video display of task demonstration to each participant. However, this is not the common delivery method used in the daily coaching and exercise prescription environment. Conversely, this is the method of choice for individuals seeking out exercise information via the internet or other digital media sources. Since a primary goal of this study was to enhance use of BULLETs for exercise professionals and clinicians, and practitioners historically have prescribed exercise in a non-digital environment, the human delivery method was selected for this study. As previously mentioned, further research is required on understanding what specific instructions and/or demonstrations of exercise tasks affect the variables associated with lower limb joint motion and loading. But once again, the purpose of this study was not to evaluate the efficacy or modification of the exercise task description delivery, but to examine differences in the net joint moments of BULLETs when participants received consistent information pre-trial regarding task performance.
While individual net joint and support moments provide general measures of lower limb loading, there are several limitations to consider when using these kinetic variables (Bryanton & Chiu, 2014). Contributions from all moment-generating tissues/structures cannot be specified with these methods, as the joint moments calculated in this study represents the “net” effect of all such tissues and structures. If it is assumed, for example, that the moment arms and effective lines-of-action of a simplified agonist and antagonist muscle pair about at a joint are equal, moments of force of the former and latter of 500 N·m and 100 N·m, versus 1000 N·m versus 600 N·m would both result in a net joint moment of 400 N·m in the positive direction of the moment of force created by the agonist. While it does provide a crude index of the minimum level of relative muscular effort required, it may underestimate the absolute moments of force the agonist and antagonist are producing. Additionally, net joint moments do not account for the complexities associated with multi-articular muscles (e.g., hamstrings) which simultaneously produce moments of force around multiple joints, and function to transfer mechanical energy amongst body segments (D. G. E. Robertson, Wilson, & Pierre, 2008; Zajac, Neptune, & Kautz, 2002). More advanced biomechanical modelling procedures with a far greater number of variable inputs and estimations are required to properly distribute the net joint moment to all of the force-producing and force-bearing structures (Dowling, 1997; Reeves & Cholewicki, 2003). Given the argument that practitioners cannot make estimations of muscle forces from visual observation (Herbert et al., 1993), net joint and total support moments provide a more crude but simpler measure of joint and limb loading that requires far less assumptions for a practitioner to infer and apply in practice. Not only have these measures been used in many of the abovementioned studies using similar tasks for similar purposes, they have been previously validated for use in this manner (S. Flanagan & Salem, 2005).

When using net joint moments with lower limb exercise tasks another limitation emerges. Greater knee joint flexion angles and resulting vertical COM displacement have been shown to increase joint moment magnitudes across multiple exercises (Cotter et al., 2013; Scaglioni-Solano et al., 2005; Singh et al., 2015). Due to this, researchers typically implement a fixed peak knee flexion angle, or depth target relating to such an angle as a means of experimental control. However, it could be argued that in some applications, this is a confounding factor as opposed to a control. First, a constraint is being placed at a single joint which can have a significant effect on other joints within the same limb (Beach et al., 2014). Second, feedback via a target would
be given after each repetition. This additional external focus of attention could cause a change in the performance after each subsequence repetition (Peh et al., 2011). Third, the selected depth may be okay for some participants, but outside of the abilities of others for either anatomical or psychological reasons. In extreme cases, this might inadvertently lead to a change in the movement and associated loading response, or potentially cause an injury by forcing the participant through a restricted ROM. While this could be screened out for a given joint angle or depth of choice through a pre-participation visit during recruitment using an orthopedic assessment of passive and active ROM, it may have limited the participant pool by placing additional constraints on ethics (e.g., hands-on human contact), sample bias (e.g., not representative of exercising population) and recruitment processes (e.g., additional voluntary visit required).

Instructing participants to perform the task using as large of a ROM as ability allows has been used previously in effort to represent a real world clinical setting (Dwyer et al., 2010; A. Khuu et al., 2016). Interestingly enough, Khuu et al. (2016) who also used SLS observed nearly identical peak knee joint angles with values of 79.2±8.7º compared to 80.1±12.1º observed within this study. While the variation was a little greater in the current study, it also included double the number of participants (half of which were men), while Khuu et al. (2016) only included 16 women as total participants in their study. The same trends of comparable peak joint angles were observed for the hip and ankle as well.

Even though the debate over a ROM control remains equivocal, a final consideration was given as a within-subject study design was used to compare the dependent variables between BULLETs. The single between-subject factor analyzed (i.e., sex) showed no significant differences between the knee joint angles of men and women. With the lack of consensus on the use of a standard joint angle or depth of ROM for control versus confounder in BULLET-type task research, future work is necessary to better understand the implications of standardizing ROM when measuring this type of kinetic data.

Studies involving kinetic data also have another limitation where consensus on methodology has not been reached. More often than not, kinetic data are normalized to body mass and/or size of the individual to account for anthropometric variability confounding the results (Moisio et al., 2003). As mentioned above, the size and shape of an individual can...
change the magnitudes observed as mass-inertial properties and segment geometries are used in
the computations of joint moments. However, some researchers have shed light on the fact that
normalization can change the nature of the relationships between variables examined (Davidson
et al., 2014; Robbins et al., 2011). While it may not be recommended for use in all cases, in our
case it does provide control for a confounding factor when trying to make inferences about sex-
based differences. As discussed in detail in the previous section, there are several limitations
when comparing biomechanical variables between men and women given a number of inherent
anthropometric differences between them. The normalization methods used herein are a widely
accepted, evidence-based procedure to account for a large portion of the variance attributed to
these differences (Moisio et al., 2003).

5.4 Future Directions

While numerous potential suggestions for further research have been provided in the
discussion of the results and limitations, emergent themes regarding future directions of this
research have appeared. First, it is clear that more exploration on the relationships between the
kinematics and the ‘internal’ kinetics of common exercise tasks is needed. While comparison of
the biomechanics between different exercise tasks is still lacking and would provide value to
practitioners, analyzing within-task differences in loading and how changing the kinematics of
the performance affects these loading conditions should be given equal attention. For example,
in certain cases utilizing a dual force plate setup to better understand how bilateral lower limb
load sharing is regulated would allow for greater apples-to-apples comparison between
exercises. With a few notable exceptions, understanding how the addition of external load and
velocity change the kinetic response to exercise tasks has been given minimal attention (S. P.
Flanagan & Salem, 2008; Frost et al., 2013; Gullett et al., 2009; Riemann et al., 2013).
Considering these are parameters that exercise professionals typically manipulate when
attempting to progress exercise, garnering a depth of knowledge will inform decision making.

Studies utilizing more advanced modelling to deduce individual muscle forces could also be
completed to provide more granular data, provided that appreciation is given to the practicality
and relevance for practitioners. For example, further efforts to bridge-the-gap between
biomechanics and motor behavior could advance understanding of how various instructions,
demonstrations, feedback and cueing affect the kinematics and kinetics of exercise tasks, which
would provide valuable and immediately actionable data for practitioners. While this has been
discussed previously regarding the pedagogy of exercise professionals (Ives & Knudsen, 2007), there remain many untapped opportunities for progress in this regard. Without a more in-depth understanding of the cause and effect of specific instruction and demonstration, it becomes difficult to analyze and make practical recommendations from the current body of research as the methodology is very inconsistent and undefined in this area.

Finally, given the vast amount of research on sex-based differences, the methodology related to controlling for confounding variables when answering questions about the biomechanics of exercise tasks is still unclear. The scientific community in human biomechanics would benefit from a consensus statement or unified approach to dealing with this problem. Until then, the common methodology of normalization to body-size and -mass, with its known issues and gaps included, will continue to prevail in kinetic analyses as it has for some time.

5.5 Conclusion

This was the first study to compare the peak and average joint moments of four different BULLETs and a DLS both within individuals and between sexes. Net hip and knee extension and ankle plantarflexion moment magnitudes significantly differed between BULLET tasks when normalized for body size and mass. These moments, when summated into single metric representing total lower limb loading known a support moment, also differed between BULLETs. Peak and average moment values showed a similar overall response, while the specific BULLET groupings by magnitude slightly differed between them. A consistent trend of rank-ordered magnitudes was observed for the knee, ankle and support moment, with the SLS producing the highest values. A slightly different trend was seen for the hip, with FL creating the highest moment magnitudes. The DLS task was the lowest magnitude observed for every joint in every task. The kinematic discrepancies are likely responsible for the kinetic outcomes given their interconnectedness and yet measured dissimilarity in performance on a joint-by-joint basis. The qualitatively similar lunge tasks always group together and/or adjacently, suggesting similarity among segment position plays a larger role than the addition of horizontal ROM when load, movement speed and individual anthropometrics are accounted for.

Sex-based differences were also observed in BULLET joint kinetics. The peak and average support moment, and the peak knee moment significantly differed between men and women. Given body size and mass were controlled for using common normalization methodology, other
factors are likely responsible for the resulting differences. However, consensus is not reached on best practice for assessing sex-based differences in kinetic variables, and a definitive mechanism responsible for these discrepancies cannot be provided.

Numerous practical applications are available from the outcomes of this study. Direct application for exercise practitioner utilizing a controlled progression to provide continuous adaptation to a resistance stimulus. The magnitude response and incremental increase in joint moment magnitudes from DLS to individual BULLETs is lower and smaller than those observed from added external load. This allows for smaller increases in joint or limb loading, while providing novice or elderly trainees, including individuals without access to equipment for additional external load a means to initiate and safely progress using these exercise tasks. For those using these tasks as part of movement screening and clinical assessments, based on lower limb moment magnitude the results would support the use of multiple BULLETs in addition to the DLS task, providing a simple and varied stimulus to expose any maladaptive movement behavior.

The health care and fitness industry rely on exercise as a primary prescriptive and assessment tool for stimulating various targeted adaptations. The more exercise tasks in general can be characterized for practitioners on the “internal” outcomes they cannot directly observe, the better informed they will be which, in turn, will add value to the intended application of BULLETs for targeted outcomes. Providing exercise professionals and clinicians with a more in-depth understanding of the kinetic and kinematic characteristics of exercise tasks through research will only aid to better help the individuals who are given exercise as an agent for change.
References


Appendices

A. Informed Consent

Study Title
A biomechanically derived progression for bodyweight unilateral lower limb exercise tasks

Background and Study Purpose
Single-leg squatting tasks are commonly used in exercise to create specific adaptations including increasing lower body strength and muscle size. Furthermore, these tasks are included in clinical tests to try and determine an individual’s risk of getting injured. The goal of this study is to compare muscular demands between five different single-leg exercises and attempt to create a progression protocol. Exercise professionals and physical therapists need this information to develop safe and effective fitness and rehabilitation programs. This study has been reviewed and approved by the University of Toronto Office of Research Ethics.

Testing Procedures
As a participant in this study, you will attend one 60-minute testing session in the Musculoskeletal Biomechanics and Injury Prevention Laboratory at the University of Toronto. The session will be scheduled at a time that is convenient for you. During the testing session you will be asked to perform a short warm-up and five repetitions of five different single-leg exercises. Your body movements will be recorded using reflective markers taped to your skin, video will be recorded using digital cameras, and a force plate will be used to measure the forces you apply to the ground when performing the exercises. Since the markers are taped to your skin, it is required that women wear a sports brassiere and men are shirtless. Both men and women are required to wear spandex bottoms and bare feet.

Risks and Benefits
The exercises you will be asked to perform in this study are considered low-risk for a healthy, active population. However, it is possible that some participants will experience muscular discomfort or soreness as a result of performing each exercise. You will be compensated $20 for completion of the 60-minute data collection session. Although there are no immediate physical benefits to you, your participation may aid in the development of exercise safety, movement screening and injury prevention tools. This research will also contribute to the body of knowledge in order to improve the quality of service provided by exercise professionals and clinicians.
Confidentiality and Privacy of Information

If you agree to participate, a unique identification code will be assigned to you. Your data will be linked with this code, but your personal information will not be linked with the data. Only the supervisor will have access to the names of participants and will keep this information locked in a personal filing cabinet for a minimum of 5 years. The data may be used in the future to design new experiments, but your name will not be included so that research assistants will not be able to link your personal information with the identification code. If you do not wish for your data to be retained after the completion of the current study, please indicate this below:

☐ If I agree to participate, only use my data for the current study
☐ If I agree to participate, allow my data to be used in this study and to design future studies
☐ If I agree to participate, my video can be used for information dissemination and knowledge transfer (i.e., presentations).

Participation

You are eligible to participate if you are a healthy, active student, at least 18 years of age, who for the previous three months, exercises at a minimum of three times per week. Your inclusion in the study also requires the absence of musculoskeletal injury and pain, for the preceding three months and upon completing the required squatting tasks. As a participant in this study, you have the right to refuse to participate and to withdraw your consent at any time without penalty. To do so, indicate this to the investigators by saying: “I no longer wish to participate in this study”. Your participation is voluntary. Upon completion of the 60-minute testing session you will receive $20 compensation for your time. If you withdraw after the testing session you will retain your remuneration.
Inquiries

If you have any further questions or would like to receive more information about this study, please contact the supervisor. If you have questions about your right as a research participant, please contact the Office of Research Ethics.

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Consent to Participate

By signing below, I confirm that:

- I have read this information form and understand the potential risks and benefits associated with participation in this study, and
- I have had the opportunity to ask questions about this study and received satisfactory answers to my questions, and
- I am aware that I can withdraw from the study without adverse consequences at any time by advising the investigators of this decision, and
- I agree to participate in this study:

_________________________________  ___________________________  _____________________
Participant Name  Participant Signature  Date
(Please Print)

_________________________________  ___________________________  _____________________
Witness Name  Witness Signature  Date
(Please Print)
B. Physical Activity Readiness Questionnaire (PAR-Q)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

<table>
<thead>
<tr>
<th>YES</th>
<th>NO</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?</td>
<td></td>
</tr>
<tr>
<td>2. Do you feel pain in your chest when you do physical activity?</td>
<td></td>
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<tr>
<td>3. In the past month, have you had chest pain when you were not doing physical activity?</td>
<td></td>
</tr>
<tr>
<td>4. Do you lose your balance because of dizziness or do you ever lose consciousness?</td>
<td></td>
</tr>
<tr>
<td>5. Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?</td>
<td></td>
</tr>
<tr>
<td>6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?</td>
<td></td>
</tr>
<tr>
<td>7. Do you know of any other reason why you should not do physical activity?</td>
<td></td>
</tr>
</tbody>
</table>

**YES to one or more questions**

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

**NO to all questions**

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:
- start becoming much more physically active — begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal — this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:
- start becoming much more physically active — begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal — this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

**PLEASE NOTE:** If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

**NO changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.**

**Note:** If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

**NAME**

**SIGNATURE**

**SIGNATURE OF PARENT/ GUARDIAN:**

**DATE**

**WITNESS**

**Note:** This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.
C. Images of Exercise Tasks (BULLETs)

Bilateral Squat (BS)

Unilateral Squat (US)

Forward Lunge (FL)
Backward Lunge (BL)

Split Squat (SS)
D. Verbal Instructions for Exercise Tasks

Task: Single-Leg Squat (SLS)

1) Initial Setup:
   - You will begin in a tall standing position, feet flat on the ground with your dominant leg on the force plate and non-dominant leg off of the force plate with feet shoulder width apart.
   - You will place your hands on your hips and they will remain there for the duration of single leg squat task.
   - You will lift the non-dominant foot off of the ground and it will remain in the air for the duration of the exercise tasks. You will keep this foot directly beside your dominant leg. Your foot cannot touch the ground or the repetition will not count and another repetition will be repeated.
   - For the duration of the task your dominant leg foot will remain flat on the force plate and you will attempt to maintain a neutral spine with your head and chest facing forward.

2) Movement:
   - You will start the single leg squat movement on the first beep sound.
   - To do this you will bend/flex your hip, knee and ankle to lower yourself towards the ground.
   - You will have 2 seconds to lower your to the bottom position. This means move as low as you can go while maintaining control of your body. Do not go beyond a range that you can comfortably control.
   - You will attempt to be at your lowest point, the bottom of the squat on the next (second) beep sound.
   - When you reach the bottom of your squat, you will proceed to straighten/extend your hip, knee and ankle to stand back up returning to the starting position.
   - The goal will be to complete the squat and return to standing on the third beep sound.
   - You will have a two second pause prior to starting the next single-leg squat on the following (fourth) beep sound.
   - To summarize you will have 2 seconds to squat down, 2 seconds to stand up and 2 seconds to pause before you repeat the single leg squat again.
   - You will repeat this task sequence five times.
Task: Double-Leg Squat (DLS)

1) Initial Setup:
   - You will begin in a tall standing position, both feet flat on the ground with your dominant leg on the force plate and non-dominant leg off of the force plate with feet shoulder width apart.
   - You will place your hands on your hips and they will remain there for the duration of double-leg squat task.
   - For the duration of the task both feet will remain flat on the ground and you will attempt to maintain a neutral spine with your head and chest facing forward.

2) Movement:
   - You will start the double-leg squat movement on the first beep sound.
   - To do this you will bend/flex your hips, knees and ankles to lower yourself towards the ground.
   - You will have 2 seconds to lower your to the bottom position. This means move as low as you can go while maintaining control of your body. Do not go beyond a range that you can comfortably control.
   - You will attempt to be at your lowest point, the bottom of the squat on the next (second) beep sound.
   - When you reach the bottom of your squat, you will proceed to straighten/extend your hips, knees and ankles to stand up returning to the starting position.
   - The goal will be to complete the squat and return to standing on the third beep sound.
   - You will have a 2 second pause prior to starting the next double-leg squat on the following (fourth) beep sound.
   - To summarize you will have 2 seconds to squat down, 2 seconds to stand up and 2 seconds to pause before you repeat the double leg squat task again.
   - You will repeat this this task sequence five times.
**Task: Split Squat (SS)**

1) **Initial Setup:**
   - You will begin moving to the initial position by taking a large step forward with your dominant leg, a similar distance to the lunge step length.
   - Your front foot should be in front of your body, flat on the force plate and your back foot should be behind your body and off of the force plate.
   - Both knees are completely straight in the starting and final positions, so the task starts and ends with both knees fully extended at the top of the motion.
   - Your back foot is allowed to come onto your toes during this task.
   - You will place your hands on your hips and they will remain there for the duration of split squat.
   - For the duration of this task both feet will remain in contact with the ground and you will attempt to maintain an upright trunk with chest and head facing forward.

2) **Movement:**
   - You will start the split squat task on the first beep sound.
   - To do this you will bend/flex both knees to lower yourself downward towards the ground.
   - You will have 2 seconds to lower your to the bottom position. This means move as low as you can, bringing your rear knee as close to the ground as possible while maintaining control of your body, but do not go beyond a range that you can comfortably control.
   - You will attempt to be at your lowest point, the bottom of the split squat on the next (second) beep sound.
   - You will attempt to be at this bottom position in the split squat on the next (second) beep sound.
   - When you reach the bottom of your split squat, you will straighten/extend both of your knees to stand back up returning to the starting position.
   - The goal will be to complete the split squat movement on the third beep sound.
   - You will have a 2 second pause prior to starting the next split squat on the following (fourth) beep sound.
   - To summarize you will have 2 seconds to squat down, 2 seconds to stand up and 2 seconds to pause before you repeat the single leg squat again.
   - You will repeat this task sequence five times.
**Task: Forward Lunge (FL)**

1) **Initial Setup:**
   - You will begin in a tall standing position, feet flat on the ground standing on the tape behind the force plate.
   - Your dominant leg should be in-line with the force plate in front of it.
   - For the duration of the task your rear foot will remain in contact with the ground and it is allowed to come onto your toes during this task.
   - For the duration of the task you will attempt to maintain an upright trunk with chest and head facing forward.
   - You will place your hands on your hips and they will remain there for the duration of forward lunge.

2) **Movement**
   - You will start the forward lunge on the first beep sound.
   - To do this you will start by taking a large lunging step forward, moving your body forward while maintaining the rear foot in contact with the ground.
   - Your dominant foot will leave the ground and remain airborne until it makes contact with the force plate. Once it makes contact with the ground you will then focus on lowering your body towards the ground.
   - You will have 2 seconds from the start of the forward lunge movement to lower yourself to the bottom position. This means move as low as you can, bringing your rear knee as close to the ground as possible while maintaining control of your body, but do not go beyond a range that you can comfortably control.
   - You will attempt to be at your lowest point, the bottom position of the forward lunge on the next (second) beep sound.
   - When you reach the bottom of your lunge, you will straighten/extend your front hip, knee and ankle to stand back up returning to the starting position.
   - The goal will be to complete the forward lunge movement on the third beep sound.
   - You will have a 2 second pause prior to starting the next forward lunge on the following (fourth) beep sound.
   - This means you will have 2 seconds to squat down, 2 seconds to stand up and 2 seconds to pause before you repeat the forward lunge task again.
   - You will repeat this task sequence five times.
Task: Backward Lunge (BL)

1) Initial Setup:
   - You will begin in a tall standing position, feet flat on the ground. Your dominant leg will be standing in the middle of the force place and your non-dominant leg shoulder width apart off of the force place.
   - For the duration of the task your front foot will remain in contact with the ground.
   - Your rear foot is allowed to come onto your toes during this task.
   - For the duration of the task you will attempt to maintain an upright trunk with chest and head facing forward.
   - You will place your hands on your hips and they will remain there for the duration of backward lunge task.

2) Movement
   - You will start the backward lunge on the first beep sound.
   - To do this you will start by taking a large lunging step backward with your non-dominant leg, moving your body backward while maintaining the dominant foot in contact with the force plate.
   - Your non-dominant foot will leave the ground and remain airborne until it makes contact with the ground behind you. Once it makes contact with the ground you will then focus on lowering your body further towards the ground.
   - You will have 2 seconds from the start of the backward lunge movement to lower yourself to the bottom position. This means move as low as you can, bringing your rear knee as close to the ground as possible while maintaining control of your body, but do not go beyond a range that you can comfortably control.
   - You will attempt to be at your lowest point, the bottom position of the backward lunge on the next (second) beep sound.
   - When you reach the bottom of your lunge, you will straighten/extend your front hip, knee and ankle to stand back up returning to the starting position.
   - The goal will be to complete the backward lunge movement on the third beep sound.
   - You will have a 2 second pause prior to starting the next forward lunge on the following (fourth) beep sound.
   - To summarize you will have 2 seconds to squat down, 2 seconds to stand up and 2 seconds to pause before you repeat the forward lunge task again.
   - You will repeat this task sequence five times.
### Table 2. Peak joint flexion angles and ranges of motion for the hip, knee, ankle and trunk during the four BULLET conditions (SLS = single leg squat; SS = split squat; BL = backward lunge; FL = forward lunge) and a baseline double leg squat task (DLS).

Data included are the mean (standard deviation) values calculated across all participants (N = 32) as no SEX*TASK interaction and no main effect for SEX was detected from two-way ANOVAs computed for each joint individually (p > 0.05). Superscript lowercase letter indicates post-hoc results as a main effect for TASK was found for each joint individually (p < 0.05).

<table>
<thead>
<tr>
<th>TASK</th>
<th>Hip</th>
<th>Knee</th>
<th>Ankle</th>
<th>Trunk</th>
<th>Hip</th>
<th>Knee</th>
<th>Ankle</th>
<th>Trunk</th>
</tr>
</thead>
<tbody>
<tr>
<td>DLS</td>
<td>81.9(11.6)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>99.4(16.2)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>29.0(6.4)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>33.8(8.5)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>75.5(11.8)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>96.4(16.2)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>30.2(5.7)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>34.0(8.3)&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>SLS</td>
<td>53.8(12.4)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>80.1(12.1)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>29.8(5.9)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>28.2(11.0)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>47.6(13.0)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>71.7(13.8)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>27.7(5.9)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>24.6(11.3)&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>SS</td>
<td>76.6(9.2)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>104.0(9.9)&lt;sup&gt;ac&lt;/sup&gt;</td>
<td>20.7(6.9)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>7.0(5.9)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>42.6(7.7)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>95.3(12.5)&lt;sup&gt;d&lt;/sup&gt;</td>
<td>43.5(9.9)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>6.1(2.2)&lt;sup&gt;c&lt;/sup&gt;</td>
</tr>
<tr>
<td>BL</td>
<td>80.3(7.2)&lt;sup&gt;ac&lt;/sup&gt;</td>
<td>101.8(11.0)&lt;sup&gt;ac&lt;/sup&gt;</td>
<td>19.5(6.5)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>17.2(6.8)&lt;sup&gt;d&lt;/sup&gt;</td>
<td>76.4(8.1)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>96.9(12.4)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>18.9(6.5)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>16.8(5.1)&lt;sup&gt;d&lt;/sup&gt;</td>
</tr>
<tr>
<td>FL</td>
<td>77.3(8.8)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>107.4(9.0)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>22.3(6.0)&lt;sup&gt;b&lt;/sup&gt;</td>
<td>8.3(6.3)&lt;sup&gt;c&lt;/sup&gt;</td>
<td>47.4(7.8)&lt;sup&gt;bc&lt;/sup&gt;</td>
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<td>37.0(13.1)&lt;sup&gt;d&lt;/sup&gt;</td>
<td>15.5(4.4)&lt;sup&gt;d&lt;/sup&gt;</td>
</tr>
</tbody>
</table>

*p-value* .0001 .0001 .0001 .0001 *p-value* .0001 .0001 .0001 .0001
### Table 3. Peak and average net joint extension/plantarflexion moments for the hip, knee, ankle and total support during the four BULLET conditions (SLS = single-leg squat; SS = split squat; BL = backward lunge; FL = forward lunge) and a baseline double-leg squat task (DLS). Data are normalized to body-mass (N∙m/kg) and included are the mean (standard deviation) values calculated for men (N = 16), women (N = 16), and all participants combined (N = 32).

<table>
<thead>
<tr>
<th>TASK</th>
<th>Hip (Men N = 16)</th>
<th>Knee (Men N = 16)</th>
<th>Ankle (Men N = 16)</th>
<th>Support (Men N = 16)</th>
</tr>
</thead>
<tbody>
<tr>
<td>DLS</td>
<td>-0.36 (0.07)</td>
<td>0.63 (0.12)</td>
<td>-0.15 (0.04)</td>
<td>1.09 (0.11)</td>
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<tr>
<td>SLS</td>
<td>-0.55 (0.23)</td>
<td>0.97 (0.21)</td>
<td>-0.68 (0.13)</td>
<td>2.12 (0.26)</td>
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<tr>
<td>SS</td>
<td>-0.63 (0.16)</td>
<td>0.74 (0.09)</td>
<td>-0.31 (0.10)</td>
<td>1.54 (0.24)</td>
</tr>
<tr>
<td>BL</td>
<td>-0.68 (0.15)</td>
<td>0.75 (0.14)</td>
<td>-0.34 (0.09)</td>
<td>1.59 (0.27)</td>
</tr>
<tr>
<td>FL</td>
<td>-0.85 (0.18)</td>
<td>0.90 (0.14)</td>
<td>-0.42 (0.14)</td>
<td>2.10 (0.35)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th></th>
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<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>DLS</td>
<td>-0.27 (0.06)</td>
<td>0.59 (0.08)</td>
<td>-0.17 (0.05)</td>
<td>0.93 (0.13)</td>
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<tr>
<td>SLS</td>
<td>-0.42 (0.19)</td>
<td>0.89 (0.14)</td>
<td>-0.66 (0.11)</td>
<td>1.80 (0.24)</td>
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<tr>
<td>SS</td>
<td>-0.60 (0.20)</td>
<td>0.63 (0.12)</td>
<td>-0.26 (0.11)</td>
<td>1.32 (0.24)</td>
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<tr>
<td>BL</td>
<td>-0.64 (0.17)</td>
<td>0.65 (0.13)</td>
<td>-0.31 (0.10)</td>
<td>1.33 (0.38)</td>
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<tr>
<td>FL</td>
<td>-0.82 (0.22)</td>
<td>0.76 (0.14)</td>
<td>-0.38 (0.14)</td>
<td>1.90 (0.03)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>TASK</th>
<th>Hip (Combined N = 32)</th>
<th>Knee (Combined N = 32)</th>
<th>Ankle (Combined N = 32)</th>
<th>Support (Combined N = 32)</th>
</tr>
</thead>
<tbody>
<tr>
<td>DLS</td>
<td>-0.32 (0.08)</td>
<td>0.61 (0.10)</td>
<td>-0.16 (0.05)</td>
<td>1.01 (0.15)</td>
</tr>
<tr>
<td>SLS</td>
<td>-0.49 (0.22)</td>
<td>0.93 (0.19)</td>
<td>-0.67 (0.12)</td>
<td>1.96 (0.30)</td>
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<tr>
<td>SS</td>
<td>-0.62 (0.18)</td>
<td>0.68 (0.12)</td>
<td>-0.29 (0.11)</td>
<td>1.43 (0.26)</td>
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<tr>
<td>BL</td>
<td>-0.66 (0.16)</td>
<td>0.70 (0.14)</td>
<td>-0.33 (0.10)</td>
<td>1.46 (0.30)</td>
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<tr>
<td>FL</td>
<td>-0.83 (0.20)</td>
<td>0.83 (0.15)</td>
<td>-0.40 (0.14)</td>
<td>2.00 (0.33)</td>
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